THE ACCELEROMETER MMG MEASUREMENT APPROACH, IN MONITORING THE MUSCULAR FATIGUE

M. Tarata*, A. Spaepen, R. Puers *****

* Medical Informatics, University of Medicine and Pharmacy, Craiova, Romania, mihai@umfcv.ro
** Laboratory of Ergonomy and Occupational Biomechanics, Catholic University of Leuven, Belgium, arthur.spaepen@flok.kuleuven.ac.be
***ESAT-MICAS, Catholic University of Leuven, Belgium, robert.puers@esat.kuleuven.ac.be

Abstract: *This research aimed to appreciate whether the mechanomyogram (MMG) may be used to monitor the muscle behavior under fatiguing exercise. The electromyogram (EMG) and MMG were simultaneously recorded from the Biceps and from the Brachioradialis muscles, with surface electrodes - EMG, and accelerometer based transducers – MMG, from sixteen subjects. The conclusions are consistent in validating the use of the median frequency (MF) MMG to monitor the development of the muscular fatigue in the isometric isotonic exercise, at 25 % maximal voluntary contraction (MVC). The same as for the MF EMG, the MF MMG steadily decreases from the very beginning of the exercise, down to exhaustion, again demonstrating the central component of the muscular fatigue.*

1. INTRODUCTION

The 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. This is the so called localized muscular fatigue (Basmajian, 1985), associated with a compression of the power spectral density (PSD) of the EMG towards lower frequencies (Stulen et al., 1981, Tarata, 1994), measured by MF computed from the EMG PSD. This is due to the reduction in the conduction velocity in direct relation with the membrane excitability and with neural adaptations, resulting in an increase of the lower frequency content of the signal. It has been shown that, in sustained motor tasks, changes at different levels, including motoneural discharge, develop before an endurance limit is reached, the phenomenon called central fatigue (Gandevia, 1998). When considering an isometric contraction, mechanical vibrations occur, due to three main processes: the inner muscular vibrations, which are the intrinsic components of the muscle contraction (Barry, 1992), oscillations of the human motor system, e.g. tremor and clonus (Iaizzo et al., 1992), and artefacts. They are located in specific frequency ranges, with a certain overlapping: the artefacts - due to large movements- the lowest frequencies, tremor - in humans has been located under 10Hz independent of subject, between 5.85 and 8.8 Hz (Comby, 1992), and the mechanical inner vibrations between 10 and 40 Hz. It has been shown that the EMG and MMG recorded simultaneously from the same muscle have similar behavior (Tarata, 1997, Tarata et al., 1997, 1999). This research aimes to appreciate whether the MMG may be used to monitor the muscle behavior under fatiguing exercise.

2. METHOD

Seven female and nine male voluntary, healthy, and motivated subjects participated, each group aged between 22 and 40. A pair of Ag/AgCl, 15 mm diameter surface EMG disk electrodes, with their centers 30 mm apart from each other were placed on abraded, clean skin, using a conductive gel interface, longitudinally, immediately under the thickest point of the Biceps. A second pair of electrodes was placed on the Brachioradialis muscle of the same arm, under the same technical conditions. One accelerometer (+- 2g, ICSensors 3031, USA) was placed on each muscle, between the EMG electrodes, to pick up the MMG on the axis orthogonal to the muscle. At first, the 100 % MVC (maximal voluntary contraction) was estimated for the subject being tested, as the average of the maxima that the subject could develop for two seconds, after a few short preliminary trials for learning. Then sustained isometric isotonic contraction was performed at 25 % MVC, until exhaustion. The signals were amplified (EMG - TEMPS, K.U.Leuven, Belgium: x 2000, 300 MΩ input impedance, 100 dB CMRR, 250 Hz antialias filter, MMG – macromodel amplifier x 5000, 100 dB CMRR, 150 Hz antialias filter). They were acquired via a computerized acquisition system (Labmaster), at 500 Hz sampling rate on all the channels simultaneously. The acquisition was performed in time segments of 500 ms, every 6 seconds. The 500 ms buffer length insures the stationarity of the data. For each buffer, the average MF was computed via the zero padding generated 1024 point FFT (Fast Fourier Transform). MF is a parameter computed in the frequency domain from the power spectral density for each data buffer, representing the frequency, which separates the power spectral density graph into two parts of equal area, and has been consistently described as a good indicator of the muscular fatigue. For all the MF evolutions the linear regression was performed, to compute the slope a1 and intercept point a0 parameters:

$$MF_e = ao + a1 \bullet t \tag{1}$$

where: a0 – intercept point; a1 – the slope; t – time [ms].

3.RESULTS

Figure 1 shows an example of EMG and MMG signals, simultaneously recorded under the conditions above described. The both signals being recorded from the same motor territory, a certain synchronism is expected between the electrical activity – as a cause and the mechanical activity – as an effect. Such a synchronism can be seen in figure 1, with the observation that mostly the negative EMG peaks are synchronous with the positive peaks of the MMG, within the limits of a few milliseconds delay.

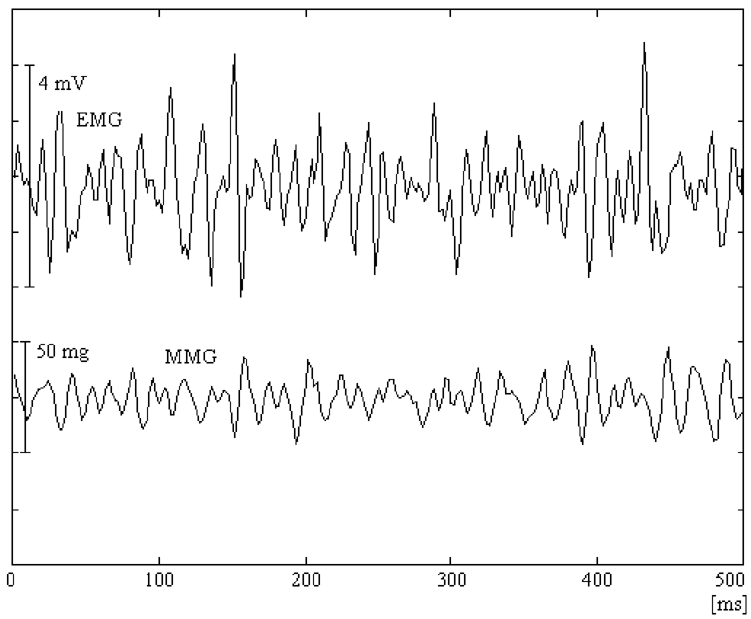

Figure 1. EMG (up) and MMG (down) signals in one 500 ms buffer. Many peaks are synchronous within the limits of a few milliseconds delay between the electrical activation and the mechanical output.

Figure 2 proves that MF decreases from the very beginning of the contraction, for the EMG and also for the MMG, for the both muscles involved in the effort at the studied joint, while the average output force slightly oscillates around 25% MVC.

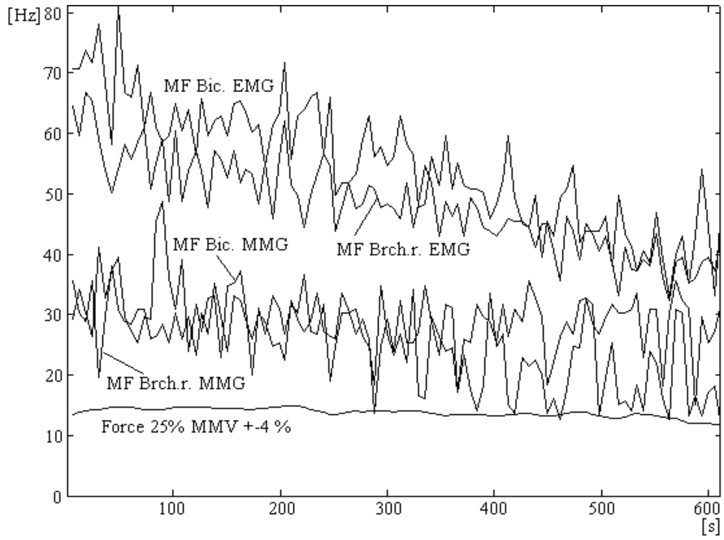

Figure 2. The evolution of the MF of the EMG and MMG signals for one subject. The force, averaged on each buffer slightly oscillates around 25% MVC. MF decreases from the very beginning of the exercise, both for the EMG and MMG, for the both muscles involved.

Table 1. The mean and standard deviation of the linear regression parameters of the MF evolution for the fatiguing exercise at 25% MVC

MF [Hz]	EMG Bic.		EMG Brch.r.		MMG Bic.		MMG Brch.r.	
	a1	a0	a1	a0	a1	a0	a1	a0
W mean	-0.034	61.017	-0.033	56.423	-0.017	25.591	-0.008	22.587
W sd (+-)	0.032	11.436	0.017	6.109	0.014	5.323	0.005	2.436
M mean	-0.025	59.471	-0.027	61.104	-0.017	32.910	-0.009	26.087
M sd (+-)	0.018	6.801	0.022	5.722	0.021	4.483	0.018	2.196
T mean	-0.029	60.147	-0.030	59.056	-0.017	29.708	-0.008	24.556
T sd (+-)	0.024	8.810	0.019	6.176	0.018	6.010	0.014	2.857

OBS. * W – women, M – men, T – men & women together, sd – standard deviation
** a1 – the slope, a0 – the intercept point

The mean and standard deviation of the linear regression parameters of the MF evolution for the fatiguing exercise at 25% MVC are shown in table 1, for women, men, and together, respectively.

Table 2. The minimum and maximum values of the intercept point a0 for the linear regression of the MF evolution for the fatiguing exercise at 25% MVC, for women and men

MF [Hz]	EMG Bic.		EMG Brch.r.		MMG Bic.		MMG Brch.r.	
	min	max	min	max	min	max	min	max
W	45.523	83.509	49.188	66.261	20.352	34.902	18.111	25.105
M	50.217	70.697	52.414	68.397	24.732	39.578	21.815	28.918

Table 2 details the minim and maxim of the MF intercept points.

4. DISCUSSION AND CONCLUSIONS

1. The work proves for the first time, a similar global behavior of the two signals MMG and EMG, respectively. Thus, the possibility of a quantitative characterization of the behavior of the muscle under fatiguing isometric exercise at submaximal levels (25% MVC), via the evolution of MF computed from MMG, is explored.
2. MMG MF decreases from the very beginning of the contraction, down to failure, proving, the same as the EMG MF, the central component of the fatigue.
3. The slopes of the MMG MF evolution are negative, for all the subjects, for the both muscles involved. The MMG MF slopes are closer to each other between men and women (Bic. -.017, -.017/Brch.r. -.008,-.009), than the EMG MF slopes (Bic. -.025, -.034/Brch.r. -.027,-.033). A reason may be the fact that the mechanical effect is a spatial summation over a larger motor territory.
4. The difference between the two muscles, in terms of MF slope, is higher for the MMG than for the EMG independently of sex.

All the results show consistency with previous research (Tarata, 1994,1997), i.e. the compression of the power spectral density towards lower frequencies, with advancing fatigue from the very beginning of the contraction, both for the EMG and the MMG signals.
From all the results above detailed emerges the conclusion that the MMG signal is at least as reliable as the EMG in monitoring the muscular fatigue. This may be important in experimental conditions where the recording of the EMG is difficult.

5. REFERENCES

Barry D.T. Vibrations and sounds from evoked muscle twitches, *Electromyogr. & Clin. Neurophysiol.*, 32, pp.35-40,1992.

Basmajian J.V., DeLuca C.J. *Muscles Alive. Their functions revealed by Electromyography.* William & Wilkins, Baltimore, 1985.

Comby B., Chevalier G., Bouchoucha M. A new method for the measurement of tremor at rest, *Archives Internationales de Physiologie, de Biochimie et de Biophysique*, 100, pp.73-78, 1992.

Gandevia S.C. et al. Neurobiology of Muscle Fatigue, Chapter 37, in FATIGUE, Gandevia S.C. et al. (eds.), Plenum Press, New York and London, 1995.

Gandevia, SC. Neural control in human muscle fatigue: changes in muscle afferents, motoneurons and moto cortical drive, *Acta Physiologica Scandinavica*, 1998:162(3):275-83.

Iaizzo P.A., Pozos R.S. Analysis of multiple EMG and acceleration signals of various record lengths as a means to study pathological and physiological oscillations, *Electromyogr. & Clin. Neurophysiol.*, 32, pp. 359-367, 1992.

Stulen F.B., DeLuca C.J. Frequency Parameters of the Myoelectric Signal as a Measure of Muscle Conduction Velocity, *IEEE Transactions on Biomedical Engineering*, vol. BME-28, No.7, pp. 515-523, 1981.

Tarata M. 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,* Nicosia, Cyprus, May, pp. 53-57, 1994.

Tarata, M., Spaepen, A., Puers, R., Hermans, Veerle. The Potential of Simultaneously Recording the EMG and Muscle Mechanical Vibrations in Monitoring the Muscle Behavior, in Proceedings of the International Conference on Measurement, Smolenice, Slovak Republic, May 29-31, 1997, pp. s1-s4.

Tarata, M. Monitoring the Evolution of the Muscular Fatigue, Via New Parameters developed from the EMG Signal, proceedings of the ECSAP'97 The First European Conference on Signal Analysis and Prediction, Prague, Czech Republic, 24-27 June 1997, p. 431- 434.

Tarata, M., Spaepen, A. A new MATLAB assisted approach to study the changes in the EMG and MMV signals in submaximal efforts, Proceedings of The 2nd Benelux Matlab Usersconference, Brussel, 24-25 March 1999, ch. 20.