# OPTIMAL SITE SELECTION FOR PROSTHETIC CONTROL

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## Abstract

We describe a microcomputer based system for the selection of optimum surface electrode sites for the Liberty Mutual Boston Elbow. Our objective was to identify the two most differentiable electrode sites for flexion and extension control of the prosthesis. While the existing patient evaluation system for the Boston Elbow has been proven successful, it is based on a tedious and inexact process requiring the repeated positioning and evaluation of a single pair of electrodes. By employing two stationary electrode arrays to monitor flexion and extension activity over two broad spatial areas, and statistically analyzing the activity at these sites, a more efficient and reliable evaluation process is possible. Although our method provides an efficient and reliable evaluation of possible control sites for the Boston Elbow, we envision that it could be easily adapted to other prosthetic devices requiring differentiable motion control.

## Introduction

Electromyographic (EMG) signals are a manifestation of the electrical activity of muscles whenever a voluntary or involuntary muscle contraction occurs. The analysis of these signals over the past decade have demonstrated the considerable promise for, among other uses, the myoelectric control of prostheses [1,2]. However, few EMG controlled artificial limbs are available. In most cases, the failure of such systems may be attributable to certain design, economic, or industrial factors. Problem areas include unnatural or complex control schemes, poor system performance, development and prosthesis costs, complex manufacturing requirements, and the difficulties associated with the development and implementation of various evaluation procedures for artificial limbs.

Relative to the last problem area noted, the fundamental objective of our research was to develop a reliable and efficient patient evaluation process that could provide a relative measure of myoelectric activity over two broad spatial areas being considered for electrode placement. Because our research was based on the Boston Elbow prothesis, the bicep and tricep muscle groups were evaluated for possible flexion and extension control sites, respectively. In particular, the Boston Elbow employs two surface myoelectrodes to monitor EMG activity associated with flexion and extension control. The objective of electrode placement is to locate two control sites that provide reliable and differentiable signals for the Elbow's locomotion. For example, myoelectric signals from flexor muscles should not cause extension motions of the prosthesis. Ongoing efforts to improve the Elbow's performance have placed a greater emphasis on the optimum positioning of the sense electrodes and the methods used to determine these sites. The prototype system that emerged served to demonstrate the validity and promise of our method.

The current method for selecting the placement location for the Boston Elbow's flexion and extension electrodes is centered around the use of an instrument called a Myotester. The Myotester utilizes three analog meters which display the output potentials of the flexion sensing electrode, the extension sensing electrode, and the magnitude difference between the electrodes. The prosthetist manually repositions the electrodes in a trial-and-error evaluation procedure to identify the optimal control sites. Since identification of the optimal control sites is based on the prosthetist's judgment, proper fitting tends to be more of an art than a science. As will be described below, our system is basically a multiple electrode implementation of a Myotester with automatic comparison between all electrode pairs.

## Model for Myoelectric Activity

Myoelectric activity as seen by surface electrodes is a complex interference pattern representing a linear, spatial, and temporal summation of motor unit action potentials (MUAPs) [4]. The model shown in Figure 1 for EMG activity recorded with surface electrodes assumes that nonfatiguing, constant-force, isometric contractions are used to generate the EMG signals, and that the operator's intent, with respect to control of the prosthesis, can be uniquely determined by observing the proper muscle sites with surface electrodes [3,4]. As illustrated in this Figure, the myoelectric signal can be effectively



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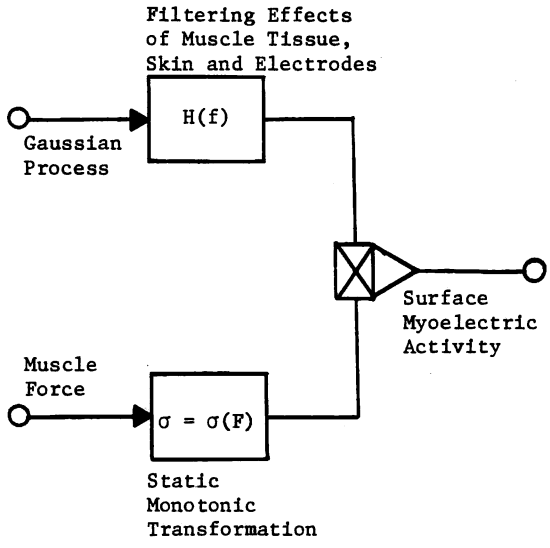


Figure 1 - Functional Model of Myoelectric Activity.[3]

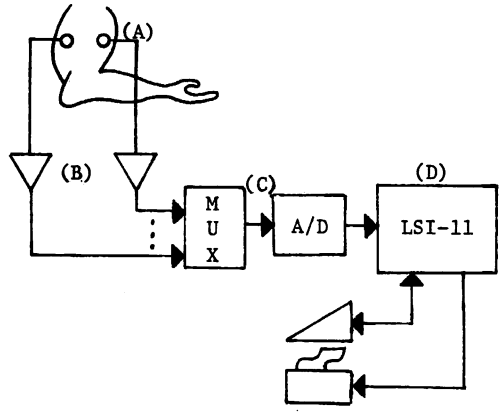


Figure 2 - System Organization. The EMG signals from potential flexion and extension electrodes (A) are amplified (B), multiplexed and sampled (C). The samples are used by an LSI-11 based user interactive (D) program to generate estimates of muscle forces for separate flexion and extension exercises.

represented as a Gaussian signal which is filtered by a linear, time-invariant filter and multiplied by a constant represented as a function of muscle force. Since a zero-mean Gaussian signal is completely characterized by its variance, the relationship between EMG activity and muscle force, as shown by the model, is through the variance. Consequently, a monotonic estimate of muscle force can be derived from the standard deviation of the recorded surface myoelectric activity. For our system's design, only the monotinicity of the function was important, not the functional relationship between the force and the resultant standard deviation of the signal.

## System Design

Referring to the system block diagram of Figure 2, our site selection process begins with the acquisition of EMG data from individual muscle points responsible for the lifting (flexion) and lowering (extension) function of the prosthesis. Arrays of electrodes are placed on the two muscle areas of interest, and the patient is instructed to perform muscle exercise trials. The generated EMG signals are sensed, amplified, and filtered by surface electrode units. The signals are then linearly amplified, multiplexed, and converted to a digital form.

The determination of the number of electrodes comprising an array is dependent upon the spatial accuracy desired. The denser the electrode packing, the greater the resulting resolution. For developmental purposes four flexion electrodes and four extension electrodes were employed. Each array is aligned in a square configuration.

The surface electrodes were developed by The Liberty Mutual Research Center. These electrodes sense myoelectric signals which have a characteristic amplitude of 25 to 1000 microvolts rms, and frequency components from 40 to 300 Hertz. To reduce ambient electrical phenomena interference, each electrode has a ground reference and two button electrode inputs to a differential amplifier with a CMRR of approximately 100 dB. An additional linear amplification stage was necessary to bring the EMG signals into the full analog-to-digital converter (ADC) voltage range, thus increasing sampling resolution.

A sampling rate of less than 100 Hz is used to obtain uncorrelated myoelectric samples necessary for computation of the standard deviation (force estimate) [5]. Following conversion, the data are ready for further processing by the site selection algorithm.

The site selection algorithm is based on an electrode by electrode comparison of the quasi-steady state force levels recorded at each of the flexion and extension electrodes. As noted above, the standard deviation (SD) of sampled myoelectric activity is a monotonic, nonlinear measure of muscle force, and thus constitutes a force level estimate.

SD force estimates are computed at all electrodes for separate levels of flexion and extension exercises. The force estimates from successive levels of exercises are multiplicatively combined to form a composite force estimate for each electrode, again with separate computations and results for flexion and extension exercises.

The multiplicative structure of the force averaging assures a consistently detectable signal at the site to be selected. Should the force estimate be particularly low at one electrode

relative to another, the former electrode's multiplicative force estimate will also be low relative to other more optimal sites.

Next, a Flexion Indicator Matrix and an Extension Indicator Matrix are formed. The elements of each matrix correspond to the ratio between each extension multiplicative force estimate and each flexion multiplicative force estimate. The numerator and denominator for each element is derived from, and corresponds to, the exercise type, i.e., flexion or extension multiplicative force estimate. The results are a measure of differentiability indicating the best flexion and extension array electrode pairs for separate flexion and extension exercises. This process is best illustrated in Figure 3.

Finally, a Composite Indicator Matrix is formed by multiplying the corresponding element of each indicator matrix. This final matrix of cross-electrode comparisons indicates the best electrode pair for myoelectric control of the prothesis (Figure 3).

## System Verification

Two square electrode arrays, each consisting of four button electrode configurations, [5] were located on the skin above the bicep and tricep muscles, and secured via an armband. The subject was seated with the arm relaxed and hanging laterally extended in front of the body. EMG signals were generated through nonfatiguing, constant-force, isometric contractions of the test arm, in an extended lateral position, against the free arm in both push and pull exercises. Bicep contractions were necessarily dominant in push exercises and tricep contractions in pull exercises.

A differential trial run consisted of the subject performing equivalent bicep and tricep dominant contractions at three arbitrary force levels. Additionally, in a co-contraction trial run the subject purposely displayed poor differential muscle control through simultaneous contraction of both the bicep and tricep muscles. Repeated differential and co-contraction trial runs were performed for a given placement of the electrode array; this constituted a trial set. The electrode array was then repositioned so that subsequent trial sets would facilitate the study of several bicep and tricep muscle areas.

An LSI-11 computer (11/02) with a data acquisition system (Data Translation DT2781) was used to develop and implement an algorithm corresponding to the analysis system shown in Figure 3. Results were printed on a DECwriter IV in the form of the Flexion Indicator Matrix, Extension Indicator Matrix, and Composite Indicator Matrix.

FIGURE 3. The following "flow chart" represents the signal processing scheme used to identify the optimal electrode pair from an array of flexion electrodes and extension electrodes. The text should be referenced for futher details.

1. $f_{ij}(n)$; $e_{ij}(n)$ = extension(e) or flexion(f) electrode sample for the $i^{th}$ electrode at the $j^{th}$ discrete isometric exercise level (flexion or extension exercise).

2. $\sigma^{e}_{ij}$; $\sigma^{f}_{ij}$ = muscle force estimate for a flexion (f) or extension(e) electrode. Computed from the formula:

$$\sigma^{\alpha}_{ij} = \left\{ \frac{1}{N} \sum_{n=1}^{N} \alpha^{2}_{ij}(n) \right\}^{1/2}$$

where $\alpha_{ij}$ is the $f_{ij}$ or $e_{ij}$ sample for the $n^{th}$ value.

3. Collect the $\sigma^{\alpha}_{ij}$ as matrices of separate measures of ext. and flex. muscle exercise level. These matrices are $\hat{\sigma}^{f}_{f}$, $\hat{\sigma}^{f}_{e}$, $\hat{\sigma}^{e}_{e}$, and $\hat{\sigma}^{e}_{f}$ where;

$$\hat{\sigma}^{\alpha}_{b} = \left[ \cdots \sigma^{\alpha}_{ij} \cdots \right]_{b} \quad \begin{array}{l} a = \text{for e electrode.} \\ b = \text{for e exercise.} \end{array}$$

4. Form vectors $\hat{\sigma}^{\alpha}_{b}(\pi)$ by multiplying terms across rows of $\hat{\sigma}^{\alpha}_{b}$; i.e.

$$\hat{\sigma}^{\alpha}_{b}(\pi) = \left[ \begin{array}{c} \prod_{j=1}^{J} \sigma^{\alpha}_{1j} \\ \vdots \\ \prod_{j=1}^{J} \sigma^{\alpha}_{Ij} \end{array} \right]_{b} = \left[ \begin{array}{c} \sigma^{\alpha}_{1}(\pi, b) \\ \vdots \\ \sigma^{\alpha}_{I}(\pi, b) \end{array} \right]$$

where: j = exercise number (max = J)
i = electrode number (max = I)

5. Form separate flexion ($R_f$) and extension ($R_e$) Indicator Matrices by forming force level estimate ratios from the $\sigma^{\alpha}_{b}(\pi)$ vectors; i.e.

$$R_f = \left[ \begin{array}{ccc} \frac{\sigma^{f}_{1}(\pi,f)}{\sigma^{e}_{1}(\pi,f)}, & \frac{\sigma^{f}_{2}(\pi,f)}{\sigma^{e}_{1}(\pi,f)}, \cdots & \frac{\sigma^{f}_{I}(\pi,f)}{\sigma^{e}_{1}(\pi,f)} \\ \vdots & & \vdots \\ \frac{\sigma^{f}_{1}(\pi,f)}{\sigma^{e}_{I}(\pi,f)}, & \cdots \cdots, & \frac{\sigma^{f}_{I}(\pi,f)}{\sigma^{e}_{I}(\pi,f)} \end{array} \right]$$

The ratios are force estimates between the flexion (numerator) and extension (denominator) electrodes for flexion exercises and extension (numerator) and flexion (denominator) electrodes for extension exercises. The $R_f$ matrices indicate the best electrode pair for separate flexion and extension force level detection.

6. The Composite Indicator Matrix (R) is formed by multiplying each element of $R_f$ by the corresponding element in $R_e$;

$$R = \left[ \begin{array}{ccc} \frac{\sigma^{f}_{1}(\pi,f)}{\sigma^{e}_{1}(\pi,f)} \times \frac{\sigma^{e}_{1}(\pi,e)}{\sigma^{f}_{1}(\pi,e)}, & \cdots, & \frac{\sigma^{f}_{I}(\pi,f)}{\sigma^{e}_{1}(\pi,f)} \times \frac{\sigma^{e}_{I}(\pi,e)}{\sigma^{f}_{I}(\pi,e)} \\ \vdots & & \vdots \\ \frac{\sigma^{f}_{1}(\pi,f)}{\sigma^{e}_{I}(\pi,f)} \times \frac{\sigma^{e}_{1}(\pi,e)}{\sigma^{f}_{I}(\pi,e)}, & \cdots, & \frac{\sigma^{f}_{I}(\pi,f)}{\sigma^{e}_{I}(\pi,f)} \times \frac{\sigma^{e}_{I}(\pi,e)}{\sigma^{f}_{I}(\pi,e)} \end{array} \right]$$

This matrix indicates the best electrode pair for combined flexion and extension force level detection.

## Results

Preliminary results have validated the site selection capabilities of the evaluation system. The results from the indicator matrices yielded an enumerative ranking of the relative utility of the 16 electrode pairs evaluated. However, due to the non-linear nature of the force level estimates, the matrices do not indicate to what extent an electrode pair's utility exceeds that of another. Initial qualitative results have demonstrated consistent and repeatable site ranking during the consecutive differential trial runs. This consistency was observed for several trial sets. The co-contraction trial runs simulated poor subject control over muscle differentiability. These trials produced Composite Indicator Matrices with homogeneous entries. Thus, co-contraction of the contrasting muscle areas correctly failed to identify any area as optimal for site selection. Each of these results were verified spatially (quantitatively) with respect to the evaluation system currently used, i.e., the Myotester.

## Discussion

The prototype system demonstrates the practical benefits of using estimation and comparison techniques for the simultaneous evaluation of several electrode pairs. However, further clinical work is needed to evaluate both the processor performance and the specific system implementation before a commercial version of this system is possible.

Although our investigation was a first-order approach towards a practical solution, a functional and readily adaptable system evolved through a careful consideration of the basic concepts involved. The present system provides an efficient and expandable method for determining the relative utility of various electrode configurations. While the prototype employed four electrodes in each of the site evaluation arrays, the selection software and hardware can be easily modified to accommodate as many electrodes as might ultimately be deemed necessary for commercial implementation. Furthermore, the use of the Fortran language for the computational software allows the application of our algorithm on computer systems which are typically available for use in this field. Only the subroutines employing system dependent file I/O operations and data acquisition should require attention when transferring the software package to a different computer system.

A basic restriction of our algorithm is its inability to spatially interpolate between electrodes within the arrays. While it would be possible to add an interpolation feature to the algorithm, the resultant tradeoffs in processing time and computer memory requirements deserve careful attention. If increased spatial resolution is found to be necessary, then a denser electrode array configuration may prove to be the more viable solution.

Although the SD muscle force estimates provide a reasonable indication of the relative muscle force levels at a recording site, further examination of force level estimates and statistical processing schemes is suggested. Since a relative measure of muscle force was desired in our system, a non-linear monotonic force estimate was sufficient. However, consideration should be given to a more precise force estimate which could be used to provide a real-time display of force levels for feedback during patient training. In addition, the statistical decision algorithm merits further investigation. Whereas our algorithm is based on a simple ratio test of hypotheses that is reminiscent of likelihood estimation, more mathmatically rigorous estimation techniques are available.

Finally, a thorough analysis of the typical contractions used to operate an EMG controlled prothesis should be conducted. With this information, it would be possible to develop a well-defined patient routine which closely emulates use of the artificial limb.

## Summary

Evidence obtained from the testing and evaluation of the prototype described indicates that this system shows great promise as a potential solution to the need for an optimal site selection process. The ability to simultaneously compare many possible electrode site pairs provides a reliable, easy, and efficient method for the bioelectric fitting of artificial limbs. The improvements in the fitting procedure will aid both the prosthetist and the patient through a more consistent and less time consuming process and, more importantly, may ultimately increase the rate of prosthetic acceptance by the amputee. Further developments of the patient evaluation system will be conducted by The Liberty Mutual Research Center.

## Acknowledgements

This work was the result of a required senior project at WPI and was supported by WPI and The Liberty Mutual Research Center. Our advisor for this project was Professor F.J. Looft.

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# Frontiers of Engineering and Computing in Health Care – 1983

## Proceedings – Fifth Annual Conference

## IEEE Engineering in Medicine and Biology Society

Hyatt-Regency Hotel, Columbus, OH • 10-12 September 1983

edited by
Glen C. Gerhard and W. Thomas Miller

**83CH1896-0**