EMG-DRIVEN MODEL TO ESTIMATE QUADRICEPS FORCES IN ISOMETRIC KNEE EXTENSION

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Abstract *This paper uses a EMG-Driven Hill-type model to estimate muscle forces of the quadriceps in isometric contractions. A group of 10 well trainned male subjects followed a protocol of 10 seconds submaximal (20% and 60%MVC) knee extension isometric contractions, separated by relaxing intervals. Torque signal and surface EMG from the vastus medialis (VM), vastus lateralis (VL) and retus femoris (RF) muscles were synchronously collected. Raw EMG signal was initially band-pass filtered (15-350 Hz), rectified and low-pass filtered. Input excitation signal u(t) for the muscle model was found by normalizing the processed test protocol EMG by MVC EMG. Vastus intermedius (VI) u(t) was described as the mean of vastus lateralis and medialis signals. The muscle model was described in Menegaldo and Oliveira (2009). Differences between simulated and Cybex*^{TM} *measured torques were calculated as the normalized Root Mean Square Error (RMSE) between the two curves. Two-way ANOVA test was applied to assess significant changes of the muscle individual estimated torque and excitation function. Significant differences among means were set as a p value of 0.05. The subjects presented a maximum torque of 337,81 ± 59,17 Nm, compatible with their training status. Mean relative muscle excitations (expressed as a percentile of MVC EMG amplitude) were similar among the muscles for both contractions intensities (13.8±4.5%, CV 32.6 % for 20%MVC and 54.3 ±11.4 %, CV 21.0 %) suggesting the sinergic whole of the quadríceps muscles, driven by a common neural drive from the CNS. On the other hand, the high variability of the mean u(t) values between subjects, specialy for low contraction level, can be atributed to changes in motor unit recruitment pattern, in addition to different load sharing strategies. The estimated relative extension torque was different among muscles, as their functional whole depends on the individual geometry and size. VL (33.8±4.6 %) contributed signifantly more to the total torque than RF (23.8 ±5.2%) and VI (24.6 ±2.4 %). VM (17.3 ±4.9 % contributed the least.). . The model resulted in better force estimation for the higher contraction level (RMSE= 15.8 ± 5.3 %) than for the lower level (RMSE =29.5± 10.1%). Possibles sources of errors may be pointed out, such as estimation of VI input signal. This muscle is reported to contribute with about 50% for low torque levels. Seeking more reliable estimations of this deep muscle EMG signal is a possible way to improve the model predictability*.

Keywords: *EMG-Driven muscle model, quadriceps, muscle biomechanics, torque sharing*

1. INTRODUCTION

This paper uses a EMG-Driven Hill-type model to estimate muscle forces of the quadriceps in isometric contractions. Muscle force increase is usually the main objective of muscular exercise. Such strength gain can be assessed by the increase in the lift-weight capacity of the subjects or, more quantitatively, by performing tests for joint torque and power in dynamometers. However, such tests do not allow estimating the individual muscle contribution to the total joint torque. This information may be very helpful, for example, for assessing muscle unbalances, which are associated to joint pathologies, such as Patello-femoral Pain Syndrome (Dionisio et al., 2008). EMG is an useful tool for quantifying muscle forces and studying motor control strategies. However, the relationship between EMG and muscle force or joint torques is not trivial, and depends, in part, on muscle dynamics. This work aims to find muscle excitations and partial joint torque contribution patterns in isometric knee extension of trained men, considering low and medium/high contractions. An EMG-driven muscle model (Menegaldo and Oliveira, 2009) was applied to estimate muscle forces from the processed EMG recordings.

2. METHODS

A group of 10 well trained male subjects participated in the study (18.75±0.71 years, 72.89±6.63 kg body mass and 1.78±0.77 m height), which was approved by the UFRJ Ethics Committee (Proc. No. 031/07 HUCFF). The subjects sat on a Norm/Cybex^{TM} Dynamometer, with the knee flexed at 80^{o}. Isometric knee extension torque associated to maximal voluntary contraction (MVC) was collected twice with two-minute rest between the trials. The highest value was selected as the maximum subject torque. Each volunteer then followed a protocol of 10 seconds submaximal (20% and 60%MVC) knee extension isometric contractions, separated by relaxing intervals. A feedback display of the actual dynamometer torque on-line output was provided to the subject, who attempted to match it to a mask of the step protocol drawn in the computer screen. Torque signal and surface EMG were synchronously collected using a Electromyography EMG 800C by EMGSystem^{TM} (Sao Jose dos Campos, Brazil), with CMRR = 106 dB and analogical band-pass filter 10-500 Hz, 2 kHz sampling rate, 16 bits A/D converter. Ag-AgCl pre-gelled electrodes were positioned on *vatsus medialis, vastus lateralis and rectus femoris* muscles, according to SENIAM recommendations, after skin preparation (Freriks et al., 1999). Reference electrode was positioned on the left lateral malleolus (Figure 1).

Figure 1. EMG-electrodes positioning on *vatsus medialis, vastus lateralis and rectus femoris* muscles.

Raw EMG signal was initially band-pass filtered (15 – 350 Hz) to remove artifacts (Merletti and Parker, 2004) and then rectified and low-pass filtered with a 2th order Butterworth filter (2Hz cut-off frequency). Input excitation signal u(t) for the muscle model was found by normalizing the processed test protocol EMG by MVC EMG. Torque signal and surface EMG from the *vastus medialis* (VM), *vastus lateralis* (VL) and *rectus femoris* (RF) muscles were synchronously collected. *Vastus intermedius* (VI) is a deep muscle, hardly accessible through surface electrodes. Thus, VI u(t) was described as the mean of *vastus lateralis* and *medialis* signals (Loyd and Besier, 2003). Muscle model parameters such as optimum fiber length and tendon slack length were taken from the “Both Legs with Muscles” model of OpenSim (Delp et al., 2007). The optimum pennation angle parameter was selected based on *in vivo* data reported by O’Brien et al. (2010). Muscles Physiological Cross-Sectional Areas (PCSAs) were obtained from the literature (Erskine et al., 2009). A specific tension of 30.3 N/cm^2 (Erskine et al., 2009) was used to calculate the maximum force parameter. The quadriceps moment arm of 0.048m was considered for torque estimation, measured *in vivo* using MRI (Erskine et al., 2009). The muscle model was described in Menegaldo and Oliveira (2009) and extensively used by our group in several studies. It is a Hill-type muscle model that uses Zajac’s musculotendon actuator concept and notation, but includes parallel elastic and viscous elements in the contraction dynamics formulation. Muscle activation dynamics from Piazza and Delp (1996) was used. The 2nd order non-linear dynamic model was integrated numerically using u(t) as the input signal. The estimated torque output was calculated as the sum of each simulated muscle force multiplied by the knee moment arm. The differences between simulated and Cybex^{TM} measured torques were calculated as the normalized Root Mean Square Error (%RMSE) between the two curves (Eq. 1).

$$\mathrm{RMSE}\ (\%) = \frac{1}{\mathrm{TM}_{\mathrm{MAX}}}\sqrt{\frac{\sum_{i=1}^{N}\left(\mathrm{TM}(i)-\mathrm{TS}(i)\right)^2}{N}} \times 100\% \qquad (1)$$

where TM is the Cybex^{TM} measured torque, TS the simulated torque, N the number of samples in the time series and TM_{MAX} the maximum dynamometer measured torque at MVC for each subject. Differences between simulated and Cybex^{TM} measured torques were calculated as the normalized Root Mean Square Error (RMSE) between the two curves, for each contraction intensity. Two-way ANOVA test (3 muscles x 2 contraction intensities) was applied to assess significant changes of the muscle individual estimated torque and excitation function. Significant differences among means were set as a p value of 0.05.

3. RESULTS AND DISCUSSION

The subjects have produced a maximum torque of 337.81 ± 59.17 Nm, compatible with their training status (Remaud et al., 2010; Aagaard et al., 2001). The subjects were able to produce the required percentiles of the maximal torque: 19.0 ± 0.5% for the 20%MVC and 57.6 ± 2.2% for the 60%MVC contraction. Figure 2 shows, for the measured superficial muscles (VM, VL and RF), the mean relative excitations (expressed as a percentile of MVC EMG amplitude) and estimated knee extension torques (expressed as percentile of the total estimated torque), for both contractions intensities. It can be observed that, for producing 19% and 57.6% of MVC torque, the mean excitation levels are predominantly lower than such levels. This fact can be also observed in results published by other groups (Watanabe and Akima, 2009).

No significant differences can be observed among the muscles excitations considering separately each step (20 and 60%, Figure 2). This result reveals a synergic excitation pattern of the quadriceps muscles components, driven by a common neural drive from the CNS, as already observed by Alkner et al. (1999). On the other hand, the high variability of the mean u(t) values among the subjects, especially for low contraction level, can be assigned to changes in motor unit recruitment patterns, besides different load sharing strategies. The estimated relative knee extension torque produced by each quadriceps component showed significant differences. This is expected since their functional role depends on the individual geometry and size. VL (33.83 ± 4.60%) contributed significantly more to the total torque than RF (23.90 ± 5.22%), which, in turn, contributed significantly more than VM (17.64 ± 4.88). This is in accordance with Zhang et al. (2003) who used inwire electrical stimulations of each of the quadriceps muscle. They showed that in submaximal contraction, the VM contributed the least (9.5 to 12%), while VL and RF contributed approximately 30%. to the total generated torque.

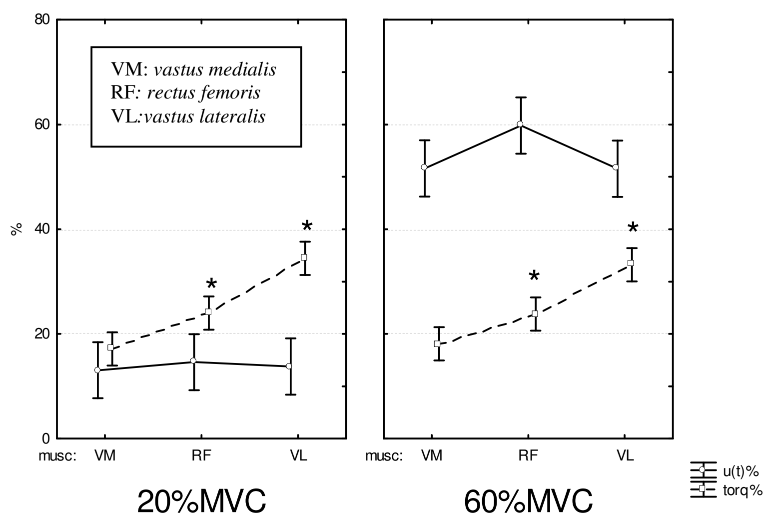

Figure 2. Mean relative muscle excitations (u(t)%) and knee extension torques (torq%) among superficial muscles for both contractions intensities (60% MVC and 20%MVC). The y-axis (%) mean both the percentile of excitation u(t) with relation to MVC u(t) and the percentile of each muscle contribution to the total torque.

For comparison purposes, we have used OpenSim to perform a static analysis with the same conditions (joint angles and activation levels). VL, VI, RF and VM contributed to the total torque, respectively, with: 36.7%, 30.7%, 1.3% and 31.3%. Opensim generated muscle fiber length for 80^{o} knee flexion and hip at 90^{o} was considered. In this condition, the normalized fiber lengths ($\widetilde{L^M}$) with relation to optimal fiber length, has been found as RF=0.544, VL=1.126, VM=1.108 and VI=1.141. Figure 3 shows that this approach resulted in significant differences in the relative torque contribution for RF and VL. The normalized fiber length is a critical model parameter, since it determines the maximum possible force exerted by the muscle, due the shape of the classical force x length relationship. With the hip flexed at 90^{o}, Opensim finds a very strong shortening of RF length, what is possibly excessive (Alkner et al., 1999), leading to very marginal torque contribution from this muscle. This question deserves more in depth studies, and probably the test protocol will require measuring such normalized length by ultrasound. Using Opensim estimated $\widetilde{L^M}$ for the initial integration conditions in the EMG-driven model, instead of considering that the muscles were initially at optimal length, resulted in a torque prediction error increased for both steps, especially for the high contraction: 24.0± 10.2% for 20%MVC and 41.3 ± 29.6% for the 60%MVC. Such increase was already expected, since the error is predominantly towards torque underestimation. In any case, the reliability of the used parameters in the EMG-driven model simulation strongly influences the force estimation results.

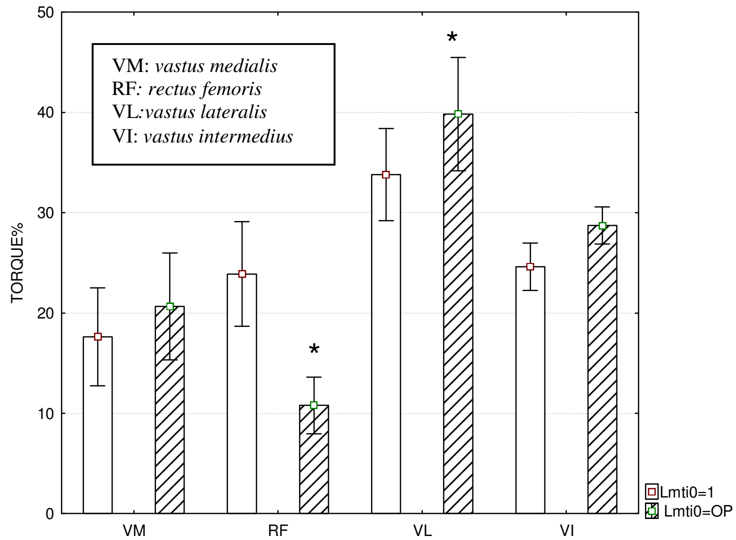

Figure 3. Mean relative estimated extension torque among muscles for both contractions intensities, using optimal length ($\widetilde{L^M}$)=1 and estimated from Opensim (OP).

The model resulted in better torque estimation for the higher contraction level (RMSE = 15.8 ± 5.3%) than for the lower level (RMSE = 29.5± 10.1%), following again a pattern observed for *triceps surae* (Oliveira and Menegaldo, 2010a). Such error is likely to be reduced by improving model formulation, such as using a non-linear version of activation dynamics (Manal and Buchanan, 2003), variable pennation angle, using ultrasound or scaling techniques to choose some muscle model parameters (Menegaldo and Oliveira, 2009; Oliveira and Menegaldo, 2010b), etc. The antagonist action from *biceps femoris* should be also taken into account. According to Alkner et al. (1999) this muscle can performs as much as 7-8% of the total agonistic torque, and to Erskine et al. (2010), 13%, both in similar conditions of the present study. Elias et al. (2006) points out the sensibility of force prediction with relation to the PCSA estimation. Finally, an important source of errors is the estimation of VI input signal. This muscle is reported to contribute with about 50% for low torque levels (Watanabe et al., 2009). Seeking more reliable estimations of this deep muscle EMG signal is also a possible way to improve the model predictability.

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4. REFERENCES

Aagaard, P., Andersen J. L., Dyhre-Poulsen, P., Leffers, A-M., Wagner, A., Magnusson, S. P., Halkjær-Kristensen, J., Simonsen, E. B., 2001, “A mechanism for increased contractile strength of human pennate muscle in response to strength training: changes in muscle architecture”, The Journal of Physiology, Vol. 534, pp. 613-623, 2001.

Alkner B.A., Tesch P A, Berg H.E., 1999, "Quadriceps EMG/force relationship in knee extension and leg press", Medicine and Science in Sports & Exercise, Vol.32, pp 459-463.

Delp, S.L., Anderson, F.C., Arnold, A. S., Loan, P., Habib, A., John, C., Guendelman, E., Thelen, D.G., 2007, “OpenSim: Open-source software to create and analyze dynamic simulations of movement”, IEEE Transactions on Biomedical Engineering, Vol. 54, pp. 1940-1950.

Dionisio, V.C., Almeida, G. L., Duarte, M., Hirata, R. P., 2008, “Kinematic, kinetic and EMG patterns during downward squatting”, Journal of Electromyography and Kinesiology, Vol. 18, pp. 134-143.

Elias J. J., Bratton D.R., Weinstein D. M., Cosgarea A.J., 2006, "Comparing two estimations of the quadriceps force distribution fpr use during patellofemoral simulation", Journal of Biomechanics, Vol. 39, pp. 865-872.

Erskine R., Jones D.A., Maganaris C.N. Degens H., 2009, “In vivo specific quadriceps femoris muscle”, European Journal of Applied Physiology, Vol. 106, pp. 827-838.

Erskine, R. M., Jones, D. A., Williams, A. G., Stewart, C. E., Degens, H., 2010, "Resistance training increases in vivo quadriceps femoris muscle specific tension in young men", Acta Physiol, Vol. 199, pp. 83–89.

Freriks, B., Hermens, H. J., Disselhorst-Klug, C., Rau G., 1999. The recommendations for signal processing methods for surface electromyography, In: European recommendations for surface electromyography – SENIAM Project Ed: Hermens H. J., Freriks B., Merletti R., Stegeman D., Blok J., Rau G., Disselhorst-Klug C. and Hâag G. Enschede: Roessingh Research and Development.

Lloyd D.G., Besier, T.F., 2003, "An EMG-driven musculoskeletal model to estimate muscle forces and knee joint moments in vivo", Journal of Biomechanics, Vol. 36, pp. 756-776.

Manal K., Buchanan, T. S., 2003 "One-parameter neural activation to muscle activation model: estimating isometric joint moments from electromyograms", Journal of Biomechanics, Vol. 36, pp. 1197-1202.

Menegaldo, L. L., Oliveira, L. F., 2009, "Effect of muscle model parameter scaling for isometric plantar flexion torque prediction. Journal of Biomechanics, Vol. 42, pp. 2597-260.

Merletti, R., Parker, P., 2004, "Electromyography: Physiology, Engineering and Noninvasive Applications". Wiley-IEEE Engineering in Medical and Biology Society, Hoboken, NJ.

O'Brien D., Reeves N.D., Baltzopoulos V., Jones A., Maganaris C.N., 2010, " Muscle-tendon structure and dimensions in adults and children", Journal of Anatomy, Vol. 216,pp. 631-642.

Oliveira L.F, Menegaldo, L.L., 2010a, "Torque sharing in submaximal isometric plantar flex-ion: an analysis based on an EMG-driven muscle model", Proceedings of XXII Brazilian Congress on Biomedical Engineering, Tiradentes-MG, Brazil.

Oliveira L.F, Menegaldo, L.L., 2010b, "Individual-specific muscle maximum force estimation using ultrasound for ankle joint torque prediction using an EMG-driven Hill-type model", Journal of Biomechanics, Vol. 43, pp. 2816-2821.

Piazza S. J., Delp, S.L., 1996, "The influence of muscles on knee flexion during the swing phase of gait", Journal of Biomechanics, Vol. 29, pp. 723-733.

Remaud, A., Cornu, C., Guével, A, 2010, Neuromuscular adaptations to 8-week strength training: isotonic versus isokinetic mode, European Journal of Applied Physiology, Vol 108, pp. 59-69.

Watanabe K, Akima, H., 2009, "Normalized EMG to normalized torque relationship of vastus intermedius muscle during isometric knee extension", European Journal of Applied Physiology, Vol. 106, pp. 665-673.

Zhang L., Wang G., Nuber G., Press J., Koh J., 2003, "In vivo load sharing among the quadriceps components", Journal of Orthopaedic Research, Vol. 21, pp. 565-571.

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