Simultaneous and proportional myoelectric control

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***Abstract*— Electromyography (EMG) is widely used as input to control scheme of myoelectric prosthetics. However, EMG signals change with limb position and thus lowers the accuracy in classification.Inclusion of the use of Inertial Measurement Units (IMU) has proved to raise the accuracy in pattern recognition methods. However, pattern recognition methods provides only control of one degree of freedom (DoF) at a time, and are computational costly. This study propose to use the combination of EMG recordings and accelerometer data in a linear regression model to overcome the slower reaction time of pattern recognition systems and to enable a simultaneous and proportional control scheme. In this study recordings from four able-bodied subjects has been collected, performing four wrist movements in three different limb positions. The data is evaluated through principal component analysis (PCA) and processed/trained with a linear regression model to classify the hand movements. One regressor is trained for each hand movement, using EMG data as well as a combination of EMG data and IMU data. The regressors are tested in a real-time visual environment on PC, measuring time to complete a target-reaching task of eight targets. The performance of the regressors are compared between using the IMU data and not using IMU data to determine the effect of including IMU data.**

I. INTRODUCTION

In recent years the development of EMG controlled prosthesis have advanced due to an increased interest in the area as well as a higher demand of better control of this prosthesis.[?] In the early years most EMG prosthetics functioned by only controlling one DOF by *on-off control*, mostly by linking antagonistic muscles to one DOF. This along with *mode switching* provided users a way to control more than one DOF, but not in a simultaneous way. However, as demands would rise, more complex methods was introduced to the EMG scene, and proportional control was brought in with pattern recognition methods. This effectively enabled simultaneous control of more than one DOF, but gave rise to new problems; a wider range of control would give less accurate movements, and training the pattern recognition methods proved difficult, as the training could over-fit, causing extended use of the prosthetics to degrade in performance. [?]. It has been proved that regression techniques can be apply as a new mapping method to achieve simultaneous and proportional control of multiple DOFs[?]. However there are still difficulties when prosthesis perform outside the clinical training environment[?]. Fougner et al.[?] noticed that majority of studies only take in account one limb position which becomes a problem since muscles create muscle-synergies to perform movements. It has been demonstrated that the variations in limb positions can have an impact on the robustness of EMG pattern recognition. In order to overcome this problem it has been suggested to combine EMG data as well as IMU data in the training sesions of the regressor to obtain simultaneous and proportional control of EMG prosthesis. Simultaneous and proportional control of two DOF's of the wrist in different limb positions, can be achieve trough the use of linear regression as control system. Combining EMG and IMU's can minimize the limb position effect when using regression as control system.

II. METHODS

A. Experimental Setup

EMG data was collected from four able-bodied subjects (three males, one female). The subjects performed four different hand gestures. This study is only focus on two DOF, which are, flexion and extension, radial and ulnar deviation of the wrist. The order in the execution of the movements was the same for each subject. EMG signals were recorded with Myo armband, this device counts with eight medical grade stainless steel surface EMG sensors. Furthermore its nine axis IMU provides information about position and orientation of the arm. The Myo armband was positioned in the right forearm of the subjects (all subjects right handed). The procedure was performed in three different limb positions. In order to avoid shoukder fatigue a relaxation period was given between trials. The subjects were instructed not to move the fingers during the data acquisition. The process was performed in a standing position.

B. Preprocessing

For this study the EMG data were filtered using a second-order Butterworth high-pass filter, cutoff frequency (f_c=10Hz).

C. Feature extraction

The features were extracted creating a sliding-window of 40 samples with an overlapping of the 50%. Two different time domain features were extracted, Mean absolute value (MAV) as well as logarithmic variance. MAV represent the amplitud of the signal. It is defined as the average of the absolute values of the EMG signal and expresed as:

$$MAV = \frac{1}{N}\sum_{i=1}^{N}|x_i| \tag{1}$$

where N is the length of the signal, and x_i is the signal of i samples. The logarithmic variance is a nonlinear transformation of the variance applied to

$$log(\sigma^2) = log(\frac{\sum_{i=1}^{N}(x_i-\mu)^2}{N}) \tag{2}$$

where N expresses the length of the signal, x_i is the i^th sample of the signal and μ is the mean.

D. Regression models

The acquired data was used to train the four different regressors that had been implemented, one for each of the movements under study. Simple linear regression had been applied as is shown in 3:

$$Y_i = \alpha + \beta X_i + \epsilon_i \tag{3}$$

where, Y is the dependent variable or response, X is the independent variable or the predictor, β is the regression coefficient or the slope, and α is the Y intercept (predicted value of Y at $X = 0$), ϵ is the error and i is the index.

E. Separability of data

In order to evaluate the quality of the features extracted from the EMG signals, Principal Component Analysis (PCA) was applied. This analysis tool was performed for each movement in each limb position and represented in three dimensional space as shown in Through this tool is possible to see significant outliers or if the clusters formed by the features can be easily distinguishable. This was done to avoid inaccurate training of the regressors.

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Fig. 1. Inductance of oscillation winding on amorphous magnetic core versus DC bias magnetic field

F. Regressor accuracy

To meassure the accuracy of the regressor, Root Mean Squared Error (RMSE) was calculated. RMSE is a calculation of the standard deviation of the residuals, that is, the difference between the estimated and the actual values.

$$RMSE = \sqrt{\frac{\sum_{i=1}^{N}(y_i - \hat{y_i})^2}{N}} \tag{4}$$

Where N is the length of the signal, y_i is the i^th variable of the actual data and $\hat{y_i}$ is the i^th output of the regressor. The RMSE will be done for the regressor of each movement.

III. MATH

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$$\alpha + \beta = \chi \tag{1}$$

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TABLE I

AN EXAMPLE OF A TABLE

One	Two
Three	Four

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Fig. 2. Inductance of oscillation winding on amorphous magnetic core versus DC bias magnetic field

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A conclusion section is not required. Although a conclusion may review the main points of the paper, do not replicate the abstract as the conclusion. A conclusion might elaborate on the importance of the work or suggest applications and extensions.

APPENDIX

Appendixes should appear before the acknowledgment.

ACKNOWLEDGMENT

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References are important to the reader; therefore, each citation must be complete and correct. If at all possible, references should be commonly available publications.

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