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Characterization of the mechanomyographic signal of three different muscles and at different levels of isometric contractions

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Purpose: Lateral (X) and longitudinal (Y) mechanical oscillations of muscle fibers that take place during muscular contraction seem to contain information additionally to the myoelectric activity, which can contribute to the interpretation of some muscle gradation force mechanisms. However, no previous study was found that had investigated the relationship between the muscle force and features associated to the mechanomyographic (MMG) signal obtained by means of a biaxial accelerometer in three different muscles. Therefore, the aim of this study was to evaluate the relationship between the force output at different load levels (20% to 100%) of the maximum voluntary isometric contraction (%MVIC) and the two signals supplied by a biaxial accelerometer and, in addition, the so-called resultant (R) acceleration signal derived from the two signals mentioned previously. Twenty seven male volunteers participated in this study. Methods: The force output related to the right *biceps brachii*, *soleus* and *gastrocnemius medialis* muscles was studied by means of linear regression models fit to log-transformed of the root mean square (RMS) values of the MMG signals in X, Y, and R axes versus each %MVIC. The phase angle of R acceleration (Phase_R) and anthropometric data were also considered. Results: The angular coefficient a and the antilog of y-intercept b from the log-transformed of MMG data values versus force output were able to distinguish partially motor unit strategies during isometric contractions in the three muscles studied. Conclusion: The findings suggest that biaxial accelerometer seems to be an interesting approach in the assessment of muscle contraction properties.

Key words: *mechanomyography, MMG signal, muscle twitch, muscle activation pattern, muscle force*

1. Introduction

Small vibrations from the skeletal muscle fiber twitches can be measured nearby the skin surface. Different transducers such as accelerometers, microphones and laser distance sensors have been used to register those vibrations, whose approach is usually named Mechanomyography (MMG) [22]. The source of the MMG signal has been mainly attributed to three main mechanisms: (i) a gross lateral muscle movement at the beginning of a contraction due the non-simultaneous activation of muscle fibers; (ii) small subsequent vibrations due to the resonant frequency of a muscle; and (iii) dimensional changes of the muscle fibers during contraction [22]. The MMG signal can be considered a mechanical counterpart of a muscle contraction and has contributed in different fields of investigations, such as motor control [1], muscle fiber typing [19], [25], and muscle fatigue [1].

With regard to accelerometers, some authors [1], [17] suggest that the main information related to the gradation of the muscle force mechanisms seems to be contained in the perpendicular direction to the muscle fibers. On the other hand, few authors have also verified a longitudinal vibration [4], [18] that may contain some evidence from those strategies, which have been under discussion with regard to the technical aspect in

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the number of axes that must be investigated for a proper interpretation of this signal [4].

The MMG has been used to investigate different muscles under different motor tasks and most reports present particular results for the relationship between muscle force and some parameters in time and/or frequency domains of this signal. The possible reasons for that are the influence of muscle architecture [1] and fiber type [19], [25], both in the amplitude and spectral features. Besides that, no previous studies were found that had used biaxial accelerometers in muscles with different muscle fiber types and architecture. Therefore, since we have found no previous studies that have investigated simultaneously the longitudinal and lateral vibration properties of muscle fibers by means of an accelerometer, the aim of this study was to evaluate the relationship between muscle gradation force of three different muscles and the correspondent MMG signals in the temporal domain. Additionally, we also evaluated the resultant MMG signal provided by the previous two signals, which are orthogonal to each other.

2. Material and methods

2.1. Subjects and muscles tested

Twenty seven volunteers, all male, without any history of neuromuscular diseases, participated in this study (age: 25.6 ± 2.4 years; body weight: 72.1 ± 8.2 kg; height: 1.74 ± 0.07 m). The muscles studied were the right biceps brachii (BB), soleus (Sol) and gastrocnemius medialis (GM). The volunteers, all right-handed, were tested by means the Edinburg Handedness Inventory [10]. Concerning the dominance of lower limbs, few questions were also applied. Those questions regarded which limb could perform tasks like kicking a ball; keeping standing and jumping on one leg; and which one they could refer to as the dominant one. The study was submitted to the local ethical committee (238/06) in accordance with the Declaration of Helsinki and was performed after each volunteer gave a written informed consent.

2.2. The acquisition system

The acquisition system was based on a computer with a 16 bits A/D converter (Spider 8 – HBM, Germany) in a range of ± 10 V. A biaxial accelerometer (ADXL202E, Analog Devices, USA) mounted on a small printed circuit board was used for the MMG signal acquisition, which resulted in a total mass of 0.0015 kg. Its sensitivity and frequency bandwidth were set at 315 mV/g (g = acceleration due to gravity) and 200 Hz, respectively. The sampling frequency was set at 9600 Hz.

A support apparatus with two dynamometer systems (2000 N; Alpha Instrumentos, Brazil) were especially built for collecting force output (“muscle force”) signals associated to the superior and lower limbs. This apparatus and body positions adopted during the MMG signal acquisition of the three muscles studied can be observed in Figs. 1, 2, and 3. This apparatus allowed keeping the volunteers sat comfortably with limbs well fasten for the acquisition procedures as can be seen in Fig. 1. Individual adjustments were possible in some joint angles (ankle, knee, hip and elbow) before collecting data. The software for the acquisition of the MMG and force output signals was built in LabView (National Instruments, USA).

Fig. 1. The positions adopted by the subjects for collecting force output and MMG signals during Sol muscle force tests

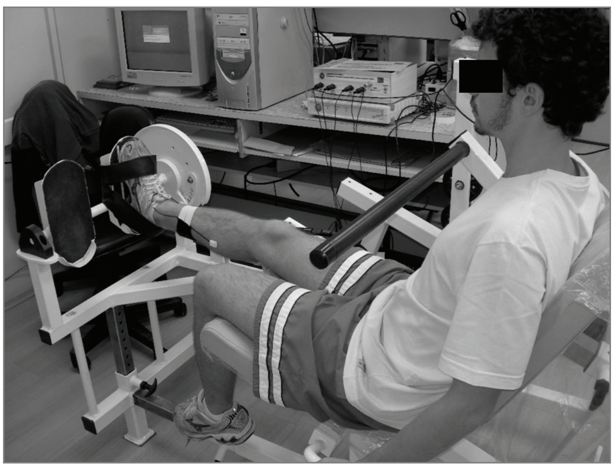

Fig. 2. The positions adopted by the subjects for collecting force output and MMG signals during GM muscle tests

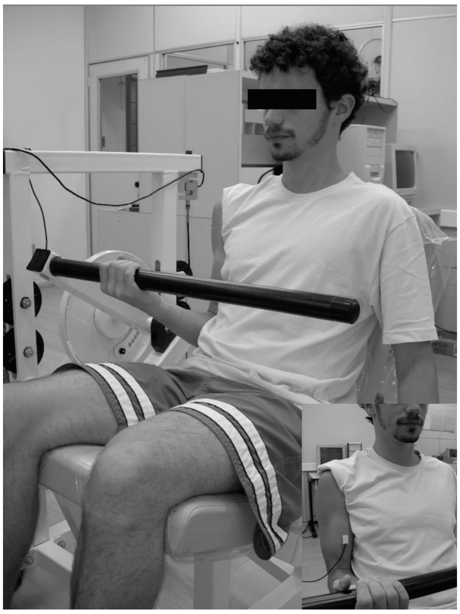

Fig. 3. The positions adopted by the subjects for collecting force output and MMG signals during BB muscle tests

2.3. Procedures prior to the MMG signal acquisition

Prior to the MMG signals acquisition, maximum voluntary isometric contraction (MVIC) of each muscle was evaluated for further determination of four different and arbitrary load levels: 20%, 40%, 60%, and 80% of MVIC. The MVIC test followed the same conditions adopted for the MMG signal acquisition. Each test consisted of MVIC for 6 s and the mean value of the three trials was considered as reference (100% of MVIC). A time interval of two minutes was adopted between each trial to avoid muscle fatigue. The MVIC test was applied arbitrarily in the following sequence to avoid fatigue of the lower leg muscles: Sol, BB and GM.

The MVIC test was conducted while the volunteers were sat on the apparatus. Sol was tested with the knee and ankle at 90° and 100° (10° plantar flexion), respectively (Fig. 1). On the other hand, GM was tested with the knee joint extended and the ankle maintained at 100° (10° plantar flexion) (Fig. 2). BB was tested with the elbow joint at 90° and the shoulder at neutral position. The forearm was maintained in supine position (Fig. 3). All the chosen joint angles were based on the capacity of those muscles generating maximum torque [11].

Prior to MMG data acquisition, skinfold thicknesses were measured from the same spots where the accelerometer was placed on all the three muscles studied, according to Norton et al. [20]. A skinfold caliper (sensitivity: 0.1 mm; CESCORF Equipamentos Esportivos Ltda. – Porto Alegre – RS, Brazil) was used by an experienced experimenter. At least two skinfold thickness measurements with a maximal difference of 10% between them were collected from each muscle spot to obtain a mean value also from each muscle.

2.4. MMG signal acquisition

After the MVIC test the accelerometer was fixed over the muscle belly by means of a double-face ad-

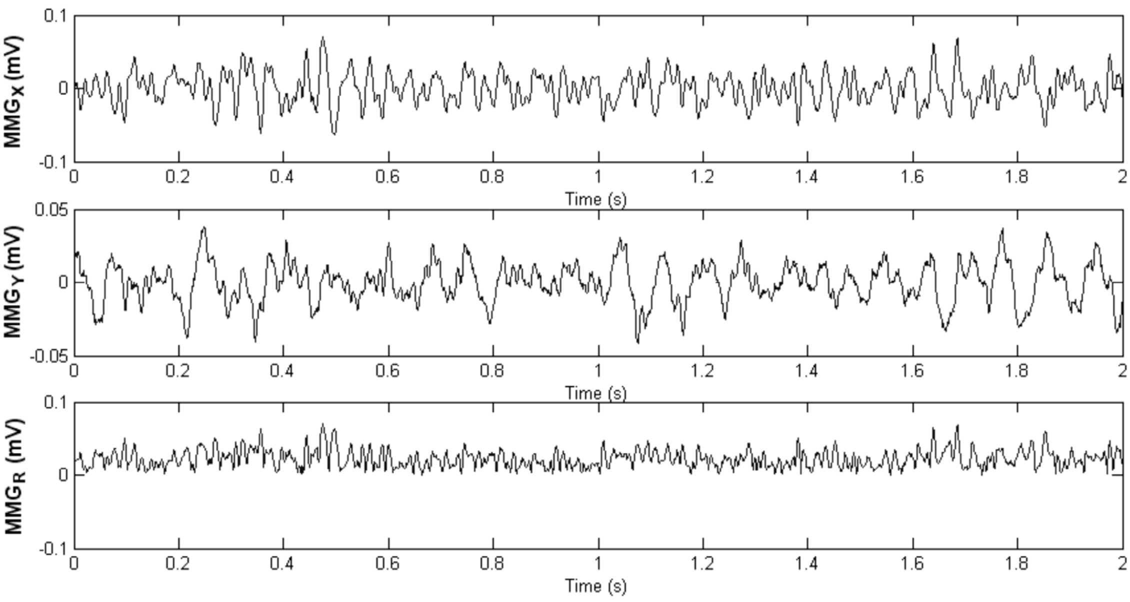

Fig. 4. MMG signal samples from BB muscle at 60% of MVIC in *X*, *Y* and *R*

hesive tape, adopting the anatomical references suggested by Hermens et al. [11] for surface electrodes placement in electromyographic acquisitions. Therefore, the accelerometer was aligned with the longitudinal muscle belly, and its parallel (Y) and perpendicular (X) directions were considered in the results and discussion.

The five load levels representing the percentages of MVIC (%MVIC) were randomised before the data acquisition and followed the same sequence of muscles and conditions applied in MVIC tests. During the data acquisition the volunteers had visual feedback through a target line shown on a video screen, which was related to the degree of force representing the %MVIC (a maximum error of 5% of the load level was considered). The volunteers were consequently requested to maintain a constant muscle force for at least 5 s with a rest time interval of two minutes as previously mentioned.

The initial two and the final one seconds were excluded from the acquired MMG signals to minimize their variability associated with the beginning and the end of the test. As cited above, the MMG signals associated to the two orthogonal acceleration directions were arbitrarily defined as X and Y, which were perpendicular and parallel or longitudinal to the muscle belly, respectively. Samples of MMG signals collected from BB muscle at 60% of MVIC are presented in Fig. 4.

2.5. MMG signal analysis

A computational routine was written in Matlab R 2011.a (Mathworks, USA) for processing the acquired MMG signals. Since the expected bandwidth of the MMG signals is smaller than 200 Hz, the signals initially acquired in 9600 Hz were resampled to 960 Hz. Subsequently, the MMG signals were band-pass filtered (2–100 Hz, Butterworth, 4th order) to attenuate the interferences [13]. Consequently, the time behaviour of the Root Mean Square (RMS) values from the MMG signals was then obtained and considered for further analysis. The temporal behaviours of this parameter, considering non-overlapped windows of 520 milliseconds of duration, were obtained for the two MMG signals supplied by the accelerometer.

The magnitude of a so-called MMG resultant signal (MMG_R) was then obtained from the longitudinal and transversal original accelerometry signals

$$\mathrm{MMG}_R = \sqrt{(\mathrm{MMG}_X)^2 + (\mathrm{MMG}_Y)^2} \qquad (1)$$

where MMG_R represents the magnitude of the so-called MMG resultant (R), and MMG_x and MMG_y represent, respectively, the two signals supplied by the accelerometer.

Similarly, a so-called phase resultant signal (Phase_R) was also computed using

$$\mathrm{Phase}_R = tg^{-1}\left(\frac{\mathrm{MMG}_Y}{\mathrm{MMG}_X}\right) \qquad (2)$$

where Phase_R represents the so-called phase of the MMG_R.

These two derived signals were calculated for all %MVIC. Mean values for the whole RMS signals (energy signals – $\mathrm{MMG}_{\mathrm{RMS}}$) associated to the MMG signals (X, Y, and R) were calculated and hereafter named $\overline{\mathrm{RMS}}_X$, $\overline{\mathrm{RMS}}_Y$ and $\overline{\mathrm{RMS}}_R$. Similarly, the mean value of the Phase_R signal ($\overline{\mathrm{Phase}}_R$) was calculated for each %MVIC.

2.6. Methodological and statistical data analyses

Since MMG signal magnitude and %MVIC relationship can be nonlinear [10] and, in addition, the slopes and y-intercepts from a log-transform model set on a subject-by-subject or a robust data group basis seem to provide a suitable approach to perform the interpretation of gradation of the muscle force mechanisms as well, we adopted the methodological approach proposed by Herda et al. [10] as a way of analysis. These authors suggest fitting equation (3) to each napierian log-transformed $\mathrm{MMG}_{\mathrm{RMS}}$ value and torque data relationship, which represents a simple linear regression model

$$\log_e[y] = a(\log_e[x]) + \log_e[b] \qquad (3)$$

where e, y, x, a and b represent, respectively, the Euler number (~2.71828), the napierian log of the $\mathrm{MMG}_{\mathrm{RMS}}$ data, the napierian log of the torque data, the angular coefficient, and the napierian log of y-intercept.

Alternatively, equation (3) can also be expressed as

$$y = bx^a\,. \qquad (4)$$

Additionally, the coefficients of determination (R^2) were calculated from the exponential model as represented in equation (4) for each volunteer. Once the model above had been fitted to the MMG $\overline{\mathrm{RMS}}_X$, $\overline{\mathrm{RMS}}_Y$ and $\overline{\mathrm{RMS}}_R$ data separately, the angular coefficient a and the antilog of y-intercept b from each signal (X, Y, and R) served as a basis for the subsequent interpretation of the likely muscle gradation force strategies required from each muscle to perform the isometric contraction tasks (ICT). Subsequently, a two-way ANOVA was used to compare, separately, each coefficient among muscles (Sol, BB and GM; factor 1) and axes (X, Y and R; factor 2). A one-way ANOVA was used to compare $\overline{\text{Phase}_R}$ among muscles (Sol, BB and GM; factor 1). In turn, the *skinfold thicknesses data were also compared among muscles* (Sol, BB and GM; factor 1) by means of a one-way ANOVA.

To evaluate possible differences, the Tukey test was adopted as a post-hoc whenever necessary. The level of significance (α) was set at 5%.

3. Results

The results are presented in Box-Whisker plots with medians, 1st and 3rd quartiles and maximum and minimum data values. Furthermore, a dotted line set at "1" was drawn in Figs. 5d, 6d and 7d with the aim of providing a reference in respect to the behaviour of MMG_{RMS} versus force output (at the five %MVIC) relationship. Thus, a coefficient $a \cong 1$ refers to a linear behaviour; while a coefficient $a < 1$ refers to a plateau; and a coefficient $a > 1$ represents an upward and fast increase relationship [10].

The figures below present the angular coefficient a and the antilog of y-intercept b from the MMG $\overline{\text{RMS}_X}$, $\overline{\text{RMS}_Y}$ and $\overline{\text{RMS}_R}$ versus force output (N) at the five levels of %MVIC from Sol (Fig. 5), GM (Fig. 6) and BB (Fig. 7).

Regarding angular coefficient a for the factor *axes*, there was only significant statistical difference between X and Y ($F_{(2,234)} = 3.39$; $P = 0.016431$) for Sol (Fig. 5d). In contrast, the antilog y-intercept b did not show any significant statistical difference ($F_{(2,234)} = 0.486$; $P = 0.61533$) among *axes* for each individual muscle (Fig. 5e, Fig. 6e, and Fig.7e).

When comparing the angular coefficient a among the three muscles, there were significant statistical differences ($F_{(2,234)} = 18{,}15$; $P < 0.05$) between Sol and GM ($P = 0.000022$), Sol and BB ($P = 0.000823$) and GM and BB ($P = 0.049950$) (Fig. 8a), while the antilog of y-intercept b (Fig. 8b) showed only significant statistical differences between BB and Sol ($P = 0.000022$) and BB and GM ($P = 0.000022$).

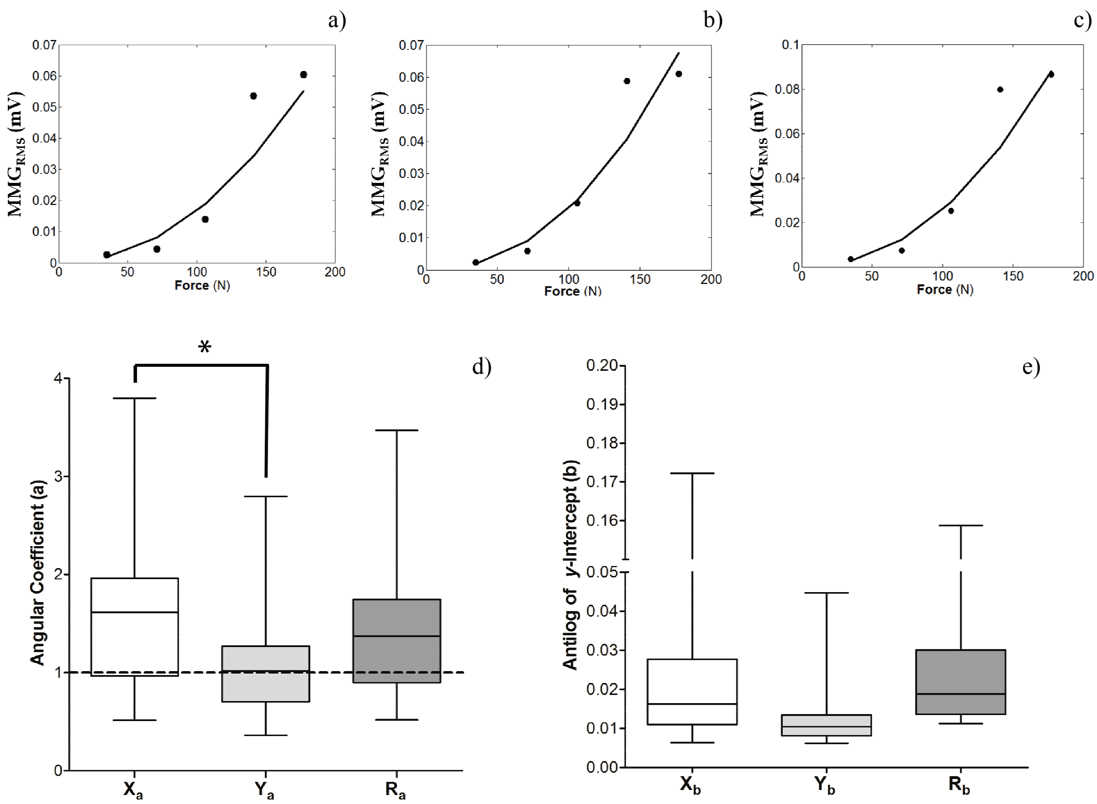

Fig. 5. Mean behaviour of MMG versus force output (N) relationship from Sol fit with equation (4) to X (a), Y (b) and R (c) data axes. Box-Whisker plots are presented for the angular coefficient a (d) and the antilog of y-intercept b (e) from the MMG signal in X, Y and R axes; $^*P = 0.016431$

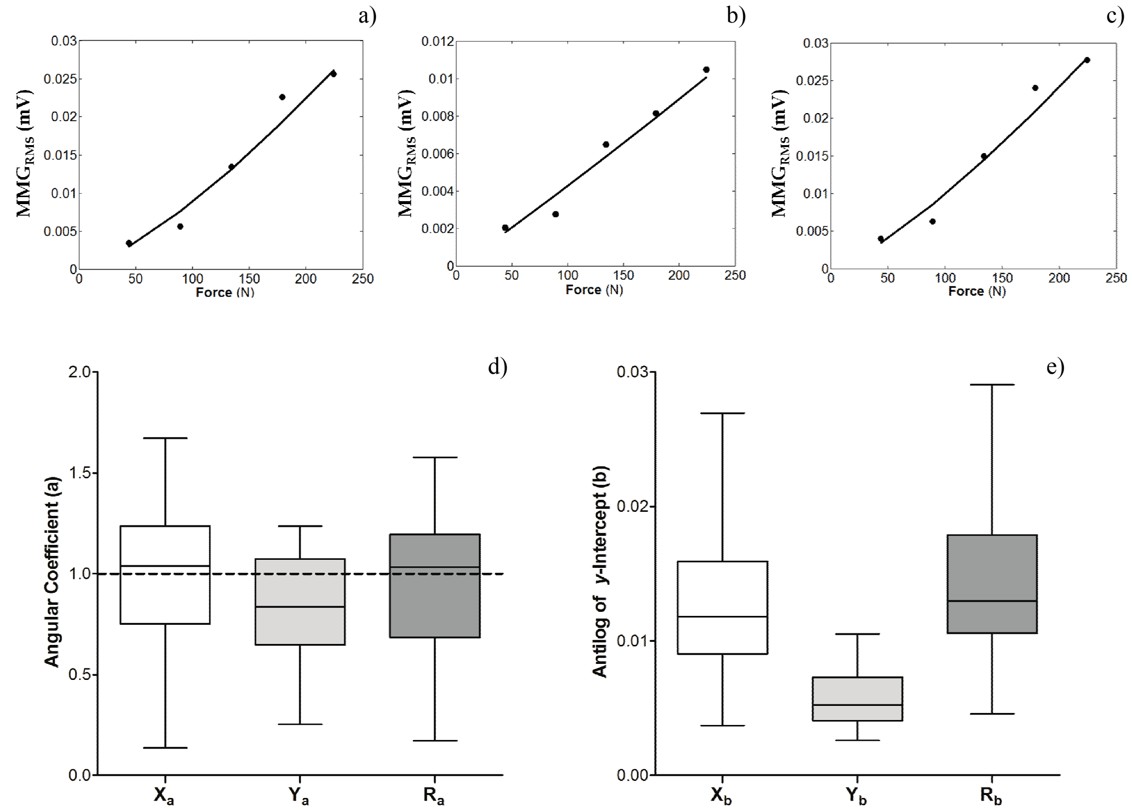

Fig. 6. Mean behaviour of MMG versus force output (*N*) relationship from GM fit with equation (4) to *X* (a), *Y* (b) and *R* (c) data axes. Box-Whisker plots are presented for the angular coefficient *a* (d) and the antilog of *y*-intercept *b* (e) from the MMG signal in *X*, *Y* and *R* axes

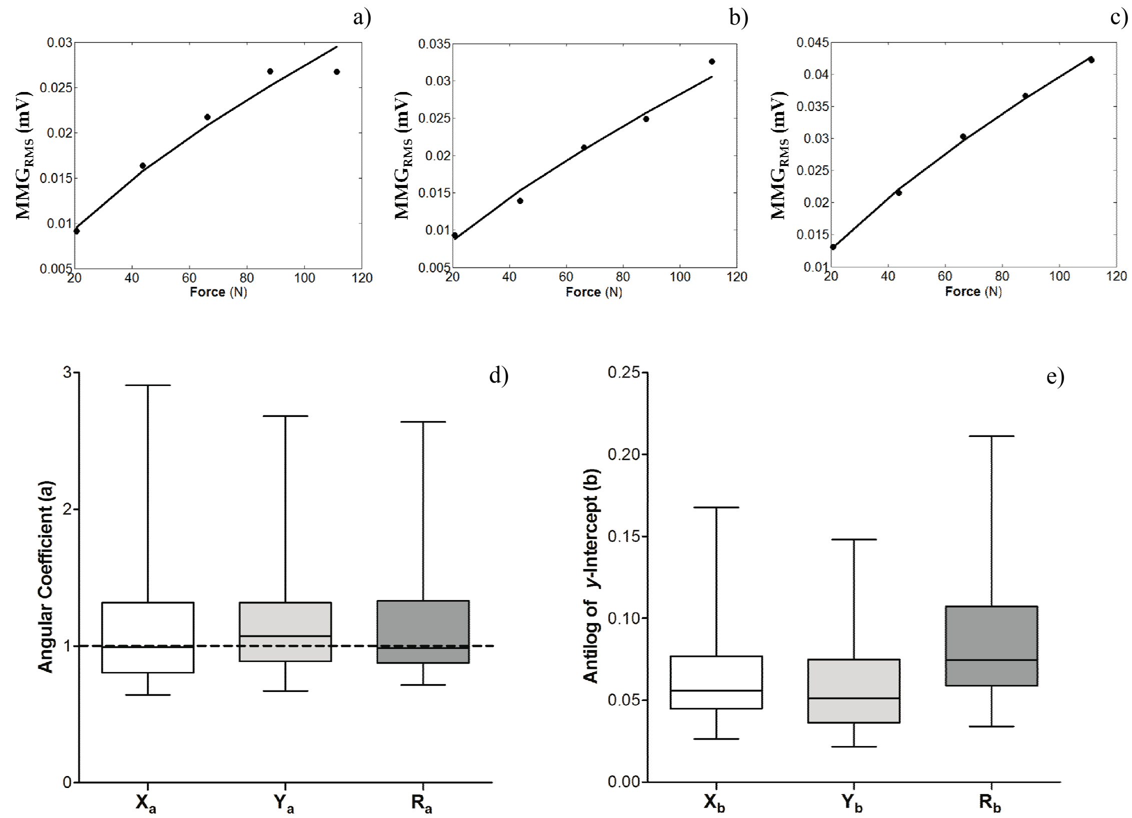

Fig. 7. Mean behaviour of MMG versus force output (*N*) relationship from BB fit with equation (4) to *X* (a), *Y* (b) and *R* (c) data axes. Box-Whisker plots are presented for the angular coefficient *a* (d) and the antilog of *y*-intercept *b* (e) from the MMG signal in *X*, *Y* and *R* axes

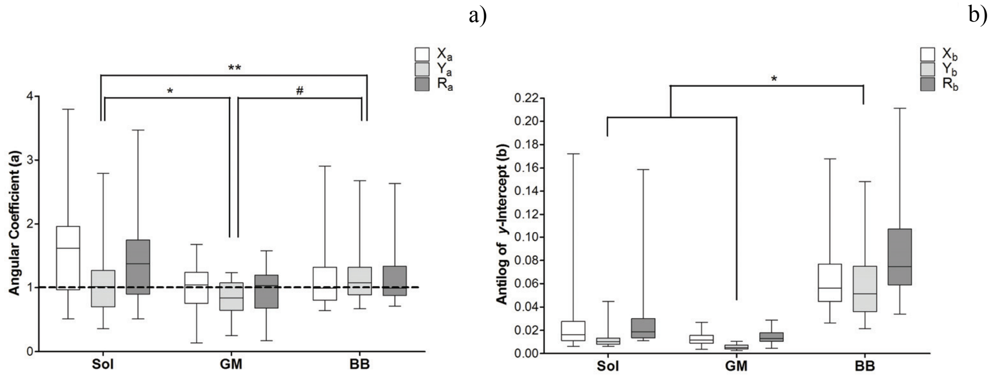

Fig. 8. Box-Whisker plots are presented for the angular coefficient *a* (a) and the antilog of *y*-intercept *b* (b) from the MMG signal in *X*, *Y* and *R* axes and Sol, GM and BB muscles; $*P = 0.000022$; $**P = 0.000823$; $^{\#}P = 0.049950$

Table 1. Means and standard deviations (± SD) of R^2 for the MMG signal in *X*, *Y* and *R* axes obtained from the three muscles (Sol, GM and BB)

Muscle	Sol			GM			BB		
Axis	*X*	*Y*	*R*	*X*	*Y*	*R*	*X*	*Y*	*R*
Mean	0.9791	0.9904	0.9847	0.9938	0.9950	0.9945	0.9957	0.9961	0.9965
± SD	0.0236	0.0143	0.0196	0.0085	0.0072	0.0077	0.0075	0.0064	0.0059

With regard to R^2 obtained from all the three muscles and axes, they were all higher than 0.97 as can be seen in Table 1.

In relation to $\overline{\text{Phase}_R}$, there were significant statistical differences between 20% and 60% (P = 0.03872), 20% and 80% (P = 0.000906), and 20% and 100% of MVIC (P = 0.000405) for Sol (Fig. 9a). Conversely, there were not significant statistical differences among any %MVIC tested for $\overline{\text{Phase}_R}$ in GM (Fig. 9b). Similarly, we also found no significant statistical differences among the levels of contraction in BB (Fig. 9c).

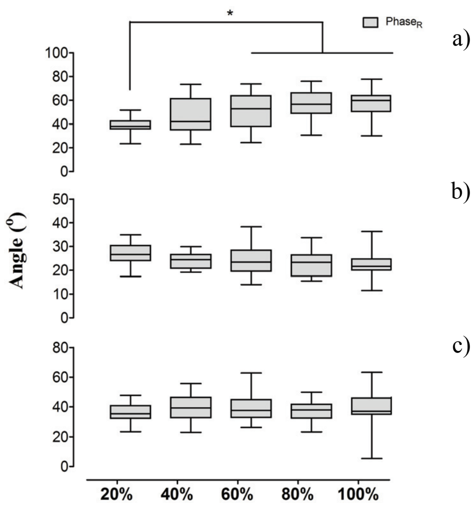

Fig. 9. Box-Whisker plots from Phase_R data.
There were significant statistical differences between 20% and 60%, 20% and 80%, and 20% and 100% of MVIC for Sol muscle (a), GM (b) and BB (c) muscles did not show any statistical differences among all the levels of MVIC; $*P < 0.04$

With respect to skinfold thicknesses, there was not significant statistical difference only between Sol and GM (P = 0.075217). In contrast, Sol and BB, and GM and BB presented significant statistical differences (P = 0.000107 and P = 0.000201, respectively).

4. Discussion

Similarly to the myoelectric activity (EMG), the MMG signal also provides additional information and further details concerning muscle force gradation mechanisms. In both cases, volunteers are usually required to perform motor tasks in steps from 0 to 100% of MVIC and some signal parameters, such as the RMS value, are obtained to further provide a magnitude-force relationship from which the results can be considered for analysis. However, Herda et al. [10] point to the dissimilarities concerning the relationship between the magnitude of the EMG and muscle force, and the relationship between the magnitude of the MMG signal and muscle force. These authors suggest that it is not possible to discriminate firing rate and recruitment of motor units from the typical surface EMG signal. Conversely, the MMG signal seems to provide hints regarding both muscle force gradation mechanisms. According to this interpretation, we intended to characterize the MMG signal magnitude derived from the longitudinal (*Y*) and perpendicular (*X*) axes to the muscle length as suggested by Matta et al. [18] and Polato et al. [24] as well as by means of a resultant MMG signal, which was then obtained from the previous two MMG signals. Moreover, three different muscles, usually described as presenting different fibre types, were studied by means of a linear regression model (equation (4)) fit to log-transformed of the RMS values of the MMG X, Y, and R signals versus each force output resulting from each %MVIC as suggested by Herda et al. [10].

4.1. Sol muscle

4.1.1. The angular coefficient *a*

Few authors have investigated the relationship between the MMG signal properties in Sol at different %MVC, even in ICT [19], [25]. Regarding the MMG signal magnitude, Yoshitake and Moritani [25] have shown a linear increase in the mean MMG_{RMS} value versus the force output from 20 to 60% of MVIC for Sol. Such an increase was marked by no significant statistical difference between 20 and 80% of MVIC and, furthermore, a different behaviour characterized by a decrease of MMG_{RMS} values at 80% of MVC. The decrease was attributed to the fusion-like state of muscle fibres, which is mentioned to be characterized by a diminishing in the dimensional changes of those already recruited and resulting in a reduction of the MMG signal magnitude [22], [25]. It has been reported that muscles predominantly composed by slow motor units (MUs), similarly to Sol [6], [25] may reach a fusion-like state at lower force output levels than those that present faster MUs [6]. In general, the achieved MU activity pattern can be observed in other muscles [19], but not necessarily at force output levels close to 100% of MVC (isometric or isotonic), from which some different tendencies may also be observed [1], [3], [17].

The findings of the present study disagree with those reported by Yoshitake and Moritani [25]. Considering the magnitude of MMG signal amplitude is mainly related to the MU recruitment up to 60–80% of MVIC [22], one must take into account the hypothesis of this mechanism occurring closer to 100% of MVIC. As a consequence, there would be a lack of the fusion-like state of muscle fibers and a corresponding MMG signal amplitude increase across almost the whole range of %MVIC. Thus, differently from previous studies, our results pointed to a different behaviour for the angular coefficient *a* from Sol. As can be seen in Fig. 5, ~75% of our volunteers reached values of the angular coefficient *a* higher than "1" in MMG $\overline{\mathrm{RMS}}_X$ and $\overline{\mathrm{RMS}}_R$ while in the MMG $\overline{\mathrm{RMS}}_Y$ we noted ~50%. Such pattern of response represents an upward and fast increasing relationship between the magnitude of the MMG signal and force output, which would be related to the MU recruitment as the predominant strategy in muscle gradation force, as suggested by Herda et al. [10]. Even though our data seem to disagree with previous studies, Oya et al. [23] performed selective recordings of EMG activity through indwelling electrodes from forty-two MUs in Sol and they observed that most MUs were recruited throughout the full range of %MVIC. Thus, according to the previous findings, which provided new insights about Sol properties, a muscle that predominantly consists of slow fibers, our results seem to be corroborated by the hypothesis of Oya et al. [23], which suggest that low-threshold MUs exhibit higher peak firing rates than high-threshold MUs. Interestingly, the observed results disagree with the expected results that would be described by the so-called "onion skin" MU control arrangement [7]. Additionally, these last authors mentioned a dependence of recruitment and firing rate of MUs in relation to the muscle functions. An example of this would be the comparison of smaller muscles with greater level of precision, such as those related to hand movements, to BB. Another important issue concerns the lower limb muscles that are still poorly investigated [23], which reinforces our data as a possible outcome of a specific strategy adopted by Sol in reaching different %MVIC in our contraction task.

The MMG_Y signal in Sol also presented lower values of angular coefficients *a* when compared to X and R signals. It has been reported to be associated to small longitudinal (Y) mechanical oscillations of muscle fibers that must be induced by the properties of passive elastic components in series highlighted in muscle mechanical models and usually represented by tendons [18].

4.1.2. Phase$_R$

We observed an increase in $\overline{\mathrm{Phase}}_R$ versus %MVIC. Due to the way the $\overline{\mathrm{Phase}}_R$ was calculated, this increase can be thought of as a differentiation in the magnitude of MMG_X and MMG_Y signals with the increase of muscle force output. The behaviour of MMG signal during muscle gradation force can be considered as minimally dependent of the sensor placement position over the muscle [5]. In the present study, the accelerometer was placed over the distal/medial portion of Sol, which is considered multipennate on this spot [5], and being subjected to variations of muscle fiber angles as the muscle force output changes [15]. Therefore, it is possible that changes in the muscle fibers angles have resulted in an interference with the MMG signal acquisition, which meant a lack of alignment between the axis of the transducer and the muscle fibers while muscle force output increased. Considering this hypothesis, the differences observed among MMG_X, MMG_Y and MMG_R can be provided by Sol muscle architecture changes while it contracts and increases its muscle force output level, which reinforces an interesting aspect concerning the use of accelerometers in evaluating also the dynamics of muscle fiber distribution and architecture.

4.2. GM muscle

4.2.1. The angular coefficient *a*

Few authors evaluated properties of the GM muscle through the MMG signal. As an example, Yoshitake and Moritani [25] evaluated the MMG signal from GM at different %MVIC by means a different transducer: a microphone. They observed a linear increase of the RMS values of the MMG signal in a range from 20 to 80% of MVIC in steps of 10%, which was marked by significant statistical differences among those levels. Since GM has been characterized as being composed by slow and fast MUs in a quite similar proportion [6], [14], differently to Sol [6], [25], Yoshitake and Moritani [25] suggested that such behaviour of increase in MMG signal amplitude would be due to a higher composition of fast MUs in GM. Despite the difference in methodological approach (microphone × accelerometer), our results are in agreement with theirs, which is corroborated by Jaskólska et al. [13] that report similar trends in MMG signals provided by both transducers with an increase in muscular contraction that, in turn, ratify the linear behaviour of the angular coefficients *a* observed in our study.

With respect to the angular coefficient *a* values, we did not observe any significant statistical difference among the directions associated to the signals *X*, *Y* and *R*, although they were statistically significantly lower than those observed in Sol and BB muscles as can be seen in Fig. 8. Additionally, the level of dispersion of GM angular coefficient *a* values seems to reveal a lower rate of increasing in magnitude of the MMG signal at levels towards 100% of MVIC, as previously reported by Herda et al. [10], in comparison with the other two muscles. Moreover, the MMG_y angular coefficient *a* values are even lower when compared to those from MMG_X and MMG_R with a high percentage of volunteers (~ 70%) reaching values below "1". We hypothesize a higher complacence of the GM tendon at low %MVIC as the main source of MMG_Y magnitude and so resulting in lower values of angular coefficient *a* values than those computed in *X* and *R* axes.

4.2.2. Phase$_R$

The results are similar to those found for the BB muscle, which were marked by no significant statistical differences. The hypothesis based on an influence of changes in the architecture that happen during a force increasing of an ICT in the mean angle phase seems not to be applicable to GM muscle. In spite that GM might be considered a muscle with pennate properties [15] but with muscle fibers obliquely distributed to the skin surface, its particular muscle architecture must be taken into consideration in respect to the MMG signal acquisition. Moreover, we must also consider the possibility of a "mechanical crosstalk" between Sol and Gastrocnemius muscles, since both muscles control primarily the plantar flexion. Therefore, it is quite possible that those variables had influenced the MMG signal acquisition and so masking changes in the mean angle and, therefore, a reliable interpretation of these data.

4.3. BB muscle

The BB muscle has been widely investigated regarding muscle gradation force and the MMG signal properties [1], [3], [13], [18], [19]. Concerning the interpretation of the angular coefficient *a*, the findings of the present study are in accordance with those observed in the literature, which report an increase in the mean MMG_{RMS} values as the muscle force level also increases up to 80% of MVIC [17], [18]. This behaviour is related mainly to the MU recruitment with the aim of increasing the isometric force level [3]. Differently from Sol, BB muscle is usually mentioned as a muscle composed by similar percentages of fast and slow fibers in a fusiform shape [6], [14]. Matta et al. [18] evaluated the MMG signals from BB muscle by means of a biaxial accelerometer and found an increased behaviour in the magnitudes of MMG_X and MMG_Y as the muscle force also increased, similarly to the present study and with a plateau from 60–80% of MVIC.

Our BB data presented quite similar distributions of angular coefficient *a* values in the three axes. However, the most interesting result is related to MMG_Y pattern, which was also observed by Matta et al. [18] although they submitted the volunteers to isometric contractions in a different muscle length – a higher one – exceeding the optimal position to reach the maximum force generation [11]. Since different lengths of muscular elongation interfere with the capacity of muscle force generation, they could also be expressed in the MMG signal. This hypothesis might explain partially the lack of significant statistical differences in MMR_{RMS} between 40% and 60%, and 60% and 80% of MVIC observed by these authors. However, this same hypothesis seems not to be suitable to our results because approximately 50% of angular coefficient *a* values obtained from *Y* axis reached "1". Thus, as suggested by Herda et al. [10], angular coefficient *a* values higher than "1" meant an increase in the MMG signal magnitude as force also increases, disregarding any decrease or attenuation, similarly to the findings of Matta et al. [18].

4.3.1. Phase$_R$

There were no significant statistical differences when we compared the $\overline{\text{Phase}_R}$ angles among the levels of contraction. This result suggests that the ratio between the magnitudes of the MMG_X and MMG_Y signals was maintained constant while the muscle force was increased up to 100% of MVIC. Beck et al. [4] found similar signal components in both directions by means of a laser transducer but in rectus femoris muscle that is bippeniform. This result could be explained by the homogeneity of the BB muscle architecture, i.e., a fusiform shape, while it increases the %MVIC. Since BB muscle is considered a muscle with some few fibers projected in a pennate direction [2], [9], it means that BB fibers can also present some minimal changes in their direction while it contracts. It is possible that those changes in the architecture are not enough to alter the correspondence among the axis and the accelerations resulting from the different %MVIC.

4.4. The antilog of *y*-intercept and the skinfold thicknesses

Considering the sources of the MMG signals, there seems to be an unequivocal influence of anthropometrical variables in their temporal and frequency domains. Skin and adjacent tissues operate as a low-pass filter for mechanical vibrations from the skeletal muscle twitches that can be measured nearby the skin surface. To our knowledge, however, few studies report the effects of anthropometrical variables on MMG signal content [13], [16], [24]. Jaskólska et al. [13], Krueger et al. [16] and Polato et al. [24] observed a significant effect of skinfold thickness on frequency domain in BB muscle. Conversely, Krueger et al. [16] and Polato et al. [24] suggest a lower susceptibility of MMG signal to the skinfold thickness in the temporal domain although a different behaviour has been observed when the transducer is a microphone [25].

Interestingly, the BB skinfold thicknesses were statistically significantly lower than those collected from Sol and GM spots (Fig. 10). Herda et al. [10] mentioned that lower values in skinfold thickness are associated with higher values of the antilog of *y*-intercept or coefficient *b*. The log transform coefficient *b* is regarded as a "gain factor", which might mean an up or a downward inflection in an exponential relationship between the MMG signal magnitude and muscular torque output, without changing the shape of the curve fitted to the data [10]. In this context, our results are in full agreement with those postulated by these authors, which is corroborated by the largest significant statistical coefficient *b* values obtained from *X*, *Y* and *R* axes in comparison to those reached by Sol and GM muscles. To our knowledge, no previous study reported a similar result and, therefore, we must take into account not speculatively the hypothesis of considering the skinfold thickness contributing to the MMG signal morphology.

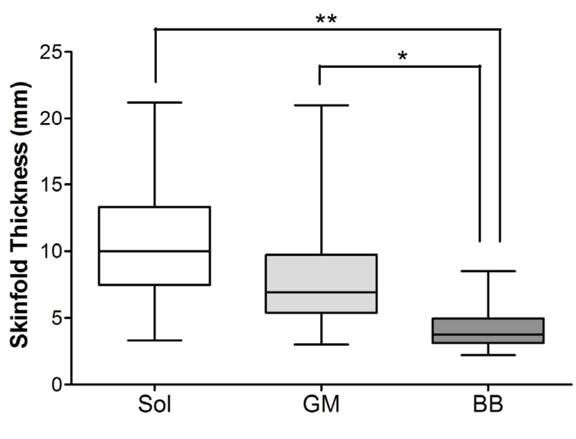

Fig. 10. Box-Whisker plots from the skinfold thicknesses of Sol, GM and BB. There were significant statistical differences between Sol and BB (**P = 0.000107), and GM and BB (*P = 0.000201)

5. Conclusion

Since we have found no previous studies that have investigated simultaneously the longitudinal and lateral vibration properties of muscle fibers by means of an accelerometer, a biaxial transducer seems to be an interesting approach in muscle mechanical studies since both axes looks like to supply different information related to isometric contraction properties. MMG signal features contained in two or more axes of an accelerometer can also provide further details not distinguished by a uniaxial transducer or even by a typical surface EMG. Even though bi or triaxial accelerometers seem to sound promising in evaluating MU strategies during contraction, MMG signals from muscles with complex architectures and fibers orientation such as GM must be carefully interpreted. Additionally, the alternative approach of MMG signal analysis proposed by Herda et al. [10] seems to better clarify the relationship between MU strategies and muscle force output. Anyhow, it is suggested that a triaxial accelerometer plus ultrasound measurements would contribute to clarify other poorly understood mechanisms that underlie the gradation of muscle force.

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