Electromyogram (EMG) Amplitude Estimation and Joint Torque Model Performance

O. Bida [a], D. Rancourt [b], E.A. Clancy [c]
[a, c] Worcester Polytechnic Institute, Electrical & Computer Engineering, 100 Institute Rd, Worcester, MA 01609 USA
[b] University of Sherbrooke, Mechanical Engineering, Sherbrooke, PQ J1K2R1, Canada

Abstract— **The relationship between surface EMG and torque about a joint has been the focus of many studies. Some of these studies however, have utilized conventional processors to obtain the EMG amplitudes. Recent advances (multiple channel combination and whitening) have demonstrated significant improvement in EMG amplitude estimation accuracy. The purpose of this study was to investigate the influence of these EMG amplitude advances into EMG-torque estimation. EMG from biceps/triceps muscles and torque about the elbow joint were collected from fifteen subjects producing constant-posture, nonfatiguing, force-varying contractions. EMG amplitudes were obtained using processors with and without the advances, and then they were related to torque using a linear FIR model. Results demonstrated that both whitening and multiple-channel combination reduced EMG-torque errors and their combination provided an additive benefit. Specifically, the EMG-torque prediction errors were reduced to an average of 8% of maximum voluntary contraction of flexion (MVC_F) when incorporating a four-channel, whitened processor.**

I. Introduction

Relating EMG amplitude estimates (i.e. the standard deviation of recorded surface EMG) to the tension of individual muscles via mathematical models is a very useful step in many applications [1]. However, the tension produced by individual muscles can not be measured non-invasively, thus there is no direct mechanical method to validate the model predictions. Many researchers have focused their efforts to relate surface EMG to torque about a joint. The net torque about a joint can be easily verified via mechanical measurements and can alleviate some of the limitations.

Over the last few years, there are clear advances in estimating EMG amplitude, yet EMG-torque modeling has not benefited from this progress. If EMG is a useful indicator of the muscular tension, it is necessary to accurately measure and interpret EMG in the mathematical models. Clancy and Hogan showed that the torque estimation error is reduced when using improved EMG amplitude processors [1]. Their experimental results were obtained using a linear model to relate EMG amplitude from biceps/triceps to the elbow joint torque in the case of constant-posture and constant-force contractions. Encouraging results were also obtained in less constrained conditions (slowly varying force), but several trial combinations that lead to unrealistic model performance were an obstacle that needed further investigation [2]. The result of the previous research inspired the focus of this project: relating the EMG amplitudes from biceps/triceps to the torque about the elbow and to determine if better EMG amplitude processing leads to better torque predictions during dynamic experimental tasks (force-varying and constant-posture contractions).

II. Methods

The experimental apparatus and methods are described in detail elsewhere [2], [3], [4]. Briefly, fifteen healthy subjects (eight male, seven female; aged 23–65 years) had four EMG electrode-amplifiers on the skin above biceps and triceps muscle groups to record EMG signal. Each performed three sets of dynamic (constant-posture, force-varying) target tracking contractions as directed by a random pursuit displayed on a computer screen. A tracking object moved from 50% MVC extension to 50% MVC flexion and had a uniform random distribution with a bandwidth of either 0.25 Hz (slow tracking) or 1 Hz (fast tracking). Subjects completed 15 (3 sets of 5) slow tracking trials and 15 fast tracking trials, each of 30 sec duration and were allowed 2–3 minutes of rest between trials to avoid fatigue. Additionally, subjects performed 5 sec of 50% MVC and rest trials (0% MVC) used to calibrate and to model noise for the advanced amplitude processors.

After processing in hardware (filtered, amplified, and sampled at 4096 Hz), the EMG data were analyzed off-line using MATLAB. Four different EMG amplitude processors were contrasted separately for the biceps and triceps muscle groups. For all processors, the EMG data were high-pass filtered at 15 Hz and rectified [2]. Processor 1 was the single-channel, unwhitened processor using recordings from an electrode located centrally on each of the muscle groups. Processor 2 was a single-channel, whitened processor (same electrode as Processor 1). Prior to rectification, each channel was whitened using the adaptive whitening technique of Clancy and Farry [3]. Processor 3 was a four-channel, unwhitened processor using four EMG signals from a muscle group that were normalized in magnitude and ensemble averaged after rectification. Processor 4 was a four-channel, whitened processor.

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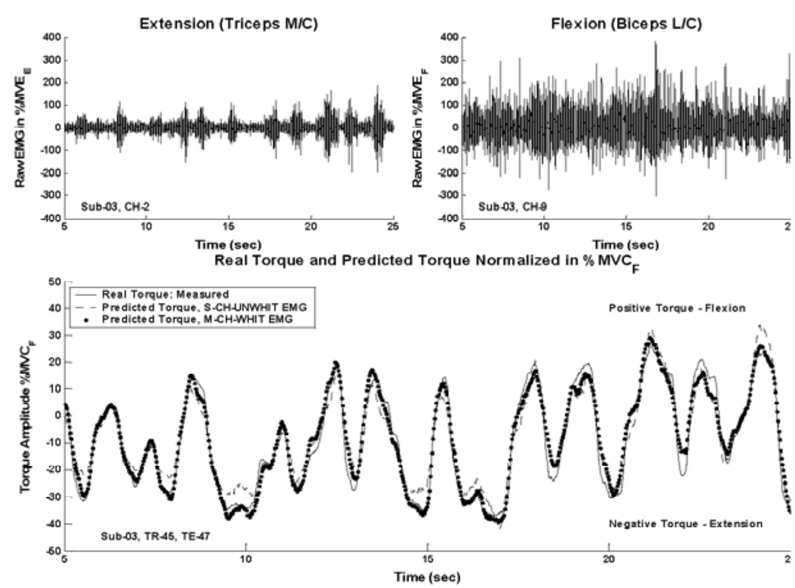

Fig. 1: Typical raw EMG data (top) and predicted torque (bottom)

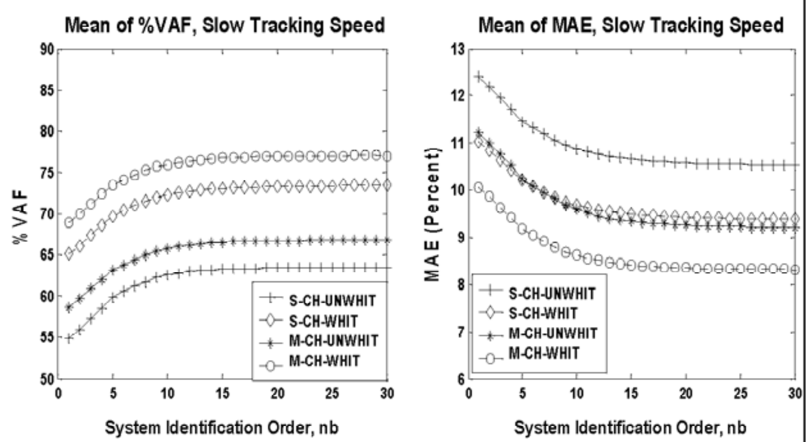

Fig. 2: Mean % VAF (left) and mean MAE (right) vs. system ID order

EMG amplitude estimates from biceps/triceps were the inputs to a system identification (a.k.a. system ID) model in which elbow joint torque was the output [4]. Prior to system ID, the EMG amplitudes were decimated by 100 to an effective rate of 40.96 Hz to avoid high variances and unpredictable system behavior in high frequencies (torque bandwidth was about 4–5 Hz). The decimation stage included an anti-aliasing low-pass filter at half the new sampling rate. The decimated flexion *F(k)* and extension *E(k)* EMG amplitudes were related to torque *T(k)* in a dynamic, linear, FIR model [5]

$$T(k) = \begin{matrix} -e_1E(k-1) - e_2E(k-2) - \cdots - e_{nb}E(k-nb) \\ + f_1F(k-1) + f_2F(k-2) + \cdots + f_{nb}F(k-nb) \end{matrix}, \quad (1)$$

where the e_i are extensor model coefficients, the f_i are flexor model coefficients and *nb* is the model order. A train-test routine was utilized in which the model coefficients were determined using linear least squares from a training trial and then used to calculate the torque from a distinct test trial [5]. An error signal was obtained from the difference between the predicted and actual test trial torque. Within each dataset, fit coefficients were estimated using one trial, and then tested on the four remaining trials. The average performance was determined from a total of 180 error signals (15 subjects x 3 sets per subject x 1 training trial per set x 4 test trials per training trial). All errors were normalized to twice the torque at 50% flexion MVC, denoted %MVC$_F$. To evaluate these errors we used the mean absolute error (MAE) computed for each trial and the percent variance accounted for (%VAF) [4].

III. Results and Discussion

Fig. 1 shows the raw data from a typical trial as well as the predicted torque from two different EMG-torque processors. The predicted torque captures most of the dynamics of the actual torque. Fig. 2 provides summary results of analysis of errors between the predicted and actual torques from all subjects, for the slow tracking speed data. The figure shows the mean MAE and %VAF, as a function of the system ID model order, for each of the four EMG amplitude processors. For all processors, the plots show a progressive increase in performance as model order is increased up to 15th order then no visible improvement occurs further. EMG-torque performance is positively influenced by whitening and multiple-channel combination. The best EMG-torque model produced an average error of 8% MVC with a %VAF of 78%. Similar results were found for the fast tracking speeds [4].

IV. Summary and Conclusions

Advances in EMG amplitude estimation were applied to the EMG-torque problem for constant-posture, non-fatiguing, force-varying contractions about the elbow. Results from 15 subjects showed that EMG whitening and multiple-channel combination both reduce EMG-torque errors and their combination provides an additive benefit. Using 15th-order and higher linear FIR models, EMG-torque errors with a four-channel, whitened processor were reduced to an average error of 8% MVC (%VAF of 78%) at the slow tracking speed from an average error of 11% MVC (%VAF of 62%) with conventional processors. The unpredictable system ID behavior outside the natural bandwidth is avoided if the EMG amplitude data are decimated to a sampling rate of ten times the system bandwidth.

References

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