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Electromyography in the Study of Muscle Reactions to Vibration Treatment

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1. Introduction

Electromyography (EMG) is a common used technique to evaluate muscular activity. Analysis of EMG recordings is important for assessing muscle activation, its relationship to the force developed during specific tasks and for evaluating fatigue processes occurring in response to physical activity.

Electromyography can be performed using different types of electrodes, depending on the specific analysis: surface (or skin) electrodes or inserted electrodes (wire and needle); the first it is used to monitor the overall activity of a muscle while the second is generally used to reveal the electrical activity of a nerve root. (De Luca, 1997, Basmajian and De Luca, 1985)

Electrode types and configurations, as well as associated instrumentation, influence the quality of the EMG signal detected and displayed, recorded or processed (Merletti et al, 2001; Saitou et al, 2000; Rainoldi et al, 2004, Nishihara et al, 2008). Various studies have been dedicated to the matter and guidelines in EMG recording are available (Basmajian and De Luca, 1985, Hermens H.J. et al, 1999).

Surface electromyography (SEMG) analysis is a largely used EMG recording method as it is non-invasive, safe, it does not cause pain and it is simple to perform. Root mean square (RMS) of the surface EMG signals is often used as a concise quantitative index of muscle activity; indeed, electromyography devices often provide EMG RMS output.

SEMG is often used for the assessment of muscle activity occurring in response to physiological or to externally applied stimuli, i.e. vibratory stimulation.

Vibration stimulus is a mechanical muscle excitation, applied generally to a tendon, a muscle or to the body as a whole, aimed to activate muscles by eliciting stretch reflexes.

Local tendon vibrations induce activity of the muscle spindle Ia fibers, mediated by monosynaptic and/or polysynaptic pathways; the reflex muscle contraction that arises in response to such vibratory stimulus has been named Tonic Vibration Reflex (TVR). (Roll et al, 1989; Bongiovanni and Hagbart, 1990; Romaguère et al, 1991; Person and Kozhina, 1992; Martin and Park, 1997)

As well as in other external stimulation, vibratory muscle activation can be examined by the analysis of electromyography recordings. Many studies report a significant increase of EMG RMS values in the lower body muscles during vibration training, these changes suggested

an increase in neuromuscular activity (Cardinale and Bosco, 2003; Verschueren et al, 2004). Specific WBV frequencies seem to produce a higher EMG RMS signal than others (Cardinale and Lim 2003).

However, as well as in every surface bio-potential recording, during local or whole body vibration treatment the EMG signal can be affected by artifacts.

Motion artifacts may in fact arise from relative motion between electrodes and skin and also between skin layers. The only skin stretch may result in a variation of electrode potential (Turker, 1993, De Talhouet and Webster, 1996; Ödman and Öberg, 1982, Searle and Kirkup, 2000, Tam and Webster, 1977).

In classical clinical EMG recordings (isokinetic, isotonic, gait, etc.), frequency content of motion artifact is considered below 10-20 Hz, then the general approach to motion artifact reduction is to apply a high-pass filter (e.g. with a cut-off frequency of 20 Hz).

During vibratory stimulation the artifact frequency contents, typically limited at vibratory frequency and its harmonics, extend within the EMG spectrum (Fratini et al, 2009) and standard high-pass filters are not suitable for filtering out this artifact.

In the majority of the cases appropriate filtering is used to remove motion artifacts before any signal analysis, while in some other they are used to characterize the mechanical response of the tissue to a specific stimulus (mechanogram) and its correlation to the stimulus itself (Person and Kozhina, 1992; Fratini et al, 2009).

With this chapter the authors aim to investigate the use and the efficacy of surface electromyography in the study of muscle response to vibration treatments. A review of vibration characterization and analysis is reported, SEMG recordings of Rectus Femoris , Vastus Medialis and Vastus Lateralis were collected and analyzed.

Specific artifacts were revealed and the role of those artifact was investigated and assessed. Since the use of vibratory stimulus produces peculiar EMG response a specific model was adopted to describe the EMG synchronization effect and its influence on the resultant recorded muscle activity (Person and Kozhina, 1992).

2. Vibratory stimulation

Different methods of vibratory stimulation have been reported in literature, however, the main are: local applied stimulus or whole body extended vibration.

Local stimulation is achieved by the use of vibrating devices directly applied on tendons of the muscle to be activated. The primary effect of this stimulation is a muscular reflex discharge of impulses some of which are locked with vibratory cycle.

Following the idea of stimulating muscle by using elicited reflex contraction like TVR, alternative methods have been proposed for vibration delivery.

Whole body vibration (vibration transferred to the body as a whole) is the most common used method of delivering vibration in the fields of sport medicine and exercise physiology for enhancing human performance.

In this mechanical stimulus delivery, individuals stand on an oscillating platform while vibrations transmit through the body to the target muscle depending on the subject posture.

Whole body vibration stimulations are produced in two main ways: by alternating rotation of a plate (tilting) or by vertical oscillation (see fig.1).

Alternative methods to deliver vibration are also reported in literature: vibratory modulation of loads during traditional dynamic exercise has also been applied to those typically seen in gymnasiums modified with a vibratory apparatus to produce localised

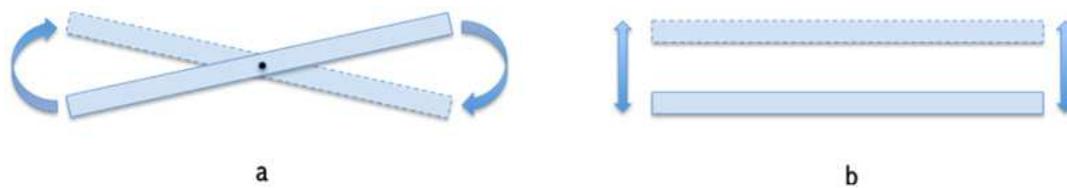


Fig. 1. Methods of whole body vibration stimulation. The figure depicts the two methods of platform oscillation: alternating rotation (a) and vertical oscillation (b).

vibrations to the body parts contacting the vibrating device (Cardinale and Bosco, 2003; Issurin et al, 1996; Yessis, 1992; Issurin and Tenenbaum, 1999; Nazarov and Spivak, 1987). However, in order to avoid misinterpretation in the communication of the experiments carried out by researchers, nomenclature has to be as standardized as possible (Lorenzen et al, 2008). Some few remarks are reported in the following paragraphs.

2.1 Magnitude of vibration

In whole body vibration studies, magnitude has been associated with displacement, amplitude, peak to peak displacement. (Lorenzen et al, 2008).

In the case of vibration, amplitude is the maximum displacement of a vibrating point from a mean position, while peak-to-peak displacement is used to describe the total vibration excursion of a point between its positive and negative extremes (see Figure 2).

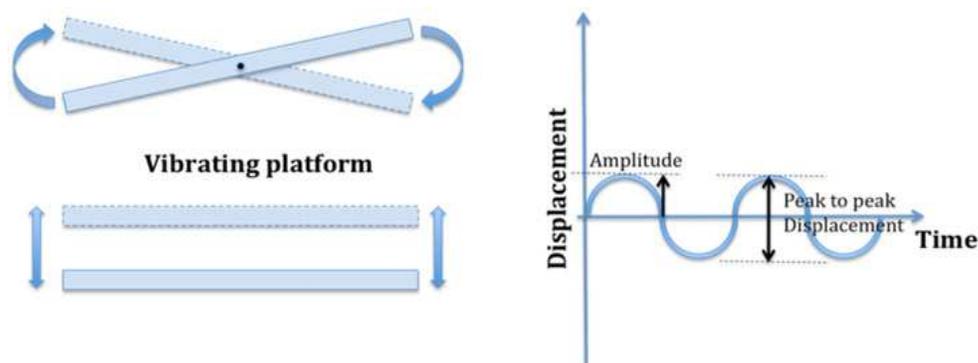


Fig. 2. Amplitude and peak to peak displacement of a vibrating object.

The movement of tilting and vertical oscillating platform can be assumed sinusoidal:

$$s(t) = A \sin(2\pi f t) \quad (1)$$

where f is vibratory frequency, A is the amplitude.

In vertical oscillating platforms all the points will move approximately at the same way; they will have the same peak to peak displacement and the same maximal amplitude.

However, in tilting oscillating platform it is difficult to deduce the real acceleration or displacement to which the subject undergoes.

As shown in figure 3 feet distance from the rotation axis vary the magnitude of the stimulus delivered to the patient as well as the acceleration. It can be computed as the double derivatives of $s(t)$:

$$a(t) = \frac{d^2 s(t)}{dt^2} = -A \cdot (2\pi f)^2 \cdot \sin(2\pi f t) \quad (2)$$

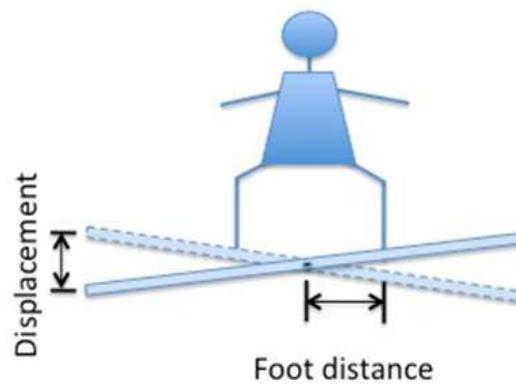


Fig. 3. Explanation of the variation in peak to peak displacement of platform fixed points depending on their distance from the axis.

2.2 Acceleration

Acceleration is not always reported as peak (maximal) acceleration, it depends on frequency and amplitude of the vibration (formula 2) and the maximal value (peak) can be easily computed by:

$$|a|_{\max} = A \cdot (2\pi f)^2 \quad (3)$$

Gravitational forces then, can be obtained by dividing the maximal acceleration by gravity g ($9,81 \text{ [m][s]}^{-2}$). Therefore, an amplitude of 2.5 mm and a frequency of 25 Hz will produce a peak acceleration of $a_{\max} = 61.68 \text{ m s}^{-2}$. In Table 1 are shown some example of estimated maximal accelerations as a function of frequency and amplitude.

Amplitude [mm]	Frequency [Hz]	Acceleration [m][s] ⁻²
2	20	31,55
2	25	49,30
3	20	47,33
3	25	73,95
4	20	63,10
4	25	98,60

Table 1. Maximum acceleration produced for various amplitude and frequency.

Acceleration (maximum), amplitude and frequency are therefore the basic parameters to be reported in vibration studies. The use of the maximal acceleration will demonstrate to the reader the real vibration intensities applied to individuals.

3. Methods

3.1 Subjects & system set-up

Different individuals were involved in the study: twenty healthy subjects, ten males and ten females, not affected by any known neurological or musculoskeletal disorders, voluntarily participated in the study and gave their informed, written consent to participate. All of the subjects were not athletically trained.

A vibrating platform (XP Multipower - TSEM S.p.A., Padova - Italy) was used to deliver vibratory stimulation to the subjects.

Vibrations impressed by the platform were exclusively vertical (there was neither horizontal shift, nor pitch, roll or yaw); platform peak to peak displacement was set to 1.2 mm with a frequency ranging from 10 to 60 Hz. Being the peak-to-peak displacement constant, the maximum acceleration impressed to the patients depends on the square of the pulsation.

3.2 Data recording & processing

Before recording, the subjects were instructed about proper positioning on the platform (hack squat with a 110° knee flexion) and they were familiarized with the device. Signals from the Vastus Medialis (VM), Rectus Femoris (RF) and Vastus Lateralis (VL) of both of the legs were collected in accordance with SENIAM Project (Hermens et al, 1999) guidelines. In addition, a couple of supplementary electrodes was mounted on the patient's patella, supposing no EMG contribution at this site, to assess nature and magnitude of motion artifact on recordings. Signals were recorded using small disc Ag/AgCl electrodes (5 mm in diameter, inter-electrode distance of 20 mm arranged in the direction of the muscle fibres). Electrode skin areas were shaved and cleaned before the placement of electrodes and conductive paste was used. All the electrodes and cables were secured to prevent them from swinging and from causing induced artifact. EMG signals were amplified using a multi-channel, isolated biomedical signal amplifier (BM623 - Biomedica Mangoni, Pisa, Italy). The gain was set to 1000 V/V and a band pass filter (-3dB frequency 10 - 500 Hz) was applied; no notch filters were used. All signals were sampled at 2048 Hz. A set of consecutive 20-second vibrations at different frequencies: 15, 20, 25, 30, 35, 40, 45 Hz, spaced with 60 seconds rest intervals, was delivered to patient.

During rest intervals the patient stood up; five seconds before stimulus he reached the hack squat position. To minimize fatigue-related effects, the vibration frequencies order was randomized. Running RMS on each signal was estimated using 500 ms time window. RMS values were computed before and after artifact suppression to quantify motion artifact influence on EMG RMS.

Motion artifact components on recorded EMG signals were filtered out using a set of standard notch filters centred on the applied vibration stimulus frequency and its harmonics.

EMG power spectrum was computed using a standard FFT algorithm with a Hanning window of 2048 points. Noise (motion artifact) power was assessed considering only five narrow bands (1.5 Hz wide) centred at f_0 , $2f_0$, $3f_0$, $4f_0$, $5f_0$ where f_0 is the applied vibration frequency. Artifacts filtering was achieved using sharp notch filter with a -3dB band of 3 Hz wide centred at f_0 , $2f_0$, $3f_0$, $4f_0$, $5f_0$.

3.3 EMG modelling

Surface electromyography signal acquired during a voluntary muscle contraction is assumed as summation of all the active Motor Units (MU) contributes (Basmajian and De Luca, 1985); it is largely dependent on the properties of MUs and their firing patterns as well as muscle innervation zones (Saitou et al, 2000).

The MUAP characteristics, i.e. shapes and distribution of amplitude and duration, are determined by morpho-functional properties of the activated muscle fibres and MUs,

together with passive and active bioelectric phenomena. The firing patterns reflect the motor control of the central nervous system and in particular situation, such as a vibratory stimulation, they may become correlated with the vibration frequency (Person and Kozhina, 1992, Lebedev and Polyakov, 1992).

3.3.1 MUAP shape

MUAPs shape was assumed biphasic and obtained by slightly modifying the type proposed by Lebedev and Polyakov described in the formula:

$$s_i(t) = a_i s(t) = \frac{a_i t}{(1.7t + 1)^6} \quad (4)$$

where $s_i(t)$ represents the i -fibre MUAP waveform, t is the time variable expressed in ms and a_i corresponds to the MUAP amplitude. The formula was adapted to change independently either time-duration and amplitude variation for each phase, positive and negative (figure 4)

$$s_i(t) = a_i s(t) = \frac{a_i t}{((t * \tau)^6 + 1)} \quad (5)$$

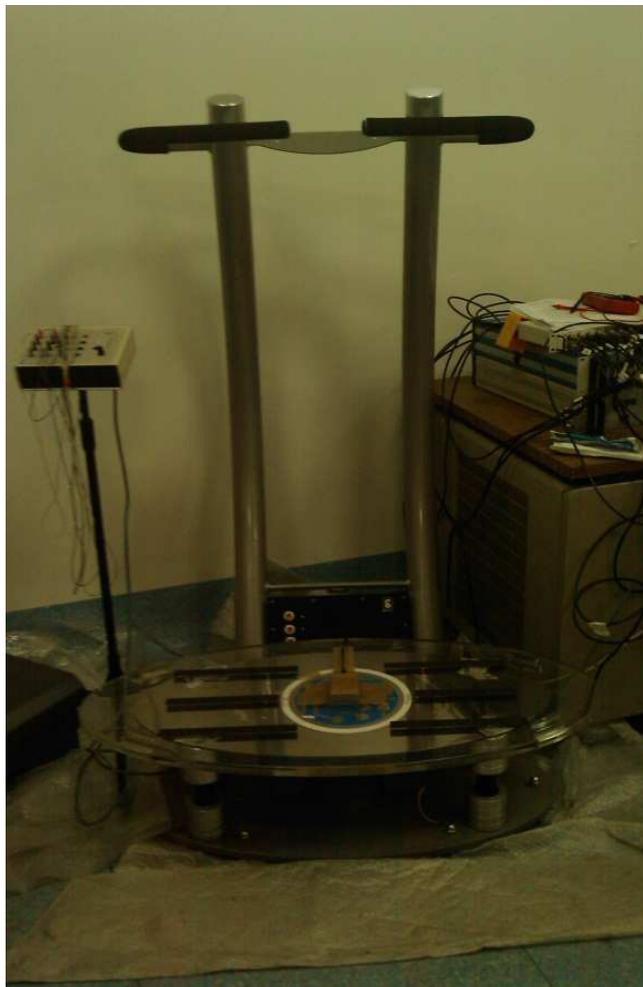


Fig. 4. Vibrating platform used to deliver vibratory stimulation (see text).

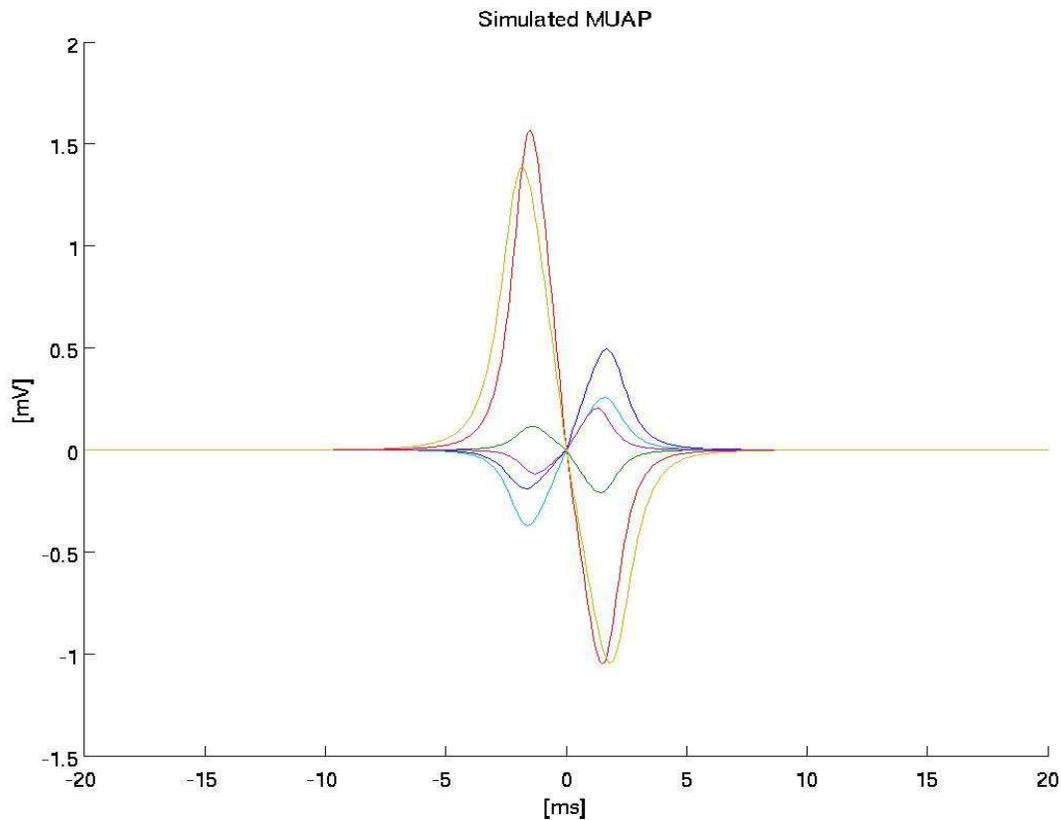


Fig. 5. Some of the different MUAPs obtained with the model.

The amplitude a_i was modeled as an uniform distribution in the range 0.01 to 0.5 mV to take in account the amplitude variability of the simulated signal.

The simulated EMG signal can be expressed as the summation of those MUAPTs:

$$e(t) = \sum_{i=1}^{N_{MU}} e_i(t) \quad (6)$$

where MUAPT, i.e. a summation of MUAPs of the same i-MU, $e_i(t)$, was described as follows:

$$e_i(t) = \sum_k s_i(t - t_k) \quad (7)$$

in which k is the number of pulses and t_k is the time instant of k -pulse of the i-MU.

3.3.2 Inter pulse interval

The general rule to describe the firing behavior of MUs is to consider their interpulse intervals (IPIs) as independent samples of a random variable.

The base distribution of the interpulse intervals, Δt_i ($t_{ki} - t_{(k-1)i}$) was then modeled as a normal distribution, with its probability density function (PDF) $g(\Delta t_i)$, with mean ΔT_i and standard deviation σ . Δt_i was considered as a random value uniformly distributed in the range 55-80 ms and σ equal to 12 ms in accordance to experimental data of Basmajian and De Luca (1985) on the Rectus Femoris.

However, different findings in literature have shown that when a vibratory stimulus is delivered to a tendon the firing rate became synchronous with the vibration cycle, that is t_{ki} is described by a specific PDF to that take into account this synchronization effect.

It is also known that, for vibration frequency under 60 Hz, the majority of MU (80 to 100%) are synchronous with the stimulus or its lower harmonics (Person and Kozhina, 1992).

The time interval between the vibratory stimulus and the activation of the i-MU (t_{cycle}) can also be characterized by a nearly Gaussian distribution $p(t_{cycle})$ with mean equal to half of the vibration period ($T_{vb}/2$) and variance (S_{cycle}) equal to 2 ms.

$$p(t_{cycle}) = Ae^{-\frac{(t_{cycle} - \frac{T_{vb}}{2})^2}{2S_{cycle}^2}} \quad (8)$$

with A is a normalizing factor and $0 < t_{cycle} < T_{vb}$

t_{cycle} is then correlated to the time variable t by:

$$\text{where } n=1,2,3,\dots \quad t = t_{cycle} + (n-1)T_{vb} \quad (9)$$

To provide simultaneously either the base variability of the interspike-intervals and the triggering effect modeled with $p(t_{cycle})$ distribution, a $h(t_{ki})$ PDF was built by multiplying the base PDF $g(\Delta t_i)$ with the $p(t_{cycle})$ PDF (these two distribution were assumed to be independent):

$$h(t_{ki}) = B \cdot g(t_{ki} - t_{ki-1}) \cdot p[-(n-1)T_{vb}] \quad (10)$$

where B is the normalization coefficient, and n is the number of cycle at which t_{ki} is referred to.

The initial value t_{1i} was generated from an uniform distribution in the range $0-\Delta T_i$; time of the kth discharge of the ith-MU t_{ki} was considered a random value, depending on the time of the preceding discharge t_{ki-1} .

If the effect of vibration is negligible, the PDF of the distribution of the IPIs would be close to $g(\Delta t_i)$. However, in this work we assumed the majority of the MUs to be synchronous with vibration stimulus.

4. Results

Although prior to the stimulus onset the recorded signal resulted very low (subject were requested to maintain the hack squat position with a knee flexion of about 110°), after vibration start, the recorded EMGs grew, arising in amplitude.

Biopotential signals were recorded from VM, RF, VL and the patella and as it is possible to see from figure 6 a common behaviour was recognized. Patellar site was chosen for its negligible muscular contribution to the electrical potential recorded; however, its spectrum showed the same frequency components peaks of recorded EMGs (figure 7) and a first harmonic equivalent to the mechanical vibration frequency of the platform as already found in other studies (Fratini et al, 2009). Vibration-induced artifact appeared clearly visible in SEMG recordings (especially on VM, in the figure) after onset.

During vibration, power spectra clearly showed sharp peaks in correspondence to the mechanical frequency and its harmonics, similarly to those recorded onto the patella.

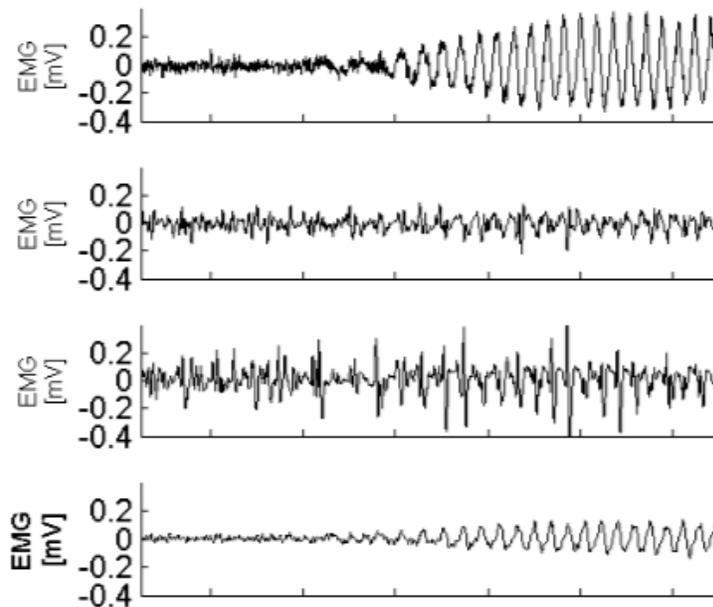


Fig. 6. Typical example of the recorded signals during a vibration onset. From the top to the bottom the first three signals represent VM, RF, VL SEMG respectively. The last is the recorded biopotential at the patellar site.

Presence of higher harmonics components could be an effect of non-linear mechanical behaviour of soft tissues.

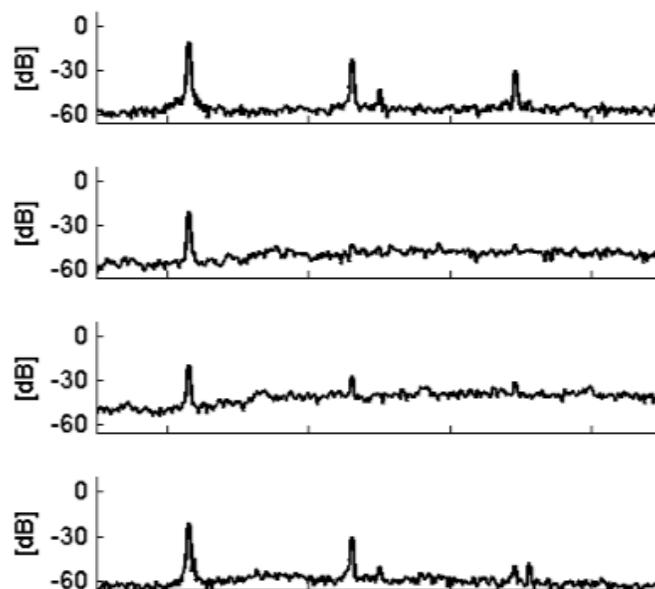


Fig. 7. Power spectrum of the previous SEMG and patella recordings, the first three signals represent VM, RF, VL SEMG spectra respectively while the last is spectrum of the biopotential recorded at the patellar site.

Previous results showed that the percentage of power content in artifacts bands represents less than 4% of total power (on average), while inter-subject standard deviation

ranged from a minimum of 0.7% to a maximum 2.2% (on average) and as confirmed by our actual results after vibration onset, power related to the three to five considered bands raised significantly.

A typical SEMG response for RF muscle at different frequencies is depicted in figure 8; some frequencies seem to induce a higher myoelectrical activity. When subject held the hack squat position; the signal is supposed to contain only the spontaneous surface EMG, while a negligible contribution was recorded at the patellar site. However artifacts related to the vibration frequency and its harmonics laid within the standard surface EMG frequency band (a standard high-pass filter would be not suitable).

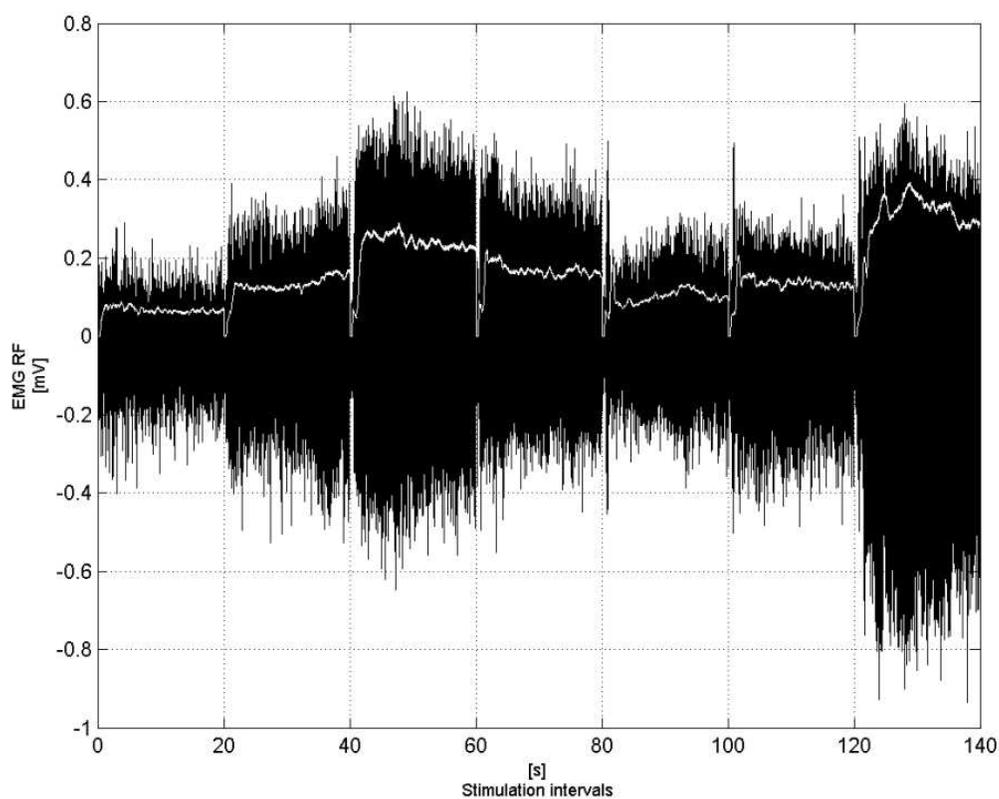


Fig. 8. Typical sequence of raw RF EMG recordings (only the stimulation phase is reported) at different stimulation frequency. A running RMS (white track) is also reported.

As previous study confirmed, the use of sharp notch filter can help in reduce uncertainty of muscle reaction analysis during vibratory stimulation.

After artifact removal signal changes significantly; results showed that the EMG RMS value computed on filtered signal (once motion artifact suppressed) can reduce up to 30% on average (up to 45% in some cases); this indicated that the power of motion artifact was not negligible with respect to SEMG activity.

The use of the considered model has helped in understanding of the behaviour of muscular complex response to vibratory stimulation: MUAP synchronization effect, distribution of IPI, etc. As result the simulated surface EMGs were very similar to those registered with surface electrodes from subjects under vibration.

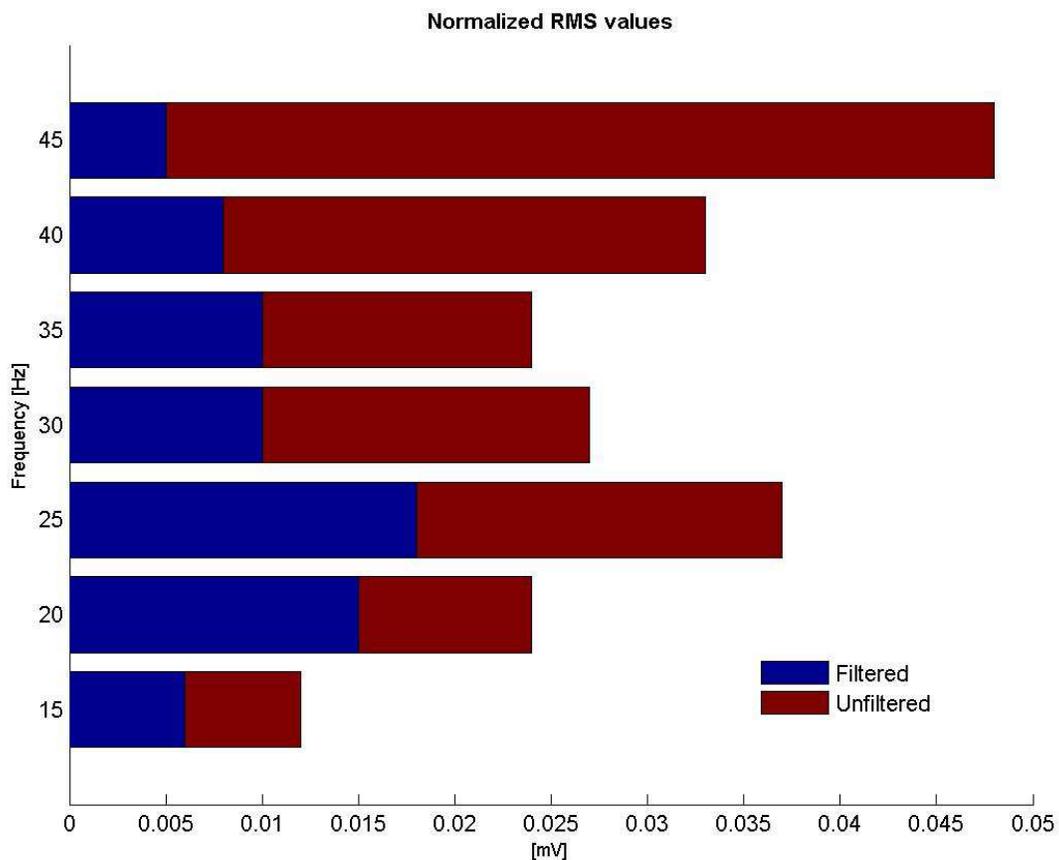


Fig. 9. Average RMS values of surface RF-EMG during vibration on all subjects: The diagram shows the difference of values between filtered (blue bars) and unfiltered signals (red bars). Signals were filtered using multiple notch to eliminate the artifact contribution (see text).

The example shown was obtained considering 600 active MUs and a 80% of them synchronous with a vibratory stimulation, and a random distance between electrodes and active end-plates. Vibration period in figure was set to 25 ms (40 Hz). As it can also be noticed from the figure, although none of the MUs is discharging at that frequency, there are peaks at 40 Hz and its multiples. It can be also noticed a slight increase of the spectrum at low frequencies, representative of the mean IPI (12-18 Hz).

Multiple test were conducted for each explored frequency, considering a random disposition of active end-plates. The power percentage of the spectrum, in a range of ± 1.5 Hz around the vibration frequency and its first four superior harmonics, was estimated and on average less than 10% was found in those bands.

Various simulations were also conducted to determine the power percentage contained in the mentioned narrow bands without vibratory stimulation. The estimated power in those bands obviously decreases, in fact the first four multiples of 40 Hz had on average 5.38% of the total power with 0.95 standard deviation.

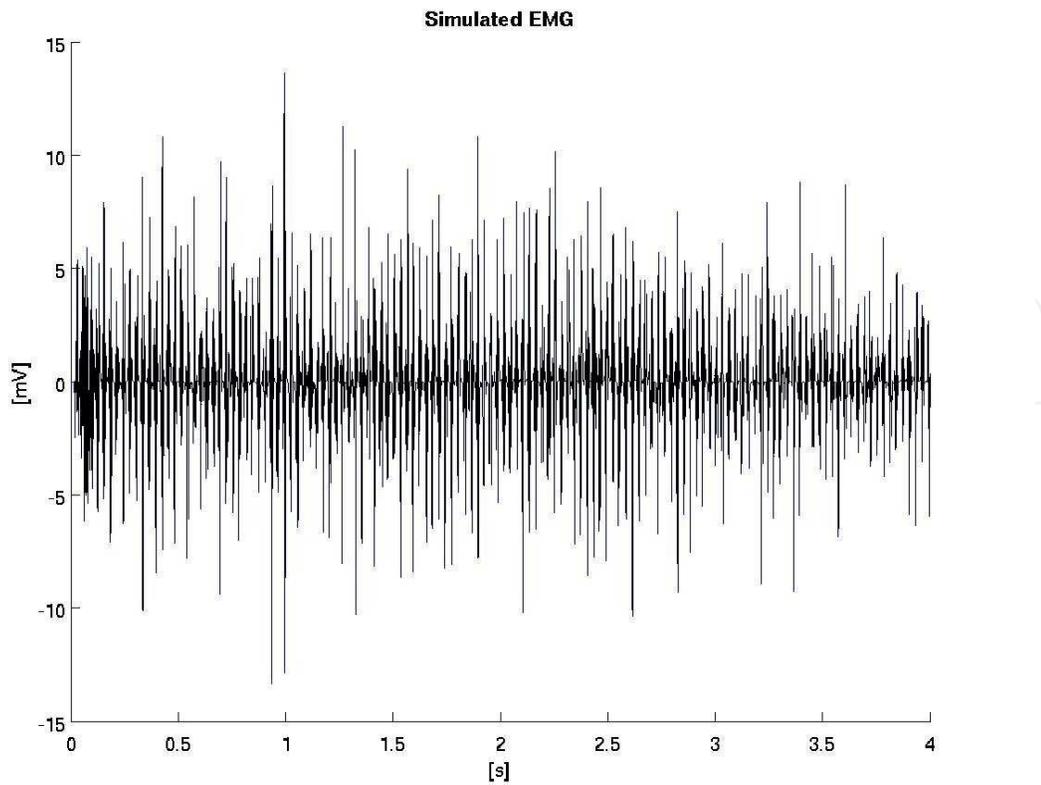


Fig. 10. An example of simulated SEMG signal obtained considering 600 active MUs and a vibratory stimulus at a frequency equal to 40 Hz.

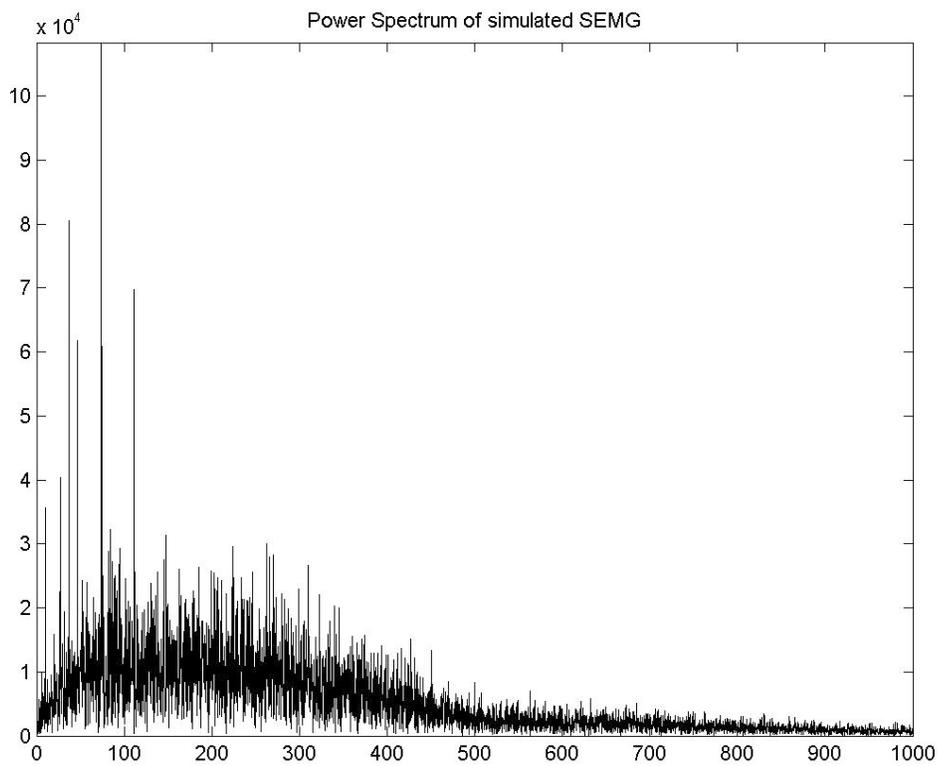


Fig. 11. An example of simulated SEMG signal obtained considering 600 active MUs and a vibratory stimulus at a frequency equal to 45 Hz.

5. Conclusion

Surface electromyography is a helpful technique for the analysis of muscle activity. However, its efficacy is related to the correct electrode positioning, the adequate skin preparation and opportune recording instrumentation. In addition, it is mandatory to recognize artifacts which may alter EMG signals and choose a particular filtering procedure before any additional analysis.

As confirmed by our results, vibrations generate peculiar, not negligible motion artifact on skin electrodes. The artifact appears in all the quadriceps muscles analysed (Vastus Medialis, Rectus Femoris and Vastus Lateralis) and in different amounts: artifact could depend on relative local muscle/electrode and skin layers motion.

Artifact spectrum only consists of the vibration frequency and its higher harmonics, its amplitude is unpredictable and depends on skin properties, electrode type and preparation, amplitude of vibration stimulus etc.

Artifact harmonics extend within the EMG spectrum, making classic high-pass filters unusable, however it is easy to get rid of the artifacts with a series of sharp notch filters centred at the vibration frequency and its superior harmonics applied to the raw EMG signal. Despite the presence of artifacts, some author consider the chance of a greater amount of true EMG appearing in the mentioned narrow frequency bands due to synchronous mechanical activation of muscles during vibration (Lebedev and Polyakov, 1992; Martin and Park, 1997).

Synchronization of motor unit (MU) during whole body vibration is a questionable argument. Surface EMG is formed by a summation of various MUAP components. Even not considering the contribution of the loosely-synchronized MU, all the MU synchronized with vibration must also be synchronized between each other in order to achieve a repeatable waveform.

Previous studies about the tonic vibration reflex concentrates on response of single motor unit (Romaiguère et al, 1991). A certain synchronization of some specific MUs, probably due to a monosynaptic reflex, has been found in response to vibration stimuli (taps) applied to proximal muscle tendon together with other much less synchronized MUs (probably polysynaptic reflexes) (Romaiguère et al, 1991). However, even the synchronized MUAP do not appear for each vibration pulse and also the latency from the tendon tap shows a standard deviation of about 1.4 ms (corresponding to a average of 22.7 ms) this implies a not exactly periodic signal. We have also pointed out differences between the case of delivery vibration stimuli directly to a specific muscle (often sequence of tapping of the distal tendon or muscle belly) and indirectly, i.e. from a sinusoidal vibrating plate (the vibration stimulus is mediated by a complex biomechanical chain before reaching a specific muscle).

The use of a specific EMG model in order to investigate and assess the actual muscle activity under vibratory stimulation was mandatory. The difference between simulated EMG signals and raw surface EMG recording, depending on the innervation zone, was found to be consistent, it can be reasonably justified by the presence of artifacts superimposed on the raw surface EMG.

However, results were find dependent on the relative position between electrodes and between electrodes and end-plates. In particular, a simulation of EMG recording considering MUAPs approaching only from a single direction would result in a much

higher power percentage in the bands in which is also present the artifact contribution. However, according to our results, it is also true that the electrodes placed onto the patella showed an amount of power associated to the considered narrow bands comparable with that of the recorded EMG on quadriceps muscles. In conclusion, analysis of muscular activity during vibration based on unprocessed surface EMG recordings may significantly overestimate muscle response: filtering out the motion artifact would prevent misinterpretation of experimental results. It is our opinion however, that we cannot exclude a-priori that the true EMG power spectrum can show peak components at the vibration frequency and its harmonics.

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