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Joint Angle-based EMG Amplitude Calibration

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***Abstract*— A calibration method is proposed to compensate for the changes in the surface electromyogram (SEMG) amplitude level of the biceps brachii at different joint angles due to the movement of the muscle bulk under the EMG electrodes for a constant force level. To this end, an experiment was designed, and SEMG and force measurements were collected from 5 subjects. The fast orthogonal search (FOS) method was used to find a mapping between SEMG from the biceps and force recorded at the wrist. Comparison between evaluation values from models trained with calibrated and non-calibrated SEMG signals revealed a statistically significant superiority of models trained with the calibrated SEMG.**

I. INTRODUCTION

The surface electromyogram (SEMG) is representative of the activation level of the underlying muscle which in turn is related to the muscle force output. Accurate muscle force estimation using SEMG is required in a number of applications including control of prostheses, ergonomic analysis, sports medicine, and human-robot interaction [1].

However, SEMG is affected by physiological and non-physiological factors which consequently imapact the accuracy of SEMG-based muscle force estimation [2]. One such factor that affects SEMG amplitude is the shift in muscle bulk as a result of joint movement, which potentially alters the relative location of the innervation zone (IZ) and the recording electrode [3]–[5]. The IZ shift under the electrode was previously believed to have a negligible effect on SEMG amplitude. However, recent research using state-of-the-art measurement equipment has shown that the shift might have considerable effect on SEMG levels [4]. Therefore, to improve SEMG-based force estimation accuracy, the IZ-shift must be taken into consideration.

According to the Hill muscle model, the output force calculated from SEMG is dependent on the muscle length and joint angle both in the static and dynamic cases. Due to the motion of the muscle bulk, SEMG amplitude for constant-force isometric contractions will vary for different joint angles not only as a result of the force-length property of the muscle, but also because of the IZ shift.

In this work, a new calibration method is introduced in which the variation in SEMG level due to the shift in relative IZ-electrode position at different joint angles is compensated. To evaluate the proposed calibration method, a dataset of SEMG recordings from the biceps brachii and forces measured at the wrist during elbow flexion is collected. An orthogonal search approach is used to predict wrist force; we demonstrate that by calibrating the SEMG amplitude based on the joint angle at which the contraction was performed, statistically significantly more accurate force predictions are achieved.

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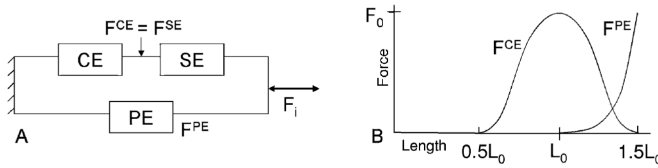

Fig. 1. (A) Structure of a Hill model. (B) Isometric contractile element force F^{CE} and parallel elastic element force F^{PE} as a function of muscle length.

II. SEMG CALIBRATION METHOD

The proposed calibration method for SEMG signals collected from the biceps brachii is based on the Hill muscle model, which will be explained briefly.

A. Hill muscle model components

The classic Hill model is composed of a contractile element (CE), a series elastic element (SE), and a parallel elastic element (PE) [6], as shown in Figure 1. The contractile force generated by the CE (F^{CE}) is equal to that in the SE (F^{SE}), and the total muscle force (F_i) is the sum of the forces in each of the two parallel sections, F^{CE} and F^{PE}.

F^{CE} can be interpreted as the activity of the contractile units within the muscle fiber, which contract and generate tension following stimulation from a motor nerve. F^{CE} can be expressed as the product of maximal isometric force (F_0), the force-length or equivalently force-joint angle (f_θ) and force-velocity (f_v) relationships, and muscle activation $u(t)$ [7], that is:

$$F^{CE} = F_0 \cdot f_\theta \cdot f_v \cdot u(t) \tag{1}$$

For an isometric contraction, f_v has no contribution i.e. $f_v = 1$. The output of F^{CE} peaks at the optimal joint angle (θ_0) and is reduced for values of θ less than or greater than θ_0. The force generated by the PE component (F^{PE}) is attributed to the stretch resistance in inactive muscle and only exerts tension when the muscle is stretched beyond its optimal angle, as illustrated in Figure 1. In our experiment, the joint angle was limited to the range $45 - 105°$ (where $0°$ is full extension) and F^{PE} for the biceps brachii was assumed to be negligibly small. The force induced at the wrist due to torque about the elbow generated by a single muscle can be expressed as:

$$F_{W_\theta} = F_0 \cdot f_\theta \cdot u(t) \cdot MA_{m_\theta}/MA_f \tag{2}$$

where MA_m is the muscle moment arm and MA_f is the forearm moment arm. Although MA_f is also a function of joint angle, MA_f variation is small over the joint angle range of our experiment and is assumed to remain constant.

B. Angle-based calibration

For an individual muscle, the amount of effort needed to generate a specific level of muscle force varies with joint angle due to the force-length property of the muscle, where, based on the Hill muscle model, minimal effort is required at the optimal joint angle, θ_0. Conversely, for a constant muscle force the SEMG amplitude will vary with joint angle. Part of this variation can be described by the change in contraction dynamics of the muscle, i.e. f_θ in the Hill model and by variation of the muscle moment arm, MA_m. Another factor, which is not related to muscle mechanics, is the movement of the muscle bulk at different joint angles, which results in a shift in the relative position of the IZ and the recording electrode [4].

Let the muscle activation, $u(t)$, be represented by the SEMG amplitude recorded at a single electrode site at a reference joint angle, θ_{Ref}. A change in joint angle introduces a modifying factor, c_θ, such that the SEMG amplitude will be $c_\theta \cdot u_1(t)$, instead of $u_1(t)$ which would have been measured had there been no shift in the relative IZ-electrode position. At θ_{Ref}, $c_{Ref} = 1$.

The main objective of this work is to correct the SEMG amplitude level by applying correction factors, c_{θ_i}, where i denotes joint angle, to compensate for the shift in IZ-electrode position. The resulting calibrated SEMG signals will be used in an EMG-force relationship model for elbow flexion.

Adding the effects of IZ shifting to equation (2), wrist force at arbitrary joint angle θ, due to flexion torque generated by isometric contraction of the biceps brachii, is:

$$F_{W_\theta} = F_0 \cdot f_\theta \cdot (c_\theta \cdot u(t)) \cdot MA_{m_\theta}/MA_f \tag{3}$$

Consider the case where a constant SEMG amplitude level ($u(t) = \overline{EMG_0}$) is measured at joint angles, $\theta_1 = \theta_{Ref}$ and θ_i, $i = 2, \cdots, n$, then the induced wrist forces at the above angles will be

$$F_{W_{\theta_1}} = F_0 \cdot f_{\theta_1} \cdot \overline{EMG_0} \cdot MA_{m_1}/MA_f \tag{4}$$

$$F_{W_{\theta_i}} = F_0 \cdot f_{\theta_i} \cdot (c_{\theta_i} \cdot \overline{EMG_0}) \cdot MA_{m_i}/MA_f \tag{5}$$

The unknown factor, c_{θ_i}, is reflected in the ratio of the measured forces which we call the calibration coefficient α_i:

$$\alpha_i = F_{W_{\theta_i}}/F_{W_{\theta_1}} = \frac{f_{\theta_i} \cdot c_{\theta_i} \cdot MA_{m_i}}{f_{\theta_1} \cdot MA_{m_1}} \tag{6}$$

Now, from (3), for a series of isometric constant force measurements at joint angles θ_i, $i = 1, \cdots, n$

$$F_{W_{\theta_0}} = F_0 \cdot f_{\theta_i} \cdot (c_{\theta_i} \cdot \overline{EMG_i}) \cdot MA_{m_i}/MA_f \tag{7}$$

where $\overline{EMG_i}$ is the measured SEMG amplitude level. In this case, the SEMG-force model tries to find the mapping β:

$$\beta = F_0 \cdot f_{\theta_i} \cdot c_{\theta_i} \cdot MA_{m_i}/MA_f \tag{8}$$

However, if $\overline{EMG_i}$ is calibrated by α_i such that $\alpha_i \cdot \overline{EMG_i}$ is mapped to the constant force $F_{W_{\theta_0}}$, then the EMG-force model will try to find the new mapping $\tilde{\beta}$ such that for each joint angle:

$$\alpha_i \cdot \overline{EMG_i} \cdot \tilde{\beta} = F_0 \cdot f_{\theta_i} \cdot c_{\theta_i} \cdot \overline{EMG_i} \cdot MA_{m_i}/MA_f \tag{9}$$

By substituting for α_i in (3), one finds the mapping

$$\tilde{\beta} = \frac{F_0 f_{\theta_1} \cdot MA_{m_1}}{MA_f} \tag{10}$$

which can be considered as constant for any given angle θ_i. The signal $\alpha_i \cdot \overline{EMG_i}$ is a calibrated version of the $\overline{EMG_i}$ in which variations of SEMG amplitude due to changes in IZ locations shift with respect to the electrode and contraction dynamics are both compensated for. Therefore, the SEMG-force model will try to find the new mapping $\tilde{\beta}$, which is a constant, simpler and is more linear than the mapping β. Two main advantages of this method compared to the Hill model is that unlike Hill model the effect of IZ shift under the electrode is taken care of and also no physiological measurements is required. Furthermore, since the nonlinearity of the EMG-force relationship is reduced, use of simpler modeling structures might be possible and the training time could be reduced.

III. DATA COLLECTION AND PROCESSING

A. Experimental Setup

The experiments were conducted on a single degree-of-freedom (1-DOF) exoskeleton testbed [8]. The apparatus holds the shoulder and wrist in a fixed position, and constrains flexion and extension of the right arm to the horizontal plane. The axis of rotation of the elbow is aligned with a pivoting aluminum bar attached to a Maxon DC motor, with an 8:1 cable driven power system. The elbow angle is equivalent to 1/8 of the motor angular position and is measured with a resolution of 1/4000 of a degree. Elbow torque expressed as force at the wrist was measured using an ATI 6-DOF Gamma force/torque sensor with a stiffness of 9.1×10^6 N/m.

B. SEMG Data collection and pre-processing

Data were collected from 5 male subjects with mean age of 25 years and no known neuromuscular deficit. All subjects provided informed consent and the study was approved by the Health Sciences Research Ethics Board, Queen's University. SEMG data were recorded from the biceps brachii, and triceps brachii of the right arm of each subject using two Invenium Technology AE100 active bipolar SEMG sensors (electrode separation of 15 mm and electrode diameter of 4 mm) for each muscle. Electrode locations were measured with respect to anatomical landmarks and recorded for each subject. All data were sampled at a rate of 1 kHz through a Quanser Quarc real-time control system and transferred to a dedicated acquisition computer. The raw SEMG signals were

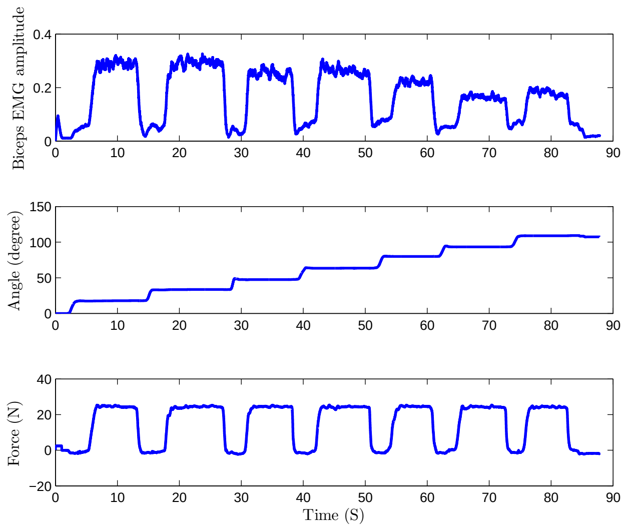

Fig. 2. A sample data trial recorded from subject 1.

processed off-line using software developed in MATLAB. DC bias was removed from the raw SEMG signals and the linear envelope (LE) was computed to estimate the signal amplitude. The LE was obtained by rectifying the SEMG and smoothing with a 400 point moving average filter with a stop-band of $0.6\,Hz$. For both the biceps and the triceps, SEMG amplitude levels from the two recording locations were averaged. The recorded force was also smoothed with a 100 point moving average filter. This resulted in a $150\,ms$ delay in the LE with respect to force. The filter length and the resulting delay were chosen to approximate the delay between the collected SEMG signal and the force generated by the muscle. For both the LE and force data records, the segments for which constant force was maintained were extracted (i.e. the transient and rest portions of the data were discarded) and concatenated into a single record.

C. Experiment Procedure

Two sets of isometric flexion tests were conducted - a constant force isometric test and a constant SEMG level test. The triceps brachii activation level was monitored. Any trials in which there was significant activation of the triceps were rejected and re-done.

Constant force isomeric tests: All five subjects completed 5 recording trials with an enforced two minute rest between trials to avoid fatigue. In each trial, the subjects were asked to generate a target flexion force at the wrist by contracting the biceps brachii at five elbow joint angles ranging from $45°$ to $105°$ in $15°$ intervals, taking full arm extension as $0°$. In all five trials, the target force was $25\,N$ as shown in Figure 2. Visual feedback of the measured wrist force was displayed to the subjects in real-time.

Constant SEMG level test: The subjects completed one recording trial following the same procedure as above and at the same joint angles. In this case, the subjects were given visual feedback of the SEMG amplitude and were asked to maintain a constant SEMG amplitude level of $0.15\,V$. LE data were calibrated using the method previously described.

D. EMG force mapping

1) FOS: Fast orthogonal search is a nonlinear identification method that approximates a system output as a weighted sum of M linear or nonlinear basis functions $p_m(n)$ and coefficient terms a_m and aims to minimize the mean square error (MSE) between the estimate and the system output [9], [10]. The FOS model takes the form:

$$y(n) = \sum_{m=1}^{M} a_m p_m(n) + e(n) \tag{11}$$

where $e(n)$ is the estimation error, $y(n)$ is the measured system output and n is the discrete time sample index. The FOS method searches through a number, N, of available candidate basis functions, where $N >> M$ and iteratively selects those functions which contribute the greatest reduction in MSE.

The FOS method is based on the principals of Gram-Schmidt orthogonal identification. Orthogonal basis functions are generated from the candidate basis functions and coefficients are found to minimize the MSE of the estimate.

2) Model identification: A FOS model with seven functions was generated for each of the five data sets for each subject. The functions were selected from pool of FOS candidate functions as in [11] which was composed of common mathematical terms such as the sigmoid function, the modified square root function, and the limited quadratic function. Models were evaluated using percent relative mean square error (%RMSE):

$$\%RMSE = \frac{\sum_{j=1}^{n}(F_{wj} - \hat{F}_{wj})^2}{\sum_{j=1}^{n} F_{wj}^2} \times 100 \tag{12}$$

where F_{wj} is the measured force at the wrist, $\hat{F}_{wj}$ is the FOS model estimate of wrist force and *n* is the discrete time sample index. Each model was evaluated four times using the data from the four remaining trials, resulting in four individual evaluation %RMSE values for each model, which were averaged to obtain a *Model* evaluation %RMSE. The best value out of five *Model* %RMSE terms was then chosen as the *Subject* evaluation %RMSE value ($RMSE_{Subject}$). This value was used as the primary value to evaluate the success of the FOS model to predict F_w.

IV. RESULTS AND DISCUSSION

Figure 3 illustrates measured and estimated force for a model trained with the fifth trial and evaluated with the first trial of recordings from subject 1. Figure 4 shows the means and standard deviations of the evaluation results for the FOS models representing each subject using non-calibrated (NC) and calibrated (C) EMG recordings. To quantify the merit of using the calibrated SEMG, the Wilcoxon signed rank test was performed between results with and without calibration over 20 values from all subjects. The boxplot in Figure 5 shows the distribution of the evaluation values for both methods. A p-value of 0.03 confirmed a statistically significant improvement in estimation results with calibration.

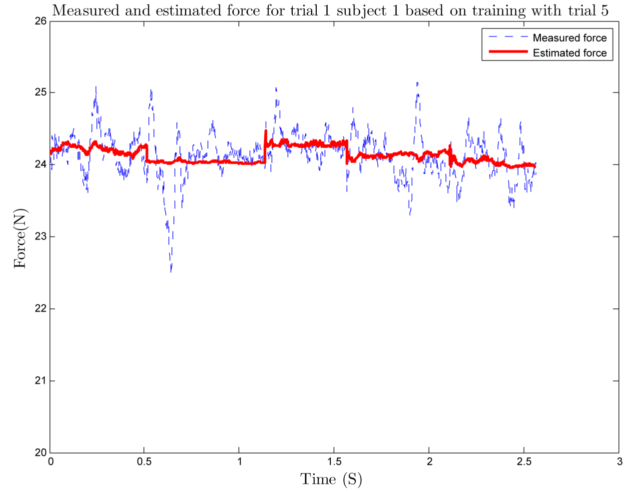

Fig. 3. Measured and estimated force for a model trained with trial 5 and tested with trial 1 of recordings from subject 1.

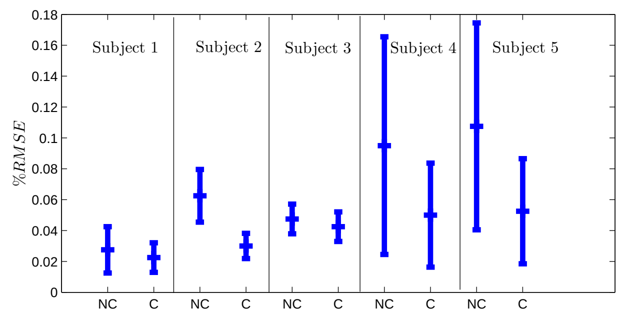

Fig. 4. NC and C represent the distribution of $RMSE$ values for non-calibrated and calibrated EMG respectively for each subject. The horizontal line is the mean value. The solid bar extends $\pm$ one standard deviation from the mean.

The main advantage of this SEMG-Force modeling is that unlike models such as Hill-based model, it incorporates a correction for the change in SEMG due to the electrode-IZ shift effect. The effects of variation in SEMG amplitude due to muscle force-length relationship and the change in muscle moment arm are also compensated with no need for physiological measurements. Furthermore, since the nonlinearity of the EMG-force relationship is reduced, use of simpler modeling structures might be possible and the training time could be reduced. Although in this study the calibration was limited to the biceps brachii, the constant SEMG tests can be conducted for other muscles such as the triceps brachii. It is important to note that the data at desired joint angles must be collected for contractions in which the muscle action is isolated as much as possible. For instance, for triceps extension contractions are required.

V. CONCLUSIONS

The goal of this research was to develop a calibration method for the amplitude of the SEMG signals collected from a single sensor installed on the biceps brachii in order to compensate for the variations in SEMG amplitude due to the changes in joint angle during isometric contractions. This was accomplished by incorporating SEMG calibration

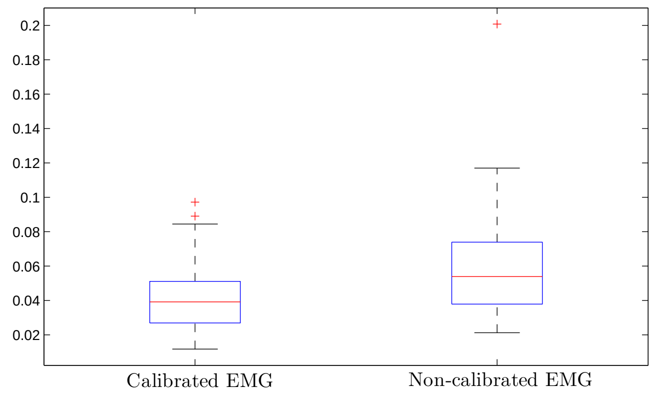

Fig. 5. Boxplot comparing the distribution of the evaluation values before and after calibration.

coefficients calculated for each angle from force (torque) recordings during constant SEMG level isomeric tests. The calibration coefficient corrects for the variations in SEMG amplitude due to changes in muscle length and displacement of the IZ. Non-parametric FOS-based SEMG-force mapping models were trained and evaluated with the collected database both for calibrated and non-calibrated SEMG. The experimental results show that forces are predicted significantly more accurately using calibrated SEMG data, than from non-calibrated data.

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