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CANCELLATION OF FORCE INDUCED ARTIFACTS IN SURFACE EMG USING FSR MEASUREMENTS

Anders Fougner[1], Marthe Sæther[1], Øyvind Stavdahl[1], Peter J. Kyberd[2] and Jeremy Blum[2]

[1]Department of Engineering Cybernetics, Norwegian University of Science and Technology, Trondheim, Norway
[2]Institute of Biomedical Engineering, University of New Brunswick, Fredericton NB, Canada

INTRODUCTION

As multifunction prostheses become increasingly common, there is a need for improved control signal quality in order to control all the functions. Most signals commonly used for prosthesis control are sensitive to sweat, motion and external forces [1], which impairs prosthesis control performance.

We have developed a prototype surface electromyogram (sEMG) sensor with three built-in force sensing resistors (FSRs) for measuring the external forces, which may be used to cancel artifacts caused by these forces. The performance of the sensor as an estimator of muscle force is presented in this paper. The sEMG and FSR signals have also been tested individually, as a reference for the performance using the combination of these signals.

MATERIALS AND METHODS

The sEMG sensor unit was built from the metal electrodes of an Otto Bock 13E125 device, mounted with the original spacing and wired to an external preamplifier.
FSRs were chosen for force sensors due to their flatness and simplicity of use. Three individual FSRs allow both magnitude and position/direction of an external force to be estimated, factors both of which may be relevant for the artifact identification. Initial tests used an FSR component that was readily available. It is anticipated that with more appropriately sized sensors, the entire device will fit into a prosthesis socket. The sensors were sandwiched between two layers of acrylic glass using soft double sided tape (Fig. 1). The electrodes were attached to this structure with the reference electrode at the centre of the FSR array.

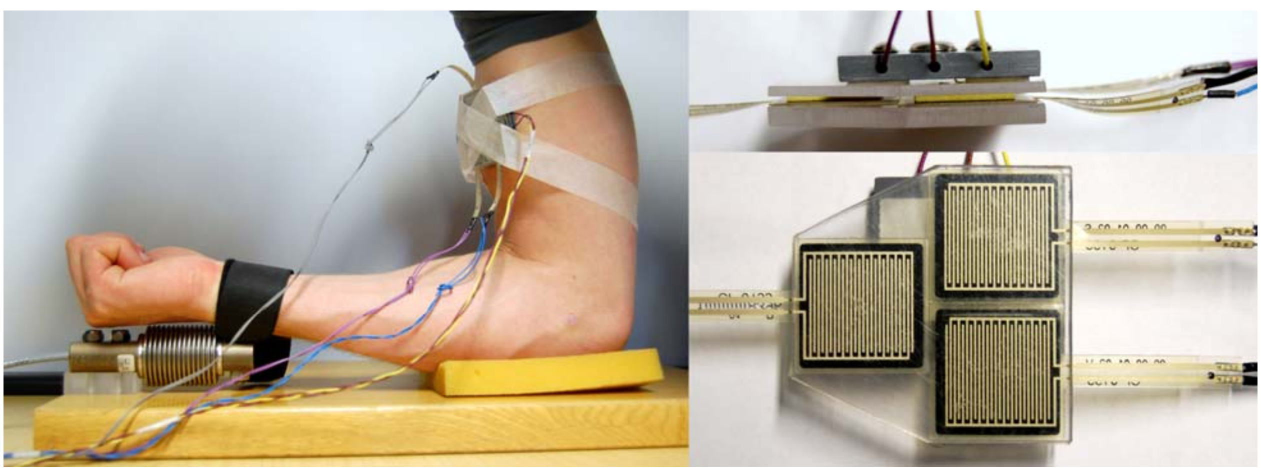

Fig. 1: Experiment setup and a close view of the sensor unit.

The device was taped to the *m. biceps brachii* of a healthy subject and tested by simultaneously measuring sEMG and FSR outputs while muscle contraction force was measured using a load cell (Fig. 1). The sEMG signal was pre-processed with a non-linear myoprocessor described in [2]. External forces in random directions were applied to the sensor during the measurements in order to induce artifacts. Data was collected at 218 Hz for approx. 50 s. Three data sets were acquired; a *training set* and a *validation set* collected immediately after each other, and a *test set* acquired after having removed and then reapplied the device to the subject’s arm.

Multilayer perceptron (MLP) networks with different numbers of hidden nodes (2-25 nodes, 10 MLP networks of each size) were employed to estimate the muscle force based on sEMG and FSR signals. Following MLP training and validation, the best 50% of the MLP networks of each size were chosen for final assessment using the test set. A linear and a quadratic mapping function were also fitted to the training set for comparison.

RESULTS

Fig. 2 presents an example data set with all recorded data. Note the two central peaks in the FSR signals, which are not accompanied by peaks in the load cell signal; these represent artifacts. The result of the force estimation, using the test set and an MLP network and a linear mapping function, respectively, is presented in Fig. 3.

Fig. 4 shows the estimated against measured force for the test set after training and validating the MLP network. Note the presense of hysteresis in the FSR based estimate and the apparent treshold levels in the sEMG based estimates. Also note the presence of force artifacts in both sEMG based graphs, evident as significant force estimate values at approximate zero load cell force.

The root mean sqare error (RMSE) rates for the different combinations of sEMG and FSR as inputs are presented in Fig. 5. No reduction in RMSE was detected when increasing the number of hidden MLP nodes beyond n=4.

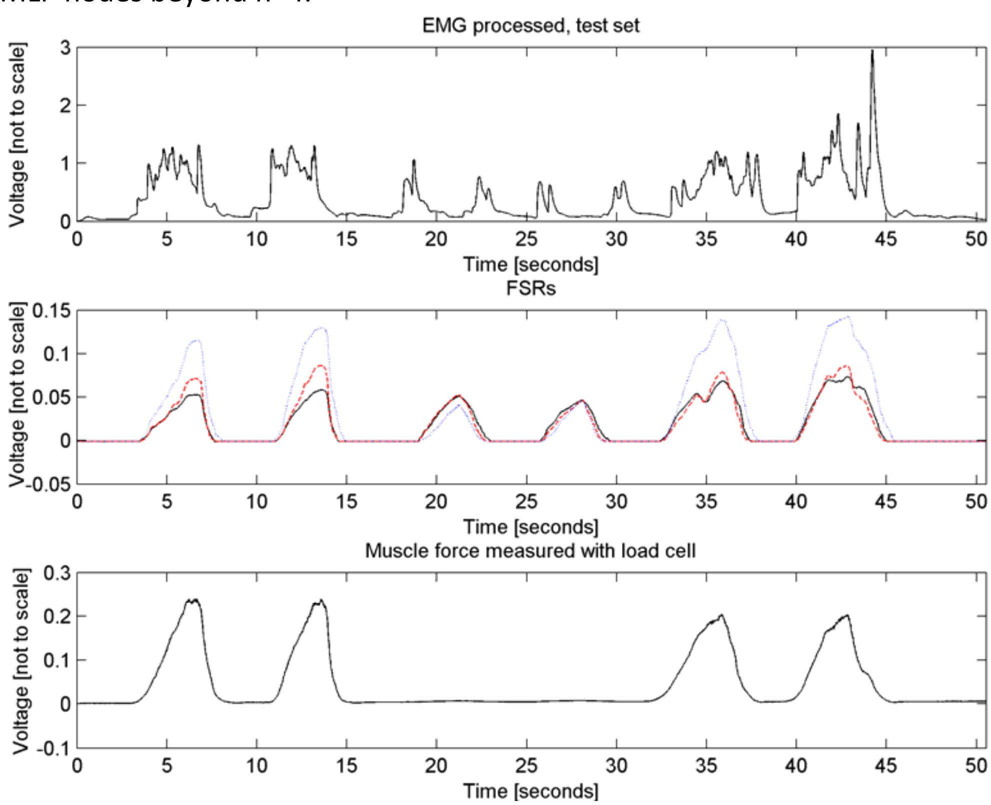

Fig. 2: Test set containing measurements from sEMG, 3 FSRs and the load cell.

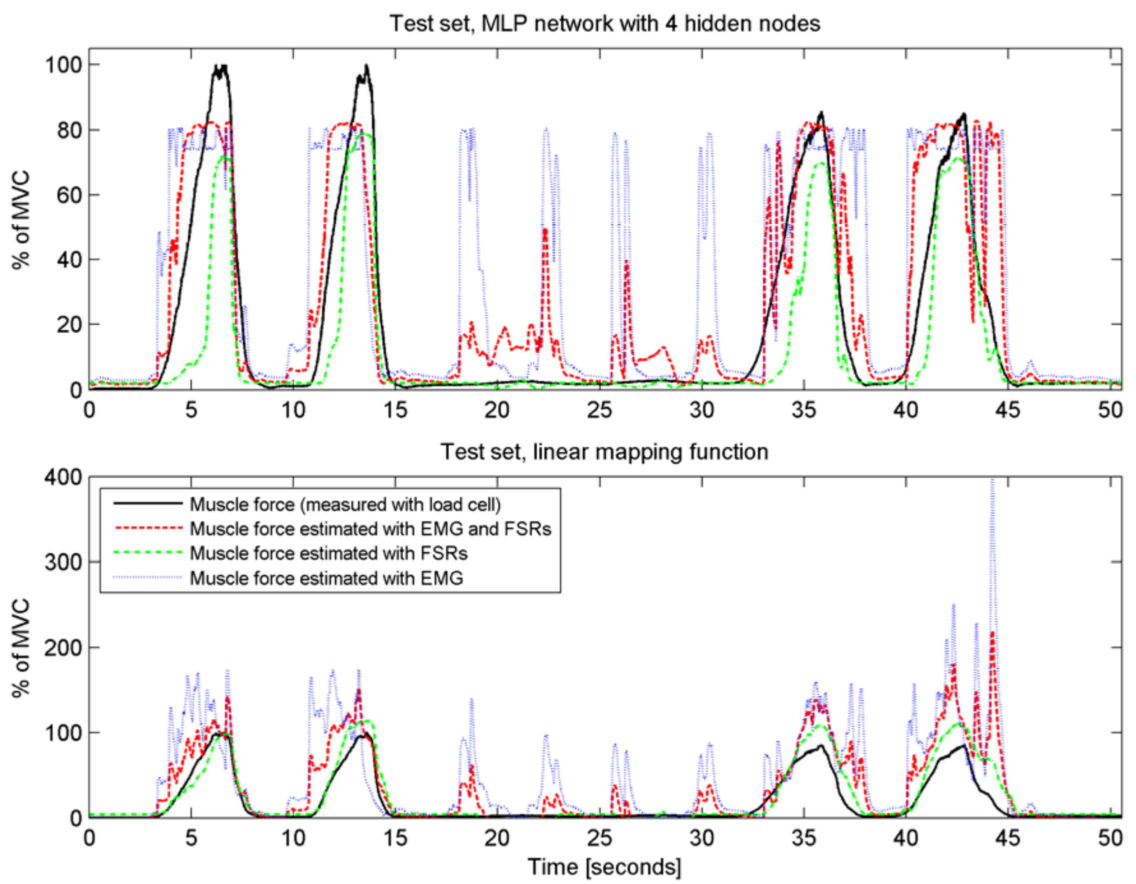

Fig. 3: Estimation results for three different test set inputs. Estimation using an MLP with 4 hidden nodes and a linear mapping function. Note different vertical scales; unit is percent of maximum voluntary contraction (MVC).

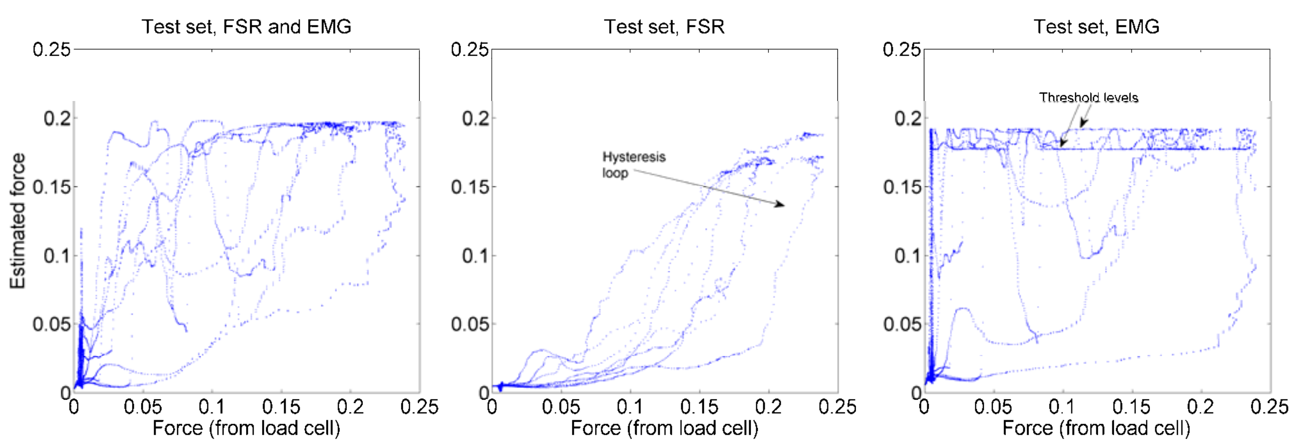

Fig. 4: Measured vs estimated force using an MLP with 4 hidden nodes. Same data set as in Fig. 3.

DISCUSSION

The results indicate that four hidden MLP nodes is a sufficient number to discriminate forces, as no improvement can be seen when increasing the MLP size beyond this point. The optimal MLP performed better than a linear estimator except when basing the estimate on FSR measurements alone, in which case the two techniques were equally successful.

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The quadratic estimator was fitted to the training set without any validation. The results indicate that this has caused "overtraining" with respect to the training data, as evident from the estimator's inferior performance when subjected to the test set (cf. the caption of Fig. 5).

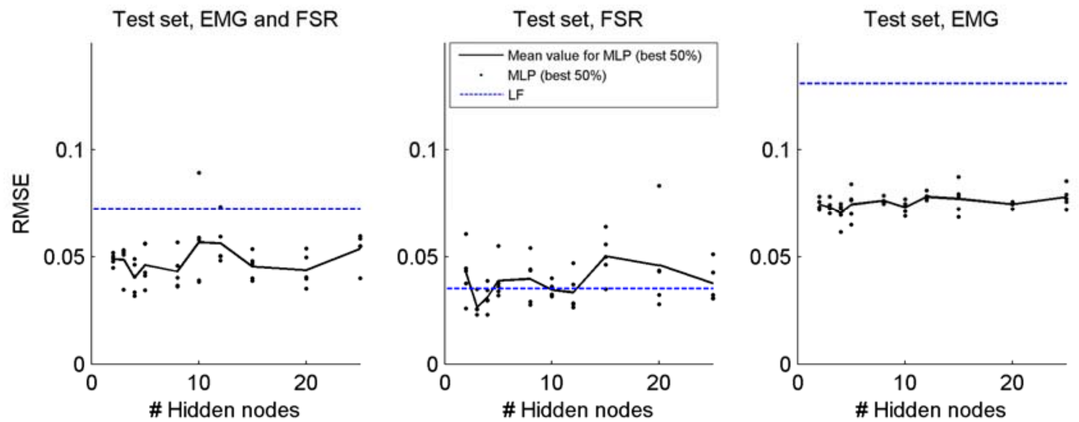

Fig. 5: RMS error rates for the same data set as in Fig. 3-Fig. 4. Corresponding values for the quadratic mapping function are 0.269 (EMG and FSR), 0.164 (FSR) and 0.152 (EMG).

It is noted that in Fig. 3 and Fig. 4, the FSR based estimates exhibit little or no artifact from external forces, which is at first a little surprising. In Fig. 2, however, it can be seen that the pure disturbance (i.e. the middle two "peaks") cause an equal response in all three FSRs, while when the muscle actually contracts, the FSRs yield different signal levels. Consequently, the estimator is able to distinguish these two signal sources.

In the upper graph of Fig. 2, the processed sEMG exhibits a transient response to the disturbance. This suggests that an optimal contraction force estimator should have a dynamic aspect rather than a purely static mapping property like the ones investigated in this study.

The results presented here are of a preliminary nature, and future study will assess the techniques using prosthetic sockets, real users and different myoprocessors. For example, the performance of a multi-FSR array inside a socket must be investigated, as the contact forces in that case may be different from those of the taped-on setup used in this study.

CONCLUSION

Measurements of contact forces exhibit promising properties for reducing force induced artifacts in conjunction with prosthesis control. The relative importance of sEMG and force measurements remain uncertain, and should be addressed in future work.

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