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Surface EMG force modeling with joint angle based calibration

Javad Hashemi [a], Evelyn Morin [a,*], Parvin Mousavi [b], Keyvan Hashtrudi-Zaad [a]

[a] Department of Electrical and Computer Engineering, Queen's University, Kingston, ON, Canada, K7L 3N6
[b] School of Computing, Queen's University, Kingston, ON, Canada, K7L 3N6

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ABSTRACT

In this paper, a calibration method to compensate for changes in SEMG amplitude with joint angle is introduced. Calibration factors were derived from constant amplitude surface electromyogram (SEMG) recordings from the biceps brachii (during elbow flexion) and the triceps brachii (during elbow extension) across seven elbow joint angles. SEMG data were then recorded from the elbow flexors (biceps brachii and brachioradialis) and extensors (triceps brachii) during isometric, constant force flexion and extension contractions at the same joint angles. The resulting force at the wrist was measured. The fast orthogonal search method was used to find a mapping between the system inputs – estimated SEMG amplitudes and joint angle – and the system output – measured force, for both calibrated and non-calibrated SEMG data. Models developed with calibrated data yielded a statistically significant improvement in force estimation compared to models developed with non-calibrated data, suggesting that the calibration method can compensate for changes in the SEMG–force relationship with changing joint angle. It was also found that the number of non-linear, joint angle-dependent terms used in the SEMG–force model was reduced with calibration. Additionally, initial inter-session analysis performed for four subjects suggests that calibration values can be used for subsequent recording sessions, and different output force levels.

1. Introduction

Accurate muscle force estimation using the surface electromyogram (SEMG) is required in a number of applications including control of prostheses, ergonomic analysis, sports medicine, and human–robot interaction (Staudenmann et al., 2010). However, even for isometric contractions, the SEMG is affected by physiological and nonphysiological factors which impact the accuracy of SEMG amplitude estimation (Disselhorst-Klug et al., 2009; Farina et al., 2004). Also, errors introduced by the estimation of physiological and biomechanical parameters affect the accuracy of the SEMG–force relationship. Changing joint angle influences the estimation of both the SEMG amplitude and muscle biomechanical parameters by altering the muscle length, the muscle moment arm and the relative location of the innervation zone (IZ) with respect to the SEMG recording electrode (Farina et al., 2001).

Different modeling methods have been used to predict muscle force from SEMG. Parametric modeling approaches often use Hill's muscle model, which explicitly incorporates muscle and joint dynamics, including the muscle force–length relationship and moment arm, in the model (Lloyd and Besier, 2003; Mountjoy et al., 2010). In general, the force–length relationship for a muscle in vivo is limited to a portion of the classic force–length relationship. The force–length relationship for the biceps brachii has an inverted-U shape (Bechtel and Caldwell, 1994; Leedham and Dowling, 1995) and the relationship for several upper limb muscles has been modeled as a Gaussian function (Cavallaro et al., 2006). Models have been developed to quantify the moment arms of the upper arm muscles with respect to joint angle (An et al., 1984; Holzbaur et al., 2005). Cavallaro et al. (2006) modeled the moment arms of several upper limb muscles based on the three-dimensional upper limb muscle models of Garner and Pandy (2001). Force–length and muscle moment arm model parameters have been used in parametric SEMG–force models for more accurate force estimation (Mountjoy et al., 2009).

A few recent studies in the literature have considered the effects of IZ shift on the SEMG using multielectrode recordings. In the biceps brachii, Martin and MacIsaac (2006) found that the IZ can shift up to 30 mm in a direction distal to the shoulder with elbow extension. Piitulainen et al. (2009) reported a shift up to 24 mm in IZ location. Beck et al. (2008) studied the effects of the relative position of the electrode and IZ on the recorded SEMG. They noted that electrode placement has an effect on the SEMG amplitude and suggested that the effect of IZ shift on SEMG amplitude is reduced by normalization with respect to the highest recorded value for each subject. Rantalainen et al. (2011) examined the effect of IZ on the SEMG–force relationship for the biceps brachii and found

* Corresponding author.
E-mail address: morine@post.queensu.ca (E. Morin).

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that proximity of the recording electrodes to the IZ results in inconsistency in the estimated force–SEMG relationship. Multi-electrode SEMG recordings can be used to compensate for this effect, but this imposes substantial hardware and computational costs.

Non-parametric single channel SEMG–force estimation methods reported in the literature, e.g. Clancy and Hogan (1997), implicitly model angle-based nonlinear effects, including force–length and moment arm effects and IZ shift. As a result, the generated models depend heavily on joint-angle. It has been suggested that multiple reference measurements are required if different joint angles will be considered in SEMG–force modeling (Disselhorst-Klug et al., 2009). Doheny et al. (2008) took maximum force and SEMG recordings at multiple joint angles and used these values to normalize sub-maximal force and SEMG recordings at the same joint angles. The results showed promise for de-correlating the SEMG–force relationship from joint angle (Doheny et al., 2008). However, the method involves collecting several maximal voluntary contractions, which may result in subject discomfort and muscle fatigue unless sufficient rest is given between each trial.

In this work, a sub-maximal calibration procedure is developed to compensate for changes in joint angle – which in turn cause changes in muscle length, muscle moment arm and IZ shift – thereby achieving an improved SEMG–force model for flexion–extension contractions of the elbow over a range of elbow joint angles. The method is developed for single-site SEMG recordings, obviating the need for multi-site recording. To this end, a dataset of SEMG recordings during elbow flexion and extension and the resulting force measured at the wrist during elbow flexion and extension was collected. Fast orthogonal search (FOS), a time domain method for rapid non-linear identification, was used to generate SEMG–force models, using both non-calibrated and calibrated SEMG amplitude data. The merit of the proposed calibration procedure was assessed by comparing the performances of the models, generated using calibrated and non-calibrated datasets, in terms of force estimation error.

2. Methods

2.1. Angle-based calibration algorithm

The Hill muscle model equations for the SEMG–force relationship are used to explain the concept of the proposed calibration method for a single muscle, in this case, the biceps brachii. The method is extended to include two muscles for which calibration values can be obtained independently.

2.1.1. Wrist force for a single contributing muscle

The force induced at the wrist due to torque about the elbow generated by a single muscle can be expressed as:

$$F_{W_\theta} = F_{muscle} \cdot M_m(\theta)/M_f \tag{1}$$

where F_{muscle} is the muscle force, $M_m(\theta)$ is the muscle moment arm and M_f is the moment arm of the loadcell measuring force at the wrist. Although M_f is also a function of elbow joint angle, M_f variation is small over the joint angle range in our experiment and is assumed to remain constant. According to the Hill muscle model, muscle force is the sum of the contractile element force (F^{CE}) and the parallel elastic element force (F^{PE}) (Winters and Stark, 1987). 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 and can be represented as:

$$F^{CE} = F_0 \cdot f_\theta(\theta) \cdot f_v(v) \cdot u(t) \tag{2}$$

where F_0 is the maximal isometric force generated by the muscle, $f_\theta(\theta)$ represents the force–length or equivalently force-joint angle relationship, $f_v(v)$ represents the force–velocity relationship, and $u(t)$ is the muscle activation (Zajac, 1989). For an isometric contraction $f_v(v) = 1$. The output of F^{CE} peaks at the optimal joint angle (θ_0) and decreases for values of θ less than or greater than θ_0. In our experiment, F^{PE} for the biceps and triceps brachii is assumed to be negligibly small for the limited range of motion chosen. As a result, $F_{muscle} \approx F^{CE}$.

For an individual muscle, the amount of activation needed to generate a specific level of force varies with joint angle, where, minimal effort is required at the optimal joint angle, θ_0. Thus, for a constant muscle force the SEMG amplitude will vary with joint angle. Part of this variation is described by the change in contraction dynamics of the muscle, i.e. $f_\theta(\theta)$ in the Hill model and by variation of the muscle moment arm, $M_m(\theta)$. SEMG amplitude is also affected by the movement of the muscle bulk as joint angle changes, resulting in a shift in the relative position of the IZ and the recording electrode (Martin and MacIsaac, 2006; Piitulainen et al., 2009), which is not related to muscle mechanics.

2.1.2. Angle-based calibration for a single contributing muscle

Consider an isometric contraction of the biceps brachii, resulting in a flexion torque about the elbow. Let the muscle activation, $u(t)$, be truly represented by the SEMG amplitude ($\underline{EMG_1}$) recorded at a single electrode site at a reference joint angle, $\theta_1 = \theta_{Ref}$, that is $u(t)_{\theta_1} = \underline{EMG_1}$. A change in joint angle introduces a modifying factor, c_{θ_i}, in the SEMG such that $u(t)_{\theta_i} = c_{\theta_i} \cdot \underline{EMG_i},\ i = 2, \ldots, n$ represents the true activation level of the muscle, where $\underline{EMG_i}$ is the amplitude of the recorded SEMG. By definition, the factor, c_{θ_i}, is due to the shift in the relative position of the recording electrode and IZ. At the reference angle $c_{\theta_1} = 1$. Thus, the force induced at the wrist at a joint angle θ_i due to the flexion torque is:

$$F_{W_{\theta_i}} = F_0 \cdot f_\theta(\theta_i) \cdot (c_{\theta_i} \cdot \underline{EMG_i}) \cdot M_m(\theta_i)/M_f \tag{3}$$

As shown in Fig. 1a, a modeling procedure, such as fast orthogonal search, attempts to find a mapping (β) between the recorded SEMG and the recorded $F_{W_{\theta_i}}$ to derive:

$$\beta(\theta_i) = F_{W_{\theta_i}}/\underline{EMG_i} = F_0 \cdot f_\theta(\theta_i) \cdot c_{\theta_i} \cdot M_m(\theta_i)/M_f \tag{4}$$

Although numerical or analytical models for $f_\theta(\theta)$ and $M_m(\theta)$ are available (An et al., 1981; Chang et al., 1999; Maganaris, 2001; Langenderfer et al., 2004; Cavallaro et al., 2006), no model for c_θ is available as c_θ is not directly observed from the recorded SEMG. However, information on c_θ at various joint angles can be observed in the measured forces F_{W_θ}. Thus, calibration values are derived from force measurements for a series of isometric constant SEMG trials at different joint angles.

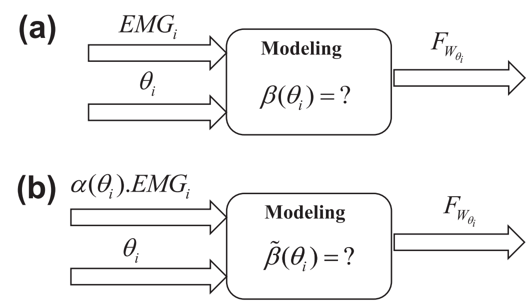

Fig. 1. Modeling the isometric SEMG–force relationship: (a) before calibration and (b) after calibration.