A sEMG-based Real-time Adaptive Joint angle Estimation and Control for a Prosthetic Hand Prototype

CHANDRASEKHAR POTLURI, MADHAVI ANUGOLU, STEVE CHIU, D. SUBBARAM NAIDU, MARCO P. SCHOEN Measurement and Control Engineering Research Center, School of Engineering Idaho State University 921 South 8th Avenue, Stop 8060, Pocatello, Idaho USA

naiduds@isu.edu

Abstract: - This paper presents a novel approach of a surface electromyographic (sEMG)-based, real-time Model Reference Adaptive Control (MRAC) strategy for joint angle estimation and control of a prosthetic hand prototype. The proposed design is capable of decoding the pre-recorded sEMG signal as well as the sensory force feedback from the sensors to control the joint angle of the prosthetic hand prototype using a PIC 32MX360F512L microcontroller. The input sEMG signal is rectified, filtered using a Half-Gaussian filter and fed to a Kullback Information Criterion (KIC) based fusion algorithm, which give the overall estimated joint angle (position). This estimated angle is given as input to the prosthetic hand prototype. Then the MRAC along with a two stage proposed embedded design is used for the position control of the prosthetic hand. For the real-time operation, the measured angle data at the joint is fed back to the controller. The data is transmitted to the computer through a universal asynchronous receiver/transmitter (UART) interface of the proposed embedded design. The experimental results in real time show good performance in controlling the angle of the prosthetic hand prototype.

Key-Words: - Prosthetic Hand, sEMG, Data fusion, Kullback Information Criterion, MRAC, Angle control

1 Introduction

For 60 years, there has been active research going on in the area of Prostheses. Due to the unavailability of prosthetic devices with full human hand functionality at an affordable price, more than 30% of upper extremity amputees chose not to use a prosthetic device on regular basis [1, 2]. Besides that, most of the available prosthetic devices do not have tactile or proprioceptive feedback [3]. Previous researchers insisted that human-centered robots must be independent with a great functionality, comfort and ease of use [4]. In rehabilitation robotics, human-machine interface, which requires a natural means of communication, is an interesting domain [5]. The electromyographic (EMG) signal is the electric potential generated by muscles. There are two kinds of EMG signals: Intramuscular (needle) EMG and the Surface EMG (sEMG). Under the assumption that amputation is transradial and which does not require surgery, we relied on sEMG signals in this work whose amplitude ranges from -5 to +5 (mV). sEMG signals are acquired from the surface of the skin, as they penetrate through various tissue layers [6]. Thus, they tend to create cross-talk, interference and noise. The sEMG is a temporally and spatially composed signal [7].

Due to the complexity and nonlinearity of the human hand, the control of a multi-fingered prosthetic hand is a challenging task [8]. From the past research, it is clear that the most widespread approach for controlling the prosthetic hand is by using a position or force control [9]. The challenging task involved in this work is the prediction of the Proximal Inter Phalangeal (PIP) joint angle corresponding to the sEMG signal.

In this paper, we used a three-by-three array of sEMG sensors and a wheel potentiometer device to acquire the EMG signal and the corresponding PIP joint angle. The data from the array of sEMG sensors were fused using a probability based fusion algorithm in order to have a better estimate [10]. A real-time embedded MRAC control system with sensory feedback was implemented to accomplish the goal of this design. Considering the probability based fused joint angle estimate as input to the real-time control system, a two stage embedded platform with a simple MRAC strategy was chosen for the PIP joint angle control of the prosthetic hand prototype.



Fig.1: Experimental set-up.

Fig.1. illustrates the experimental design used in this work for capturing the sEMG signal and the corresponding joint angle of the PIP joint of the index finger. Prior to placing the sEMG sensors on the skin surface, the test subject was prepared according to ISEK standards [11]. The motor point was identified by using a wet probe muscle point stimulator (Richmar HV-1000). Nine DE-3.1 sEMG sensors of the DELSYS Bagnoli-16 EMG system were arranged on the skin surface in a three-by-three array. The middle row of sensors was placed on the motor point of the index finger. An angle measurement device was designed using a 10k-Ohm wheel potentiometer shown in Fig.1. It was used to measure the PIP joint angle. Force Sensitive Resistor (FSR) was mounted on the stress ball and was used to measure the force. All the data was acquired at a sampling rate of 2000 samples/sec.

3 Proposed Design

The objective of the proposed design is to track a joint angle of the prosthetic hand as closely as possible. Here, the joint angle signal is inferred from the sEMG signals obtained from the array of the sEMG sensors located at the arm.

Sensor fusion is done in the time domain for the sEMG data using a probability based Kullback information criterion (KIC) fusion algorithm. The data from the nine sensors were collected around the corresponding individual motor unit at the transradial arm location (flexor digitorum superficialis) and before fusing the sEMG signals are rectified and filtered using a Half -Gaussian filter, as given by (1).

$$p(EMG|x) = 2 \times \frac{\exp\left(-\frac{EMG^2}{2x^2}\right)}{\sqrt{2\pi x^2}},$$
(1)

where p(EMG|x) is a conditional probability density function, x is a latent driving signal.

A System Identification (SI) tool was used to determine the dynamic relationship between the input-output data. The sEMG signal was used as input and the corresponding PIP joint angle as output. In particular, nonlinear Wiener Hammerstein one-dimensional polynomial models were obtained for the sEMG/joint angle data.

The mathematical representation of the modeling is given by,

$$w(t) = f(u(t)) \tag{2}$$

$$x(t) = \frac{B_{i,j}(q)}{F_{i,j}(q)} w(t - nk) + e(t)$$
(3)

$$y(t) = h(x(t)) \tag{4}$$

where u(t) is the sEMG signal and y(t) is the PIP joint angle. f and h are nonlinear functions, w(t)and x(t) are internal variables, $B_{j,i}(q)$ and $F_{j,i}(q)$ are polynomials, q is the back shift operator, and e(t) is the output error. From (3), the WH model structure utilizes a linear OE model, which is given by,

$$y(t) = \frac{b(q)}{F(q)}u(t - nk) + e(t)$$
 (5)

where *nk* is the system delay and *t* is time index

The resulting nine WH models were fused together by using a probabilistic KIC algorithm [12]. It assigns a particular probability to each model [13]. The overall estimated model output and the individual models outputs were compared by using the Pearson's correlation coefficient.

From the previous research, it was shown that the KIC algorithm performs better than the other criteria [10, 12]. KIC is an asymmetric measure of the models' dissimilarity which can be obtained by the sum of the two directed divergences known as J-divergence or Kullback's symmetric [14]. It is given by

$$KIC(p_i) = \frac{n}{2} \log R_i + \frac{(p_i+1)n}{n-p_i-2} - n\psi\left(\frac{n-p_i}{2}\right) + g(n)$$
(6)

Where $g(n) = n \log(n/2)$, ψ - digamma function.

The probability based fusion algorithm given by [12, 13] is applied to the nonlinear models obtained by using an SI technique with the following steps:

1: The WH models are obtained by considering the sEMG data u(t) as input and joint angle Y as output,

2: Calculate the residual square norm i.e.,

$$R_i = Y - \Phi_i \hat{\theta}_i = ||Y - \hat{Y}||$$

where
$$\hat{\theta}_{i} = \left\{ \Phi_{i}^{T} \Phi_{i} \right\}^{-1} \Phi_{i}^{T} Y$$
, and

$$\Phi = \begin{bmatrix} Y^{T} & u_{p-1}^{T} & Y_{p-1}^{T} & u_{1}^{T} \\ Y_{p+1}^{T} & u_{p+1}^{T} & Y_{p}^{T} & u_{2}^{T} \\ \dots & \dots & \dots \\ Y_{n-1}^{T} & u_{n-1}^{T} & Y_{n-2}^{T} & u_{n-p}^{T} \end{bmatrix}$$

3: Compute the model criteria coefficients using (6).

4: The model probability can be computed using

$$p(M_{i}(t) \mid Z) = \frac{e^{-l_{i}}}{\sum_{j=1}^{k} e^{-l_{j}}}$$

5: The fused model output can be calculated using



Fig.2: Block diagram representation of Model Reference Adaptive Controller (MRAC)

A simple MRAC based scheme (Fig. 2) is utilized for compensating the dynamics of the prosthetic hand. During the development of the artificial hand, changes were being undertaken to the mechanical design and drive trains of the hand that affect the dynamics of the finger motion of the prosthesis. In addition, the uncertain characteristics of the kinematic and actuator interaction could have led to a different performance than expected. Therefore, a simple MRAC controller was developed in order to maintain some performance stability. For updating the controller parameter θ , the MIT rule was used.

$$\frac{d\theta}{dt} = -\gamma e Y_m \quad . \tag{7}$$

According to the MIT rule [15], the gain parameter γ was selected to achieve the desired performance. In this work γ was taken as 3.0 and

the error $e = Y - Y_m$ was calculated by the difference of the model reference output Y_m and the actual output (joint angle generated by the prosthetic hand). Different MRAC stability issues were addressed in [14, 15].

Reference Model Derivation

The dynamic equations of motion for the hand are obtained from the Lagrangian approach as [16]

$$\dot{\mathbf{x}}(t) = \begin{bmatrix} 0 & \mathbf{I} \\ 0 & 0 \end{bmatrix} \mathbf{x}(t) + \begin{bmatrix} 0 \\ \mathbf{I} \end{bmatrix} \mathbf{u}(t). \tag{8}$$

The control input vector u(t) was given by

$$u(t) = -M^{-1}(q(t)) [N(q(t), \dot{q}(t)) - \tau(t)].$$
(9)

As the prosthetic hand is required to track the desired angle profile $q_d(t)$ described under the reference angle model, the tracking error e(t) is defined as

$$e(t) = q_d(t) - q(t).$$
 (10)

Here, $q_d(t)$ is the desired angle vector of the joints and can be obtained by the reference angle model [17]; q(t) is the actual angle vector of the joints. Differentiating (10) twice, we get,

$$\dot{e}(t) = \dot{q}_d(t) - \dot{q}(t)$$
, $\ddot{e}(t) = \ddot{q}_d(t) - \ddot{q}(t)$. (11)

Substituting $\dot{q}(t)$ into (11) gives

$$\ddot{e}(t) = \ddot{q}_d(t) + M^{-1}(q(t))[N(q(t), \dot{q}(t)) - \tau(12)]$$

From (12) the control function u(t) can be defined as

$$u(t) = \ddot{q}_{d}(t) + M^{-1}(q(t))[N(q(t), \dot{q}(t)) - \tau(t)].$$
(13)

This is often called the feedback linearization control law, rewriting (13) as,

$$\tau(t) = M(q(t)) [\ddot{q}_d(t) - u(t) + N(q(t), \dot{q}(t))]$$
(14)

Using (11) and (13), the state vector $x(t) = [e'(t)\dot{e}'(t)]'$, the state-space model can be represented as

$$\dot{\mathbf{x}}(t) = \begin{bmatrix} 0 & \mathbf{I} \\ 0 & 0 \end{bmatrix} \mathbf{x}(t) + \begin{bmatrix} 0 \\ \mathbf{I} \end{bmatrix} \mathbf{u}(t).$$
(15)

Now, (15) is in the form of a linear system such as

$$\dot{x}(t) = Ax(t) + Bu(t)$$
. (16)

Implementation:

The proposed control design was implemented on a PIC 32MX360F512L microcontroller in two stages: "Signal Processing" and "Motor Actuation". The Signal Processing stage enables the execution and implementation of real-time control strategies. A dsPIC block set was used to generate the C code for the PIC 32 from Simulink[®]. The dsPIC block set generates a .hex file, and this file was imported in MPLAB[®] to program the PIC 32.

Signal Processing Stage:

From the many modules available on PIC 32, the following four modules were utilized for the implementation of the signal processing stage: Analog Input Module, Digital Output module, Output Compare module, and URAT module.

The sensory feedback joint angle data from the wheel potentiometer was acquired through the Analog Input module. Based on the selected control strategy for the motor actuation stage, the digital control signals were generated by the Digital Output module of the PIC 32. The changes in the reference/command signal were detected by this module. A pulse width modulated (PWM) wave with a specific duty cycle was generated by the Output Compare module based on the error. The angle data from the microcontroller to the PC was transmitted by the UART module via serial communication. In this particular design, the data was fed to the computer by a virtual com port via USB cable. The signals from the ports were read by MATLAB[®].

Motor Actuation Stage:

In this stage, the motor was actuated with a SN754410 quadruple half-H driver [18] with the corresponding control signal. This proposed design was tested on the index finger of a prosthetic hand prototype. Fig. 3 shows the test bed for the proposed design.





Mechanical structure of the robotic hand prototype:

The prosthetic hand prototype finger has three degrees of freedom stimulated by two Pololu 35:1 mini metal gear motors. Biologically inspired parallel actuation is the main characteristic of this robotic hand prototype. It is based on the behaviour/strength space of the Flexor Digitorum Profundus (FDP) and the Flexor Digitorum Superficialis (FDS) muscles [19]. Using a belt transmission system, the DC motor in the metacarpal phalange of the finger triggers the Proximal Inter Phalangeal (PIP) joint. It also drives the DIP (Distal Inter Phalangeal) joint. The Meta Carpo Phalangeal (MCP) joint is actuated by the DC motor located at the base of the finger as shown in Fig. 4.



Fig. 4. Finger Actuation Scheme

4 Results and Discussion

Data was acquired from the microcontroller through UART channel2 of the PIC 32 micro controller by a virtual com port via USB at a 57600 baud rate. The data from the microcontroller was converted into unit16 data type before it is transmitted through the UART. It ran at an external clock frequency of 8MHz with internal scaling enabled. Fig. 5 depicts the experimental results of the proposed design. The actual angle output from the position sensor (i.e.*Y*)

closely follows the fusion algorithm estimated angle \hat{Y} . Fig. 6 shows the fusion estimated angle profile \hat{Y} and actual angle from the position sensor (Y) plotted for a separate experiment. Fig. 7 shows the repeated experimental results. The proposed control strategy was tracking angle profile and matching the actual angle profile with the model estimated angle (\hat{Y}).



Fig. 5. Fusion based angle estimate and actual angle from angle sensor during the grasp action.



Fig. 6. Fusion based angle estimate and actual angle from angle sensor during the grasp action (for experiment-2).



Fig. 7. Fusion based angle estimate and actual angle from angle sensor during the grasp action (for experiment-3).



While conducting the experiments, the following observations were made. The DC motors currently employed have the primary task of moving the prosthetic fingers. As the project is on-going, the Shape Memory Alloy (SMA) actuation scheme has not yet been implemented because of the slow response of the SMA's, and also because SMA's have a long relaxation time. Therefore it is difficult to track a randomly changing angle profile with a slowly responding actuation system. However, as the prosthetic hand prototype is designed to use the parallel actuation of these DC motors and SMA's, the DC motors alone cannot produce a fusion algorithm estimated angle profile.

In this work, the DC motors are solely responsible for motion actuation. Since the DC motors have small gear heads, the usual characteristic of gear driver actuation cores occurs: gear backlash. In addition, the DC motors employed were slow in responding to the changes in the force profile. Hence some gaps and delay were observed in the measured angle signal. However, the profile of the measured angle from the angle sensor has a similar pattern as the fusion algorithm estimated angle (\hat{Y}) . Thus we can conclude that apart from those mechanical transmission problems, the implemented control scheme produced promising results. These problems will be considered in a new prototype design that we are currently developing.

From the results, it is also evident from Figs. 5, 6 and 7 that the controller is tracking the slow changes in the angle profile very well, where as a small delay was observed while tracking the angle profile. In order to test the precision of the proposed control strategy, 15 different experiments were conducted. The mean of the Pearson's correlation coefficient for the fusion algorithm estimated angle (\hat{Y}) and the actual angle from the angle sensor (Y) in all the 15 experiments is 0.86. Because of the above mentioned transmission problems and the slow response of the gears, slight variability is observed in the correlation coefficients for the 15 experiments. Hence the difference in tracking the angle profile is observed in Figs. 6 and 7. Fig. 8 depicts the validation plot with a different fusion algorithm angle estimate \hat{Y} obtained by feeding a different sEMG.

5. Conclusion and Future Work

In this paper, based on sEMG signals, we proposed a two-stage real-time embedded MRAC strategy for a prosthetic hand prototype. The proposed design was based on tracking a reference angle profile and it gives good performance when tested on a prosthetic hand prototype. This design enables the transmission of the data from the microcontroller to the

computer. This design allows the control engineer to increase the accuracy and performance of the design by implementing various novel control strategies and also enables fast trouble shooting.

For future work, we are planning to use the above mentioned embedded platform to implement online model-based skeletal muscle angle estimation along with controller designs that address the above listed mechanical shortcomings. It will be valuable to acquire the sEMG signal directly from the arm of a healthy subject and transmit to our embedded system rather than using pre-recorded sEMG signals, which will be investigated as well in the future.

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