The Electromyogram and Mechanomyogram in Monitoring Neuromuscular Fatigue: Techniques, Results, Potential Use within the Dynamic Effort

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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 analysed, illustrated and discussed, on the international arena and through original work. An original technique - the Area/Amplitude Ratio (Raa) - is also comparatively discussed, as a practical alternative instrument in monitoring the neuromuscular fatigue in a dynamic contraction - important issue in real life activity - , e.g. work in difficult environments, pilots in mission, difficult work in normal environments.*

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

1. 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. The actual muscle force is built up from individual muscle fiber twitches, the smoothness of the force output being enhanced by the firing rate / recruitment interaction within the motor unit pool [1] and the mechanical filtering effect of the tissue, compensating for the discrete nature of the process, The variation of the firing rates of the motor units occurs simultaneously for all the motor units within a muscle, and even in different muscles acting on the same joint, according to the phenomenon of ’common drive’ [2], therefore small variations of the output force may occur. The control loops involving Ia, Ib afferents are also responsible for the occurrence of certain peaks within the output force in the isometric exercise, and others are due to the activation from higher levels [3].

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' [4]. The failure in maintaining the motor task defines the endurance limit; endurance time is the total duration of the task up to the endurance limit. 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 [5]. 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' [6, 7]. 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, due to three main processes: (i) the inner muscular vibrations, which are the intrinsic components of the muscle contraction [8], (ii) oscillations of the human motor system, e.g. tremor and clonus [9], and (iii) artefacts. They are located in specific frequency ranges, with a certain overlapping: the artefacts - due to large movements - show the lowest frequencies, possible tremor contributions are below 10 Hz, in healthy subjects usually between 5.85 and 8.8 Hz [10], and the mechanical inner vibrations affect the range between 10 and 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. The MMG may be considered to reflect the mechanical muscle vibrations generated by the spatio-temporal summation of the individual muscle fiber twitches which are evoked through 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 [11, 12]. 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 [13]. 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 explored the use of the Raa parameter (Area/Amplitude Ratio) [14, 15], 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.

2. 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 [16] 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.

The SEMG and MMG recorded simultaneously from the same muscle (Figure 1) have similar behavior [11, 17], i.e. the median frequencies of SEMG and MMG decrease from the very beginning of the contraction, and their RMS values increase.

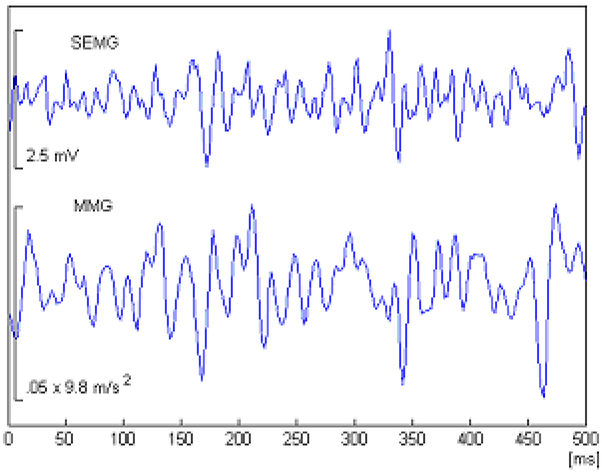

Fig. 1. SEMG and MMG recorded simultaneously from the same location of the Biceps muscle (Courtesy [11]).

Recent research showed the MMG to be a reliable signal in studying the development of fusion [18] and the changes in muscle contractile properties during repetitive unfused contractions [19], as well. Following preliminary work with refined techniques using piezoresistive silicon accelerometers in a surface mount package [15, 16], while the use of accelerometers has been further validated [20] for recording MMG, 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 [11].

3. 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 [21, 11, 22, 23, 24, 25]. 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 [26, 27, 11].

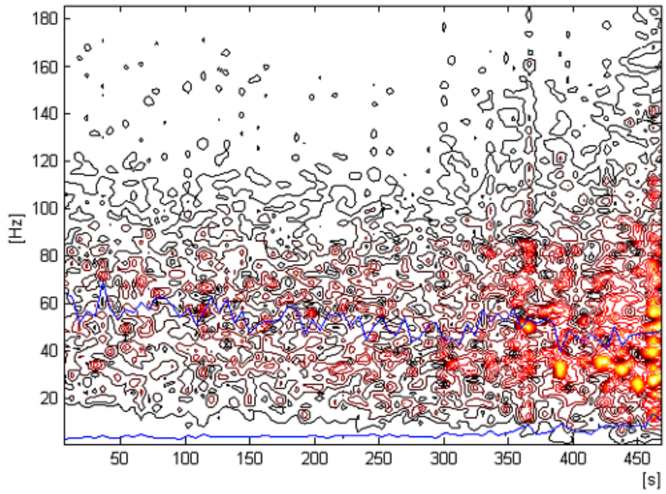

Fig. 2. SEMG power spectral density evolution in time (Biceps muscle) – contour plot of 15 levels isolines of the matrix containing the succession of the spectra, from the beginning to the end of the contraction (black – zero level; red, yellow - higher power spectral density peaks). The power spectral density compression towards lower frequency and higher peaks by the end of contraction, can be seen. The middle trace (blue) shows the SEMG MDF evolution; MDF decreases with increasing fatigue, thus proving the power spectral density compression. The lower trace (blue) shows the SEMG RMS evolution; RMS increases with increasing fatigue (Courtesy [11]).

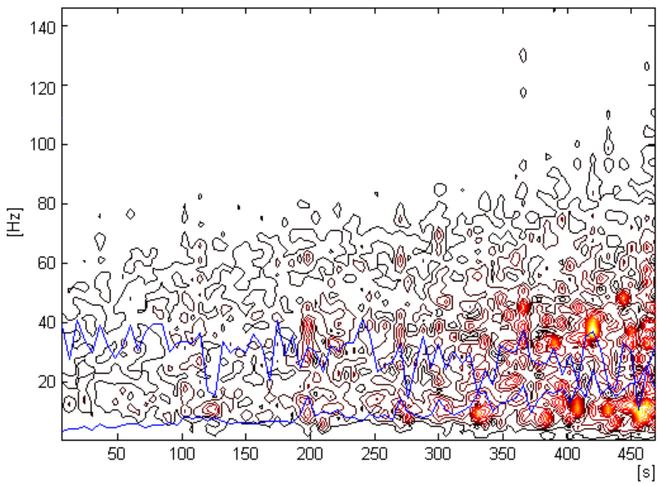

Fig. 3. MMG power spectral density evolution in time (Biceps muscle) – contour plot of 15 levels isolines of the matrix containing the succession of the spectra, from the beginning to the end of the contraction (black – zero level; red, yellow - higher power spectral density peaks). The power spectral density compression towards lower frequency and higher PSD peaks by the end of contraction, can be seen. The middle trace (blue) shows the MMG MDF evolution; MDF decreases with increasing fatigue, thus proving the power spectral density compression. The lower trace (blue) shows the MMG RMS evolution; RMS increases with increasing fatigue (Courtesy [11]).

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 [28, 29, 30, 31]. 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 [13].

Previous work [11] showed a compression of the spectra toward lower frequencies, with advancing fatigue of the spectra, both for SEMG and MMG (Figures 2, 3). 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 (Figures 2, 3, 4).

Even when the wide sense stationarity is satisfied during an isometric steady contraction for time windows of 500 – 1000 ms, while the FFT approach stands in this case, other techniques are necessary to allow the exploration of the signals within time windows shorter than 500 ms, e.g. in order to study local variations in activation, with not speaking of the dynamic contraction associated with real life work, when the FFT approach fails.

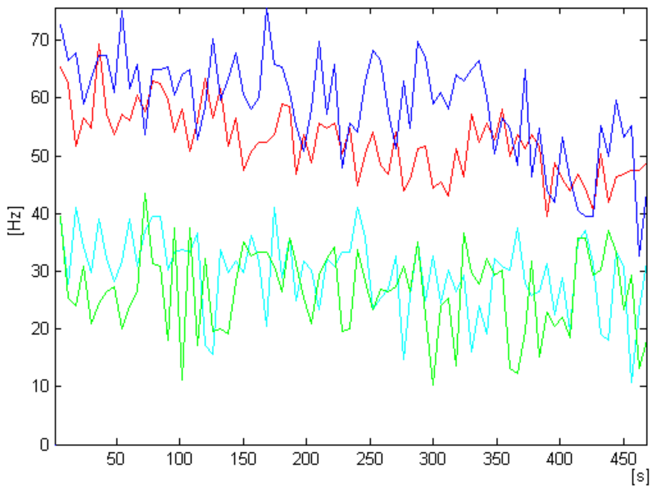

Fig. 4. SEMG MDF and MMG MDF evolutions with increasing fatigue, both for the Biceps and Brachioradialis muscles (red – Biceps SEMG MDF, blue – Brachioradialis SEMG MDF, cian – Biceps MMG MDF, green – Brachioradialis MMG MDF). An overall decrease can be noticed starting at the very beginning of the contraction, which proves the central component of the fatigue (Courtesy [11]).

The WT approach could successfully demonstrate the compression of the spectra toward lower frequencies in neuromuscular fatigue, on epochs of enough time length to allow the computation of FFT based parameters, too.

To explore shorter epochs 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 acquired from the same muscle (Biceps Brachii) under voluntary contraction. 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 [14,15], which increases with increasing fatigue, from the very beginning of the contraction.

Raa – Average Area /Amplitude Ratio, with a dimension of time [ms], is computed from the signal in the time domain, as an average of the Area/Amplitude ratios over the considered epochs, calculated between consecutive transversals of the isoelectric line, called 'phases' [14, 15]:

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

with

n - the number of phases within the current epoch,

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.

The advantage of Raa is its computational efficiency, comparing to using FFT or Wavelet techniques. As an example, figure 5 shows the evolution of the Area/Amplitude Ratio (Raa), IMS and IMedS with advancing fatigue, for both the SEMG and MMG – computed on 100 ms epochs -. All three parameters increase with advancing NMF.

Raa, IMS and IMedS show positive slopes in all subjects (SEMG and MMG), from the beginning of the contraction, for all the epoch lengths. This proves the central component of the fatigue.

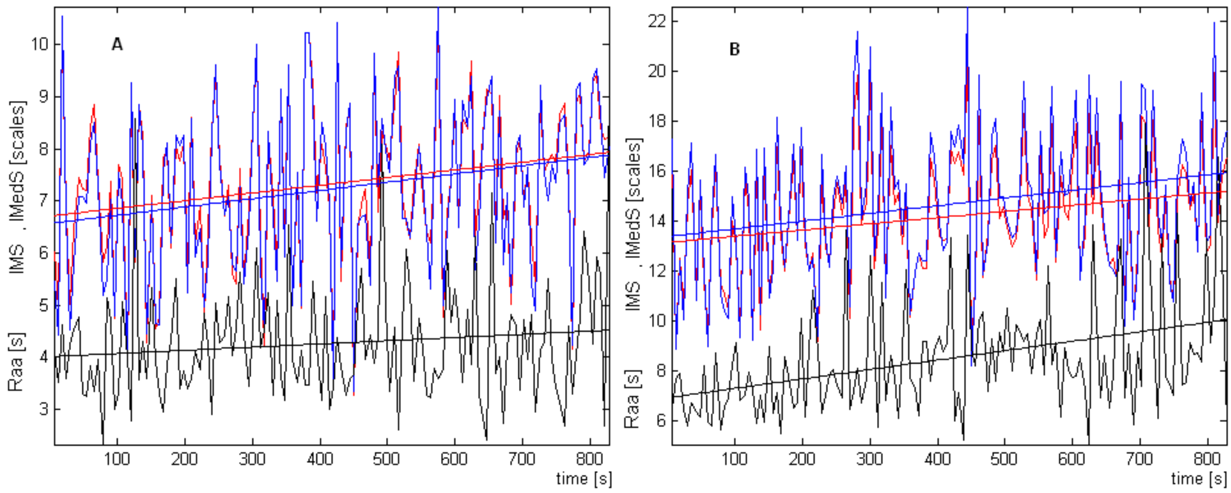

Fig. 5. The evolution of Area/Amplitude Ratio (Raa), Instantaneous Mean Scale (IMS – red), Instantaneous Median Scale (IMedS - blue) with advancing fatigue, for the SEMG (A) and MMG (B), on PTW 100ms.

The two-way analysis of variance (ANOVA), performed separately on the SEMG and MMG, shows (i) no significant differences between the slope of the linear interpolation of the evolution means either among the epochs or between sexes and (ii) no interaction between any epoch and any of the subject sexes - the probability values were greater than 0.05 -, for Raa, IMS, and IMedS as well. Raa approach was 5.52 ±0.97 times quicker than the WT. For the given number of scales considered (30) the memory required for IMS, IMedS is 285±57 times larger than for Raa.

4. Discussion and 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 [33], possibly explained by the withdrawal of the tonic fusimotor driven spindle-support via the fusimotor loop [34], mechanism responsible for up to 30 % intervention, as the muscle afferents provide up to 30% excitation to the MNs. Other contributors may be (i) the presynaptic inhibition, which affects the Ia and Ib afferents, (ii) the group II non-spindle muscle afferents which have increased discharge rates with increasing fatigue, further contributing to the inhibitory effect, (iii) group III and IV muscle afferents, activated by the accumulation of metabolites in the fatiguing muscle, thus centrally inhibiting the alpha MNs. Within this context, the critical role of somatosensory feedback from working muscles on the centrally mediated determination of central motor drive and power output has been emphasized [35], so that the development of peripheral muscle fatigue is confined to a certain level, by having shown that attenuated afferent feedback from exercising locomotor muscles results in an overshoot in central motor drive and power output normally chosen by the subject, thereby causing a greater rate of accumulation of muscle metabolites and excessive development of peripheral muscle fatigue.

Together with our findings, other reports concluded that (i) during dynamic muscle actions, 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 [36, 12], (iii) MMG MNF (IMS, IMedS in our case) can vary during fatiguing sustained isometric muscle action, despite a constant torque level, thus providing different information from the torque signal, and decreases in MMG mean or median frequency may reflect reductions in the global motor unit firing rate, rather than decreases in the firing rates of individual motor units [37], (iv) fatigue induced decreases in MMG MNF may reflect reductions in MU firing rates and/or derecruitment of fast twitch MU [38].

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. While the potential for cross-talk in surface MMG is relatively small even for muscles close to each other and having a common innervation [39] and the MU activation strategy might be estimated in more detail by the MMG than by the SEMG [40], the use of MMG is furthermore encouraged and justified.

It was recognized that SEMG and 'MMG signals recorded during dynamic muscle actions could require different signal processing methodologies when compared to isometric muscle actions' [32]. It was to the purpose of finding other processing techniques to cope with the non-stationarity inherent to the both signals, mainly in the dynamic contraction, that the FFT approach has been progressively replaced by the WT approach.

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.

Different authors used the DWT [29, 30] with different mother wavelets: Daubechies db2, db3 [29], db4 [13], db5 [30], db10 [32], Morlet [31], Sym 4, Sym5 [29], as alternatives to FFT, to be able to deal with stronger nonstationarities which were observed in the SEMG and MMG signals from a dynamic contraction.

We chose to use CWT to compute IMS and IMedS because it works at every scale and preserves all the information within the signal [41]. Provided the exact selection of the mother wavelet is not critical in CWT given that each MW gives approximately the same qualitative results [42], 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 [43].

Concluding on the use of WT, one of the problems - specific to using this approach - is the complexity of the mathematical instrument, associated with the need to use appropriate criteria to choose a proper mother wavelet, making such methods over-complete (redundant) and therefore inefficient in terms of computational time or storage space [28].

The Raa parameter overcomes such problems by its simple definition [14, 15] and proves a similar behavior in monitoring NMF, with IMS and IMedS computed via WT. Our experiments demonstrate that Raa has a similar behavior with IMS and IMedS (Figure 5), with increasing fatigue, with no significant differences between slopes, neither among the PTWs, nor between sexes , while being quicker than the WT approach and requiring less memory. Therefore it is a convenient alternative to monitor the development of neuromuscular fatigue, mostly in a dynamic contraction, due to the possibility to be computed on short epochs (100 ms).

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

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