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An EMG-driven model to evaluate quadriceps strengthening after an isokinetic training

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Abstract

This paper uses a Hill-type EMG-Driven model to estimate isometric quadriceps forces of 4 males subjects after a strength training period in an isokinetic device (Cybex-Norm), aiming for muscle hypertrophy. Before (PRE) and after (POS) training, quadriceps PCSAs were estimated from ultrasound measurements. Each individual followed a protocol of 10s submaximal (20% and 60%MVC) knee extension isometric contractions. Knee torque and surface EMG from three superficial quadriceps components were synchronously collected. EMG signal from *vastus intermedius* was estimated from other muscles. After training, the subjects presented an increase of 14.3 ± 5.1% of the maximum isometric torque, while the gain of muscle volume was 6.0 ± 3.0%. RMS error between EMG-driven and dynamometer joint torques, for the best estimation condition, were: 20%MVC PRE= 10.8 ± 3.8%, POS=11,0 ± 4.9%; 60%MVC PRE = 12.1 ± 3.2%, POS = 8.3 ± 2.5%. These results are analyzed under the viewpoint of modeling and experimental reliability.

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Keywords: muscle biomechanics; EMG-driven model; *quadriceps femoris*; strength training.

1. Introduction

EMG-Driven muscle models are becoming increasingly popular in the biomechanics community for estimating muscle forces in vivo. This approach presents a number of advantages when compared to other

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techniques for the same application: is non-invasive (if superficial EMG is the technique selected), does not require the definition of a more or less arbitrary *a priori* cost function, as in the case of optimization [1] or optimal control [2], does not require excessive analytical and numerical efforts, nor any special attention to dynamical or numerical stability problems, can easily cope with subject specific characteristics, can virtually be used in any motor task etc.

Applications of EMG-driven muscle models for determining muscle forces in ankle [3], knee [4], back [5], and upper limb as well, for normal and pathological situations [6] are found in literature. However, applying correctly and systematically the methodology in specific applications is still a challenge, even in motor tasks as simple as isometric contractions. Among the most important methodological questions, the following can be highlighted [7]: activation and contraction dynamics formulation and its parameters, number of EMG electrodes and its placement, EMG input normalization, filters shape, relative displacement of the electrode and whether using or not model parameter optimization techniques, to force the fit between the measured and estimated joint torques. These questions are being progressively elucidated in the literature, and for a broader class of applications, such as pathological conditions and physical training, particular studies are still required. This paper applies the EMG-driven technique to a strength training protocol of *quadriceps femoris* (QF). The research is focused on methodological questions, regarding how the strength gain and muscle architecture changes can be incorporated in the EMG-driven muscle model.

1.1. Strength gain and its implications in muscle architecture

Literature reports several works addressing the relationship between muscle use or disuse and force increase [8]. However, the biological mechanisms behind strength gain are not completely elucidated. The most important determinant of the amount of force a muscle can exert is its Physiological Cross-Sectional Area (PCSA) [cm^2]. This parameter is calculated using the well-know formula [9]:

$$\mathrm{PCSA} = \mathrm{MV} \cdot \cos(\alpha) / l^m \qquad (1)$$

where MV [cm^3] is the muscle volume, l^m [cm] the muscle fiber length and α the fiber pennation angle. Maximum force (F^{om}) [N], which is an important model parameter for defining muscle behavior, is found by multiplying the PCSA by maximum muscle specific tension (σ_m) [N/cm^2]. Strength training can modify slightly α and l^m [10], thus most of PCSA increase can be attributed to volume gain. Muscle PCSA is a primary indicator of the number of sarcomeres, i.e., of contractile material. On the other hand, it has been observed that force increase following strength training is often proportionally larger than PCSA gain [11], [8]. As a consequence, other factors can be alleged to explain such discrepancy [10]: neural facilitation, decrease of antagonist co-contraction, change of the angle of insertion of the muscle fibers in the tendon, tendon compliance, fiber type and sarcomeres force-length relationship transformations, among others. When excessive muscle hypertrophy is observed, as in physiculturists, a greater amount of non-contractile materials is expected to be infiltrated into the muscle, leading to a loss of force relative to the volume [12].

The most important factors, other than muscle size augment, seem to be those related with neural adaptations, which emerge in the beginning of the training period, up to 6-8 weeks. Agonist activity grows, related to neural motor firing frequency increase, better motor unities (MU) synchronization, recruiting of fast MU with bigger thresholds and increase in medullar motor neurons excitation [13]. In addition, reduction in presynaptic Ia, Ib and II afferent inhibition is also present [14] and antagonist coactivation has been also likely to decrease [15]. However, such coactivation patterns depend on the kind of movement and population group. For example, high velocity or alternating movements requires

naturally higher joint stiffness to prevent damage. For example: according to Häkkinen [16], old women presents higher coactivation compared to young adults. However, the elder group was able to decrease coactivation with training, that was not observed in young individuals.

The amount of force gain not resulted from PCSA increase is accounted in the maximum muscle specific tension parameter (σ_m). For entire muscles, σ_m values varying from 10 to 100 N/cm^2 are reported in literature [17]. In an important reference [18], which gave origin to maximum force values that are used in SIMM [19] and, more recently, in Opensim lower limb open-source model [20] , specific tension was considered as a scale factor that produced the best fit between the experimental and the simulated joint extensor moment. The resulting σ_m values varied from 30 to 73 N/cm^2, depending on the PCSA data source. Erskine et al. [17] tested twenty-seven adult males for estimating QF specific tension, using muscle volume and patellar tendon moment arm extracted from Magnetic Resonance Imagining data and remaining muscle architecture parameters from ultrasound. The authors calculated patellar tendon force from dynamometer MVC tests and found the specific tension dividing the total patellar tendon force by the entire QF PCSA. The obtained σ_m = 30.3 N/cm^2 was used as a reference value throughout this paper.

Here, an EMG-driven model of human QF is used to estimate knee torque in isometric contractions, applying a protocol that comprises low and medium-high, 10 seconds sustained contractions. The muscle model and the protocol have been extensively tested by our group, either for *triceps surae* (TS) and QF, in a number of papers, e.g. [3], [21]. The reference value of σ_m found by [17] was used to find maximum force (F^{om}) parameter, while PCSA was estimated from US measurements. The simulations were repeated using the same EMG input data, but substituting the nominal specific tension parameter by individualized values found from dynamometer MVC tests. The knee torque errors among the two approaches were compared. This work is a preliminary run with a small number of subjects (n=4), to investigate how to fit the EMG-driven model approach to strength gain training tests.

2. Methods

A group of 4 young health males (19 years old, 69,8 ± 12.16 Kg, 176.0 ± 7.15 cm), participated in the study. They underwent a long-term strength training, not periodized, in an isokinetic Cybex Norm dynamometer. The training protocol comprised 13 weeks of 2 weekly sessions (minimum of 48 hours interval, total 26 sessions). Each session included three sets of 10 repetitions of knee extension with a constant speed of 60 °/ s, with 1 minute interval between the sets. The range of motion was 100º to 10º, (full extension at 0 °), comprising both the concentric and eccentric phases.

Quadriceps volume was estimated with an ultrasound device (EUB-405, Hitachi, Japan with a linear probe of 7.5 MHz). The right QF muscle thickness (MT) was scanned and the volume was estimated using a regression equation [22] validated with magnetic resonance:

$$MV = (MT \times 311.732) + (TL \times 53.346) - 2058.529 \quad (2)$$

where MV is the muscle volume [cm^3], MT the muscle thickness [cm] and TL the thigh length [cm]. We have demonstrated that this approach improves model prediction in the case of the *triceps surae* muscle [21].

The total volume was divided into individual QF component volumes, with the proportions of 34% for *vastus lateralis* (VL), 28% for *vastus intermedius* (VI), 24% for *vastus medialis* (VM) and 14% for *rectus femoris* (RF) [23]. The specific PCSA from each muscle was calculated by multiplying the respective volume by the pennation angle and divided by the fiber length, reported in the literature for a group with the same functional and anthropometrical characteristics [23]. Two different values of specific tension

were used to calculate the Maximum Force parameter: of 30.3 N/cm^2 as reported by [17] and an individual value calculated by the volume/MVC torque ratio:

$$\sigma_m = \frac{\tau}{PCSA_{QF} r_{QF}} \tag{3}$$

where τ is the MVC knee torque measured by the dynamometer and r_{QF} patellar tendon moment arm (4.8 cm, according to [10]).

Each subject was evaluated for the maximal voluntary contraction (MVC), and thereafter performed a protocol of 10 seconds submaximal (20% and 60%MVC) knee extension isometric contractions, separated by relaxing intervals, at the same dynamometer, trying to follow a visual torque signal feedback in the computer screen (see protocol in Fig. 4). Torque signal and surface EMG from the VM, VL and RF muscles were synchronously collected (Fig. 1). Raw EMG signal was initially band-pass filtered (15-350 Hz), rectified and, finally, low-pass filtered (6th order digital Butterworth, 2 Hz). Input excitation signal u(t) for the muscle model was found by normalizing the processed test protocol EMG by MCV EMG. VI is a deep muscle which EMG cannot be collected using surface electrodes. Its excitation u(t) was estimated as the average between *vastus lateralis* and *medialis* signals [4].

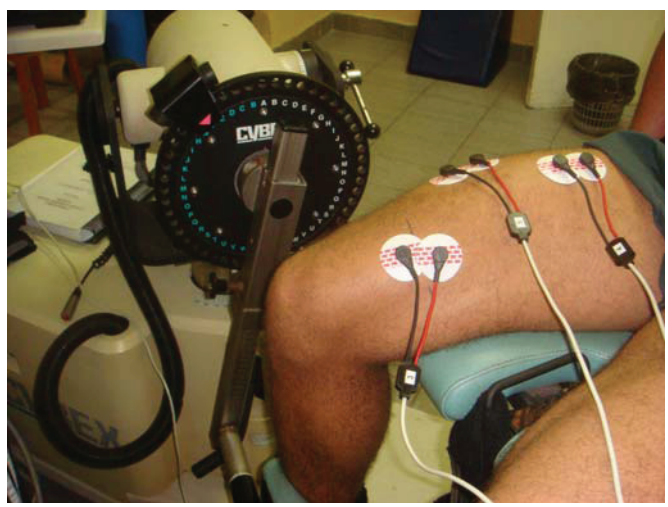


Fig. 1. Experimental setup with the subject seated on the dynamometer, showing the electrodes placement.

Muscle contraction dynamics was described in [3] and extensively used by our group. It is a Hill-type muscle model that uses Zajac's musculotendon actuator concept and notation [24], but includes parallel elastic and viscous elements in contraction dynamics formulation. Model parameters, other than those measured by the ultrasound, were taken from Opensim [20] 'Both Legs' model.

Activation dynamics from [25] was used (eq. 4), associated with the non-linear correction proposed by Manal and Buchanan [26]. The so-called "A-Model" consists of a non-linear algebraic expression between the "muscular activation" a(t) and "neural activation" a(t) (Fig. 2) The curve is composed by a logarithmic and a linear part, but is continuous and differentiable throughout the dominion (Fig. 3). For small values of the non-linear shape parameter ('A' parameter), the 'nonlinearization' is almost negligible. If a greater value for A is chosen, the curve gets more bulged in the small activation part. Several values for the non-linear shape parameter ('A parameter') were tested for each subject, selecting

the value that produced the least mean torque error for an entire protocol run, comprising both steps and relaxing intervals.

$$\dot{a} = (u - a)(k_1 + k_2) \tag{4}$$

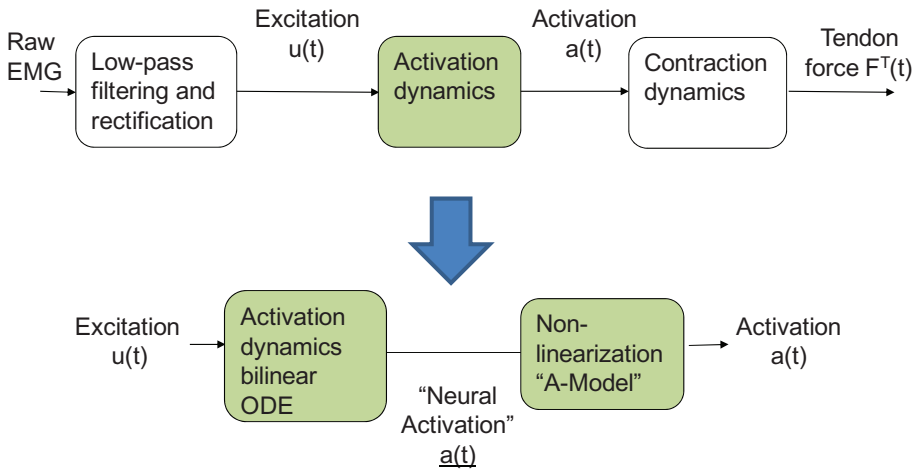


Fig. 2. The A-Model introduces an additional step in the classical EMG-to-force processing when using the EMG-driven modeling approach. The result of the classical activation dynamics is called 'neural activation', which is algebraically non-linearized to produce the muscular activation, which will be input in the contraction dynamics equations.

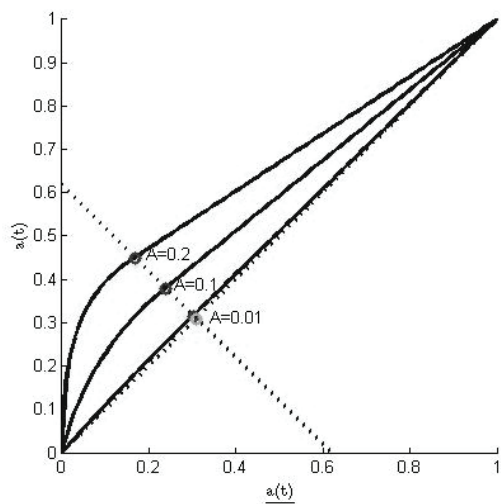


Fig. 3. A-Model relationship between neural activation a(t) and muscular activation a(t). Three different A parameter values are shown, resulting in different degrees of non-linearity. If A=0, the relationship is one-to-one and the A-correction makes no longer effect in the activation dynamics modeling.

Differences between simulated and Cybex™ measured torques were calculated as the normalized Root Mean Square Error (%RMSE) between the two curves. Differences between pre and post-training data were tested by a non-parametric Wilcoxon test and considered significant at a p-value of 0.05 (Statistica, 7.0, StatSoft Inc.)

3. Results and discussion

Figure 4 shows the dynamometer measured and the EMG-model estimated torque curves for the four studied subjects. It can be observed in Table 1 that QF volume, strength (MVC torque), PCSA and estimated σ_m increased in the post-training condition, as expected. The results shown in Table 2, however, suggest that mean increase in MVC torque (14.26%) is relatively greater than QF volume gain (5.97%). This difference is reflected in the grow of the maximum tension parameter value (9.96%). Although torque and maximum tension increases resulted only marginally significant (p=0.067), the small sample of subjects requires cautious interpretation. A larger number of subjects will probably result in a definition of the p value towards significance. In any case, the increase in the estimated σ_m can be attributed to the whole ensemble of physiological effects in the muscle contraction properties induced by training, as described in the Introduction.

Table 1. Architectural and functional *quadriceps femoris* parameters, before and after 13-week strength-gain training (n=4)

Parameter	Pre-training				Post-training			
	Mean	Min	Max	SD	Mean	Min	Max	SD
QF volume (cm^3)	2067.4	1799.2	2238.9	188.8	2196.4	2004.1	2366.8	152.1
MVC torque (Nm)	319.7	282.5	342.1	26.0	374.2	328.8	413.3	41.4
QF PCSA (cm^2)	226.9	197.5	245.7	20.7	241.7	219.9	259.8	16.6
σ_m (N/cm^2)	32.3	30.8	34.0	1.4	35.5	32.4	38.9	2.8
Optimal A-parameter σ_m from literature	0.085	0.075	0.090	0.007	0.076	0.065	0.080	0.007
Optimal A-parameter σ_m from MVC test	0.078	0.055	0.090	0.016	0.050	0.030	0.080	0.021

Table 2. Q*uadriceps femoris* volume and MVC knee torque increase.

% of increase	Mean	Min	Max	SD
QF volume	5.98	3.40	10.22	2.96
σ_m	9.96	2.94	18.08	7.39
MVC torque	14.27	7.60	19.89	5.09

High %RMSE between the estimated and measured torques can be observed in Fig. 5(a), when no non-linear activation dynamics (A-model) was applied. Smaller errors for 60% MVC step has been observed, when compared to 20% MVC. Training might be associated to %RMSE variability decrease in the medium/high activation step. When the A-model is applied, %RMSE strongly decreases in all cases,

as can be observed in Fig. 5(b). In this case, the error levels become more uniform, regarding the activation level. In Fig. 5(c) the %RMSE errors are shown when the model was integrated using the value for maximum specific tension from literature σ_m=30.3 N/cm^2 [17]. Compared to case in which σ_m was estimated experimentally form the MVC tests, a small variation in the %RMSE and in the error dispersion was observed: from 10.56±3.61% to 11.86±4.42%. Nonetheless, such error levels are comparable to the hamstrings coactivation expected contributions (~13 to 16%) [10], [17].

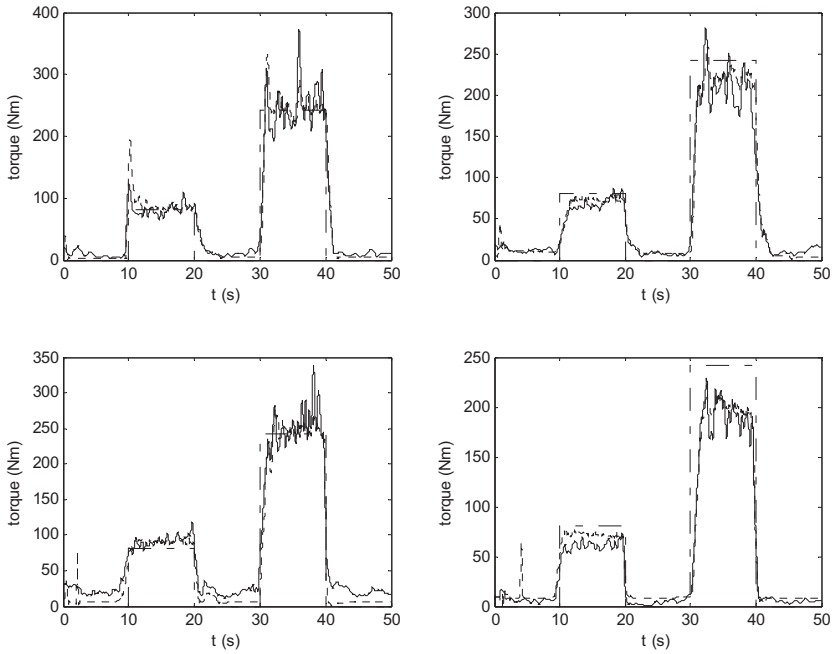
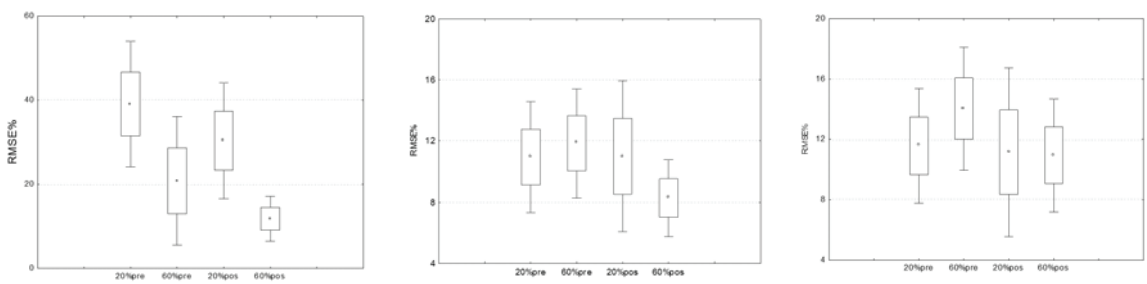


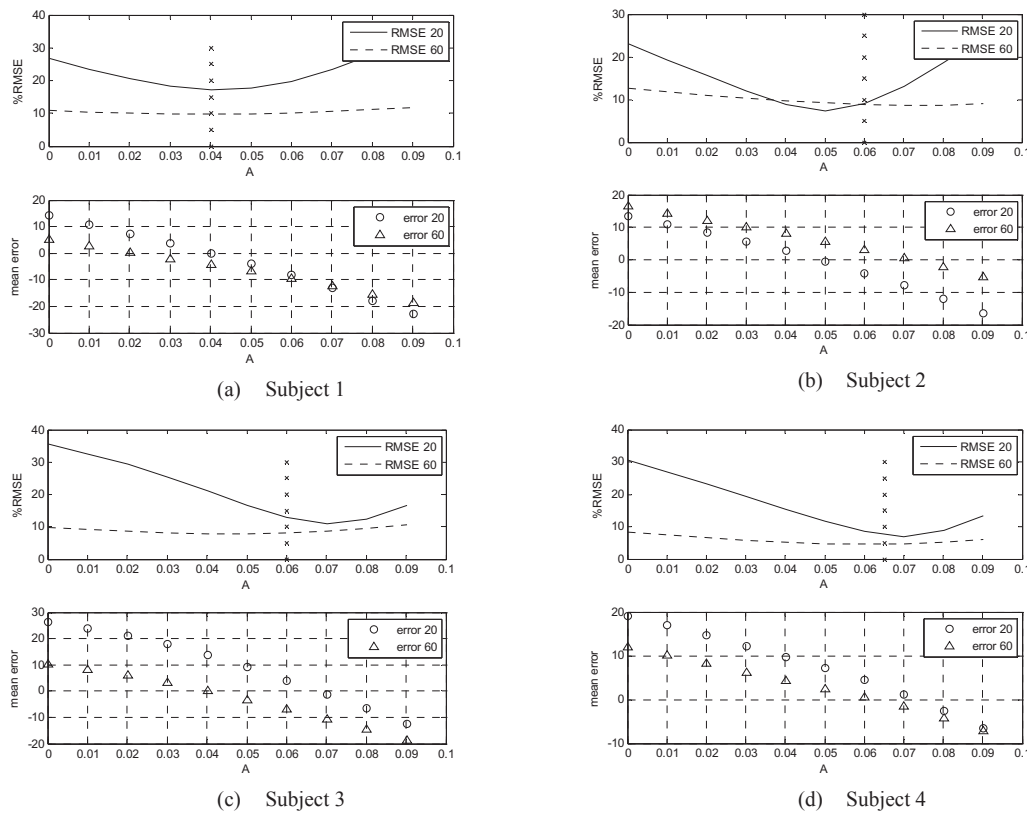
Fig. 4. Torque curves for the four studied subjects, post-training: EMG-driven model estimated (continuous line), dynamometer-measured (dotted line), target (20% MVC and 60% MVC steps and intermediate relaxing intervals, dash-dotted line).



(a) (b) (c)

Fig. 5. Root Mean Square Error between estimated and measured knee torques (%RMSE), before and after strength training. (a) A=0, σ_m estimated from MVC tests; (b) A- optimal, σ_m estimated from MVC tests; (c) A-optimal, σ_m= 30.3 N/cm^2 [17].

In Figure 6, the variation of the %RMSE (upper) and arithmetical error (lower) with relation to A value are shown, for the four studied subjects. Only the post trained results are shown, but for pre training the patterns look alike. The A parameter values that gives the least %RMSE for 20% and 60% MVC steps are different. Thus, the average between the two values of the A parameter is chosen as the optimum for the whole muscle operation. The arithmetic mean difference between the measured and estimated torques are also calculated, and plotted against A. In general, when A=0 (no non-linear correction in the activation dynamics) the errors are positive, what means subestimation of the torque by the EMG-driven model. For A=0, RMSE error for 20% MVC step is always larger than 60% MVC. By increasing A value, the arithmetic error drifts towards overestimation. The %RMSE on both activation levels presents a convex shape, with a clear different minimum value for each level. It can be observed a tendency for the optimum A to assume a value that corresponds to a compromise between an overestimation for one step and a underestimation for the other. Table 1 shows also the optimal A values for pre and post training, considering both sources of maximum muscle tension. The value of optimal A decreases with training, and slightly further when the σ_m, obtained from the MVC test is used to calculate maximum force.

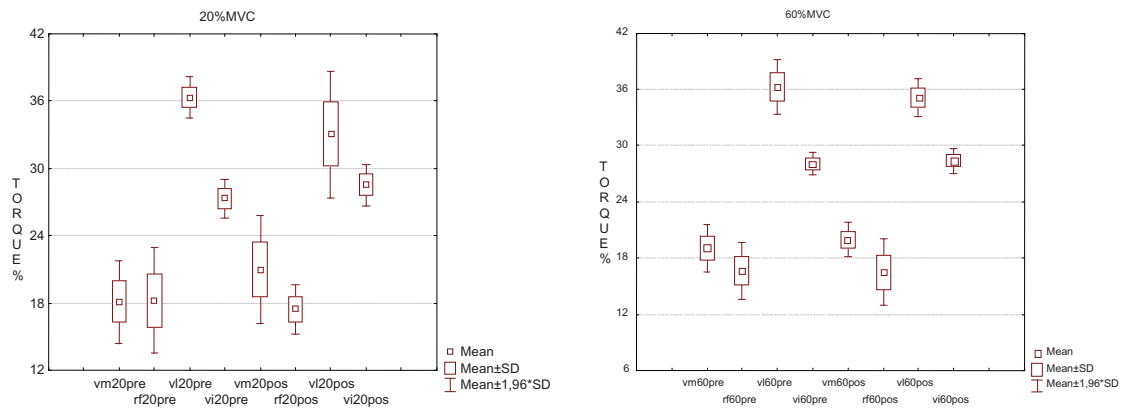


(a) Subject 1 (b) Subject 2

(c) Subject 3 (d) Subject 4

Fig. 6. Relationship between A parameter value, %RMSE (Root Mean Square Error) between the measured and estimated torques and mean arithmetic error, to verify if the model overestimate or underestimate the torques. Figures (a) – (d) refers to the four studied subjects. The results are shown for the pos-trained case. The vertical 'x x x x x' line corresponds to the optimal A parameter, chosen from the average of the minimum-error A for 20% and 60% of MVC steps.

In Figure 7, the patterns of force sharing among the QF components are shown. Fig. 7(a) presents the results for 20%MCV step and (b) 60%. In all cases, there is a pattern of individual torque contributions that is essentially repeated, in a decreasing order: VL, VI, VM and RF. The pattern does not seems to change with training nor with force increase. Zhang et al. [27], in a series of experiments using selective electrical stimulation, found a different sequence for this pattern: VI, VL and VM. Furthermore, VI relative contribution decreased with activation level increase. According to [17], the sequence for muscle volumes, from grater to smaller is VL, VI, VM and RF, while for the PCSA it follows VL, VI, RF and VM. This problem requires furthers studies. In the case of bi-articular muscle RF, the hip angle influences muscle length and thus the muscle force generating capacity.



(a) (b)

Fig. 7. Torque sharing distribution among *quadriceps fermoris* components. pre: before training; pos: after training; vm: *vastus medialis*; rf: *rectus fermoris*; vl: *vastus lateralis*; vi: *vastus intermedius*: (a) 20% MVC; (b) 60% MVC.

4. Conclusions

An EMG-driven model of the quadriceps fermoris muscle has been implemented and tested using a protocol consisting of two 10s sustained contractions, corresponding to 20% and 60% of MVC torque. The test protocol and the analysis were performed before and after an isokinetic strength-training period of 13 weeks. It has been observed an increase in MVC torque relatively greater than muscle volume and PCSA gain, what can be related to a change in the muscle maximum specific tension parameter. Using either the individually estimated (eq. 2) or literature σm produced essentially the same %RMSE, with a possible small advantage to the estimated tension. No differences among the force sharing patterns were observed, for the proposed task, with training, nor with activation level. The estimation error levels, when the A-model activation dynamics is used, is comparable to the coactivation levels reported in the literature. The present work is a run test study, and more conclusive results will require analyzing a greater number of subjects.

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