NONINVASIVE MONITORING OF NEUROMUSCULAR FATIGUE: TECHNIQUES & RESULTS

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Keywords: Area/Amplitude Ratio, Neuromuscular Fatigue, Electromyography, SEMG, Mechanomyography, MMG, Wavelet

Abstract. *The paper provides a functional, practically oriented overview of the concept and technical possibilities of monitoring the neuromuscular fatigue via the electromyographic and the mechanomyographic signals, as essentially related to the muscle contraction and intimately mirroring muscle activation and contraction mechanisms. The Fast Fourier Transform-based and Wavelet Transform-based techniques are critically analyzed, illustrated and discussed, on the international arena and through original work.*

INTRODUCTION

Muscle in voluntary contraction produces force, based on two mechanisms: (i) the firing frequency and (ii) the recruitment of the motor units. A motor unit is a functional entity consisting of a motor neuron and the whole set of muscular fibers it innervates. To increase the muscle force, either the firing frequency or the number of recruited motor units has to be increased [1]. The myoelectric activity detected with surface electrodes, surface electromyogram (SEMG), may be considered as the summation of the electrical signals generated by a number of motor units, active within the same motor territory in the proximity of the electrodes. The SEMG signal is a convenient mean to study the muscle behavior under fatiguing exercise, as it proves time-dependent changes, provided care is taken to prevent cross talk from adjacent muscles.

Sustained muscular contractions externally associated with not being able to maintain a certain force lead to physiological fatigue, tremor or pain, localized in the specific muscle (localized muscular fatigue). Fatigue is defined as 'any reduction in the force generating capacity, measured by maximum voluntary contraction (MVC), regardless of the task performed' [2]. Fatigue is associated with a compression of the power spectral density of the SEMG toward lower frequencies, from the very beginning of the voluntary contraction [3]. This is due to the reduction in the conduction velocity in direct relation with the muscular fiber membrane excitability and with neural adaptations, resulting in an increase of the lower frequency content of the signal. Fatigue is also associated with higher amplitudes of the SEMG signal toward the end of the exercise. It has been shown that, in sustained motor tasks, changes at different levels, including motoneural discharge behavior, develop before an endurance limit is reached - phenomenon called ‘central fatigue’ [4,5]. Central and peripheral fatigue develop together, and have to be seen not as a result, but as complementary elements of a complex strategy striving to insure the optimality of the motor behavior within the framework of available resources.

Under muscular contraction, mechanical vibrations occur [6] in the range 10 - 40 Hz. As a signal complementary to SEMG, the mechanomyogram (MMG) reflects the mechanical muscle vibrations generated by the spatio-temporal summation of the individual muscle fiber twitches, evoked through motor unit (MU) activation by the motor neurons.

SEMG and MMG, recorded simultaneously from the same muscles under steady contraction, show a compression of the spectra toward lower frequencies since the beginning of the contraction [7,8]. After an initial approach based on Fast Fourier Transform (FFT) techniques, in order to study transitional phenomena in muscle contraction and to monitor neuro muscular fatigue in a dynamic contraction, the use of the Wavelet Transform (WT) has been investigated, via Instantaneous Mean Frequency and Instantaneous Median Frequency [9]. The use of WT was shown on a rather limited scale until now, only for epochs where FFT can also be consistently used, i.e. windows of signal where no acceleration or deceleration occurs, situation which may occasionally happen in a steady contraction or in some isokinetic exercise. As an alternative, the author defined the Raa parameter (Area/Amplitude Ratio) [10,11], together with the Instantaneous Mean Scale (IMS) and Instantaneous Median Scale (IMedS) with the purpose to validate its use and to assess its computational efficiency in terms of speed and required memory space.

1. RECORDING SEMG AND MMG

The SEMG signal is usually recorded via two disposable self adhesive surface EMG conductive gel electrodes, (e. g. 22.5 x 22.5 mm H59P, MVAP, USA), with their centres 25 mm apart from each other, placed on abrased, clean skin, longitudinally, immediately under the thickest point of the muscle or close to the innervation point.

The SEMG signals were amplified (x 2000, 100 MΩ input impedance, 100 dB CMRR, 500 Hz antialias filter, Beckman R611, USA) and acquired together with the MMG signals via a computerized acquisition system (DAP1200 Microstar Laboratories USA), at 1000 Hz sampling rate on all the channels simultaneously.

The MMG was recorded with acceleration or sound transducers and recently with laser distance sensors. The technique of recording MMG was refined using piezoresistive silicon accelerometers in a surface mount package [12] to provide a reliable acquisition of MMG. An accelerometer (+-2g, ICS Sensors, model 3031,USA) and author-made original amplifier (x 50000, 10-250 Hz band pass filter, 250 Hz antialias filter) picks up the MMG. The accelerometer was placed between the SEMG electrodes to pick up the maximal MMG, orthogonal to the muscle from the same motor territory.

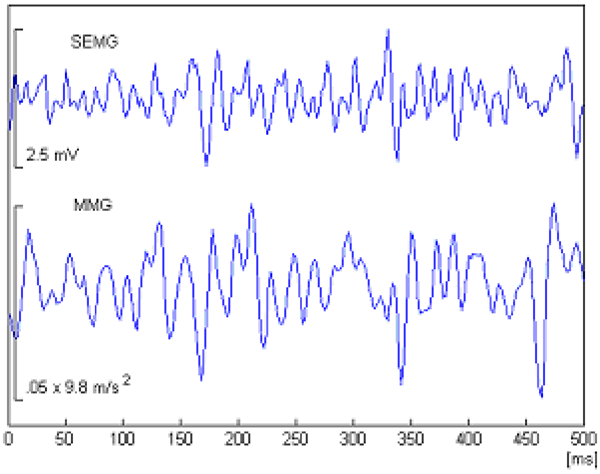

Fig. 1. SEMG and MMG recorded simultaneously from the same location of the Biceps muscle (Courtesy [11]).

The SEMG and MMG recorded simultaneously from the same muscle (Figure 1) have similar behavior [7,13], i.e. the median frequencies of SEMG and MMG decrease from the very beginning of the contraction, and their RMS values increase. Dedicated work, exploring the isometric steady contraction of different muscles, showed that the SEMG and MMG, recorded simultaneously from the same muscle, have similar behavior [7].

2. TECHNIQUES AND RESULTS

Until now, the power spectral density obtained via FFT was consistently used to compute mean (MNF) or median (MDF) frequencies to monitor activation in isometric steady contraction. A time window of 500 – 1000 ms was currently used, either for SEMG or for MMG in this type of contraction, thus fulfilling one major restriction in accurately using FFT to compute power spectral density from these signals, to ensure consistency across the data, according to the wide sense stationarity.

Because of this demand, FFT has to be cautiously used to compute power spectral density from SEMG or MMG. Therefore, to avoid the necessity to satisfy the wide sense stationarity condition, the use of WT has been investigated to monitor the dynamic contraction via Instantaneous Mean Frequency and Instantaneous Median Frequency. Equivalence of this approach with the use of FFT was shown on a limited scale until now, only for epochs where FFT can also be consistently used, i.e. windows of signal where no acceleration or deceleration occurs, which may happen in some isokinetic exercise [9].

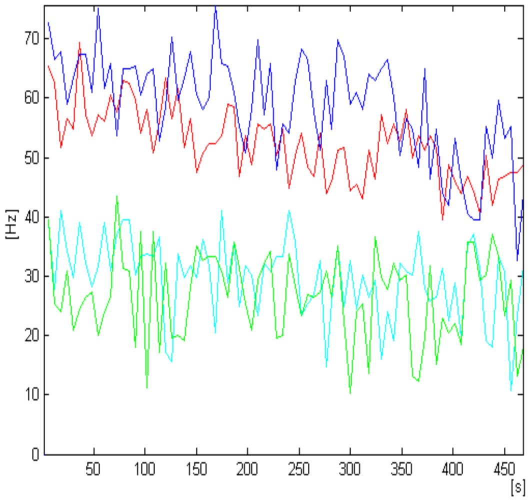

Fig. 2. SEMG MDF and MMG MDF evolutions with increasing fatigue, both for the Biceps and Brachioradialis muscles (red – Biceps SEMG MDF, blue – Brachioradialis SEMG MDF, cian – Biceps MMG MDF, green – Brachioradialis MMG MDF). An overall decrease can be noticed starting at the very beginning of the contraction, which proves the central component of the fatigue (Courtesy [11]).

Previous work [7] showed a compression of the spectra toward lower frequencies, with advancing fatigue of the spectra, both for SEMG and MMG (Figure 2). MNF, MDF, computed on a time window of 500 ms via the FFT from SEMG and MMG acquired from Biceps Brachii and Brachioradialis muscles under voluntary, steady contraction, decreased in all the subjects, from the very beginning of the contraction to its very end - when the task could not be sustained any more -, thus proving the consistence of SEMG and MMG signals in monitoring central fatigue.

We performed preliminary work on exploring the behavior of the IMS (Instantaneous Mean Scale) and IMedS (Instantaneous Median Scale) computed via the WT on epochs as short as 100 ms, from the SEMG and MMG signals (Figure 3), then on a sample_by_sample basis.

The Raa and IMedS were computed from the original signals (SEMG) for all the subjects, using a rectangular window.

Raa – Area /Amplitude Ratio, with a dimension of time [ms], is computed from the signal in the time domain, as the Area/Amplitude ratios over the considered epoch, calculated between consecutive transversals of the isoelectric line, called ‘phases’ (Tarata, 1997, 1994):

$$Raa = \frac{1}{n}\sum_{i=1}^{n}\frac{S_i}{A_i} \qquad (1)$$

with

n - the number of phases within the current epoch – on a limit case, each phase of the signal is considered an epoch -

S_i - current phase area, the integral of the i^{th} phase of the signal within the current signal segment,

A_i - the maximal amplitude of the i^{th} phase of the signal within the current signal segment, selected on all m samples within the current phase.

IMS (Instantaneous Mean Scale), IMedS (Instantaneous Median Scale) are computed via the Continuous Wavelet Transform (CWT). According to CWT a signal can be expressed as a linear combination of a particular set of wavelets, by shifting and dilating one unique function called ‘mother wavelet’ (MW). Different MWs can be used to approximate a given function, with efficiency depending on the chosen wavelet. Contrary to the FFT, where the localization of the events in time gets lost, specific to the use of wavelets is the time-frequency localization, because most of the energy of the wavelet is concentrated within a finite time interval. We use CWT as it works at every scale and preserves all the information in the signal *x(t)*.

With CWT, the Ψ(t) mother wavelet is smoothly shifted over the full domain of the analyzed signal x(t):

$$CWT_x(s,\tau)=\int x(t)\Psi^{*}_{s,\tau}(t)dt \qquad (2)$$

where

s - scale,

τ - translation (time or space shifting),

$\Psi^{*}_{s,\tau}$ - is obtained by scaling the Ψ(t) mother wavelet over τ and s; higher scales correspond to lower frequencies.

The power density function or Scalogram is:

$$SCAL(\tau,s)=|CWT_x(s,\tau)|^2 \qquad (3)$$

where

$SCAL(\tau,s)$ - is the time dependent scalogram.

IMedS - The instantaneous median scale of the scalogram, localized at a specific time moment, computed as the actual scale value separating the scalogram into two surface regions of equal area:

$$\int_0^{IMedS} SCAL(\tau,s)ds=\int_{IMedS}^{scal} SCAL(\tau,s)ds \qquad (4)$$

where

scal - is the maximal scale considered within the transform.

IMS - The instantaneous mean scale of the scalogram, localized at a specific moment *τ*, is:

$$IMS(\tau)=\frac{\int_0^{scal} s\cdot SCAL(\tau,s)ds}{\int_0^{scal} SCAL(\tau,s)ds} \qquad (5)$$

where

scal - is the considered maximal scale considered within the transform,

$SCAL(\tau,s)$ - is the time dependent scalogram.

This allowed a comparison with the evolution of Raa (Area /Amplitude Ratio), an original parameter computed from the time domain, with a dimension of time [10,11], which increases with increasing fatigue, from the very beginning of the contraction. Raa, IMS and IMedS show positive slopes in all subjects (SEMG and MMG), from the beginning of the contraction (Figure 4). This proves the central component of the fatigue.

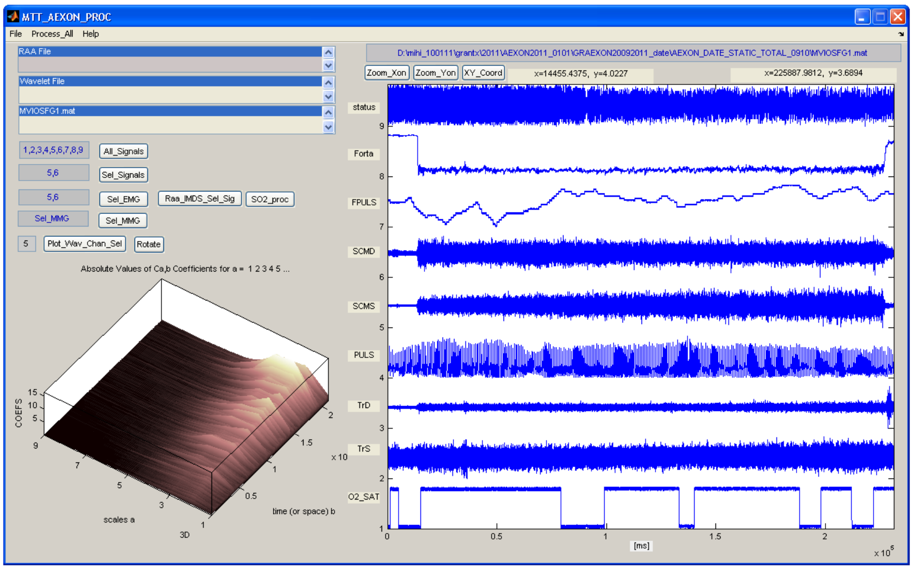

Fig. 3. MTT_AEXON_PROC GUI to process IMS (Instantaneous Mean Scale), IMedS (Instantaneous Median Scale) and Raa from the SEMG signal.

Experiments run on 12 female and 12 male subjects showed a steady increase (positive slopes) of IMedS (2.88e-003+-3.89e-003 [scale/s]; 3.95 +-0.29 [scales]), Raa (1.23e-003 +-1.491e-003 [ms/s]; 1.88+- 0.22 [ms]) and a decrease (negative slope) of SaO_2 (-3.3299e-003+- 9.5170e-003 [%/s]; 97.92+- 0.95 [%]) from the beginning of the contraction up to exhaustion.

3. CONCLUSIONS

The Raa, IMS and IMedS increase with advancing fatigue may be the effect of the centrally generated progressive alteration of the activation of individual MNs [14], possibly explained by the withdrawal of the tonic fusimotor driven spindle-support via the fusimotor loop [17], mechanism responsible for up to 30 % intervention, as the muscle afferents provide up to 30% excitation to the MNs.

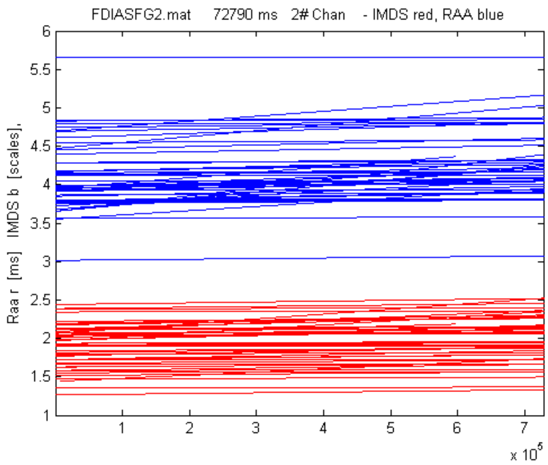

Fig. 4. The Raa and IMedS (IMDS) increase with advancing fatigue

During dynamic muscle actions, (i) the MMG signal provides valid information regarding muscle function, (ii) SEMG and MMG provide complementary information about the electrical and mechanical activity of the muscle [16,17].

All these findings advocate the use of MMG alone or together with SEMG, further underlining the intimate link between the SEMG and MMG as functionally related signals, both witnessing the neural activation of the muscular fibers and its mechanical effect, respectively.

Characterizing the dynamic contraction is only possible via parameters computed on very short epochs, thus being able to appropriately explore transitional episodes of muscle contraction or relaxation, which naturally alternate within the muscle function during normal work.

WT is able to meet such challenges, yet some comments are needed regarding whether to use the Discrete (DWT) or the Continuous Wavelet Transform (CWT) and what mother wavelet to use. We chose to use CWT to compute IMS and IMedS, because it works at every scale and preserves all the information within the signal.

Our choice was the 'Mexican Hat', chosen from a set of wavelets (coif5, db3, db4, gaus5, mexhat, meyr, morl, rbio3.5) after a selection based on sensitivity, by computing the ratio of variation of IMS and IMedS, over their maximal value. The 'Mexican Hat' mother wavelet gave an average ratio of 15%±4% comparing to 9%±4% for the others, therefore showing a higher sensitivity, while satisfying the admissibility condition and showing excellent localization in time and frequency.

This work opens way to monitoring neuromuscular fatigue in any type of exercise, in difficult environments and activities.

4. REFERENCES

[1]. De Luca CJ. Control properties of motor units. J. of Experimental Biology, 115:125-136, 1985.
[2]. Bigland-Ritchie B, Woods JJ. Changes in muscle contractile properties and neural control during human muscular fatigue. Muscle & Nerve, 7: 691-699, 1984.
[3]. Stulen FB, De Luca CJ. Frequency parameters of the myoelectric signal as a measure of muscle conduction velocity. IEEE Trans. Biomed. Eng., BME-28: 515-522, 1981.
[4]. Bigland-Ritchie B et al: Central and peripheral fatigue in sustained maximum voluntary contractions of human quadriceps muscle. Clinical Science and Molecular Medicine, 54: 609-614, 1978.
[5]. Gandevia SC et al. Central fatigue: critical issues, quantification and practical implications. Adv. Exp. Med. Biol., 384: 281-294, 1995.

[6]. Barry DT. Vibrations and sounds from evoked muscle twitches. *Electromyogr. &Clin. Neurophysiol.*, 32: 35-40, 1992.

[7]. Tarata MT. Mechanomyography versus Electromyography, in monitoring the muscular fatigue. Biomed Eng Online, 2:3, 2003.

[8]. Madeleine P, Jørgensen LV, Søgaard K, Arendt-Nielsen L, Sjøgaard G. Development of muscle fatigue as assessed by electromyography and mechanomyography during continuous and intermittent low-force contractions: effects of the feedback mode. Eur J Appl Physiol, 87:28–37, 2002.

[9]. Beck TW, Housh TJ, Johnson GO, Weir JP, Cramer JT, Coburn JW, Malek MH. Comparison of Fourier and wavelet transform procedures for examining the mechanomyographic and electromyographic frequency domain responses during fatiguing isokinetic muscle actions of the biceps brachii. J Electromyogr Kinesiol, 15(2):190-199, 2005.

[10]. Tarata MT. The Average Area/Amplitude Ratio (Raa), A Consistent Parameter in the Quantitative Analysis of the Electromyogram. In Proceedings of the International Conference on Medical Physics & Biomedical Engineering MPBE'94 May 5-7. Nikos Milonas, Nicosia, 1994, 53-57.

[11]. Tarata MT. Monitoring the Evolution of the Muscular Fatigue, Via New Parameters developed from the SEMG Signal. In Proceedings of the ECSAP'97 The First European Conference on Signal Analysis and Prediction 1997 June 24-27. ICT Press, Prague 431-434.

[12]. Tarata MT, Spaepen A, Puers R. The Accelerometer MMG Measurement Approach in Monitoring the Muscular Fatigue, MEASUREMENT SCIENCE REVIEW, ISSN 1335-8871, http://www.measurement.sk, 1: 47-50, 2001.

[13]. Tarata MT. Sensorimotor interactions within the context of muscle fatigue. In "Sensorimotor Control" (Dengler R., Kossev A., eds.), NATO Science Series, Series 1: Lifeand Behavioural Sciences 2001, 326: 84-91.

[14]. Orizio C, Gobbo M, Diemont B, Esposito F, Veicsteinas A. The surface mechanomyogram as a tool to describe the influence of fatigue on biceps brachii motor unit activation strategy. Historical basis and novel evidence. *Eur. J. Appl. Physiol.*, 90:326–336, 2003.

[15]. Hagbarth KE, Macefield VG. The fusimotor system. Its role in fatigue. In Gandevia SC editor. Fatigue. Plenum Press: New York, 1995; 259-70.

[16]. Madeleine P, Bajaj P, Søgaard K, Arendt-Nielsen L. Mechanomyography and electromyography force relationships during concentric, isometric and eccentric contractions. *J. Electromyogr.Kinesiol.*, 11:113–21, 2001.

[17]. Beck TW, Housh TJ, Johnson GO, Cramer JT, Weir JP, Coburn JW, Malek MH. Does the frequency content of the surface mechanomyographic signal reflect motor unit firing rates? A brief review. J. Electromyogr. Kinesiol., 2007; 17(1):1-13.