

Triaxial MMG investigation during FES-induced cycle movements

M C Gelain¹, G N Nogueira-Neto², P Nohama³ and P M Gewehr⁴

¹Master's student of Electrical Engineering and Industrial Informatics at UTFPR (Rehabilitation engineering laboratory phone: +55 (41) 3310-4489), Brazil.

²PhD in Biomedical Engineering, researcher and professor of Computer Engineering at Pontifícia Universidade Católica do Paraná (PUCPR), Brazil (e-mail: nogueira.g@pucpr.br).

³PhD in Biomedical Engineering, researcher and full professor at the PPGTS *stricto sensu* program at Pontifícia Universidade Católica do Paraná (PUCPR), Brazil (e-mail: percy.nohama@gmail.com).

⁴PhD in Medical Physics Bioengineering and professor of Biomedical engineering at UTFPR, Brazil (e-mail: gewehr@utfpr.edu.br)

E-mail: gelainfisio@gmail.com

Abstract. Spinal cord injury is damage to the spinal cord that results in loss of functions, causes muscle, joint and bone deficits, moreover it increases the muscle tone level. FES-cycling is an alternative rehabilitation process to conserve the musculoskeletal system affected by the injury in the functional state. The aim of this study is to get the legs bilaterally synchronized by FES and compare MMG signal responses. The preliminary protocol consists of five consecutive contractions. The first and the last contractions were discarded and the three intermediate contractions were analyzed. A ratio between MMG values of muscle responses during propulsion and return of pedalling phases was calculated for each cycle and each leg. The applied FES profile consisted of pulse and burst on time of 250 μ s and 5 ms and frequencies of 1000 Hz and 20 Hz, respectively. Results showed that greater MMG RMS values tend to occur during the propulsion phase than in the return phase. MMG Z axis of RLL and MMG X axis of LLL were the axes that presented the greatest ratios. The MMG RMS signal showed that greater values tend to occur during the propulsion phase than in the return phase of the same thigh.

1. Introduction

Spinal cord injury (SCI) is damage to the spinal cord that results in loss of functions, causes muscle, joint and bone deficits and, moreover, it increases the muscle level [1]. Such tissues could be preserved through the long-term rehabilitation utilizing functional tasks such as treadmill training, leg cycling and functional electrical stimulation (FES) that also optimizes the plasticity mechanisms [2]. The association of leg cycling with FES is an alternative strategy to train the impaired muscles, allowing improvements in muscle mass and strength, blood flow, bone density, decreasing the abdominal adipose tissues by cardiopulmonary fitness and influencing the psychological health [3] and social integration [4, 5].



However, it is possible to raise at least three complications in this rehabilitation strategy: (i) finding appropriate parameters to provide the synchronization timing that generates a coordinated pedalling movement [6]; (ii) producing pedalling movements elicited by FES while minimizing the effects of muscle fatigue that acts as a muscle performance suppressor mechanism [2, 7, 8]; (iii) determining the required stimulation amplitude to contract the muscles accordingly without causing tissue damage and pain [9, 10]. Regarding the third complication, it is particularly important to be aware of this problem when healthy persons participate in the research.

The SCI classification (paraplegic or quadriplegic and complete or incomplete) is important to the rehabilitation process, as well as the injury extension like Peripheral Nervous System (PNS) integrity, the specification of lower motor neuron, allowing the lower limb movements by neuromuscular stimulation [9].

The employed system [11] was already used in other studies, analyzing primarily isometric contractions without bilateral stimulation.

Mechanomyography (MMG) is a non-invasive technique used for registering the muscle vibrations sensed on the muscle surface. Inside the muscle, the myofibrils contract generating intra-tissular pressure waves. Conversely to an electromyography (EMG) device, an MMG acquisition system can be easily attached to an electrical stimulator because the high voltage electrical stimuli do not interfere with the vibrations recorded by the accelerometer embedded in the sensor, a problem that hinders continuously monitoring signals during FES application. However, MMG is employed with isometric contractions, instead of dynamic contractions. Regarding leg bilateral stimulation to continuously propel a cycle ergometer's crown, MMG can be used to investigate the muscle contractive activity during both FES-elicited contraction and return phases. Therefore, the aim of this study is to produce pedalling movements using synchronized FES bilaterally applied to the lower limbs and compare the MMG signal amplitude response.

2. Methods

The preliminary protocol was performed at PUCPR, Brazil, in the Rehabilitation Engineering Laboratory, using Movement's[®] LX130 cycle ergometer. Three healthy male participants were seated on the ergometer and the required adjustments were made according to their leg size. Because recumbent cycle ergometers did not provide the best ergonomics to develop strength in the propulsion phase [12], the seat height was increased i.e. the hip and knee angles were set around 100 degrees of flexion, increasing the length of rectus femoris.

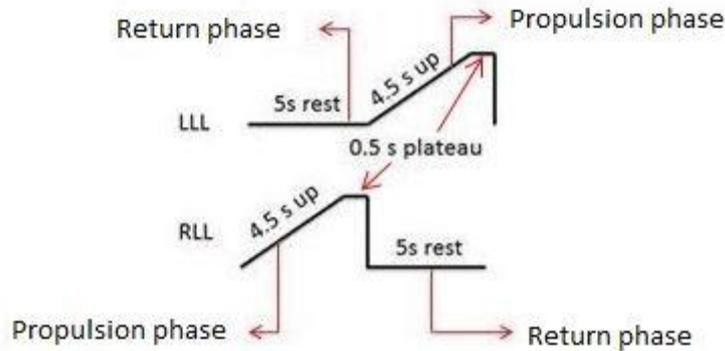
Three electrodes were used, one placed over the belly of rectus femoris (between the superior border patella and anterior superior iliac spine) [13], other placed over femoris nerve (femoral triangle) [14] and the last one over the belly of gluteus maximus (between the great trochanter and sacrum) [13].

The protocol consisted of two parts. In the first one (warm up), individuals cycled during 2 min. In the second part, they ceased pedalling and FES was applied over rectus femoris and gluteus maximus muscles of both legs, to allow five consecutive contractions. Data obtained in the first and the last contractions were discarded. The amplitudes of MMG signals of both legs were compared.

The FES profile applied consisted in pulse and burst on time of 250 μ s and 5 ms, respectively, pulse and burst frequencies of 1000 and 20 Hz, respectively, square wave, four channels. Multichannel FES reduces the number of electrodes to be employed for eliciting electrical stimulation once it requires one single common return electrode near the motor point for each leg [9].

The amplitude modulated FES profile consisted of 10 s divided in three phases: propulsion, consisting of ramp up phase (4.5 s); plateau (0.5 s); and return phase representing the last 5 s, as illustrated in Fig.1. The FES amplitude values were set according to two factors: (i) sensitivity response of participants i.e. how affordable they perceived the sensation and (ii) the required energy to let the leg artificially propel the cycling.

Figure 1 – FES amplitude modulated profile.



LLL = Left Lower Limb; RLL = Right Lower Limb

The MMG sensor uses a triaxial accelerometer (Freescale MMA7260Q MEMS [13x18 mm, 0.94 g] with sensibility equal to 800 mV/G in 1.5 G [G is gravity acceleration]). The hardware amplification was 2.2x and a Butterworth 3rd order bandpass filter was built. The software has been developed in the NI LabVIEW[®] platform to acquire and process all signals. The acquisition system consisted in an acquisition board (NI-USB 6221), using a frequency of 1 kHz to signal acquisition [11].

Data were analyzed for each contraction during four continuous seconds that corresponded to the increasing stimulation ramp and plateau phases. The three intermediate contractions were taken from the five contractions and a simple ratio between MMG RMS (propulsion phase / return phase) was calculated for each cycle and each leg.

3.Results

The second, third and fourth propulsion phase contractions of persons 1, 2 and 3 are indicated in Tables I and II by the MMG RMS ratios calculated for each cycle and each leg. One can visualize in Table II the mean of the three subjects.

Table I. Second, third and fourth contraction using FES of Person 1 and 2

			2nd	3rd	4th	Mean
Person 1	RLL	X	1.56	1.48	0.31	1.11
		Y	1.64	1.30	3.45	2.13
		Z	2.55	2.90	2.52	2.66
	LLL	X	1.79	2.14	1.01	1.65
		Y	1.48	1.41	0.64	1.18
		Z	1.48	1.41	0.64	1.18
Person 2	RLL	X	1.00	1.63	1.88	1.50
		Y	0.93	1.64	2.36	1.64
		Z	1.64	2.21	4.24	2.70
	LLL	X	1.95	1.17	1.19	1.44
		Y	1.14	1.23	0.60	0.99
		Z	1.14	1.23	0.60	0.99

RLL = right lower limb; LLL = left lower limb; X, Y and Z represents the three sensor axes.

Table II. Second, third and fourth contraction using FES of Person 2 and Mean

			2nd	3rd	4th	Mean
Person 3	RLL	X	0.78	0.84	0.89	0.84
		Y	0.71	0.64	0.81	0.72
		Z	1.26	0.84	0.99	1.03
	LLL	X	2.47	1.67	1.64	1.93
		Y	1.59	2.18	1.89	1.89
		Z	1.59	2.18	1.89	1.89
Mean	RLL	X	1.10	1.29	0.64	1.01
		Y	1.06	1.18	1.94	1.39
		Z	1.79	1.80	2.25	1.95
	LLL	X	2.03	1.60	1.28	1.64
		Y	1.35	1.52	0.91	1.26
		Z	1.35	1.52	0.91	1.26

RLL = right lower limb; LLL = left lower limb; X, Y and Z represents the three sensor axes.

The results in Tables I and II lead to the assumption that greater MMG RMS values tend to occur in the contraction phase in spite of the return phase of the same thigh. This can be observed by values greater than one. MMG Z axis to right lower limb (RLL) and MMG X axis to left lower limb (LLL) were the axes that presented the greatest ratios.

4. Discussion

FES-cycling might be an option to socially include SCI people, providing recreational activities as proposed by Hunt et al. who developed a tricycle [4], or activities of daily living (ADL) using a wheelchair with pedals like the one built by Watanabe et al. [5]. Before FES-cycling become a marketable product it is necessary to focus on research outcomes.

An important factor is the electric stimulation perception by subjects. Besides not being presented in the table, individuals informed a perception of an intensity difference between legs during FES application. The sensation was stronger on the right thigh than on the left thigh, however, the stimulus intensity needed to be adjusted until the sensation on both legs seemed equal. The same fact was observed by [5] and the delay of sensation would cause a delay on muscle answer, being necessary to adjust the pedalling time and velocity. Gfohler and Lugner [6] emphasized that the sensibility between legs occurs due to physiological differences and each individual has his/her own parameters profile for FES-cycling.

From a sensor's triaxial perspective, it is possible to observe that there were differences between the responses of MMG signals between legs. In Table I and II, the mean column and mean rows show that the Z axis presented greater average RMS values for the pedalling propulsion phase than for the returning phase to the RLL. To the LLL, conversely, the X axis presented similar response. FES intensity adjustments could be the reason for the existence of such particular differences since they were performed according to individual perception [10, 15]. However, another factor can be strength asymmetry [16] in both thighs and gluteus, as well as position, size and shape of the electrodes [17].

The adjusted seat position of the participants shortens the rectus femoris and since it is a biarticular muscle this fact compromises the propulsion phase, moreover it caused more discomfort during FES application. The difference between angles and muscles involved in pedalling with recumbent and vertical cycle ergometers are explained in [18]. However, Lopes et al. [19] affirm that differences in muscles activation do not exist using the two types of cycle ergometer.

The most correct posture for pedalling is obtained with a vertical cycle ergometer [20]. During a kinematic test the cyclist posture and joint position were verified, between 5-10 initial minutes, the fatigue was observed, because there were some joint alterations changing the initial posture. The adopted new musculoskeletal body position was trunk lean towards the front, increasing the hip flexion to facilitate the gluteus maximus contraction, consequently altering the knee joint and decreasing the knee flexion angle on the pushing phase [12], finally promoting an easier propulsion.

One difficulty that happens when working with FES application to healthy people is whether the contraction is happening solely due to the electric stimuli or also with the help of the participant. So there are two physiological reflexes to FES: H-reflex and flexion afferent reflex. During a muscle contraction elicited by FES, the H-reflex mechanism occurs due to external stimuli, sending an afferent signal to the spinal cord which in turn returns the signal by efferent pathway. However, two responses to neutralization of the H-reflex signals could be received when there are high levels of stimulus intensity: (i) major firing quantity of motor fibers inhibiting the motoneurons and (ii) the efferent response occurring at the same time as the afferent stimuli, creating a collision between them [21]. Another possibility is the flexion afferent reflex [22], because the organism could interpret the stimulus intensity as harmful and block the voluntary movement. The inverse can also happen, when a healthy individual aids the movement voluntarily blocking the reflex. Variations in the efferent stimulus conduction velocity can vary between 20-80 ms [22] and to the reflex between 20-30 ms [21], further the FES OFF time (pedalling return phase) is shorter than the time it takes for the CNS stimulus to adapt, therefore the CNS can interrupt or sum with H-reflex signal.

5. Conclusion

The MMG RMS signal from FES stimulation showed that greater values tend to occur during the propulsion phase than in the return phase of the same thigh. The signals recorded by different MMG axes can respond with different trends. Future research will occur with SCI individuals using personalized amplitude patterns in order to avoid the muscle unbalance as a result of disproportionate stimulation intensity.

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