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Study of muscle fatigue in isokinetic exercise with estimated conduction velocity and traditional electromyographic indicators

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Abstract **Introduction**: In the study of human biomechanics, it is often desirable to evaluate fatigue in the muscles that are involved in performing a particular task. Identifying the phenomena responsible for this condition is a problem that in most cases is complex and requires appropriate research mechanisms. Isokinetic dynamometry (ID) and surface electromyography (SEMG) are two techniques widely used in studies on strength and muscle fatigue. Their effectiveness is conditioned upon a good understanding of their limitations and the adoption of procedures to fully exploit the potential of each one. The main goal of the present study is to verify whether the electromyographic parameters, especially the conduction velocity (CV), are sensitive to the fatigue instauration process within sets of maximal isokinetic contractions. CV is a basic physiological parameter directly related to muscle activity and still little explored in experiments combining ID and SEMG. **Methods**: Instrumentation architecture that combines ID and SEMG was used to estimate electromyographic and biomechanical parameters in protocols of maximum intensity isokinetic knee extension exercises. This architecture allows for limiting the parameter estimates to a specific region of isokinetic exercise, called the isokinetic load range (ILR), where one can consider that the angular velocity is constant and the SEMG signals are cyclo-stationary. Electromyographic signals were acquired using an array of electrodes. **Conclusion**: The results suggest that CV and the other SEMG parameters, including amplitude and frequency descriptors, are sensitive to detect a fatigue process only in protocols that restrict the analysis to ILR and that also bring the subject to a state of fatigue quickly.

Keywords Muscular fatigue, Dynamic contraction, Isokinetic dynamometry, Surface electromyography.

Introduction

Surface electromyographic signals have attracted great attention from areas that address sports and orthopedic medicine, physiotherapy and biomechanics. One of the main reasons for the wide interest in surface electromyography (SEMG) is based on the possibility of accessing the muscular structure and function through a non-invasive process (Merletti and Parker, 2004). In sport sciences, particularly in the area of high performance sports, research into muscular fatigue phenomena during dynamic activities is the subject of interest (Clarys, 2000). Obtaining the parameters indicative of muscular fatigue can help in proposing new experimental protocols and developing new techniques for evaluating sport performance.

In general, fatigue may be defined as a reduction in the ability of a muscle to generate force or power in an induced exercise (Gandevia et al., 1996; Søgaard et al., 2006). Identifying the phenomena responsible for this condition is a problem that most of the time it is complex. The physiological issues that cause muscle fatigue vary from the accumulation of metabolites in muscle fibers to the possibility of an incorrectly generated command by the motor cortex (Enoka and Duchateau, 2008). Therefore, the main challenge of SEMG studies is to properly identify the electrical manifestation related to such phenomena.

In static contractions, the accumulation of biochemical byproducts within the muscle is the main fatigue phenomenon, which is known as localized fatigue and results in myoelectric signal non-stationarities, such as the frequency scaling of the power spectrum (Basmajian and De Luca, 1985). This behavior of the myoelectric signal is called slow non-stationarity (or quasi-stationarity), given that the SEMG signal may preserve its characteristics during several seconds (Bonato et al., 2001), after which new muscle fiber recruitment starts changing the signal properties (Farina, 2006). It is well established in scientific literature on fatigue analysis that under quasi-stationary conditions, the amplitude parameters (e.g., average rectified value and root mean square) tend to increase, whereas frequency indicators (e.g., mean frequency and median frequency) tend to decrease during sustained fatiguing contraction (Lloyd, 1971; Moritani et al., 1982; Arendt-Nielsen and Mills, 1988; Kallenberg et al., 2007). However, under dynamic

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conditions the assumption of quasi-stationarity does not hold because the frequency content of the signal continuously changes over time. Changes in muscle length, force, and electrode position contribute to the fast non-stationarities of the SEMG signal, which are unrelated to localized fatigue. Therefore, certain techniques are used to minimize fast non-stationarities by setting out a range in which the mechanical changes of muscles are cyclic, i.e., they are repeated periodically (Bonato et al., 2001). According to Farina (2006), the study of fatigue in the dynamic case must necessarily go through identification of the proper conditions in which SEMG signals may be assumed to be relatively stationary. Hence, specific protocols must be developed in order to limit and isolate the factors that cause fast non-stationarities.

A suitable technique that aids the study of muscle fatigue in SEMG signals in avoiding fast non-stationarities is isokinetic dynamometry (ID), because it makes it possible to (1) perform controlled cyclic dynamic contractions, (2) isolate each constituent phase of the cycle, and (3) estimate the respective biomechanical variables. The isokinetic dynamometer is a device that controls the speed and execution of the exercise. More force being exerted on the dynamometer lever increases the energy absorbed from the member in movement by the control mechanism and returned as additional resistance to the movement (Brown et al., 1995a, 1995b). Therefore, the movement occurs at a predetermined constant speed during a range of movement called the load range (LR) (Brown, 2000), where the dynamometer imposes an external load in opposition to the movement. In fact, the speed is constant only in the LR part where the imposed load matches the muscular force exerted. Schwartz et al. (2010) called this part isokinetic load range (ILR) and they showed that ILR is the range where the SEMG signal has the highest level of stationarity (Schwartz et al., 2012). There is another range of LR called velocity overshoot (VO), prior to the ILR, where the speed oscillates when the dynamometer is still adjusting to the force applied. Figure 1 illustrates the stages of the knee extension isokinetic exercise.

There are many fatigue studies that estimate amplitude and frequency indicators combining the SEMG and ID techniques given their recognized applicability. Horita and Ishiko (1987) verified that the median frequency of SEMG signal decreased during isokinetic knee extensions performed for 30 s and 60 s. Molinari et al. (2006) studied the manifestations of muscle fatigue during concentric extension-flexion and eccentric isokinetic movements of the knee; they also observed an accentuated decrease in MNF value due to fatigue process. Kellis (1999) assessed the SEMG in a knee extension isokinetic endurance test and reported an increment in the signal amplitude of the middle and the last repetitions in comparison to the values of the first contractions. Gerdle et al. (2000) examined SEMG signals in individuals who performed 100 isokinetic knee extensions and verified that frequency parameters are more correlated to torque reduction than to amplitude variables.

However, the effectiveness of the conduction velocity indicator to describe fatigue in experiments combining ID and SEMG seems to be little explored. The conduction velocity (CV), defined as the velocity with which the action potential spreads through the muscle fibers (Merletti et al., 1990), is a basic physiological parameter that has been proven to be related to muscle fatigue during cycling activities (Farina et al., 2004; Lenti et al., 2010; Pereira et al., 2013). González-Izal et al. (2010) investigated the CV behavior during four sets of isokinetic concentric and

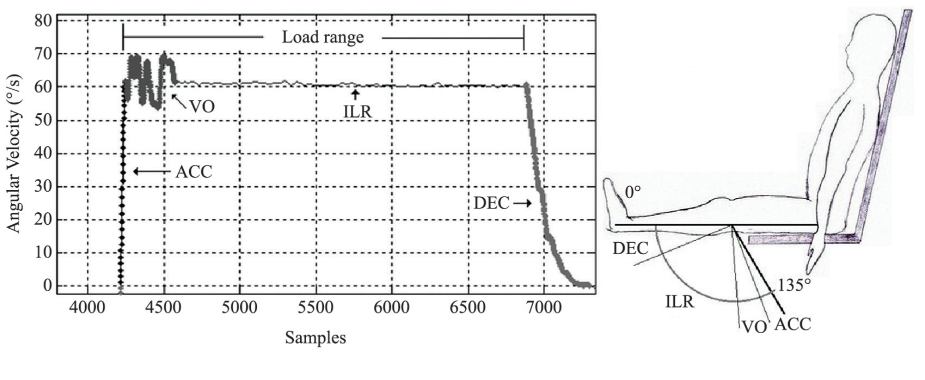

Figure 1. The angular velocity signal with the respective phases of the isokinetic knee extension exercise at 60°/s. Acceleration (ACC), velocity overshoot (VO), isokinetic load range (ILR) and deceleration (DEC).

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eccentric contractions. They have found that CV presented a significant decrement between the first and the last 10 repetitions of the test. To the best of our knowledge, this is the only study that has evaluated the CV behavior in isokinetic exercise.

During the development of the present research, no studies were found in scientific literature that estimated CV in multiple sets of isokinetic contractions at different angular velocities. Moreover, no other study has also considered the estimate of CV and traditional electromyographic parameters only within the ILR segment. According to Schwartz et al. (2012), the ILR is the segment where the best levels of stationarity of SEMG signal are found. Based on the aforementioned, the main goal of this paper is to present an experimental protocol for investigating muscle fatigue to consider, with the aid of an integrated instrumental architecture, these two factors that were never found together in related experiments: (1) the estimation of CV at different angular velocities of isokinetic contractions, and (2) the estimation of SEMG fatigue parameters within the ILR. Additionally, biomechanical descriptors were also estimated in ILR to provide greater consistency to the analysis. The main expectation was to observe, under what are supposed to be the best experimental conditions, if the trends of these descriptors are coherent with those normally reported for the case of static exercise (or isometric contraction).

Methods

Sixteen healthy male subjects participated in the experiment on fatigue investigation, aged 26.8 ± 4.7 years old, height 1.76 ± 0.05 m and body mass of 79.2 ± 9.4 kg, with no history of orthopedic diseases. They read and voluntarily signed a consent form before participating in the experiment, which was approved by the Health Sciences College Ethics Committee at the University of Brasília (UnB).

The instrumentation architecture (Figure 2) proposed by Schwartz et al. (2011) was assembled in the laboratory to allow electromyographic and biomechanical signals to be acquired in a synchronized way with equal sampling rates.

An isokinetic dynamometer (Biodex Corporation, Biodex System 3 Pro model) was calibrated and prepared for performing the knee joint controlled exercises in the isokinetic concentric mode (Brown, 2000). The dynamometer control software was the System 3 (Biodex Medical Systems, v. 3.4 17/05/2006) application. The knee joint was passively moved to a 0° position of extension, which corresponds to the maximum possible extension. After, the knee was flexed approximately 5° to 10° to a comfortable position, recorded in the dynamometer control system as the maximum extension point for the experiment (extension mechanical stop). Then, the maximum flexed point (flexed mechanical stop) was defined to guarantee an 85° range of motion. The analysis of gravity compensation was performed by Biodex System 3

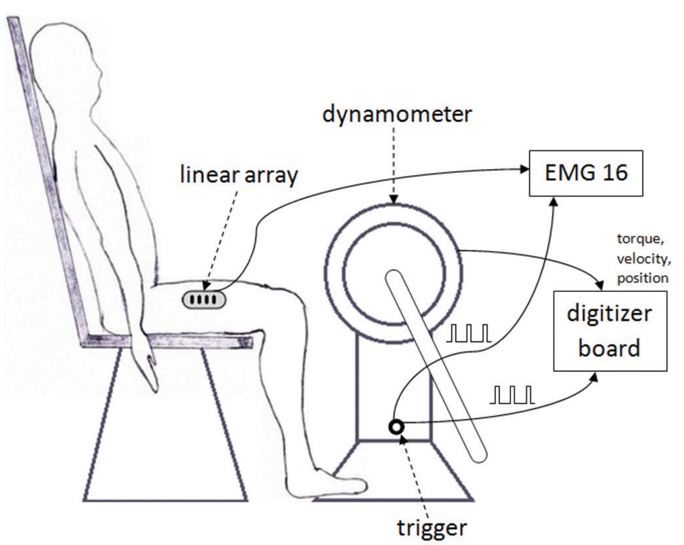

Figure 2. Instrumentation architecture.

Pro software control. The DB-15 interface of the dynamometer was connected to an adapter built with three BNC connections (one for each biomechanical signal) connected to an A/D converter (National Instruments, model BNC-2120) which digitized the biomechanical signals to 2,048 samples/s and 12 bit resolution, with the aid of an experiment control application based on a LabVIEW® v. 5.1 (National Instruments) tool. This procedure was established to match the temporal resolution of the biomechanical signals to the SEMG signals – a necessary condition for a joint analysis – given that the internal A/D conversion of the isokinetic dynamometer takes place only at 100 samples/s. The SEMG signals were acquired during *vastus lateralis* muscle contractions, through a flexible linear array (Ottino Bioelettronica) with eight electrodes (5 mm between electrodes) positioned between the innervation zone and the tendon region (De Luca, 1997; Masuda et al., 1985). The electrode array was attached to a multichannel electromyograph (LISiN–OT Bioelettronica Snc, EMG-16 model) configured for digitizing the signals at a rate of 2,048 samples/s (12 bit resolution) with a gain of 2,000. The EMG-16 also filters the signal acquired with a bandpass filter (4th order Bessel filter with –3 dB bandwidth = 10-500 Hz). The EMG-16 control software was Emgacq v. 1.0 (Centro di Bioingegneria, Politecnico di Torino). A position sensor composed of a magnetic switch (attached to the body of the dynamometer) and a magnet (attached to the movement rod) recorded the exact time of each knee extension. This information was sent simultaneously to the electromyograph and to the A/D converter, which enabled the subsequent synchronization of the biomechanical signals to the SEMG signals.

Once the equipment was set up, each subject performed three (3) sets of ten (10) repetitions of maximal concentric extensions at 60º/s with one minute of rest between sets. After 20 minutes, which is reported as time enough to provide full recovery of muscle function in isokinetic tests (Bilcheck et al., 1993; Pinciveiro et al., 1998), the three sets were repeated at 180º/s, with the same rest period between sets. A consistent and moderate verbal encouragement (without shouting) was given during the execution of each set in order to get the maximum performance from each subject, and no visual feedback on the torque produced in each repetition was available to the participants (McNair et al., 1996).

With the aid of a computer interface (Schwartz et al., 2008) developed to perform the algorithms for signal analysis proposed by Schwartz et al. (2011), the segments of the SEMG signal related to the VO and ILR steps (see Figure 1) were extracted from the 10 repetitions in all of the three sets, for both angular velocities (60º/s and 180º/s) and for all of the subjects. The mentioned interface is responsible for, among other things, making the gravity compensation for the analog signals.

The criteria for choosing the SEMG channel for estimating the electromyographic descriptors took place by verifying the signal-to-noise ratio through the idle-channel noise method (Schwartz et al., 2011). In this technique, the noise signal was captured with the electrode array already attached to the muscle with the subject, however, in a relaxed state (without voluntary muscle contraction). Only channels with a signal-to-noise ratio greater than 20 dB were considered.

The fatigue indicators studied were the temporal amplitude parameters (Average Rectified Value (ARV) and Root Mean Square (RMS)), the frequency parameters (Median Power Frequency (MDF) and Mean Power Frequency (MNF)) (Farina and Merletti, 2000), and the Conduction Velocity **(**CV) (Farina and Merletti, 2004). Since CV is the main parameter investigated, it is thoroughly described below.

Conduction velocity

The representative model for an array of SEMG sensors with M channels can be represented according to Farina and Merletti (2004) as

$$x_m[n] = x_1[n - m\zeta] + e_m[n];\ m = 0,1,\ldots M-1;\ n = 0,1,\ldots,N-1 \quad (1)$$

where $x_m[n]$ corresponds to a delayed version of the sequence $x_1[n]$, ζ is the time delay between two adjacent channels of SEMG in the arrangement of sensors, N is the number of samples in the observation window, M is the number of channels acquired in SEMG array of sensors and $e_m[n]$ corresponds to m-channel additive white noise with Gaussian probability density function with variance σ_e and average $\mu_{e.}$

The problem is modeled by the technique of maximum-likelihood estimation (MLE). Since the distance between successive SEMG electrodes is known, calculating the parameter ζ is desired (delay in samples between adjacent channels) to estimate the muscle conduction velocity. Therefore, a squared error ξ^2_{MLE} function must be established. The solution of the problem is obtained by minimizing the ξ^2_{MLE} in function of the variable ζ. The squared error is defined as

$$\xi^2_{MLE} = \left(1 - \frac{1}{M}\right)\sum_{m=0}^{M-1}\xi^2_m \quad (2)$$

and

$$\xi^2_m = \sum_{n=0}^{N-1}\left|x_m[n] - \frac{1}{M-1}\sum_{r=0,r\neq m}^{M-1} x_r\left[n + (m-r)\zeta\right]\right|^2 \quad (3)$$

Applying the Discrete Fourier Transform (DFT) in the second component of the equation results in

$$\xi_m^2 = \frac{2}{N}\sum_{k=0}^{\frac{N}{2}-1}\left|X_m[k] - \frac{1}{M-1}\sum_{r=0,r\neq m}^{M-1} X_r[k]\, e^{\frac{2\pi j}{N}(m-r)k\zeta}\right|^2 = \frac{2}{N}\sum_{k=0}^{\frac{N}{2}-1}\left|\gamma[k,\zeta]\right|^2 \quad (4)$$

where DFT of length N samples $x_m[n]$ is defined as

$$X_m[k] = \sqrt{\frac{2}{N}}\sum_{n=0}^{N-1} x_m[n]\, e^{\frac{-2\pi j}{N}kn} \quad (5)$$

The first derivative of ξ_{MLE}^2 is of the form

$$\frac{\partial \xi_{MLE}^2}{\partial \zeta} = \frac{4}{N}\sum_{k=0}^{\frac{N}{2}-1}\left[Re\{\gamma[k,\zeta]\}\frac{\partial Re\{\gamma[k,\zeta]\}}{\partial \zeta} + Im\{\gamma[k,\zeta]\}\frac{\partial Im\{\gamma[k,\zeta]\}}{\partial \zeta}\right] \quad (6)$$

and the second derivative of ξ_{MLE}^2 is shown in Equation 7.

$$\frac{\partial^2 \xi_{MLE}^2}{\partial \zeta^2} = \frac{4}{N}\sum_{k=0}^{\frac{N}{2}-1}\left[\left(\frac{\partial Re\{\gamma[k,\zeta]\}}{\partial \zeta}\right)^2 + Re\{\gamma[k,\zeta]\}\frac{\partial^2 Re\{\gamma[k,\zeta]\}}{\partial \zeta^2} + \left(\frac{\partial Im\{\gamma[k,\zeta]\}}{\partial \zeta}\right)^2 + Im\{\gamma[k,\zeta]\}\frac{\partial^2 Im\{\gamma[k,\zeta]\}}{\partial \zeta^2}\right] \quad (7)$$

To obtain the minimum ξ_{MLE}^2 as a function of ζ the Newton's iterative method is applied

$$\zeta_{i+1} = \zeta_i - \frac{\frac{\partial \xi_{MLE}^2}{\partial \zeta}}{\frac{\partial^2 \xi_{MLE}^2}{\partial \zeta^2}}, i = 1,2,\ldots \quad (8)$$

where i is the i-th iteration of the algorithm. The process should stop when it reaches a threshold error, when the parameter ζ increasing is not significant from one iteration to the next, or when a set amount of iterations is reached. The conduction velocity can finally be estimated by

$$CV = \frac{d}{\zeta}(cm/s) \quad (9)$$

where d is the distance between successive SEMG electrodes.

Fatigue indices

In the experimental procedure, the acquired SEMG signals are segmented into windows of 256 samples (125 ms). Successive windows have an overlap of 64 samples (31.25 ms). At each analysis window, the temporal and frequency parameters are calculated, temporally arranged and plotted. For this set of points, a linear regression function is estimated to describe the behavior of the respective parameter as the experimental protocol is performed (see Figure 3a).

The fatigue analysis is usually based on indices that reflect the evolution, during contraction time, of measurable quantities such as force and angular velocity, which are biomechanical parameters, or variables associated with the SEMG signal (Merletti and Parker, 2004). In the present investigation, the evolution of the SEMG parameters was performed by determining regression lines from the signal acquired during isokinetic repetitions. From this point, a fatigue

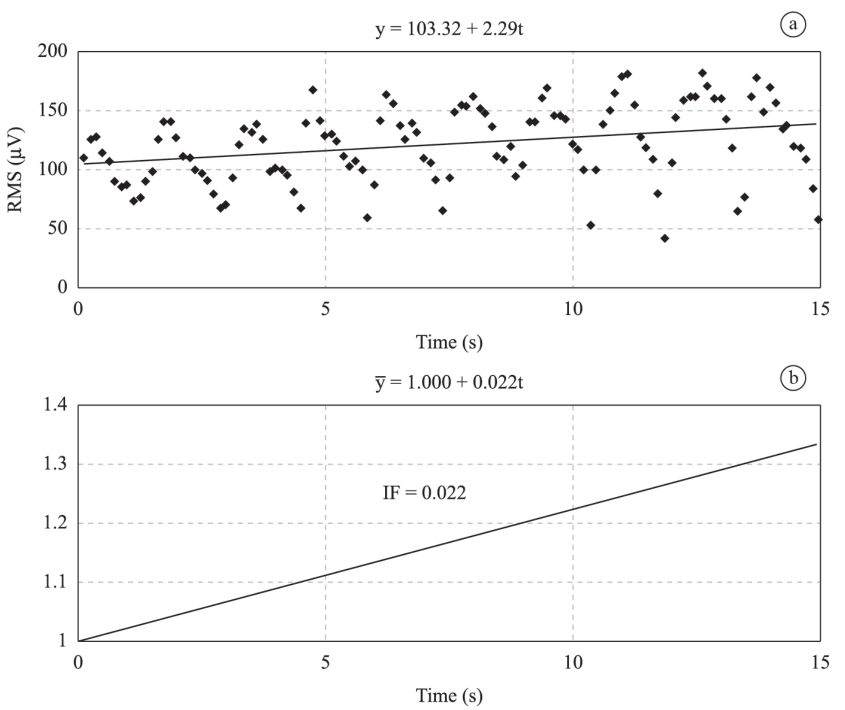

Figure 3. (a) Linear regression for the RMS parameter and (b) the respective index of fatigue (IF).

index was obtained by considering the corresponding regression line slope (Merletti et al., 1990). The angular coefficient (slope) of the regression linear function is normalized by the respective linear coefficient (initial value). Then, it was used as an index of fatigue (Bolgla and Uhl, 2007; Merletti et al., 1990) as showed in Equations 10a, 10b, and 11.

$$y = a + bx \tag{10a}$$

$$\bar{y} = 1 + \frac{b}{a}x \tag{10b}$$

$$IF = b / a \tag{11}$$

where "x" is associated with the runtime of experimental protocol, "b/a" is the slope that represents the index associated with the muscle fatigue phenomenon, and "y" represents the amplitude of temporal or frequency parameter. Figure 3 illustrates the linear regression function and the respective index of fatigue (IF).

Two biomechanical variables, PT (peak torque) and PTBW (peak torque to body weight), were also estimated to aid in analysis. In the first case, PT was considered for each of the 10 repetitions in the isokinetic set and the linear regression line in each set was determined. From here, the fatigue index of PT was calculated in the same way as Equations 10a, 10b, and 11. This index allows for the observation of the behavior of the variable within the set. In the second case, the maximum value of PTBW was identified among the 10 repetitions of a set and the average was calculated among all of the subjects for the same set. In this case, the index permits the observation of the behavior among the sets.

Statistical analysis

Statistical comparisons of fatigue indicators were carried out among the sets at the same angular velocity, with a significance level of 0.05 (double-tailed) and 95% confidence interval. A Shapiro-Wilk (De Sá, 2007) test for normality was used, which is reliable for small sample numbers (approximately 10). For comparing the averages between two datasets, the t-Student test for dependent samples was utilized for normal distributions and the Wilcoxon Signed-Rank test was used for non-normal ones. When comparing the averages among three datasets, the One-Way ANOVA test together with Tukey's HSD (*post hoc*) test was used for normal distribution whereas the Friedman test was used for the non-normal case.

Results

Based on the signal-to-noise ratio (> 20 dB) criterion, two subjects were discarded from both the 60º/s and the 180º/s sets. Hence, the average fatigue index was calculated from 14 subjects. Table 1 shows the fatigue index averages for the SEMG descriptors. Figure 4 shows the fatigue index averages for the peak torque and illustrates the averages, in the sets, of the maximum PTBW value of each subject. Table 2 presents statistical analyses where the fatigue index and the PTBW variable are compared among the sets at the same angular velocity. Table 3 compares the fatigue indices and the PTBW variable between the angular velocities (60º/s and 180º/s), in the same isokinetic set.

Discussion

For this study, the strategy of analyzing only the fixed range of the myoelectric signal corresponding to the ILR stage of the isokinetic exercise was adopted. This approach was taken in order to minimize non-stationary effects, common in dynamic contractions, caused by changes in muscle length, by variation in the force applied during the different cycles, and by the movement of the electrode on the skin relative to the muscle (Bonato et al., 2001). The ILR range was chosen due to the satisfactory levels of stationarity presented in the previous study by Schwartz et al. (2012).

In the first observation in Table 1 for the 60º/s velocity, the strategy adopted appears to confirm the initial expectation that the SEMG descriptors may have, under certain conditions, the same behavior for both

Table 1. Average fatigue indices for the SEMG descriptors.

Velocity	Set	SEMG Descriptors (10^{-3})				
		CV	**RMS**	**ARV**	**MDF**	**MNF**
60º/s	1	–16.4 (18.5)*	14.0 (15.0)	14.7 (15.3)	–15.8 (6.8)	–15.0 (6.7)
	2	–18.8 (22.9)*	8.1 (11.2)	8.0 (11.6)	–12.3 (7.4)	–11.9 (7.3)
	3	–51.6 (158.5)*	4.1 (10.2)	3.3 (11.1)	–13.7 (7.9)	–13.2 (6.4)
180º/s	1	103.5 (336.9)*	37.8 (44.2)	38.1 (50.9)	–5.0 (42.0)*	–13.5 (27.8)
	2	–179.5 (688.1)*	43.8 (47.3)	46.4 (52.4)	15.6 (65.0)*	–5.9 (26.8)
	3	–1.9 (56.1)*	15.9 (33.4)	15.5 (39.0)	–2.1 (26.7)	–4.7 (21.6)

Note. The values are represented by the average (± SD). * Significantly non-normal ($p < 0.05$) – Shapiro-Wilk test.

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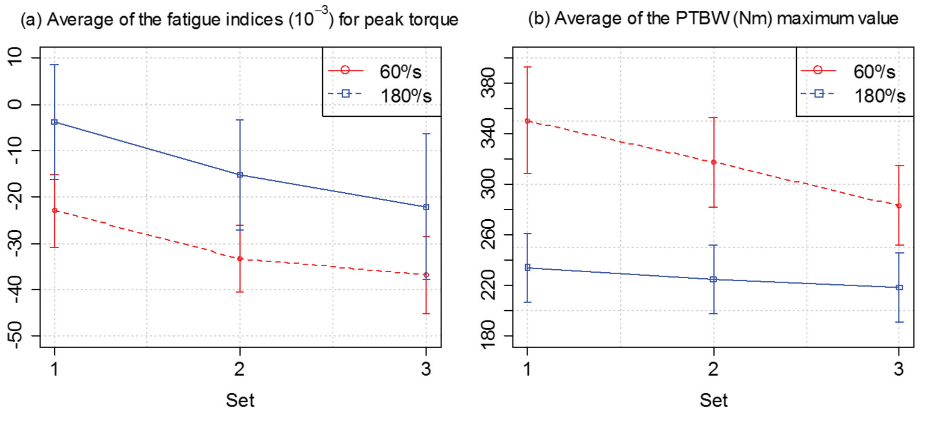

Figure 4. Behavior of biomechanical variables.

Table 2. Statistical comparisons of the indices of fatigue (IF) and the PTWB variable among the sets with the same angular velocity.

Variable	Results	
	60°/s	180°/s
IF_{CV}	(C) S1 = S2, S2 = S3, S1 = S3	(C) S1 = S2, S2 = S3, S1 = S3
IF_{RMS}	(A) S1 = S2, S2 = S3, S1 = S3	(A) S1 = S2, S2 = S3, S1 = S3
IF_{ARV}	(A) S1 = S2, S2 = S3, S1 = S3	(A) S1 = S2, S2 = S3, S1 = S3
IF_{MDF}	(A) S1 = S2, S2 = S3, S1 = S3	(C) S1 = S2, S2 = S3, S1 = S3
IF_{MNF}	(A) S1 = S2, S2 = S3, S1 = S3	(A) S1 = S2, S2 = S3, S1 = S3
IF_{PT}	(*) (A) (B) S1 > S2, S2 = S3, S1 > S3	(*) (A) (B) S1 = S2, S2 = S3, S1 > S3
PTBW	(*) (A) (B) S1 > S2, S2 > S3, S1 > S3	(A) S1 = S2, S2 = S3, S1 = S3

(A) One-Way ANOVA, (B) Tukey (*post hoc*), (C) Friedman Test. In all comparisons H_0 was accepted ($p > 0.05$), except in those marked with an asterisk(*).

Table 3. Comparison of the indices of fatigue (IF) and the PTWB variable between the 60°/s and 180°/s angular velocities in a single isokinetic set.

Variable	Set 1	Set 2	Set 3
IF_{CV}	(2) $IF_{CV60} < IF_{CV180}$	(2) $IF_{CV60} = IF_{CV180}$	(2) $IF_{CV60} = IF_{CV180}$
IF_{RMS}	(1) $IF_{RMS60} = IF_{RMS180}$	(1) $IF_{RMS60} < IF_{RMS180}$	(1) $IF_{RMS60} = IF_{RMS180}$
IF_{ARV}	(1) $IF_{ARV60} = IF_{ARV180}$	(1) $IF_{ARV60} < IF_{ARV180}$	(1) $IF_{ARV60} = IF_{ARV180}$
IF_{MDF}	(2) $IF_{MDF60} = IF_{MDF180}$	(2) $IF_{MDF60} = IF_{MDF180}$	(1) $IF_{MDF60} = IF_{MDF180}$
IF_{MNF}	(1) $IF_{MNF60} = IF_{MNF180}$	(1) $IF_{MNF60} = IF_{MNF180}$	(1) $IF_{MNF60} = IF_{MNF180}$
IF_{PT}	(1) $IF_{PT60} < IF_{PT180}$	(1) $IF_{PT60} < IF_{PT180}$	(1) $IF_{PT60} < IF_{PT180}$
PTBW	(1) $PTBW_{60} > PTBW_{180}$	(1) $PTBW_{60} > PTBW_{180}$	(1) $PTBW_{60} > PTBW_{180}$

(1) *t*-Student, (2) Wilcoxon Signed-Rank.

the dynamic case and the static case. The three sets revealed a decreasing trend for CV, MNF and MDF and a growing trend for RMS and ARV, which has been widely reported on in the scientific literature as resulting from fatigue in static contractions (Lloyd, 1971; Moritani et al., 1982; Arendt-Nielsen and Mills, 1988; Kallenberg et al., 2007). This also indicates that in the maximum intensity exercise at 60°/s, with 10 repetitions, the fatigue state is already reached in the execution of a single set. Figure 4a reinforces this fact by the accentuated negative slopes of PT. A similar effect was observed by González-Izal et al. (2010) when studying the leg press action in a sitting position. Regarding the behavior of amplitude and frequency parameters, Gerdle et al. (2000) reported positive correlations among them and PT, i.e., as PT decreases, MNF and RMS also decrease. This result corresponds to a similar behavior to that found in this study for MNF, but not for RMS. In fact, Gerdle et al. (2000) tested the 90°/s velocity during 100 consecutive isokinetic knee extensions. Thus, the recruitment of new motor units can occur in a different moment from that at 60°/s, which could explain the different trend.

When the parameters of the SEMG were compared among the three sets (see Table 2), no significant difference in the slopes was found. This finding seems to corroborate the results reported by Larsson et al. (1999) that have also evaluated three sets of ten repetitions of isokinetic knee extensions and found no difference among SEMG descriptors of the beginning and the end of the sets. One might expect an increase in the fatigue level at each subsequent set and this would have been reflected in some change in the slopes when comparing the three sets, but in fact this did not occur. Nevertheless, the drop in the PTBW at 60º/s, which is approximately 10% at each set (Figure 4b), shows signs of increased fatigue with a significant difference among the sets (see Table 2). This suggests, for the experimental protocol used, that the degree of fatigue increases with each set and that the 1-minute rest is capable of providing considerable recovery of muscular activity among the sets (approximately 90% of the previous set's capacity).

In other words, it may be said that, due to the rest, the increase in the fatigue level from one set to another was not very intense and is not reflected in the SEMG variable behavior. In any case, these parameters vary within a range in which, having reached the limits, they lose sensitivity for detecting greater degrees of fatigue. Farina et al. (2002) reported, for example, that the MDF and the MNF reflect the recruitment of new motor units – progressively larger and faster – with an upward slope until the recruitment of all of the units, whereupon the trend begins to decrease until it reaches a constant value. In the sets at 60º/s, given that they are maximum intensity exercises, the total recruitment already happens within the set. This phenomenon is also made evident in a similar result, but with an inverse relationship. This may be seen in Table 2; the slope in set 1, in which the muscle is in better condition for producing maximum torque, is greater than the slope in set 2, given the capacity for greater muscle power output (type IIb fibers) in the initial repetitions of set 1 (McCartney et al., 1983; Sargeant, 1994). Having lost the burst capacity, the difference between the intensity of the first and last repetitions becomes lower, resulting in lower and matching slopes in sets 2 and 3.

When the focus turns to the velocity of 180º/s, there seems to be no correspondence in the results with the static case. Nevertheless, at higher velocities, new questions need to be addressed. Firstly, the levels of force produced at high speeds are significantly lower than those reached at lower velocities (Parcell et al., 2002), which may be verified by the PTBW variable in Figure 4b and by the PTBW and IF_{PT} variables in Table 3. In establishing a relationship from Figure 4a, it may be noted that the peak torque at 60º/s is approximately 50% greater than the peak torque at 180º/s. From Figure 4a it may be seen that the potential in torque production falls 20% (7%) at 60º/s (180º/s). Therefore, it is reasonable to assume that the first set of exercises at 180º/s does not have enough intensity to bring the muscle to a state of fatigue. At such a velocity it may be necessary to sustain the effort for a longer period to identify the influence of the fatigue in SEMG parameters. Horita and Ishiko (1987) have verified that approximately 25 knee extensions at 180º/s are necessary to produce high levels of acidosis and, hence, significant changes in the SEMG indexes.

The positive slope of the CV in the first set (Table 1) is an indicator that there are still motor units to be recruited (Farina et al., 2002). This difference in relation to 60º/s also appears in Table 3 ($IF_{CV60} < IF_{CV180}$). Given that at each new set the accumulation of metabolites adds a greater degree to the state of fatigue, it only becomes evident during the sets, as seen in Table 2 in which significant differences for PT fatigue indicator are only detected between sets 1 and 3 (S1 > S3) and not observed between consecutive sets. Taking into account that at 180º/s the state of fatigue is only clearly detected in set 3, a comparison with the static case should only be considered in this set, where the similarity in behaviors is actually confirmed. This is consistent with the results from Andrade (2006) and Andrade et al. (2006), which suggest that these patterns observed in the isometric contractions are only observed in dynamic contraction protocols which quickly bring the subject to fatigue.

Under conditions of quasi-stationarity provided by the proposed experimental protocol and those in which the production of force has enough intensity to produce significant levels of fatigue, the behavior of the SEMG descriptors in the dynamic case is quite similar to the isometric case and very likely governed by the same physiological factors. These results open a new perspective to the study of fatigue and suggest the ILR segment as an appropriate way of controlling fast non-stationarities. However, this mechanism was effective only at maximum contraction intensity when all the muscle fibers are recruited. For velocities greater or equal to 180º/s, it may be assumed that a greater number of sets and/or repetitions would be necessary for producing fatigue and hence, for observing the same patterns. In future studies, appropriate levels of fatigue may be produced by the criteria – common in the area of isokinetic dynamometry – in which a set should only be interrupted when the subject carries out three repetitions with intensity 50% lower than the highest peak in the set.

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