The *Rectus Femoris* Muscle Fatigue through Mechanomyographic Time-Frequency Response in Paraplegic Subject
Preliminary Results

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Abstract: The purpose of this study is the evaluation of mechanomyographic (MMG) time-frequency response of rectus femoris muscle of a paraplegic subject during an isometric electrically-elapsed fatigue protocol. An accelerometer sensor was used to measure the vibration of muscle during voltage-controlled functional electrical stimulation application at 1 kHz pulse frequency (20% duty cycle) 70 Hz modulated frequency (20% duty cycle). A load cell (50 kgf) measured the force signal with the participant seated on a bench with the hip and knee angle set to 90º. During the protocol the electrical output voltage was adjusted to keep the force at 30% of maximal stimulated contraction (MSC). When the electrical stimulation was unable to keep the force above approximately 10% of MSC the protocol was ceased. Ten seconds with unfatigued (initial period) and fatigued (final period) muscle, MMG signal was processed with Cauchy wavelet transformation (bandpass 5-100 Hz). For fatigue conditions of paraplegic subject, MMG signal presents concentration energy to lower frequencies mainly to 11.31 Hz band frequency.

1 INTRODUCTION

Muscular fatigue produces electromechanical modifications on neuromuscular tissue as the incapacity to generate force required in contraction (Cè et al., 2013). Muscular fatigue of people with complete spinal cord injury (SCI) can be detected by using force information (Gerrits et al., 1999); however, to elicit the muscular contraction the main way is applying the functional electrical stimulation (FES).

Unfortunately, the stimulating electrical current produces interference on the acquisition of feedback signals such as electromyography (Faller et al., 2009) that registers the bioelectrical signal of neuromuscular tissue. Alternative way to acquire the muscular response of paraplegics with complete lesion is by means of mechanomyography (MMG) (Nogueira-Neto et al., 2013), which is a technique based on mechanical oscillations of muscles during contraction. So, it is immune to electrical interference yielded during FES application (Seki et al., 2003).

The goal of the research discussed in this paper is to verify if there is any frequency band of time-frequency response that has higher energy in *rectus femoris* muscle of a paraplegic subject during an isometric electrically-elapsed fatigue protocol to improve the knowledge of muscle physiology.

2 METHODS

2.1 Volunteer

This study was approved by Human Research Ethics Committee of Pontifícia Universidade Católica do Paraná (PUCPR) under register 2416/08. The participant involved in the research was 38 years old, weight: 88 kg, height: 179 cm and diagnosed with incomplete spinal cord injury at T3-T4 radicular level acquired twenty years earlier by gunshot. The left lower limb of the volunteer underwent a physical evaluation classifying it as level B (without voluntary contraction) in accordance with the American Spinal Injury

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Association (AIS) impairment scale (from A to E). In the day of test, the volunteer did not use any drug that could change his motor condition.

2.2 Electrical Stimulation

The custom electrical stimulator produced monophasic rectangular wave with amplitude controlled stimulation pulses. The parameters configured were pulse frequency: 1 kHz (20% duty cycle); burst (modulating) frequency: 70 Hz (20% duty cycle).

2.3 Instrumentation

The MMG system used had a triaxial accelerometer sensor (Freescale® MMA7260Q MEMS with 800 mV/G sensitivity at 1.5 gravitational acceleration - 13x18mm, 0.94 g) and the Z-axis (transverse to the muscle belly displacement) was used as the signal process. The electronic circuits allowed 2.2x amplification. A load cell (50 kgf ≈ 500 N), of S shape aluminium body, with four strain gages in full Wheatstone bridge, was used to measure force. After the skin preparation (trichotomy and cleaning), MMG sensor was positioned over the belly of rectus femoris (RF) muscle using double-sided adhesive tape. The placement was equidistant between the anterosuperior iliac spine and the top of the patella. The self-adhesive electrodes with different sizes were positioned on the thigh over the knee region (anode with 5 x 9 cm) and over the femoral triangle (cathode with 5 x 5 cm) to stimulate the quadriceps muscle via femoral nerve.

2.4 Experimental Protocol

The volunteer was positioned seated on a bench with the hip and knee angle set to 90º. After the placement of the FES electrodes on the left limb a minimum of 10 min rest was respected to skin-electrode impedance balance (Reilly, 1992). The MMG sensor was positioned over the *rectus femoris* muscle belly. The load cell was attached on the leg through band strips and a Velcro strap belt was positioned in trunk for volunteer stabilization. The instrumentation layout is illustrated in Figure 1.

Firstly, the maximal stimulated contraction (MSC) force was performed by increasing the electrical stimulating magnitude (approximately 3 V/s to avoid motoneuron adaptation/habituation) until the force started to level off. Twenty-minutes rest interval was respected to avoid interference on fatigue protocol. The protocol was initiated with 0V of electrical stimulating, them it was controlled to keep the force in 30% of MSC (30MSC). When the electrical stimulating was unable to keep the force over 30% of 30 MSC (approximately 10% of MSC) the protocol was ceased (Gerrits et al., 1999).

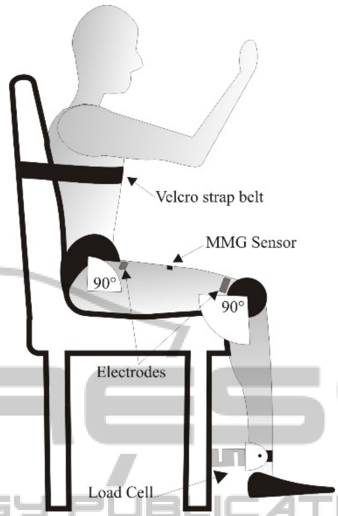

Figure 1: Experimental set up and instrumentation layout.

2.5 Data Processing and Analysis

The signal process was realized by custom-written MatLab® software version R2008a. Ten seconds window was chosen at initial instant when the 30MSV was reached (unfatigued muscle). Ten seconds window was chosen at final instant before 30% of 30MSV force (fatigued muscle). The MMG signal was processed with a third-order Butterworth filter with bandpass of 5-100 Hz (spectral range of muscular vibration). The signal was processed in eleven bands of Cauchy wavelet (CaW) transform (von Tscharner, 2000) (2.07, 5.79, 11.31, 18.63, 27.71, 38.54, 51.12, 65.42, 81.45, 99.19 and 118.63 Hz) and the root mean square (analysis window length: 1s) was computed for each CaW band. Data were normalized by the first second at the initial moment to each frequency band in order to show signal variations.

3 RESULTS

The volunteer did not show any spastic event during the protocol application. In order to reach the MSC force (31.5 kgf), the output stimulator voltage was set to 190 V. The green line in Figure 2 shows the target force (30% MSC), the blue line indicates the threshold of 10% MSC and the red line indicates the response of the force obtained during the protocol. After 50 s of applying the electrical stimuli, one may observe the inability of the muscle to sustain its strength in the target area even with the increase of the electrical stimulation.

Figure 3 shows the MMG time-frequency response: at first stage (first 10 s – A and B) two frequency bands are prominent, at 11.31 Hz and at 65.42 Hz; the latter one is probably due to myofibers oscillation in accordance with burst frequency (70 Hz – tetanic frequency). At last stage (last 10 s – C and D) under fatigue conditions, the frequency band around 11.31 Hz shows an enhancement in energy concentration (seen as higher intensity value) near the final setpoint protocol.

Figure 4 illustrates normalized data. At first stage (A and B), the lower frequency components show greater energy content (from 5.79 to 11.31 Hz); at final stage (C and D), under fatigue conditions, the same components (from 5.79 to 11.31 Hz) also show greater energy concentration near the final set point protocol. The frequency components around 65.42 Hz in normalized data do not show variation along the protocol, possibly because few myofibers achieve oscillation in this vibration frequency.

4 DISCUSSION

At initial stage (unfatigued), the frequency components around 65.42 Hz show content concentration, as well as the band at 11.31 Hz, probably due to myofibers oscillation in accordance

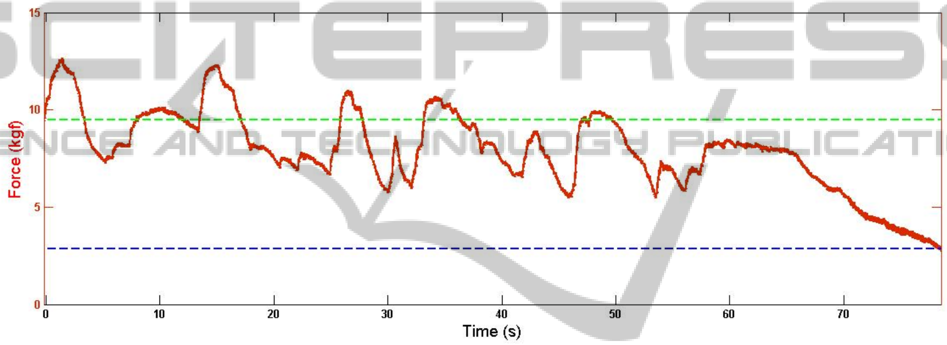

Figure 2: Force (red line) during the protocol. Force goal control (green line), inferior force threshold (blue line).

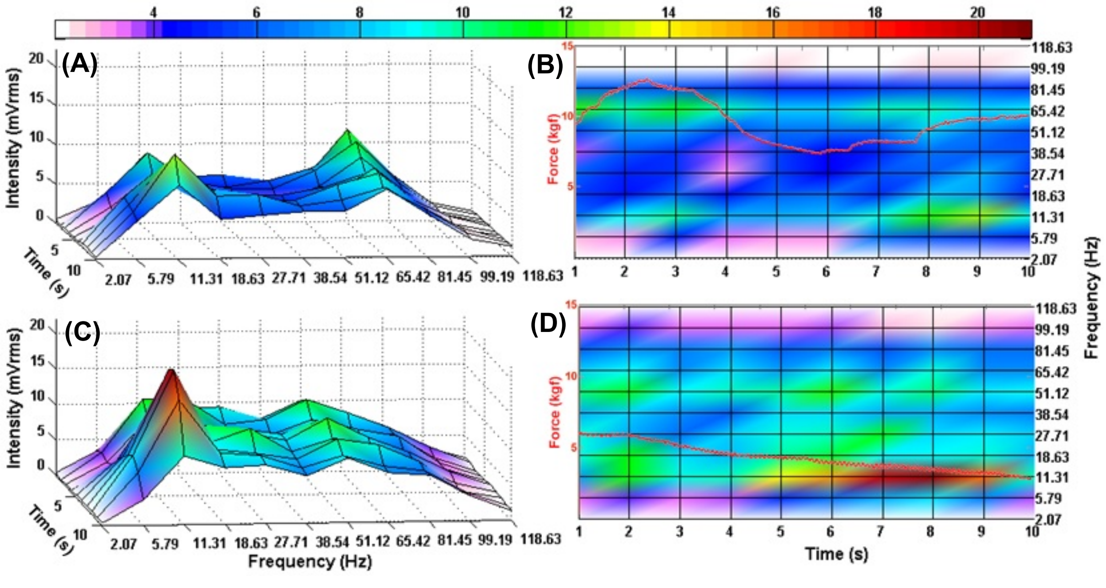

Figure 3: 3D (A and C) and 2D (B and D) MMG time-frequency responses of rectus femoris muscle. A and B: ten initial seconds of FES-force-controlled application (unfatigued muscle). C and D: ten final seconds of FES-force-controlled application (fatigued muscle). The color bar shows the maximum value (21 mVrms) in dark red color and the values below 10% of maximum (0.21 mVrms) are in white color. Force: red line in 2D image.

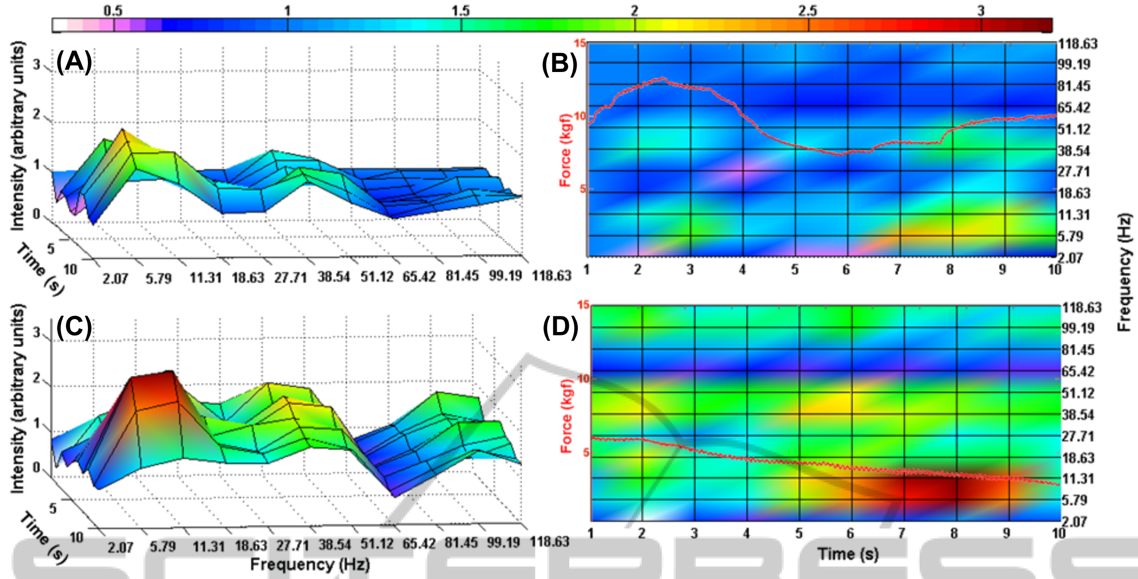

Figure 4: Normalized 3D (A and C) and 2D (B and D) MMG time-frequency responses of rectus femoris muscle. Data normalized by the first second. A and B: ten initial seconds of FES-force-controlled application (unfatigued muscle). C and D: ten final seconds of FES-force-controlled application (fatigued muscle). The color bar shows the maximum value (3.2 normalized data) in dark red color and the values below 10% of maximum (0.32 normalized data) are in white color. Force: red line in 2D image.

with modulated frequency (70 Hz tetanic frequency). The same event occurs when modulated frequency is set to 50 Hz. At this situation the signal shows a peak energy at the range of 51.12 Hz and other peak at 11.31 Hz (Krueger et al., 2013b) as in our previous study. These events on signal are not due to FES interference, because the electrical stimulator intensity is increased along the protocol application and the energy variation around 65.42 Hz keeps practically the same content (normalized data in Figure 4).

At final stage (fatigued), both Figures 3 and 4 indicate the increase of energy at lower frequencies (mainly 11.31 Hz) when the muscular fatigue occurs (Tarata, 2003; Gandevia et al., 1995). Using the FES-control based on knee joint angle rather than on force the result is the same (Krueger et al., 2013a). This lower frequency energy concentration may be due to motor units coherence (Yao et al., 2000) when myofibers vibrate in phase during contraction in fatigue state. Moreover, the increased energy for low frequency bands in the last seconds of protocol may be associated with a decrease in muscle fibre conduction velocity, that is a peripheral factor, related to the occurrence of muscle fatigue as occurs with able-bodied subjects (Schillings et al., 2003).

5 CONCLUSIONS

During fatigue conditions the mechanomyographic signal of rectus femoris muscle from a paraplegic subject presents a greater energy concentration at lower frequencies (mainly 11.31 Hz band frequency). This fatigue information in paraplegic subject (without voluntary contraction) is helpful to knowledge's improvement regarding muscular physiology after spinal cord injury.

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