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Muscle fatigue assessment by mechanomyography during application of NMES protocol

Avaliação da fadiga muscular pela mecanomiografia durante a aplicação de um protocolo de EENM

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Abstract

Background: Neuromuscular electrical stimulation (NMES) is a widely used technique for rehabilitation in physical therapy, however it causes muscle fatigue more rapidly than does voluntary contraction. In clinical practice, it becomes necessary to monitor muscle fatigue during NMES protocols to adjust the parameters of electrical current stimulation and, thus, increase stimulation time. **Objectives:** The aim of this study is to use mechanomyography (MMG) as a means of evaluating peripheral muscle fatigue during the execution of an NMES protocol. **Methods:** An MMG signal acquisition system and an experimental protocol were developed. During *in vivo* tests, 10 participants performed maximal voluntary contractions (MVCs) for knee extension. A maximization phase was conducted with dynamic contractions generated by NMES at 10% of MVC (100 Hz, 400 µs) on the quadriceps muscle, and the main NMES protocol occurred at 30% of MVC (50 Hz, 400 µs). Simultaneously, MMG_{RMS} (amplitude) and MMG_{MPF} (frequency) signals of the rectus femoris and the knee extension torque were acquired. **Results:** The tendency line of the MMG_{RMS} was descendant, indicating that MMG_{RMS} correlates with torque amplitude. However, MMG_{MPF} did not show a significant correlation with torque for the present NMES protocol. **Conclusions:** MMG is a technique that can be simultaneously applied to NMES because there is no electrical interference and it can be used during functional movements in the NMES-generated muscle contraction.
Article registered in the Australian New Zealand Clinical Trials Registry (ANZCTR) under the number ACTRN12609000866202.

Key words: neuromuscular electrical stimulation; muscle fatigue; mechanomyography; rectus femoris muscle.

Resumo

Contextualização: A estimulação elétrica neuromuscular (EENM) é uma técnica bastante utilizada na área da reabilitação em fisioterapia, porém a instalação da fadiga ocorre de maneira mais rápida se comparada à contração voluntária. Na prática clínica, torna-se necessário monitorar a fadiga muscular em protocolos de EENM, visando adequar os parâmetros da corrente elétrica e prolongar o tempo de estimulação. **Objetivos:** O objetivo deste estudo foi utilizar a mecanomiografia como meio de avaliação da fadiga muscular periférica durante a aplicação de um protocolo de EENM. **Métodos:** Um sistema de aquisição de sinais mecanomiográficos (MMG) e um protocolo experimental foram desenvolvidos. Durante os ensaios *in vivo* com 10 voluntários, foram realizados testes de contração voluntária máxima (CVM) para extensão do joelho. Realizou-se uma fase de potencialização com contrações dinâmicas produzidas por EENM a 10% da CVM (100 Hz, 400 µm) no músculo quadríceps femoral, e o protocolo de EENM propriamente dito ocorreu a 30% da CVM (50 Hz, 400 µm). Simultaneamente, foram adquiridos os sinais de MMG_{RMS} (amplitude) e MMG_{MPF} (frequência) do músculo reto femoral e de torque (amplitude) para a extensão do joelho. **Resultados:** A linha de tendência da MMG_{RMS} foi descendente, indicando que a MMG_{RMS} relaciona-se à amplitude do torque. Porém, a MMG_{MPF} não teve uma boa correlação com o torque para este protocolo de EENM. **Conclusões:** A MMG pode ser aplicada simultaneamente à EENM, pois não ocorre interferência elétrica, e pode ser utilizada na realização de movimentos funcionais na contração muscular gerada por EENM.
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Palavras-chave: estimulação elétrica neuromuscular; fadiga muscular; mecanomiografia; músculo reto femoral .

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Introduction

Neuromuscular electrical stimulation (NMES) is a technique by which an electrical current is applied to induce muscle contractions and produce functional movements in individuals with neurologic diseases or to promote muscle strengthening in order to improve their physical performance[1-3]. NMES is believed to provide muscle strengthening by means of a mechanism that is different from voluntary contraction[4,5]. In a voluntary muscle contraction, the smaller motor neurons that innervate the slow-twitch fibers are activated first, followed by the larger motor neurons that innervate the fast-twitch fibers. However, the recruitment sequence of muscle fibers is inverted during the application of the NMES, i.e. the fast-twitch fibers (less resistant to fatigue) are recruited first[6]. Furthermore, NMES stimulates a fixed set of motor units, resulting in the precipitation of muscle fatigue when the skeletal muscles are electrically activated[5].

It is important to measure muscle strength in its relationship with the muscle's contractile characteristics during continuous electrical stimulation so that a better control of NMES can be achieved[3]. In the clinical practice of physical therapy, muscle fatigue must be assessed during the application of the NMES protocols to adapt the current parameters to the stimulated muscles and minimize the harmful effects caused by the onset of muscle fatigue (pain, fibrillation, and torque reduction)[7], thus optimizing the benefits of the technique. Lactate analysis, electromyography (EMG) and mechanomyography (MMG) have already been described in the literature as techniques to assess the variations that occur in the muscle before and during the onset of fatigue[3,8,9]. However, the first two techniques have limitations concerning the assessment of synaptic fatigue and the application during the NMES. Lactate analysis only indicates metabolic fatigue[9], and EMG suffers interference from the NMES signal, leading to saturation of the EMG signal.

MMG is a non-invasive method that does not suffer any interference from the electrical stimulation device while the muscle signal is being collected[10-12]. By using this technique, the signal of the muscle vibration can be captured and used to study mechanical aspects involved in the voluntary muscle contraction[12] and to assess muscle fatigue[13]. The assessment of muscle fatigue through MMG during the application of an NMES protocol has already been described by Gobbo et al.[13], however only the parameters of the mechanomyographic root mean square amplitude (MMG_{RMS}) and torque were assessed. The parameter related to signal frequency was not included in the evaluation.

In the present study, we hypothesized the existence of changes in the mechanomyographic mean power frequency (MMG_{MPF}), as well as variations in the MMG_{RMS}, concomitantly to changes in torque during the onset of peripheral muscle fatigue in the application of an NMES protocol. Thus, the objective of the study was to assess the behavior of the MMG_{MPF} and MMG_{RMS} during an NMES protocol.

Methods

The experimental protocol was applied to ten healthy male individuals aged 26.7±5.35 years (mean±SD), 1.77±0.06 m in height, weighing 79.60±9.73 kg and with a BMI of 25.17±2.37. All participants were aware of the protocol to be carried out and signed an informed consent form. The study was approved by the Human Research Ethics Committee of Pontifícia Universidade Católica do Paraná (PUCPR), CAAE:001.0,084,000-05 (CEP 514, FR 055738). The experimental protocol was composed of three phases: isometric assessment, potentialization, and application of the NMES protocol.

Isometric assessment

This phase aimed to measure the maximal voluntary isometric contraction (MVIC) to serve as a reference for the potentialization and for the application of the NMES protocol. Initially, the participants warmed up on the cycle ergometer (Caloi®, Caloicicle Eletronic Pulse), for 5 minutes[14]. After the warm-up, they were placed on the isokinetic dynamometer chair (Cybex®, Norm 7000) for the isometric assessment. The hip was positioned and fixed at 100º, and the axis of the knee to be assessed was positioned on the same axis of the dynamometer at a 60º flexion, as described by Ruiter et al.[1], in order to obtain the stabilization of the segments and better torque. All tests involved the knee of the dominant lower limb (right side for all participants).

The isometric assessment consisted of two submaximal and one maximal contraction[14,15] of the knee extensors, each contraction lasting five seconds, so that the participant could learn the movement, become familiar with the apparatus, and recruit the motor units[15], with a five-second interval between contractions. Next, three more MVICs were performed, each repetition lasting five seconds, considering the maximal torque generated among the three repetitions. A two-minute rest was observed between the tests (resting time also used in studies carried out by Ebersole et al.[16] and Beck et al.[17]) and before the potentialization phase.

Potentialization

Prior to the application of the NMES protocol, a potentialization phase was carried out by means of low-intensity,

short-duration stimulation of the quadriceps femoris. The potentialization aimed to promote motor-unit activation and to reduce stimulation time and intensity in order to achieve the desired contractile force. In this phase, the intent was not to produce muscle fatigue. For the muscle stimulation (both for the potentialization and application of the NMES protocol), two self-adhesive electrodes were used (5 cm x 9 cm). The skin's electrical impedance was reduced by shaving and by cleaning the skin with alcohol to remove dead cells and oil. A stimulation electrode was positioned 10 cm below the iliac crest and another 5 cm above the superior edge of the patella[18,19] (Figure 1), promoting local stimulation[20].

The potentialization was carried out in the same position on the isokinetic dynamometer, so that the torque could be observed in this phase. It started with a stimulation of 10% of MVIC. The stimulation current was generated by an electrical stimulator (KLD®, Endophasys NMS0501) in FES PAM mode (modulation by pulse amplitude). The stimulation current was a biphasic square wave pulse of 400 μs duration[21] and 100 Hz frequency[2,13], with five-second stimulation and five-second rest for three contractions. After the potentialization, a two-minute rest was observed, followed by the fatigue protocol itself, with NMES of 30% of MVIC.

Application of the NMES protocol

The NMES protocol aimed to promote peripheral muscle fatigue by means of continuous muscle contraction without rests for the metabolic fiber recovery, so that the MMG signal and torque variables could be observed. The protocol consisted in applying the NMES and capturing the MMG and torque signals for comparison. In the application of the developed stimulation protocol, the electrodes were kept in the same position as used in the potentialization, but the burst frequency was set at 50 Hz (studies report that force peaks are generated at 50 Hz and reach a plateau above this frequency[2,22]), until 30% of MVIC[23]. When the stimulator's output current intensity was reached to generate 30% of MVIC, it was maintained, and the stimulation time began (120 seconds; time used in NMES protocols on the quadriceps femoris to support a paraplegic in a standing position[24]). A sustained isometric contraction was carried out for 120 seconds, and the participants did not perform any MVICs associated with the application of the NMES. The MMG and torque signals were measured throughout the application of the NMES protocol, and only the signals of this phase were processed and analyzed.

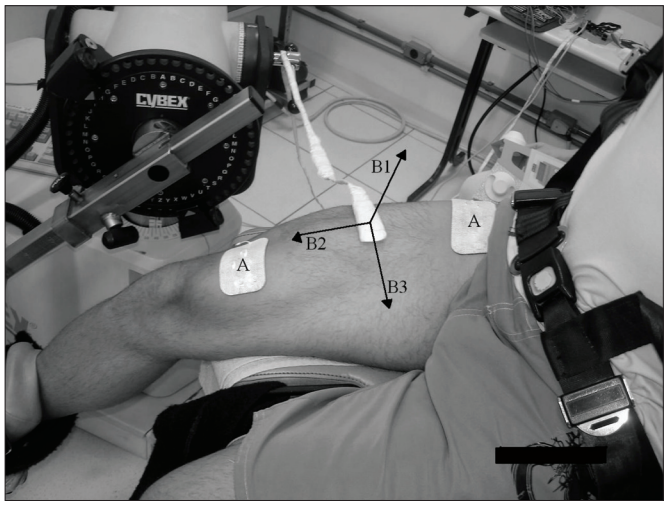

Figure 1. A) Placement of stimulation electrodes, B1) Axis 1, B2) Axis 2 and B3) Axis 3 of mechanomyography sensor.

MMG signal acquisition system

A system of MMG signal capture was developed to meet the need to assess muscle fatigue concomitantly to the NMES application. An electrical circuit for MMG signal acquisition was designed using sensors with a triaxial accelerometer[25,26]. This accelerometer has a sensitivity adjustment that considers the value of the normal acceleration of gravity (g) as a basis for the calculation. The sensor output was set to 800 mV/V and pre-amplified with a gain of 100 V/V. Nevertheless, only the oscillations were recorded, and the static acceleration was discarded by filtering the lower frequencies. The system also captured variations in three axes (Figure 1), which better represents the oscillations that occur in the muscle during the contraction.

The system with the sensor was calibrated by employing a sine-wave signal generator (PASCO Digital Function®, model PI-9587C) connected to a PASCO SF-9324 mechanical wave driver. The generator produced a sine-wave signal at various frequencies (10, 11, 12, 13, 14, 15, 16, 17, 18, 19, 20, 25, 30, 35 and 40 Hz with two-decimal-place precision). The MMG sensor was positioned on the mechanical wave driver and satisfactorily measured all the frequencies, suffering only decimal-place variations and the influence of the low-frequency harmonics. During a muscle contraction, either voluntary or prompted by the NMES, the frequencies in the various motor units overlap in the captured signal. Therefore, the emergence of harmonics does not affect the final analysis of the MMG signal. In order to test the reproducibility, the test was repeated with the generator connected to a subwoofer (PIONEER®, model TSW256DVC). The test with the subwoofer was even more precise than with the mechanical wave driver.

During the application of the NMES protocol, the signals of the three MMG axes and the torque signal were captured simultaneously to the application of the electrical current. The MMG sensor was positioned on the skin by using double-

sided adhesive tape on the muscle belly of the rectus femoris[25]. The MMG and torque signals were converted into 12-bit digital signals and processed using the software LabVIEW (National Instruments®, Austin, TX). By using this software, the digital MMG signal was filtered with a 4-40 Hz 10th order band-pass Butterworth filter[27]. The Fast Discrete Fourier Transform (FDFT) with a 1-second Hanning Window was applied to obtain the signal in the frequency domain. The power spectral density (PSD) was also corrected after this windowing. The torque signal was not digitally filtered. The sample frequency was 1 kHz, both for the MMG and torque signals. After the conversion of these parameters to Excel®, it was possible to observe the data and analyze the patterns found in the signals.

Statistical analysis

The parameters of the MMG signal chosen for analysis were the root mean square amplitude (RMS) and mean power frequency (MPF) of the MMG signal (MMG_{RMS} and MMG_{MPF}, respectively), and also the torque signal. In order to compare the MMG_{RMS} and MMG_{MPF} values of each axis, the cross-correlation coefficient between these two parameters was calculated to determine which axis of the MMG signal would be the most significant and from which the analysis would be made. Finally, the tendency curve with a 6th order polynomial adjustment was plotted on Excel® to understand the behavior of the MMGRMS, MMGMPF and torque during the application of the stimulation protocol based on the normalized mean values found in all participants. The 6th order polynomial adjustment was used for all the variables to form the tendency curve because it was the adjustment that best represented the signals and had the highest coefficient of determination when compared to the linear, exponential, or polynomial adjustments of different orders.

Results

Based on the analysis of the cross-correlation coefficients between the MMG_{RMS} and MMG_{MPF} values, it was observed that, for the rectus femoris muscle, the most significant axis for capturing the muscle vibration signal was axis 3 with a value of 0.654, whereas axis 2 had 0.559. Axis 1 had very low MMG_{RMS} and was, therefore, excluded from the study. The following results are based on the analysis of axis 3 of the MMG and torque, which were captured simultaneously.

As regards the MMG_{MPF}, the mean values oscillated widely. However, when represented by the tendency curve, they were reduced at the beginning of the protocol, coinciding with the moment of rise in stimulation intensity, until the moment the torque values dropped (MMG_{RMS} values in approximately 55% of the plateau of the MMG_{MPF} value; Figure 2). After this period, there was a stabilization of the MMG_{MPF} during the stimulation. However, after approximately 100 seconds of constant stimulation, with continuous current intensity, the MMG_{MPF} gradually decreased to values below 40% of the MMG_{MPF} plateau, unlike the tendency curves of the MMG_{RMS} and torque values. However, the coefficient of determination for the mean tendency curve of the normalized MMG_{MPF} resulted in a very low value (R^2=0.195) for the mathematical model of this parameter for the population observed.

With regard to the amplitude of the MMG_{RMS}, when looking at the tendency curve of the normalized mean values, the signal increases at the beginning of the protocol until it reaches 100% of the amplitude of the MMG_{RMS} signal (which coincides with the moment that the stimulation intensity increases) and peaks when it reaches the current intensity needed to generate 30% of MVIC. After this peak, there is a reduction in the amplitude for values below 60% and a subsequent torque reduction (Figure 2). At a given point of the stimulation, there was a variation in the signal's tendency curve, coinciding with the visualization of fasciculations in the anterior medial portion of the thigh, followed by torque increase in knee extension. This fasciculation increased the amplitude signal of the MMG_{RMS} value (78% of the MMG_{RMS}). At the end of the protocol, the MMG_{RMS} values decreased, as did the torque signal (below 40%). The coefficient of determination calculated for the tendency mean curve of the normalized MMG_{RMS} (R^2=0.354) was low for the mathematical model of this parameter for the evaluated population, however the tendency curve of the MMG_{RMS} better represented the phenomenon that occurred in the signal, accompanying the torque signal in the time domain.

During the application of this protocol of stimulation and fatigue, the torque behaved similarly to the tendency curve of the MMG_{RMS}. At first, there was an increment in the torque amplitude by up to 100% of the measured torque (meaning an increase in strength) during the increment in the intensity of the current generated by the stimulator. After the peak reached by the current to generate 30% of MVIC, there was a decrease and stabilization of the torque values during the application of the electrical stimulation (below 10%). The moment that fasciculation in the anterior medial portion of the thigh and knee extension were detected, there was a period of torque increase (around 15%). At the end of the protocol, there was a decrease in the tendency torque curve (Figure 2). The coefficient of determination calculated for the tendency torque mean curve (R^2=0.8963) is a high value and allows the mathematical modeling of this parameter for this population and the application of the same protocol.

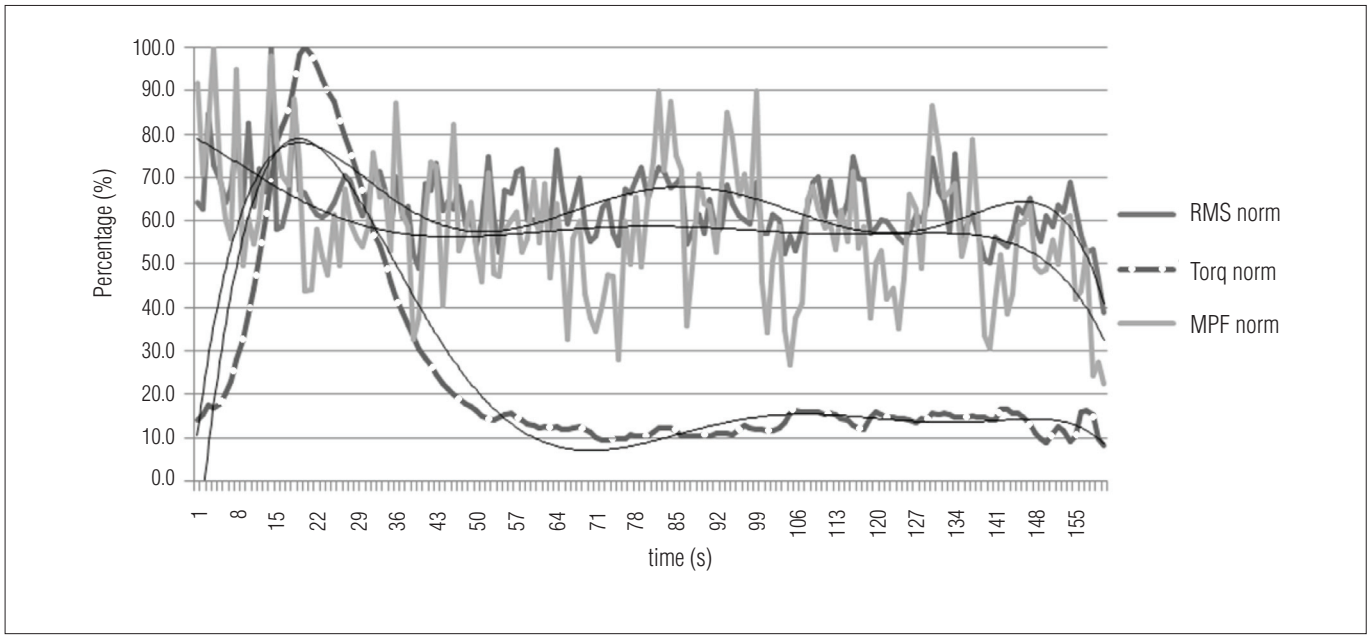

Figure 2. Tendency curves of MMG signals (RMS and MPF) in relation to torque and normalized to 100%.

Discussion

The strong and prolonged contraction of a muscle leads to muscle fatigue, characterized by the inability of the contractile and metabolic processes of the muscle fibers to maintain the same work production (torque reduction)[28]. Synaptic fatigue is described as the process in which there is insufficient accumulation of neurotransmitters in the synaptic cleft to trigger and propagate the action potential by means of the T-tubules, therefore the motor units are not activated[28]. There are several methods of monitoring muscle fatigue, however some of them have limitations concerning their application. The local lactate measurement method allows the assessment of metabolic changes after prolonged NMES, but it does not allow momentary measurement of synaptic fatigue[9]. In contrast, isokinetic dynamometry is capable of assessing local muscle fatigue by measuring torque[29], however it cannot be used in field research outside the laboratory where the equipment is located. EMG has already been described in the literature as a method of assessing muscle fatigue during NMES application[30,31]. Nevertheless, preliminary EMG tests displayed saturation of the EMG circuit preamplifier due to interference of the NMES signal.

In a study conducted by Tarata[27], the author mentioned that the EMG can be contaminated by electrical noise and carried out an extensive comparison between EMG and MMG in voluntary isometric contraction. In this study, the author detected a similar evolution both in the EMG and in the MMG, supporting the hypothesis that MMG can also represent a decrease in muscle activation (characterized by torque reduction) and that it can be used to monitor the development of muscle fatigue. The MMG signal reflects the mechanical muscle vibrations by the spatio-temporal summation of individual muscle fiber twitches, which are evoked by motor unit activation by the motor neurons[27]. Because it is a non-invasive method to assess muscle function and does not suffer interference from the electrical signal, MMG can be used in a vast array of places, such as those under the interference of electrical noise, and during the application of NMES protocols. Moreover, the mechanical components of the muscle (such as torque and contraction speed) are more closely related to the muscle function than to its electrical characteristics[32].

There is a relationship between the MMG signal and torque during voluntary contraction[17], and most studies only describe the use of MMG in this kind of contraction[17,33-35]. Only a few studies describe the use of MMG during NMES[13,23], therefore some parameters of MMG application described in the methodology of this article are based on studies that used the technique during voluntary muscle contraction. Ryan et al.[25] and Al-Zahrani et al.[36] described the application of MMG in the assessment of the rectus femoris muscle, suggesting that the accelerometer be positioned on the muscle belly and the use of double-sided adhesive tape to attach the sensor during the isometric contraction of the quadriceps.

In the present study, we assessed the MMG signal during the isometric contraction generated by NMES, using the amplitude and frequency parameters of this signal (MMG_{RMS} and MMG_{MPF}, respectively). A triaxial accelerometer was employed, which allowed the comparison of the parameters of all axes and the conclusion that, for these conditions and for the muscle

being assessed, axis 3 was the most significant (Figure 1). Most of the studies in the literature used less complex sensors to detect monoaxial acceleration[10,37,38]; however, Al-Zahrani et al.[36] have more recently described the use of triaxial accelerometers, which capture the variations in three axes and allow the identification of the axis with the highest acceleration variation and its correlation with the anatomy of the muscles under assessment.

Gobbo et al.[13] carried out a study assessing the changes detected in the MMG signal during dynamic contractions generated by NMES, however this study differs in that it only used the parameter related to the amplitude of the MMG signal and because it used dynamic contractions. Gobbo et al.[13] also carried out a phase of potentialization at 100 Hz prior to the stimulation protocol itself (using a 50 Hz current). This phase serves to increase the phosphorylation of the myosin filaments, causing the contractile filaments of the muscle to become more sensitive to calcium[39]. According to Chou et al.[2], approximately 300 pulses are necessary to potentialize the human quadriceps femoris.

Blangsted et al.[23] carried out a research which differs from the present study in that it used low-frequency electrical stimulation (the stimulation protocol consisted of a 10-second-train at 1 Hz, two 2.5-second trains at 20 Hz, and two 2-second trains at 100 Hz, with a 30-second rest between the trains) and dynamic contractions. The authors observed that the MMG_{RMS} signal was better correlated to the torque than to the EMG_{RMS} torque in the low-frequency stimulation.

According to the results put forward in the present article, there was a significant increase in the MMG_{RMS} value at the beginning of the contraction (reflecting an increase in the activation of the muscle fibers, possibly due to the growing number of active motor units combined with a synchronous firing[27]) followed by a progressive reduction at the end of the effective contraction which leads to muscle fatigue. This reduction in the MMG_{RMS} value was also reported by Bajaj et al.[32] at the onset of muscle fatigue.

During the monitoring of the MMG_{MPF}, fatigue was associated with the compression of the PSD of the MMG signal for the lower frequencies[27,40]. This displacement of the power spectrum to the lower frequencies has been attributed to the decline in the action potential conduction speed due to the decline in intracellular pH or to the accumulation of extracellular K^+ and synchronism in the motor unit action potentials[40]. The analysis of the MMG_{MPF} tendency curve displayed in the results of the present study indicated a downward trend of the mean normalized values in the development of muscle fatigue. However, because the coefficient of determination of the adjustment was very low (R^2=0.1949), this parameter could not be considered an indication of muscle fatigue during the application of the NMES under the same conditions of this study. Beck et al.[41] described similar results concerning the coefficient of determination (R^2=0.258), also suggesting that the MMG_{MPF} parameter would hardly be modeled. According to Tarata[27], the MMG_{MPF} suffers great oscillation in its mean values in the presence of muscle fatigue[27]. This oscillation was also observed in the present study, making it difficult to model the MMG_{MPF} parameter. In the literature, there was no description of the MMG_{MPF} evolution in the development of muscle fatigue during the application of NMES and no justification for not using the MMG_{MPF} as an evaluative parameter.

The results put forward in this article partially confirm the hypothesis of the existence of changes in the MMG parameters during the muscle fatigue, because the MMG_{RMS} had a similar change to that of the torque. Nevertheless, the data found concerning the MMG_{MPF} were inconclusive. A better understanding of the variations that occur in the MMG signal, and its relationship with the characteristics of muscle contraction, will allow the use of this innovative technique in the evaluation of NMES rehabilitation programs. Although new research is still needed to corroborate the relationship between the parameters obtained from the signal and from muscular physiology, the MMG technique has been shown to be a promising alternative to assess muscle strength and fatigue.

By applying the protocol developed for the present study, it was possible to conclude that MMG may be applied concomitantly to NMES because it does not suffer electrical interference and can be used during functional movements and muscle contraction generated by NMES. Changes were identified in the MMG signal and in torque in the presence of fatigue. However, the MMG_{MPF} had little significance in determining muscle fatigue during the application of this NMES protocol (low coefficient of determination).

During the conduction of this study, limitations were found in the NMES equipment. The commercial equipment displayed limitations when applied to scientific purposes, such as difficulty manipulating the parameters and manual increment of current intensity. Nevertheless, this equipment was the most similar to those used in rehabilitation programs. There was also difficulty in recruiting healthy individuals with similar body type who were willing to take part in the research involving NMES.

References

1. de Ruiter CJ, Kooistra RD, Paalman MI, de Hann A. Initial phase of maximal voluntary and electrically stimulated knee extension torque development at different knee angles. J Appl Physiol. 2004;97(5):1693-701.

2. Chou LW, Ding J, Wexler AS, Binder-Macleod SA. Predicting optimal electrical stimulation for repetitive human muscle activation. J Electromyogr Kinesiol. 2005;15(3):300-9.

3. Tepavac D, Schwirtlich L. Detection and prediction of FES-induced fatigue. J Electromyogr Kinesiol. 1997;7(1):39-50.

4. Delitto A, Snyder-Macker L. Two theories of muscle strength augmentation using percutaneous electrical stimulation. Phys Ther. 1990;70(3):158-64.

5. Binder-Macleod SA, Halden EE, Jungles KA. Effects of stimulation intensity on the physiological responses of human motor units. Med Sci Sports Exerc. 1995;27(4):556-65.

6. Scott O. Ativação dos nervos motores e sensitivos. In: Kitchen S, Bazin S (editor). Eletroterapia de clayton. 10ª ed. São Paulo: Manole; 1998. p. 59-79.

7. Pincivero DM, Gear WS, Sterner RL. Assessment of the reliability of hight-intensity quadriceps femoris muscle fatigue. Med Sci Sports Exerc. 2001;33(2):334-8.

8. Yoshitake Y, Moritani T. The muscle sound properties of different muscle fiber types during voluntary and electrically induced contractions. J Electromyogr Kinesiol. 1999;9(3):209-17.

9. Nohama P, Macedo RM. Avaliação da fadiga muscular como função do lactato em paraplégicos submetidos à estimulação elétrica neuromuscular. III Congresso Latino Americano de Engenharia Biomédica; 2004 Set 22-25; UFPB. João Pessoa - PB: IFME Proceedings; 2004.

10. Beck TW, Housh TJ, Johnson GO, Cramer JT, Weir JP, Coburn JW, et al. Comparison of the fast fourier transform and continuous wavelet transform for examining mechanomyographic frequency versus eccentric torque relationships. J Neurosci Methods. 2006;150(1):59-66.

11. Beck TW, Housh TJ, Johnson GO, Weir JP, Cramer JT, Coburn JW, et al. Comparison of a piezoeletric contact sensor and an accelerometer for examining mechanomyographic amplitude and mean power frequency versus torque relationships during isokinetic and isometric muscle actions of biceps brachii. J Electromyogr Kinesiol. 2006;16(4):324-35.

12. Orizio C. Muscle sound: bases for the introduction of a mechanomyographic signal in muscle studies. Crit Rev Biomed Eng. 1993;21(3):201-43.

13. Gobbo M, Cè E, Diemont B, Esposito F, Orizio C. Torque and surface mechanomyogram parallel reduction during fatiguing stimulation in human muscles. Eur J Appl Physiol. 2006;97(1):9-15.

14. Preis C. Método de dinamometria isocinética para diagnóstico complementar na osteíte púbica asséptica não traumática [dissertação]. Curitiba (PR): Pontifícia Universidade Católica do Paraná; 2006.

15. Dvir Z. Isokinetics: muscle testing, interpretation and clinical applications. New York: Churchill Livingstone; 1995.

16. Ebersole KT, Housh TJ, Johnson GO, Evetovich TK, Smith DB, Perry SR. MMG and EMG responses of the superficial quadriceps femoris muscles. J Electromyogr Kinesiol. 1999;9(3):219-27.

17. Beck TW, Housh TJ, Johnson GO, Weir JP, Cramer JT, Coburn JW, et al. Mechanomyographic amplitude and mean power frequency versus torque relationships during isokinetic muscle actions of the biceps brachii. J Electromyogr Kinesiol. 2004;14(5):555-64.

18. Delitto A, McKowen JM, McCarthy JA, Shively RA, Rose SJ. Electrically elicited co-contraction of thigh musculature after anterior cruciate ligament surgery. A description and single-case experiment. Phy Ther. 1988;68(1):45-50.

19. Liebano RE, Alves LM. Comparação do índice de desconforto sensorial durante a aplicação da estimulação elétrica neuromuscular com correntes excitomotoras de baixa e média frequência em mulheres saudáveis. Rev Bras Med Esporte. 2009;15(1):50-3.

20. Low J, Reed A. Eletroterapia explicada: princípios e prática. 3ª ed. São Paulo: Manole; 2001.

21. Eser PC, Donaldson Nde N, Knecht H, Stüssi E. Influence of different stimulation frequencies on power output and fatigue during FES-cycling in recently injured SCI people. IEEE Trans Neural Syst Rehabil Eng. 2003;11(3):236-40.

22. Binder-Macleod SA, McDermond LR. Changes in force-frequency relationship of the human quadríceps femoris muscle following electrically and voluntary induced fatigue. Phys Ther. 1992;72(2):95-104.

23. Blangsted AK, Sjogaard G, Madeleine P, Olsen HB, Sogaard K. Voluntary low-force contraction elicits prolonged low-frequency fatigue and changes in surface electromyography and mechanomyography. J Electromyogr Kinesiol. 2005;15(2):138-48.

24. Rabischong E. Surface action potentials related to torque output in paraplegics' electrically stimulated quadriceps muscle. Med Eng Phys. 1996;18(7):538-47.

25. Ryan ED, Cramer JT, Egan AD, Hartman MJ, Herda TJ. Time and frequency domain responses of mechanomyogram and electromyogram during isometric ramp contractions: a comparison of the short-time fourier and continuous wavelet transforms. J Electromyogr Kinesiol. 2008;18(1): 54-67.

26. Faller L, Salles FA, Müller RW, Nohama P, Nogueira Neto GN, Button VLSN. Mecanomiografia por acelerometria em análise isocinética: resultados preliminares. 1º Encontro Nacional de Engenharia Biomecânica; 2007 Mai 23-25; Centro General Ernany Ayrosa - Itaipava. Petrópolis (RJ): INEBI; 2007.

27. Tarata MT. Mechanomyography versus electromyography, in monitoring the muscular fatigue. Biomed Eng Online. 2003;2:3.

28. Guyton AC, Hall JE. Tratado de fisiologia médica. 9ª ed. Rio de Janeiro: Guanabara Kogan; 1997.

29. Gerdle B, Larsson B, Karlsson S. Criterion validation of surface EMG variables as fatigue indicators using peak torque: a study of repetitive

maximum isokinetic knee extensions. J Electromyogr Kinesiol. 2000;10(4):225-32.

30. Winslow J, Jacobs PL, Tepavac D. Fatigue compensation during FES using surface EMG. J Electromyogr Kinesiol. 2003;13(6):555-68.

31. Pires KF. Análise do efeitos de diferentes protocolos de eletroestimulação através da frequencia mediana. Rev Bras Ciênc Mov. 2004;12(2):25-8.

32. Bajaj P, Madeleine P, Sjogaard G, Arendt-Nielsen L. Assessment to postexercise muscle soreness by electromyography and mechanomyography. J Pain. 2002;3(2):126-36.

33. Tarata M, Spaepen A, Puers R. The accelerometer MMG measurement approach, in monitoring the muscular fatigue. Measur Sci Rev. 2001; 1(1):47-50.

34. Ebersole KT, O'Connor KM, Wier AP. Mechanomyographic and electromyographic responses to repeated concentric muscle actions of the quadriceps femoris. J Electromyogr Kinesiol. 2006;16(2): 149-57.

35. Coburn JW, Housh TP, Cramer JT, Weir JP, Miller JM, Beck TW, et al. Mechanomyographic and electromyographic responses of the vastus medialis muscle during isometric and concentric muscle actions. J Strength Cond Res. 2005;19(2):412-20.

36. Al-Zahrani E, Gunasekaran C, Callaghan M, Gaydecki P, Benitez D, Oldham J. Within-day and between-days reliability of quadriceps isometric muscle fatigue using mechanomyography on healthy subjects. J Electromyogr Kinesiol. 2008;[Epub ahead of print].

37. Madeleine P, Farina D, Merletti R, Arendt-Nielsen L. Upper trapezius muscle mechanomyographic and electromyographic actovity in humans during low force fatiguing andn non-fatiguing contractions. Eur J Appl Physiol. 2002;87(4-5):327-36.

38. Madeleine P, Jorgensen LV, Sogaard K, Arendt-Nielsen L, Sjogaard G. Development of muscle fatigue as assessed by electromyography during continuous intermittent low-force contractions effects of the feedback mode. Eur J Appl Physiol. 2002;87(1):28-37.

39. Ding J, Storaska JA, Binder-Macleod SA. Effect of potentiation on the catchlike property of human skeletal muscles. Musc Nerve. 2003;27(3):312-9.

40. Yoshitake Y, Ue H, Miyazaki M, Moritani T. Assessment of lower-back muscle fatigue using electromyography, mechanomyography, and near-infrared spectroscopy. Eur J Appl Physiol. 2001;84(3):174-9.

41. Beck TW, Housh TJ, Cramer JT, Weir JP, Johnson GO, Coburn JW, et al. Mechanomyographic amplitude and frequency responses during dynamic muscle actions: a comprehensive review. Biomed Eng Online. 2005;4(1):67.