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Correlation between spectral and temporal mechanomyography features during functional electrical stimulation

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Abstract **Introduction**: Signal analysis involves time and/or frequency domains, and correlations are described in the literature for voluntary contractions. However, there are few studies about those correlations using mechanomyography (MMG) response during functional electrical stimulation (FES) elicited contractions in spinal cord injured subjects. This study aimed to determine the correlation between spectral and temporal MMG features during FES application to healthy (HV) and spinal cord injured volunteers (SCIV). **Methods**: Twenty volunteers participated in the research divided in two groups: HV (N=10) and SCIV (N=10). The protocol consisted of four FES profiles transcutaneously applied to *quadriceps femoris* muscle via femoral nerve. Each application produced a sustained knee extension greater than 65° up to 2 min without adjusting FES intensity. The investigation involved the correlation between MMG signal root mean square (RMS) and mean frequency (MF). **Results**: HV and SCIV indicated that MMG_{RMS} and MMG_{MF} variations were inversely related with $-0.12 \geq r \geq -0.82$. The dispersion between MMG_{MF} and MMG_{RMS} reached $0.50 \leq r^2 \leq 0.64$. **Conclusion**: The increase in MMG_{RMS} and the decrease in MMG_{MF} may be explained by the motor units coherence during fatigue state or by motor neuron adaptation (habituation) along FES application (without modification on parameters).

Keywords: Functional electrical stimulation, Mechanomyography, Muscular physiology.

Introduction

From 1960 on, people with spinal cord injury or some other types of movement disorders have been benefiting from artificial muscle contraction elicited by functional electrical stimulation (FES) (Kesar et al., 2010). One of the available techniques to monitor the contractions evoked by FES is electromyography (EMG). However, as a drawback, the electrical pulses of FES may cause interference on electromyographic signals due to electronic limitations (Seki et al., 2003). Alternatively, mechanomyography (MMG) allows the measurement of mechanical oscillations produced by muscle contraction without electromagnetic interference yielded by FES (Krueger et al., 2014). In this sense, MMG signals may also be applied in clinical settings (Cè et al., 2015) to control myoelectrical prostheses, orthoses (Prociow et al., 2008) or neuroprostheses (Chen et al., 2016; Popovic and Thrasher, 2004). Therefore, MMG enables the evaluation of FES-induced muscle contraction.

MMG has been employed in several areas, including those in which the EMG is already consolidated, but emphasizing voluntary contraction. Generally, the signal analysis methods involve the time and frequency domains or both. As an example, Tarata (2003) correlated the MMG signal between time and frequency domains concluding that MMG_{RMS} and MMG frequency responses present a negative correlation along time for voluntary contractions.

Despite the feasibility of MMG for the evaluation of FES-induced muscle contraction there are few studies in the literature about MMG response during FES elicited contractions in spinal cord injured subjects. An investigation of muscle activity caused by FES application to people with spinal cord injury can be achieved using different MMG parameters.

Hence, the goal of this study is to correlate spectral and temporal MMG features during FES application to healthy and spinal cord injured volunteers.

Methods

Subjects

This study received the approval of the Pontifícia Universidade Católica do Paraná (PUCPR) Human Research Ethics Committee under register number 2416/08 according to the Helsinki Declaration of 1975 as revised in 1983. The participants signed a consent form to take part in the study. The experimental study was carried out in the Rehabilitation Engineering Laboratory without temperature control.

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Ten healthy volunteers (HV) without neurological or orthopaedic disorders (28.30 ± 6.58 yrs) and sixteen spinal cord injured volunteers (SCIV) (32.06 ± 9.68 yrs) were chosen to participate in the study. All participants were male. Before the application of the first stimulation protocol, the SCIV underwent medical examination so as to verify the inclusion/exclusion criteria. The inclusive criteria were clinical stability after spinal cord injury, no metallic implant in thigh or neoplastic tissue. Six SCIV were excluded from the initial group of volunteers, once they either did not tolerate the sensation caused by the application of electrical current or due to denervation of motor units. Table 1 presents the participants demography and motor response parameters reflecting their neuromuscular condition. Temperature and relative humidity in the research room were 22.9 ± 3.7 °C and 62 ± 5.9%, respectively, during the application of protocols.

Sensors

The developed MMG instrumentation included sensors which were built using Freescale MMA7260Q MEMS triaxial accelerometers with sensitivity set to 800 mV/G at 1.5 G (G: gravitational acceleration). MMG signals are bipolar and therefore, symmetric-voltage source supplied the electronic circuits that amplified the MMG signals by 10x, whereas a 4-40Hz Butterworth third order filter conditioned their spectral content. The system acquired and processed the signals via Data Translation™ acquisition board and National Instruments™ LabVIEW™ program. The sampling rate was 1 kHz.

Electrical stimulation and sensors layout

The stimulatory current was a monophasic rectangular wave of 1 kHz pulse frequency (10% duty cycle) and 50 Hz burst frequency (15% duty cycle). After trichotomy and skin cleaning, self-adhesive electrodes (5×9 cm) were positioned over the knee region (anode) and over the femoral triangle (cathode) to stimulate the *quadriceps* muscle. The MMG sensors were positioned on the *rectus femoris* (RF) and the *vastus lateralis* (VL) muscle bellies. A single axis electrogoniometer acquired the knee joint angle data.

Research design

The volunteers were seated on an adapted chair with the hip and knee angles set to 70º (Matsunaga et al., 1999) and 90º, respectively. For each volunteer, the intensity set to the electrical stimuli was determined experimentally in an individual basis, and consisted of the smallest voltage required to make the limb move and cause the knee angles to vary from ~90º to ~40º. To avoid muscle damage, the rest periods were adjusted to 2 and 5 min, respectively, for HV and SCIV. Four stimulations were performed and the movements were artificially elicited by FES (with 5 s rise). Two criteria were established to cause each contraction to end: time limit (until 120 s) in case the knee angle was always

Table 1. Demography of spinal cord injured volunteers.

Vol	Age	Spinal cord injury				Sensibility	Motor system			Deliverance	
		Etiology	Com	Inc	Months	L1-L2	Force	Reflex	Spasticity	A	NA
A	25	Gunshot	T8		24	-	0	0	0		X
B	46	Automobilist	T8		31	-	0	2	0	X	
C	30	Gunshot	T6		84	-	0	2	+1	X	
D	28	Automobilist		T12	48	+	0	1	0	X	
E	29	Automobilist		T12	108	+	1	3	2		X
F*	26	Gunshot	T10-11		168	-	0	2	+1		X
G	34	Automobilist		T4-5	84	-	0	2	1	X	
H	24	Gunshot	T12		24	-	0	0	0		X
I	25	Automobilist	T12		18	-	0	0	0		X
J	37	Diving		C5-6	162	-	0	3	1	X	
K	19	Gunshot		T10	12	+	0	3	1	X	
L	48	Fall		T11	60	+	3	3	2	X	
M	52	Other		L4	60	+	4	2	0		X
N	26	Automobilist		C6-7	28	-	0	3	2	X	
O	28	Automobilist	T3		60	-	0	0	1	X	
P	36	Other		L1	132	+	4	2	0	X	

Vol: volunteer; Com: complete; Inc: incomplete, nociceptive sensitivity; "-" absent; "+" present; Force: Force scale [0 (absent muscular tonus) to 5 (normal force)] (Cipriano, 2003); Reflex: Wexler scale [0 (absent reflex) to 5 (sustained clonus)] (Cipriano, 2003); Spasticity: Ashworth modified scale [0 (without spasticity tonus) to 4 (rigid affected part)] (Bohannon and Smith, 1987); A: accepted in this research; NA: not accepted in this research. *withdrew from the study.

under 65°, or if the knee angle was beyond 65° the elicited contraction was ceased (Figure 1). For each FES stimuli, three analysis windows (1-initial, 2-middle and 3-final) were extracted from the MMG and angle signals response. The initial window started 8 s after the beginning (the first five seconds was rise time); the final one was triggered 3 s before the end (125 s or angle greater than 65º) to avoid spurious signal due the initial and final of contraction. Moreover, the middle was the equidistant window between the initial and final ones.

Data acquisition and analysis

All signals and volunteers data were saved into European Data Format (EDF) files. The acquisition system contained a DT300 series Data Translation™ board working at 1 kHz sampling rate. The *rectus femoris* muscular shape is bipenate (Blemker and Delp, 2006) and the displacements of muscle fiber oscillations during contraction occur in several directions; therefore, the resultant (or modulus) of three axes (X, Y and Z) was used, because it represents quite rightly the entire event. The analysis window length (AWL) was 1 s, and the Hanning window was applied to the signals before the spectral feature extraction. For every AWL, root mean square (RMS) and mean frequency (MF) features were computed from the MMG acquired signals. Then MMG_{RMS} and MMG_{MF} moduli were calculated and analyzed.

All data were normalized by means of the values of the initial window (1I). A Kolmogorov-Smirnov test was performed in order to evaluate the normal distribution of data. Paired-sample t-test was applied to compare MMG features between HV and SCIV.

Linear trend line was added to a representative SCIV data, to show the determination coefficient (r^2) to MMG_{RMS} and MMG_{MF}. Pearson's correlation coefficients were computed between the values of MMG_{RMS} and MMG_{MF} obtained from MMG sensors. Scatter plots were traced for the middle and final points showed a variation trend such as a cubic function.

Results

According to Kolmogorov-Smirnov test all data have Gaussian distribution and linear correlation analysis was performed. The stimulator output amplitude was adjusted in 82 ± 16 V for HV and 161 ± 36 V for SCIV. Table 2 shows the angular changes during the protocols applied to HV and SCIV. Table 3 presents the mean and standard deviation (along all the protocol) of MMG_{RMS} and MMG_{MF} normalized for

Table 2. Magnitudes of knee angle flexion obtained during the application of protocols to healthy and spinal cord injured volunteers.

Series	HV (°)	SCIV (°)
1I	37.71±20.59	27.05±23.94
2I	42.80±20.11	46.98±19.97
3I	48.25±19.85	50.54±16.91
1II	67.23±13.12	50.88±19.46
2II	57.86±18.53	50.78±23.04
3II	60.78±19.17	51.15±23.32
1III	63.53±11.83	53.43±17.77
2III	60.38±12.77	55.85±17.86
3III	61.78±10.49	58.71±17.92
1IV	67.59±6.41	57.14±15.18
2IV	64.58±8.71	55.66±17.49
3IV	64.96±7.68	58.82±17.26

HV: healthy volunteer; SCIV: spinal cord injured volunteer.

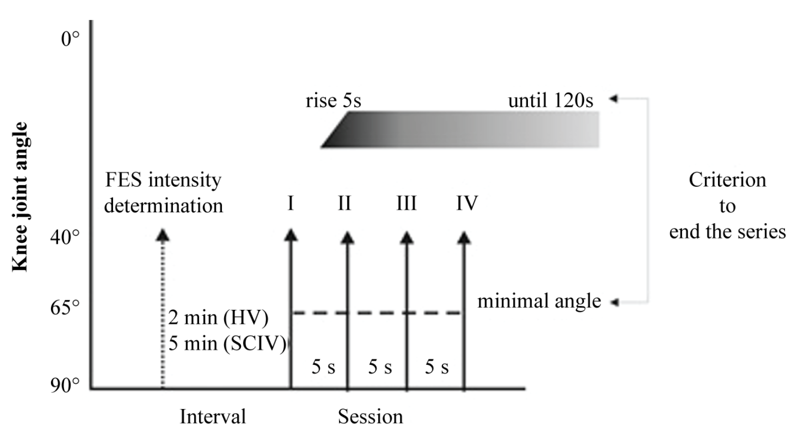

Figure 1. Timing scheme of FES application and MMG acquisition. FES intensity determination to reach 40°, Interval I (2 min – HV and 5 min – SCIV), Session (four contractions I, II, III and IV with 5 s interval between consecutive contractions).

HV and SCIV as the paired-sample t-test p value, where just the MMG_{MF} to VL muscle was different between the groups.

Figure 2 illustrates the MMG responses of RF and VL muscles of a single volunteer (K – see Table 1) submitted to FES application period of 32 s during the first series (I). The plotted trend lines (Linear MMG_{MF} and Linear MMG_{RMS}) indicate a divergence between MMG_{RMS} and MMG_{MF} variation rates along knee angle decrease.

Table 3. Normalized mean values of MMG_{RMS} and MMG_{MF} for healthy and spinal cord injured volunteers.

Vol	MMG_{RMS}		MMG_{MF}	
	RF (Norm)	VL (Norm)	RF (Norm)	VL (Norm)
HV	0.98±0.33	0.98±0.36	1.05±0.19	1.00±0.18
SCIV	1.01±0.56	0.89±0.41	0.96±0.19	1.01±0.16
p*	0.678	0.063	0.001	0.625

Vol: volunteer; RMS: root mean square; MF: mean frequency; Norm: normalized values by first point of analysis (1I); HV: healthy volunteer; SCIV: spinal cord injury volunteer; RF: *rectus femoris* sensor; VL: *vastus lateralis* sensor. *paired-sample t-test.

Table 4. Pearson's correlation coefficients (r) between MMG_{RMS} and MMG_{MF}.

Series	HV	SCIV
I	–0.82*	–0.12
II	–0.72*	–0.60*
III	–0.68*	–0.64*
IV	–0.75*	–0.58*

HV: healthy volunteer; SCIV: spinal cord injured volunteer. *$p< 0.05$.

Table 4 lists the Pearson's coefficients calculated. Negative correlation values corroborate that MMG_{RMS} and MMG_{MF} features tend to diverge during the FES application. Those coefficients varied from –0.12 up to –0.82.

Figure 3 shows scatter plots of the middle and final data for all FES series. Since all values were normalized by first series (I), the initial values were not included in those figures since they were all equal to unity. For HV and SCIV, the trend lines of MMG_{MF} and MMG_{RMS} presented determination coefficients (r-squared) from 0.50 to 0.64 indicating a moderate correlation.

Discussion

In this paper we determined the correlation between spectral and temporal MMG features during FES application to healthy and spinal cord injured volunteers. Our results indicate a negative correlation between the analyzed parameters (MMG_{RMS} and MMG_{MF}) to both groups as found by Tarata (2003) with HV during voluntary contraction. Therefore the negative correlation indicates that MMG_{RMS} and MMG_{MF} values diverge due to muscle fatigue and/or motoneuron adaptation. With a similar research aim, Merletti and Lo Conte (1995) used EMG to identify human *tibialis* anterior muscle fatigue and their results show that the EMG_{RMS} magnitudes increased and EMG_{MDF} (median frequency of EMG signal) values decreased along time.

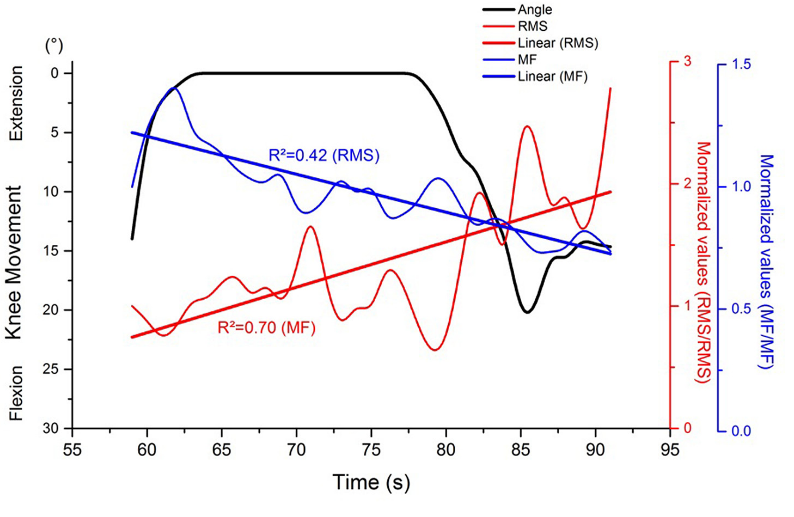

Figure 2. MMG response of *rectus femoris* muscle of spinal cord injured volunteer (K) during 32 s of the first FES application (I). Straight lines show MMG_{MF} and MMG_{RMS} data tendencies by means linear regression.

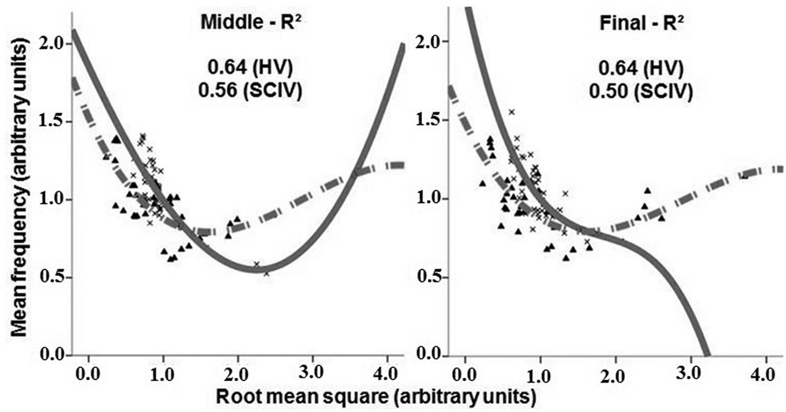

Figure 3. Scatter plots of MMG_{MF} and MMG_{RMS} for healthy [HV – filled line (x)] and spinal cord injured volunteers [SCIV –dashed line (▲)].

Regarding to the MMG temporal feature, Smith et al. (1997) found that with stronger muscular force, there is an increase in temporal features, thus raising the intensity of muscle vibration.

In the present research, the MMG sensor was attached on the skin. As the protocol involved dynamic movements, it was expected that the skin movement has changed the sensor position relative to the innervation zone. In that way, the MMG signal could be contaminated as the EMG signal in some cases (Artuğ et al., 2016), although Malek and Coburn (2011) showed that the MMG signal is not contaminated by the innervation zone for time and frequency domains.

Blangsted et al. (2005) suggested that the increase in MMG_{RMS} signal is due to the intramuscular pressure increase. However, Søgaard et al. (2006) showed that the increase in intramuscular pressure does not interfere in the amount of MMG_{RMS}. Akataki et al. (2003) investigated the increase in muscular force and its relationship to MMG. Until 40% of maximal voluntary contraction (MVC), the MMG_{RMS} magnitude in the first dorsal *interosseous* muscle tends to increase initially and to decrease later. This increase is similar to that occurring on *biceps brachii* muscle, followed by a decrease in MMG_{RMS} after 60% of MVC. Esposito et al. (2005) showed for the VL muscle an increase in MMG_{RMS} magnitude before of 80% MVC, and after this, the MMG_{RMS} variation trended down. Similar results were found by Stock et al. (2009) to RF, VL and VM (vastus medialis) muscles. These results indicate that the relationship between MMG_{RMS} and muscle strength may be non-linear. We did not measure muscular force directly (we measured the knee joint angle variation), therefore it was not possible to compare MMG_{RMS} and muscle strength.

In the present study, the magnitude of electrical current applied by means of a voltage-controlled stimulator to the quadriceps muscle contraction was just enough to keep a knee joint extension, consequently, evoking a contraction not so strong that could increase the lactate level (that was not measured) of the participants. However, we had two sample groups, one of SCIV and other of HV who have the integrity of their neuromuscular systems. There had been just significant MMG_{MF} difference for VL sensor (Table 3) between the groups, despite of their different neuromuscular condition and FES intensity levels.

Ebersole et al. (2006) recorded EMG signals while HV subjects performed fifty voluntary concentric repetitions of *quadriceps* muscle and obtained a decrease in the mean power frequency along the time. Carrying out a protocol to evaluate muscle fatigue during 10 s of 30% of MVC, Jansen et al. (1997) showed that there is a relationship between the onset of muscle fatigue, indicated by a decrease in EMG_{MDF} and the increasing of blood lactate.

In voluntary contractions, Ebersole and Malek (2008) stated that the MMG_{RMS} increases could be attributed to the recruitment of new motor units. Our results showed that along the time, with the application of electrical current, occurred a decrease in muscle oscillatory frequency (our results). This event could be explained by the increase in the motor units firing rate threshold caused by motor neuron adaptation (Merletti and Lo Conte, 1995; Spielmann et al., 1993). We think that some motor units tend to oscillate at the same frequency of other motor units. When motor units firing rate have a close frequency, the muscle performs an in phase contraction due to motor units coherence (Yao et al., 2000). This could explain the MMG_{RMS} increase and MMG_{MF} decrease.

In conclusion, during contraction evoked by FES, MMG_{RMS} and MMG_{MF} variations presented are inversely related, with coefficients reaching –0.82, which is represented by a negative correlation between them. These results were similar to healthy as to spinal cord injured subjects. The rise of MMG_{RMS} could be explained by the motor units coherence increasing the mechanical wave amplitude. The reduction of MMG_{MF} might be due to the increase in the motor units firing rate threshold caused by motor neuron adaptation (habituation). Future studies will be necessary in order to identify a technique to differentiate the timing among muscular fiber events along the FES application.

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