EMG-Driven muscle model parameter scaling improve isometric plantar flexion torque prediction

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Abstract. *This paper uses a EMG-Driven Hill-type muscle model to estimate individual muscle forces of the triceps surae in isometric plantar flexion contractions. A group of 20 young physical-active adult males was instructed to follow a specific contraction protocol with low (20% MVC) and medium-high (60% MVC) contractions, separated by relaxing intervals. The torque calculated by summing the individual muscle forces multiplied by the respective moment arms was compared to the torque measured by a dynamometer. Three different "correction factors" were applied on some of the parameters, based on anthropometric and dynamometric measurements. Model torque agreement with dynamometer was recalculated with the parameter scales. It was observed that the relative torque estimation error decreased when all factors were applied simultaneously (12.92 $\pm$ 4.94 % without scaling to 10.12 $\pm$ 1.73 %), resulted mainly from the correction of the maximal muscle force parameter.*

Keywords: *muscle model, Hill-type muscle, ankle joint, triceps surae, plantar flexion, dynamometry*

1. INTRODUCTION

One of the most important problems in biomechanics is the measurement or estimation of muscle forces (Eredemir et al., 2007). Usually, these forces are estimated from theoretical or semi-empirical models. Measured external forces, kinematics and electromyography (EMG) signals may be used as external inputs to an underdetermined problem, often solved by optimization. In an EMG-Driven model, muscle dynamics, normally Hill-type, is fed by an excitation signal considered equivalent to the rectified and low-pass filtered EMG and tendon force is the output. However, in practice, no human joint is spanned by a single muscle. Therefore, direct validation of such system is not usually possible in humans, as muscle force measurement in situ is still a highly invasive procedure. Such validation can be inferred by measuring associated physical phenomena such as joint torques. For a mono-articular 1 DOF problem, muscle force can be assessed as the output of a system where the inputs are the EMGs and the output the joint torque.

In this work, a Hill-type EMG-Driven muscle model will be used, modified by Menegaldo et al. (2003). Model parameters available in literature comes from cadavers studies (Keeple et al., 1998) or from medical imaging (Blemker et al. 2007). Some authors proposed practical approaches attempting to scale muscle parameters from more or less simple and non-invasive subject measurements (Manal and Buchanan, 2004; Winby et al., 2008).

This paper evaluates experimentally the error level that can be expected if a set of parameters from literature is applied to solve a forward dynamics EMG-Driven model of isometric plantar flexion contractions, with the ankle in neutral position. Additionally, we investigate the role of applying in such simulations some simple muscle model parameters scale factors. The associated errors were evaluated with and without applying the proposed scale factors to some of the muscle model parameters.

2. Methods

A homogeneous group of 20 male subjects (age:18.43 0.51 years, mass: 68.64 8.29 kg and height: 175.26 8.38 cm) participated in the study. The volunteers were selected among the military personnel of the Physical Education School of the Brazilian Army, Rio de Janeiro, engaged in the regular regimen of physical activity. All participants provided written consent and did not relate any history of osteomyoarticular injuries at the right knee or ankle. This experiment was approved by the Federal University of Rio de Janeiro Ethical Committee under the approved project HUCFF 031/07. Anthropometric measures consisted of the leg length (from lateral knee interline to lateral malleolus) and the bimalleolar diameter.

Subjects laid prone on a Norm/CybexTM Dynamometer, with the knee extended and the ankle at neutral (90o) position. The right foot was firmly fixed to the foot adaptor. Plantar flexion torque associated to maximal voluntary contraction (MVC) was collected twice with sufficient rest among the trials, and the highest value was selected as the maximum subject torque. Each volunteer was instructed to follow a protocol consisting of two 10 seconds steps of submaximal loads of 20 and 60% MVC each, separated by 10 seconds relaxing intervals. A feedback display of the actual force output was provided to the subject who attempted to match it to a mask of the protocol.

Torque signal and surface EMG were synchronously collected using a Electromyography EMG 800C by EMGSystemTM (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 gastrocnemius medialis, gastrocnemius lateralis and soleus muscles according to SENIAM recommendations, after skin preparation (Freriks et al., 1999). Reference electrode was positioned on the left lateral malleolus. The EMG signal from each muscle was initially band-pass filtered (10 - 350 Hz) to remove artfacts (Merletti and Parker, 2004) and then low-pass filtered with a 2th order Butterworth filter (2Hz cut-off frequency). The resulted enveloped signal was normalized by the respective mean EMG-MVC signal, processed by the same way, and considered the muscle model excitation signal u(t).

The torque output was found by the sum of each simulated muscle force multiplied by its respective ankle angle moment arm, using the polynomial equations from Menegaldo et al. (2004).

The difference between the CybexTM measured torque (TM) and simulated torque (TS) was calculated as the mean square error (RMSE) between the two curves and the result was expressed as relative to the maximal measured torque value (RMSE(%)).

The simulations without applying any correction factors to the muscle parameters ('Standard' Case) were performed using the parameters values available in the "Both Legs with Muscles" model of OpenSimm (Delp et al. 2007).

Four different and simple "correction factors" were tested separately, as well as all together, and the torque error curves recalculated. The proposed scale factors were based on two anthropometric measures (leg length (LL) and bimalleolar diameter (BD)) and on the maximum measured torque (TM_{MAX}). Such factors were determined by dividing the individual measurement by the respective mean value of the entire group of volunteers. The maximal muscle force (F_0^M) was scaled by (TM_{MAX}), the tendon slack length (L^{ST}) by LL and the moment arm by BD. A one-way ANOVA statistical test was applied to assess significant changes of the RMSE(%) according to different scaling procedures, after testing for normal distribution (Kolmogorov Smirnoff test). The Fisher LSD Post Hoc test was applied to identify significant difference between means with p = 0.05 (Statistica 6.0 - StatSoft, Inc.).

3. Results and Discussion

Figure 1 shows a typical example of the gastrocnemius medialis (GM), gastrocnemius lateralis (GL) and soleus (SOL) muscles excitation functions (u(t)) for one subject. In this case, the normalized EMG signals showed less GL participation for the 20%MVC step but similar muscle contributions to the 60%MVC target. This is one of the possible different strategies observed among the subjects to achieve the torque demands. In Figure 2, the ankle torque contributions from each individual muscle and the overall sum of such torques and the torque measured by CybexTM, are shown for one subject, as an example. Table 1 shows the mean values of the scale factors and the respective RMSE(%) resulted from its application. There were statistical differences between the Standard Case (No Correction - NC) and the (TM_{MAX}) scaling factor, as well as the corrections applied altogether. Scale factors based on anthropometric measures, when tested alone, did not resulted in torque estimation improvements.

Table 1: Mean (standard deviation) values of the scaling factors and the effect on the RMSE% between estimated and measured joint torques.

Scaling factor	Mean(SD)	**RMSE (%) (SD)**
NC	-	12.92 (4.94)
TM_{MAX} (Nm)	104.15 (18.53)	10.32 (2.06)*
BD (cm)	7.19 (0.47)	12.19 (4.37)
LL(cm)	41.13 (2.78)	12.94 (5.04)
ALL	-	10.12 (1.73)*

NC=No Correction; TM_{MAX}=Maximum Measured Torque; BD=Bimalleolar Diameter; LL=Leg Length; ALL=All corrections applied simultaneously. * $p<0.05$.

4. Conclusions

This paper compares the torque errors between the EMG-Driven model and dynamometric measurements of isometric ankle plantar flexion. A homogeneous group of subjects (age, sex and physical activity) was selected to participate on the study. By applying simple scale factors of muscle model parameters, based on anthropometry and on individual performance, it was possible to reduce significantly, in some cases, the error between the two approaches.

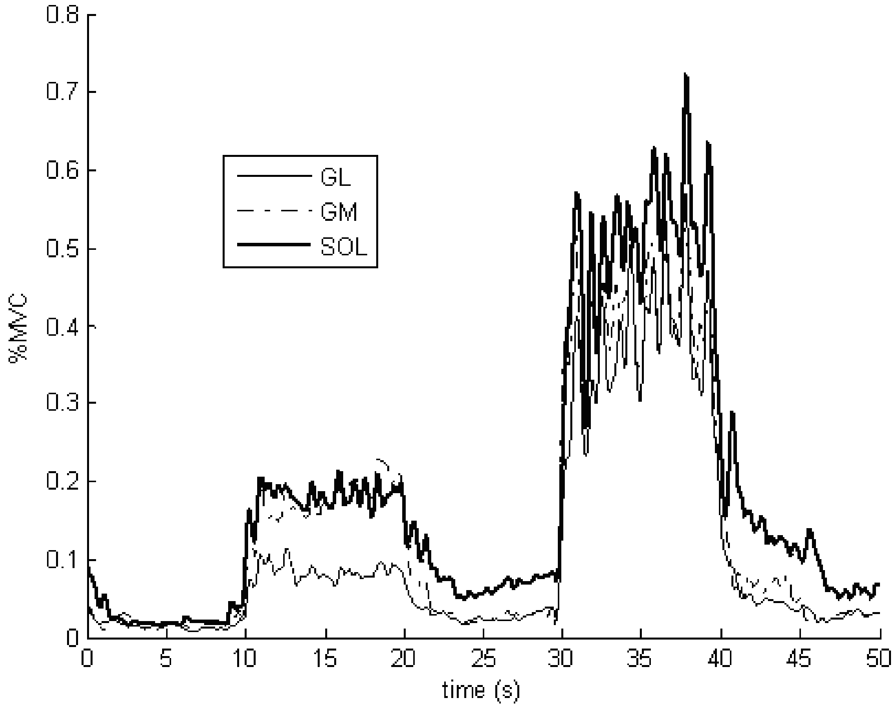

Figure 1. Normalized and filtered EMG signals of gastrocnemius medialis, gastrocnemius lateralis and soleus muscle for one subject.

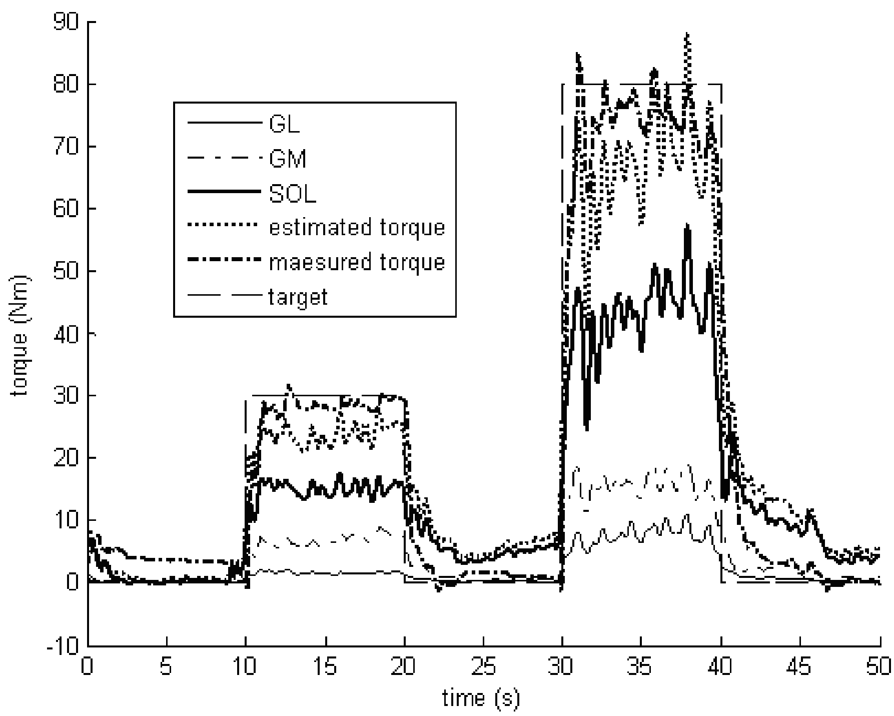

Figure 2. Ankle torque estimated for each individual muscle, overall sum of such torques and the torque measured by CybexTM, for the same subject. The dashed line represents the target shown on-line to the subject with the torque demands, superimposed with the measured torque from the dynamometer.

The proposed protocol adequately demonstrated the capacity of the muscle model to reproduce low and moderate-high muscle activation levels, as well as the rise and fall of the contraction curves. Model behavior is qualitatively coherent, and the errors are moderate, although with high dispersion among the subjects.

The proposed parameter corrections gave no RMSE% reduction when applying each anthropometric factor individually. However, the maximum torque has shown to be a promising scale, reducing significantly the RMSE% and the associated standard deviation. Appling simultaneously the anthropometric and torque factors led to further decrease of RMSE and SD.

Although a very simple situation was accessed here (isometric, mono-articular contraction), the experiment was highly controlled, and the calculated torque errors are limited mainly by the dynamometry, that is a gold standard. This work is contribution for improving reliability and expanding the applicability of such EMG-Driven models to a broader class of problems.

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