34th Annual International Conference of the IEEE EMBS
San Diego, California USA, 28 August - 1 September, 2012

Enhanced Multi-Site EMG-Force Estimation Using Contact Pressure

Javad Hashemi, Evelyn Morin, Parvin Mousavi and Keyvan Hashtrudi-Zaad

***Abstract*— A modification method based on integrated contact pressure and surface electromyogram (SEMG) recordings over the biceps brachii muscle is presented. Multi-site sEMGs are modified by pressure signals recorded at the same locations for isometric contractions. The resulting pressure times SEMG signals are significantly more correlated to the force induced at the wrist (F_W), yielding SEMG-force models with superior performance in force estimation. A sensor patch, combining six SEMG and six contact pressure sensors was designed and built. SEMG, and contact pressure data over the biceps brachii and induced wrist force data were collected from 5 subjects. Polynomial fitting was used to find a mapping between biceps SEMG and wrist force. Comparison between evaluation values from models trained with modified and non-modified SEMG signals revealed a statistically significant superiority of models trained with the modified SEMG.**

I. INTRODUCTION

An accurate determination of individual muscle forces is desired in a number of fields such as ergonomics, sports medicine, prosthetics and human-robot interaction. However, muscle forces cannot be directly measured without using invasive methods. Therefore, surface electromyogram (SEMG) based muscle force estimation is frequently used as an alternative non-invasive approach.

The linearity of the SEMG-force relationship is affected by physiological and non-physiological factors, where muscle force is controlled by two physiological parameters: firing rate modulation and motor unit (MU) recruitment. The increase in force with MU recruitment is predicted by modeling to be more than proportional (size principal) [1]. The increase in MU size with increasing force suggests a more than proportional increase in SEMG amplitude as well. The increase in force with increasing firing rate has been predicted by modeling to be less than proportional (due to phase cancelation) [1]. This holds for the increase in SEMG amplitude, because of greater phase cancelation with increasing firing rate [2]. Other factors such as load sharing with unequal contributions of the synergistic muscles at different contraction levels [3] and variation in joint stiffness [4] are reported to contribute to the nonlinearity of the SEMG-force relationship. Non-physiological factors such as the relative shift between innervation zone and recording site have also been reported to deteriorate the SEMG-force linearity [5], [6].

This work is supported by FedDev Ontario, Ontario Centre of Excellence and Invenium Technology.

J. Hashemi, K. Hashtrudi-Zaad, E. L. Morin are with the Department of Electrical and Computer Engineering, and P. Mousavi is with the School of Computing, Queen's University, Kingston, ON K7L 3N6, Canada. javad.hashemi@queensu.ca

Novel methods such as multi-channel monopolar SEMG recording yield better estimates of muscle force by: (1) reducing the effects of phase cancelation, and (2) providing an adequate representation of the heterogeneous activity of motor units within a muscle, i.e. reducing the effects of the size principle. Monopolar recordings have a larger pick-up area and contain more power from the far-field potentials of distant MUs than bipolar recordings [7]. Although this helps to record small MU activity deep in the muscle at low contraction levels, the risk of recording co-contraction activity in neighboring muscles increases. Using multi-channel bipolar SEMG recording can alleviate this issue.

Surface muscle pressure (SMP) have been recently used as an alternative to SEMG signals to study the behavior of the muscles [8] as well as signature of the muscle generated grip force [9]. SMP recordings are not affected by parameters such as phase cancelation and size principal and therefore can have complementary information to SEMG in estimating muscle force.

In this work, a new multi-channel SEMG recording approach based on a grid of active bipolar electrodes with integrated contact pressure sensors is presented. The integrated signal is recorded over the biceps brachii. Contact pressure will vary with both contraction level and elbow joint angle as the muscle changes configuration and bulges against a cuff wrapped around the upper arm over the integrated sensor patch. Since bipolar SEMG sensors are used, the risk of recording co-contraction activities at higher contraction levels is reduced. The relationship between the recorded signal modalities - SEMG, surface muscle pressure, and SEMG modified by pressure - and force measured at the wrist was explored using simple polynomial fitting which provides fast and computationally efficient force estimation.

II. MATERIALS AND METHOD

A. Subjects

Data were collected from 5 subjects (two female and three male, mean age 24.75 years, with a standard deviation of 2.6 years) with no known neuromuscular deficit. All subjects provided informed consent and the study was approved by the Health Sciences Research Ethics Board, Queen's University.

B. Data collection protocol

The experiments were conducted on a single degree-of-freedom (1-DOF) exoskeleton testbed (Figure 1(a)) which has been described previously [10]. The apparatus holds the shoulder and wrist in a fixed position, and constrains flexion and extension of the right arm to the horizontal

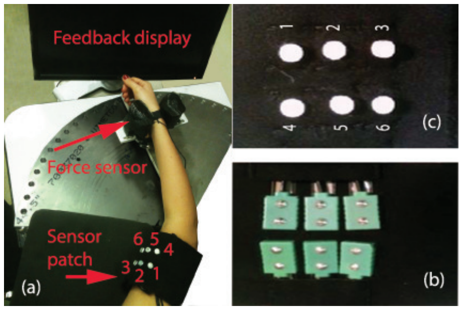

Fig. 1. The 1-DOF testbed used to collect SEMG and force data (a). For isometric contractions the pivoting bar is locked mechanically at the desired joint angle. The designed sensor patch composed of six active bipolar SEMG sensors arranged as two rows of three sensors (b). Six contact pressure sensors were mounted on top of the EMG sensors (c).

plane. The axis of rotation of the elbow is aligned with a pivoting aluminum bar. Data were collected in two sessions. In the first session biceps brachii SEMG and wrist force data were recorded for maximum voluntary contraction (MVC) at 90° elbow flexion for two trials with 5 minutes rest between trials. The MVC force for each subject was obtained by averaging the two recordings. The second session was performed 15 minutes after the first session. Subjects were asked to follow a half-cycle sinusoidal force pattern varying from zero to a maximum force level of 50% MVC, displayed on a computer monitor. Visual feedback of the measured wrist force was provided in real time. Subjects generated isometric contractions at 90°. Subjects completed five sets of sub-maximal isometric flexion tests ranging from 0 to 50% MVC. SEMG, SMP and wrist force data were sampled at 1000 samples per second with 14-bit resolution through a National Instrument $PCI6225$ dedicated data acquisition board.

C. Data collection and instrumentation

SEMG data were recorded from the biceps brachii muscle of the right arm of each subject using a patch composed of six Invenium Technology AE100 active bipolar SEMG sensors (electrode separation of 15 mm and electrode diameter of 4 mm) arranged as two rows of three sensors across the width of the muscle as shown in Figure 1(b). Six FlexiForce A201 contact pressure sensors were mounted on top of the SEMG sensors as shown in Figure 1(c). The location of the sensors were indexed as shown in Figure 1(a). The intersection of the vertical and horizontal midline of the electrode patch was placed on the location suggested by SENIAM [11] for the biceps brachii measured with respect to anatomical landmarks for each subject. A cuff similar to a regular blood pressure cuff (without the inflating tube and compartments) was wrapped around the upper arm over the sensor patch so that a change in muscle shape due to contraction or a change in joint angle pushes the contact pressure sensors against the cuff and generates a contact pressure signal. The cuff was tightened such that the maximum force on the contact pressure sensors which occurred at MVC did not exceed the sensors' linear range (4 N). The force applied to the pressure sensors during the 5 sub-maximal contractions varied between 1 to 2 N, which is within the linear range of the sensors. Elbow torque expressed as force at the wrist was measured using an ATI 6-DOF Gamma force/torque sensor.

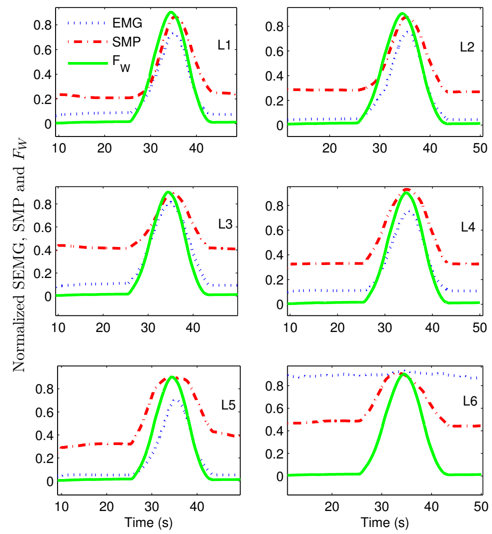

Fig. 2. Sample dataset from one trial collected from six locations (L1 to L6). For demonstration purpose the normalized values of the signals are shown with respect to their maximum values. It should be noted that the SEMG signal of the lowest right plot had a very small value before normalization.

D. Data analysis

The raw signals were processed off-line using software developed in MATLAB (Mathworks, Inc, MA, USA). DC bias was removed from the raw SEMG signals and the linear envelope (LE) was obtained by rectifying and smoothing with a 400 point ($400\,ms$, $0.6\,Hz$) moving average filter to estimate the signal amplitude. The readings from contact pressure sensors were calibrated using calibration coefficients obtained prior to the test. The contact pressure readings and the recorded force at the wrist were smoothed with a 100 point moving average filter. 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.

To examine the inter-relationship between the input modalities, cross-correlation coefficients for the SEMG and SMP signals were computed. As well, cross-correlation coefficients were calculated for SEMG, SMP and SEMG scaled by SMP versus force measured at the wrist. Values were calculated for all six recording locations and all five trials across all subjects. Averaged SEMG and surface muscle pressure (SMP) signals were also obtained by averaging the corresponding waveforms from all six recording locations and the correlation coefficients were calculated. To verify the effects observed based on results of distributions of the cross-correlation between input and output modalities, second degree polynomial fitting was used to train SEMG-force models. For the five trial recordings, five models were

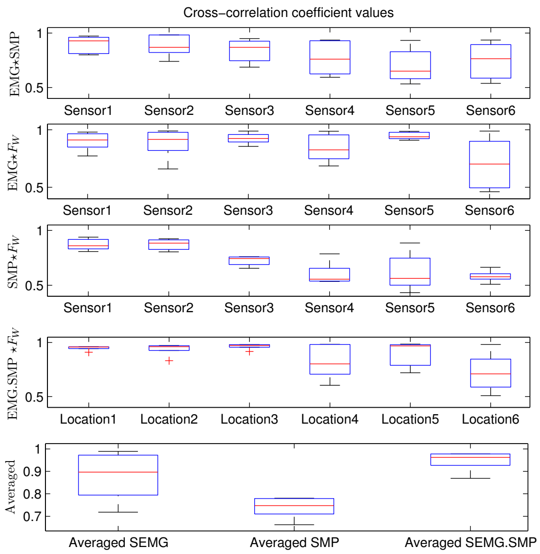

Fig. 3. Boxplots of cross correlation coefficient values between input modalities recorded over the biceps brachii and the resulting output wrist force averaged across subjects and trials for each sensor location (Location 1 to 6), and averaged across the six sensor locations (Averaged). Top: SEMG; middle: SMP; bottom: SEMG · SMP. Note that ⋆ symbol represents the cross-correlation operator.

trained and evaluated for each subject using the leave one out method i.e. one trial was used for training and the other four were used for evaluating. This resulted in twenty (5*4) $\%RMSE$ evaluation values for each subject and a total of 100 $\%RMSE$ values for each of the input signals. Where $\%RMSE$ was defined as

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

one-way ANOVA test was performed on evaluation values of the models trained and evaluated using different input modalities.

III. RESULTS AND DISCUSSION

A. *cross-correlation coefficient*

Figure 2 shows a sample trial recording from all locations. SEMG, SMP and force recordings at each location were normalized with respect to their maximum value only for demonstration purpose. The relationship between SEMG and force is affected by the size principle and phase cancelation due to spatial and temporal summation of motor unit potentials. There is also the effect of the muscle shifting under the electrodes as muscle force increases and the muscle bulk shifts. In single site recording this can cause variation in the pool of motor units detected and consequently in the SEMG-force mapping. Spatial averaging of the SEMG should improve the estimate of muscle activation providing better force prediction than for single site recording. However, since a large area of the muscle is covered by the electrodes, there is a possibility that some of the sensors are located close or even over the IZs resulting in extremely low SEMG recording values. This might explain the close to zero SEMG level in the 6^{th} location in Figure 2 which became artificially high in the plot due to the normalization. As for the SMP-force relationship, the hypothesis is that as a muscle contracts (considering the isometric case only), the muscle bulk will shift, resulting in increased pressure in regions where there is more muscle bulk (possibly over the central region of the muscle, which is likely the case for the biceps) and decreased pressure where there is less muscle bulk (near the end of the muscle or close to the muscle tendon).

In order to check the linearity of SEMG-force and SMP-force relationship the cross-correlation between SEMG and SMP with force was calculated. Boxplots for the cross-correlation coefficients calculated for the different input signals (SEMG and SMP) and their combination (SEMG · SMP) with measured wrist force are shown in Figure 3. The first plot shows SEMG and SMP were highly correlated in locations 1 to 3. Since the correlation coefficient does not change with scaling, the large variation in the median values of the coefficients between different recording sites (in second and third plots) indicates significant differences in signal morphology recorded from different locations for both SEMG and SMP recordings. The large range of variance in the coefficient values at specific locations (such as locations 4 and 6 in the second plot and location 5 in the third plot) could indicate the dependency of the values on subject specific anatomic features of the biceps brachii muscle. For the SEMG⋆F_W plot, locations 3 and 5 have the highest cross-correlation coefficients.

Looking at the SMP-force correlations (third plot), SMP is correlated with force for locations 1 and 2, indicating that the recorded pressure increases as force (or muscle contraction level) increases. It can be inferred that the muscle bulk at these locations is increasing with contraction level, resulting in a correlated pressure profile. Correlation drops for locations 3-6, indicating that the pressure signals (and hence muscle bulk) are not tracking with force as well as signals at 1 and 2.

The comparison between corresponding boxplots from the SEMG and SMP signals indicates that both signals are highly correlated with force at locations 1 and 2 only. SEMG is more correlated with force at locations 3 to 6 compared to SMP. This could be due to the large bias of the SMP signals as can be observed in Figure 2. On the other hand, unlike SEMG, SMP does not suffer from phenomena such as phase cancelation, size principal, and shift between the innervation zone and the recording location at different contraction levels.

Spatial averaging of the SEMG across multiple locations should result in an improved force prediction (sixth plot in Figure 5). However, considering the spread of the grid, some of the recorded SEMG signals will be close to the edges of the muscle and suffer from cross-talk, and some will be closer to the terminal tendon and suffer from low signal amplitude and/or end effects. Thus, we want to apply weighing factors such that the contributions, to the spatial average, from SEMGs recorded near the edges or end of the muscle are reduced relative to the contributions of SEMG recorded over the central bulk of the muscle. The hypothesis

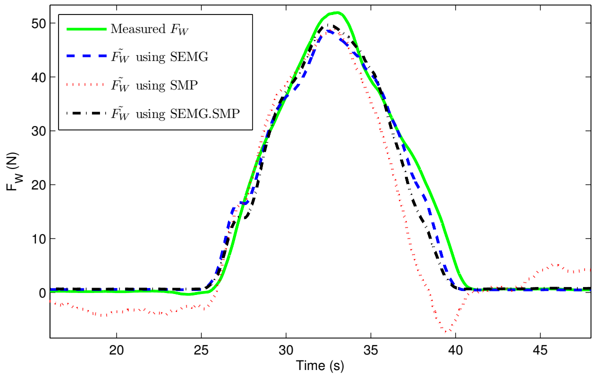

Fig. 4. Measured force at the wrist and estimated force using SEMG, SMP and SEMG $\cdot$ SMP signals.

is that the contact pressure measurements will provide a reliable means of generating these weighting factors.

In order to benefit the smaller bias of SEMG signals and more linearity of the SEMG-force in some locations, we created a new set of signals by multiplying SEMG by SMP signals and calculated the cross-correlation between the new data with the force as shown in the fifth plot in Figure 5. In 4 out of 6 locations SEMG·SMP had higher median values with smaller variance compared to SEMG alone. The sixth plot shows the results for SEMG, SMP, and SEMG·SMP averaged over the six location. The correlation coefficients for the signal averaged across locations are higher than for the SEMG or SMP signals alone. The averaged SEMG signals were more highly correlated with force than the averaged SMP signals. The correlation coefficients for locations 1 to 3 and for the signal averaged across locations are higher than for the SEMG or SMP signals alone, where the median values for the correlation coefficients for SEMG, SMP and SEMG $\cdot$ SMP are 0.89, 0.74 and 0.96. Since the averaged SEMG and averaged SEMG $\cdot$ SMP had satisfactory results they were more investigated by modeling.

B. Force estimation error

Figure 4 shows the measured force at the wrist and the estimated force using SEMG and SEMG·SMP as model input. As expected, the SEMG·SMP signal has a better performance in higher contraction levels and almost similar performance as SEMG signals in lower levels. A one-way ANOVA statistical test was performed on the three sets of $\%RMSE$ values for the force estimator models generated using the averaged SEMG, averaged SMP and averaged SEMG·SMP signals. Figure 5 shows the ANOVA boxplot comparing the evaluation results. The ANOVA test confirms that there is a significant improvement ($\alpha = 0.05; p < 0.001$) in evaluation results obtained using the SEMG·SMP data versus the SEMG data or the SMP data alone.

IV. CONCLUSIONS

The goal of this research was to develop a method for adjusting the amplitude of the SEMG signals obtained from a grid of active bipolar electrode such that the signal is less influenced by physiological (phase cancelation and size principal) and non-physiological (relative location of electrode and signal sources and shift in IZ) factors affecting the linearity of SEMG-force estimation. This was accomplished by using a sensor patch combining six SEMG and six SMP sensors. Since the SMP signals were not affected by the mentioned factors, they were used as a modifying factor for the SEMG amplitude recordings at 6 different locations on biceps brachii. The modified SEMG signals are believed to have better performance in SEMG-force estimation. This was verified by polynomial fitting and comparing the results. The modeling results show that forces are statistically predicted more accurately using modified SEMG amplitude data than SEMG amplitude data by itself.

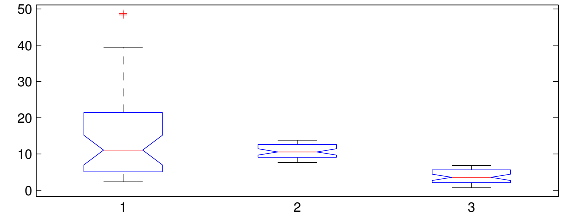

Fig. 5. ANOVA boxplots comparing $\%RMSE$ values for EMG-force models generated using averaged SEMG values versus averaged SEMG $\cdot$ SMP values.

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