Spring 2016

Assessment of optimized electrode configuration in Electrical Impedance Myography study using genetic algorithm via Finite Element Model

Somen Baidya

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ASSESSMENT OF OPTIMIZED ELECTRODE CONFIGURATION IN ELECTRICAL IMPEDANCE MYOGRAPHY STUDY USING GENETIC ALGORITHM VIA FINITE ELEMENT MODEL

by

SOMEN Bайдя

ABSTRACT

Electrical Impedance Myography (EIM) is a neurophysiologic technique in which high-frequency, low-intensity electrical current is applied via surface electrodes over a muscle or muscle group of interest and the resulting electrical parameters (resistance, reactance and phase) are analyzed to isolate diseased muscles from healthy ones. Beside muscle properties, some other anatomic and non-anatomic factors like muscle shape, subcutaneous fat (SF) thickness, inter-electrode distance, etc. also impact the major EIM parameters and thus affect the EIM analysis outcomes. The purpose of this study is to explore the effects of variation in some of these factors impose on EIM parameters and propose an optimum electrode configuration which is least affected by these anatomic and non-anatomic factors without compromising EIM’s ability to detect muscle conditions. In this study, genetic algorithm was applied as an optimization tool in order to find out an optimized electrode setup, which is less prone to these factors other than muscle properties. The results obtained suggest a particular arrangement of electrodes and minimization of electrode surface area to its practical limit, can overcome the effect of undesired factors on EIM parameters to a larger extent.

INDEX WORDS: EIM, Optimized Electrode Configuration, FEM, Genetic Algorithm.
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by

SOMEN BAIDYA

B.S., Bangladesh University of Engineering and Technology, Bangladesh, 2012

A thesis Submitted to the Graduate Faculty of the Georgia Southern University in the Partial
Fulfillment of the Requirements for the Degree of

MASTERS OF SCIENCE

STATESBORO, GEORGIA
ASSESSMENT OF OPTIMIZED ELECTRODE CONFIGURATION IN ELECTRICAL IMPEDANCE MYOGRAPHY STUDY USING GENETIC ALGORITHM VIA FINITE ELEMENT MODEL

by

SOMEN BAIDYA

Major Professor: Mohammad Ahad
Committee: Rocio Alba Flores
Danda B Rawat

Electronic Version Approved:
May 2016
DEDICATION

To my beloved family
ACKNOWLEDGEMENTS

First of all I express my profound gratitude to my thesis supervisor, Dr. Mohammad Ahad, Department of Electrical Engineering, Georgia Southern University for enlightening me about the utmost importance of bio-instrumentation. Without his continuous supervision, guidance, thoughtful suggestions and valuable advice, this work would never be possible.

I am grateful to all of those with whom I have had the pleasure to work with during this research. I also thank Khandokar Fazle Rabbi, former research assistant under Dr Ahad, Georgia Southern University. I am extremely grateful and indebted to him for his cooperation as a lab partner.

I also thank my parents and all family members for their encouragement and love during my course of work.

Lastly, I express my gratitude to GSU library. Many reference papers and books which are necessary for our thesis work have been issued from the library.
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<table>
<thead>
<tr>
<th>Symbol</th>
<th>Explanation</th>
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<tbody>
<tr>
<td>(R)</td>
<td>Resistance</td>
</tr>
<tr>
<td>(X)</td>
<td>Reactance</td>
</tr>
<tr>
<td>(\Theta)</td>
<td>Phase</td>
</tr>
<tr>
<td>(V)</td>
<td>Voltage</td>
</tr>
<tr>
<td>(I)</td>
<td>Electric Current</td>
</tr>
<tr>
<td>(Z)</td>
<td>Impedance</td>
</tr>
<tr>
<td>(\sigma)</td>
<td>Conductivity</td>
</tr>
<tr>
<td>(\epsilon)</td>
<td>Permittivity</td>
</tr>
<tr>
<td>(K)</td>
<td>Geometric factor</td>
</tr>
<tr>
<td>(d)</td>
<td>Distance between voltage electrodes</td>
</tr>
<tr>
<td>(A)</td>
<td>Cross-sectional area</td>
</tr>
<tr>
<td>(\Omega)</td>
<td>Applied frequency</td>
</tr>
<tr>
<td>(C)</td>
<td>Capacitance</td>
</tr>
<tr>
<td>(G)</td>
<td>Conductance</td>
</tr>
<tr>
<td>(\Re)</td>
<td>Real part of complex number</td>
</tr>
<tr>
<td>(Im)</td>
<td>Imaginary part of complex number</td>
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</table>
CHAPTER 1

1 Introduction

1.1 Neuromuscular Disease

The human body is comprised of three major muscle types, which are cardiac muscle, smooth muscle and skeletal muscle, the latter being responsible for 40% of our body mass. Skeletal muscles are responsible for the voluntary control, with some exceptions, like the tongue and pili arrector muscle in the dermis. Neuromuscular disorder is a common term that is used to classify many different syndromes and diseases that either directly or indirectly hamper the function of the skeletal muscles. More than a million people in the United States are affected by some form of neuromuscular disease and 40% of them are under age 18. Almost all neuromuscular disease is progressive in nature and all result in muscle weakness and fatigue. Although muscle wearing is not painful, the resultant weakness can cause cramping, stiffness, joint deformities and sometimes the tightening and freezing of joints. Congenital Muscular Dystrophy (CMD), Dystrophinopathies, Duchenne Muscular Dystrophy (DMD), Becker Muscular Dystrophy (BMD), Facioscapulohumeral Muscular Dystrophy (FSHD), Spinal Muscular Atrophy (SMA), Amyotrophic Lateral Sclerosis (ALS) are some of the very common forms of neuromuscular disorders that have been encountered by the physicians so far.

1.2 Diagnosing Neuromuscular Disease

During the past century, researchers have developed a few effective means to diagnose neuromuscular diseases. In many cases, diagnosing a neuromuscular dystrophy effectively can
involve more than a single test. Below is a brief description of some of the most commonly used diagnosing tools by the physicians for neuromuscular disease detection.

1.2.1  **Muscle Biopsy**

Muscle biopsy is a procedure that involves estimation of muscle proteins to ensure the presence of neuromuscular disease. Currently, two types of biopsies are present, namely- *Needle biopsy*, which involves inserting a needle into the muscle and obtaining tissue sample from the tip of the needle and *open biopsy*, which involves performing a minor surgical operation under local anesthesia to remove a small sample of muscle tissue from the region of interest (Pfenninger & Fowler, 2010). The major drawback of this method is obviously its invasive nature. Besides, although it can detect the presence of muscular dystrophy, it cannot classify the exact disorder type.

1.2.2  **Electromyography (EMG)**

Electromyography (EMG) is a technique that exploits the fact that voluntary muscle activities are controlled by the electrical impulses generated from the brain. Observation of these signals can help identify neuromuscular diseases since they impair the electrical signal conduction. Two types of EMG techniques are currently in practice: *needle EMG*, which involves inserting a needle electrode directly into the muscle to record the electrical activity; and *surface EMG* involves placing the electrodes on the skin. The technique allows the observer to see muscle energy at rest and changing continuously over the course of a movement. The limitation of EMG technique is it requires lot of preparation of the acquired signal and vulnerable to the phenomenon known as “cross talk” in which energy from one muscle group travels over into the recording field of another muscle group (Criswell, 2010).
1.2.3. **Nerve Conduction Velocity (NCV) Test**

NCV, also known as a nerve conduction study or NCS, is a technique that is associated with the concept of sEMG. In case of NCS one of the electrodes is stimulated by external electrical impulse while the other electrode placed at a distance along the limb from the first electrode measures the speed of impulse transmission by determining the time needed for the electrical signal to pass from one electrode to another (Griggs, Jozefowicz, & Aminoff, 2011). A healthy nerve signal travels at speeds of up to 120 miles per hour. A decrease in the speed of nerve conduction indicates a nerve disease.

![Hand NCS Setup](image)

*Figure 1: Hand NCS Setup (Diagnosis and Assessment of Neuromuscular Diseases, 2016)*

1.2.4 **Electrical Impedance Myography (EIM):**

Electrical Impedance Myography (EIM) is a tool that is yet to be established for clinical use. But, regarding the potential it possesses, it can overcome all the shortcomings of the existing diagnosis tools used in neuromuscular disease detection. It is a non-invasive four electrode measurement tool that works in the same way as NCV test, but, instead of measuring the velocity, it measures
the very fundamental response of an alternating current - impedance. EIM technique accentuates on the anisotropic property of muscle. Muscle anisotropy changes with neuromuscular disease. Changes in the property are also highlighted by the EIM parameters encountered.

1.3 Background of this study:

Because of unreliability and painful approach of current techniques to assess and diagnose neuromuscular disease, researchers have come up with the idea of EIM in order to provide a reliable, quantitative and relatively painless diagnosis tool (Shiffman, Aaron, & Rutkove, 2012). A high frequency low intensity current is injected inside the muscle or muscle group of interest via one of the corner surface electrodes while the other corner electrode acts as a ground. Assessing the voltage difference between two middle electrodes, the major EIM parameters, i.e. resistance, reactance and phase are obtained and analyzed for neuromuscular disease detection (Esper, Shiffman, Aaron, Lee, & Rutkove, 2006). Besides being prone to change in muscle properties, because of its non-invasive nature, EIM parameters also vary significantly with change in muscle thickness, subcutaneous fat thickness, electrode alignment over the muscle region, and also on the inter-electrode distance (Jafarpoor M., Li, White, & Rutkove, 2013).

1.4 Objective of the research:

This study concentrates on the application and development of EIM technology in neuromuscular disease detection. Being a non-invasive measurement tool, EIM also incorporates the electrical property of other body tissues beside muscle anisotropy. EIM parameters also vary significantly
based on the subcutaneous fat thickness and muscle thickness variation. The electrode setup and area of the electrode used also affect the EIM parameters. Because of these reasons, variation in EIM parameters do not necessarily indicate the presence of neuromuscular disorder. Researchers over the world are trying to propose a more generic idea of measurement that can either be a parameter that only reflects the muscle anisotropy or a specific technique of measurement that only highlights the muscle structure, so that regardless of the changes in other anatomic or non-anatomic factors EIM parameters are fixed over the frequency range and only prone to changes in muscle property. Our study also concentrates on the same focus. In this study, taking advantage of the fact that EIM parameters vary with the electrode configuration and setup, we propose an optimized electrode configuration that can be used as a generic procedure to determine neuromuscular disease detection using EIM. In summary, this study is solely concentrated on the development of EIM technique in order to make it more user friendly and appropriate for clinical use. For that purpose, the following goals were set for the research

- Design a 3-D model of human upper arm for FEM study
- Compare the outcome of different electrode shape and electrode setup to propose an optimized electrode configuration
- Application of Genetic algorithm techniques to propose an optimized electrode setup to EIM less affected by the anatomic factors other than muscle
1.5 Outline of the Thesis:

**Chapter 2** Elaborately covers the literature review of this research. It explains the EIM fundamentals, experiment done on rat to predict human outcome, FEM model of rat leg limb and analysis of the data obtained from previous work.

**Chapter 3** Outlines the methods of this study. It mainly focuses on the Comsol software simulation and the experiment on human subject to validate the simulation.

**Chapter 4** Discusses the results yielded from the FEM model, EIM experiment, proposed electrode configuration from genetic algorithm (ga) technique.

**Chapter 5** Final conclusion on the current study and based on the shortcomings it will provide guidance for future studies.
CHAPTER 2

2 Literature Review

2.1 Electrical Impedance Myography (EIM) Experiment

Human civilization was first introduced to the concept of Electrophysiology by the discovery of Luigi Galvani’s experiment on a frog muscle to make it contract all by passing electricity through it during the 1780’s. In the late 19th and 20th century, researchers became more interested in measuring the electrical properties of tissues rather than exploring its self-generating electrical characteristics. The basic idea of Electrical Impedance Myography is very much adapted from “Bioelectrical Impedance Measurement” (BIA), which was only confined to nutritional assessment before 1950s (Rutkove, 2009).

Like all other impedance methods, EIM relies on the basic concept of alternating current application to a substance. As current passes through the substance, it loses energy dissipated to overcome the substance’s inherent resistance it encompasses along the path. Loss of electrical energy results in lower amplitude of the applied current and can also introduce a phase shift in the later depending on the composition of the substance. The drop in amplitude is proportional to the resistance or impedance of the substance and can be expressed by the Ohm’s law:

\[ V = IZ \]

(1)

Where \( V \) is voltage, \( I \) is current flow and \( Z \) is the impedance. Measured complex impedance from the experiment can be written as

\[ Z = R + j (X_L - X_C) \]

(2)
However, $X_L$ is considered to play a minimal role in standard biomedical measurements (Rutkove, 2009). We can also assess the phase of the resulting current from the relationship

$$\theta = \tan^{-1}\frac{X}{R}$$  \hspace{1cm} (3)

Thus, when discussing bioimpedance measurements, one of several values can be assessed: the resistance ($R$), reactance ($X$), both of which are measured in ohms, or their combination, either as the impedance ($Z$) (also measured in ohms) or as the phase angle ($\theta$), measured in degrees (Rutkove, 2009).

The basic concept of biological impedance can be described more clearly by the simplified three element model of (Rutkove, 2009). Figure 2 illustrates the three model with two resistances—depicting the intracellular and extracellular matrices and a capacitor, consisting of the lipid bilayer that makes up the cell membranes.

![Figure 2: The “standard” basic equivalent circuit of bioelectrical impedance (Rutkove, 2009)](image)

Under alternating current excitation at high frequencies, the intracellular branch will become more conductive because of the presence of the capacitor. Lower frequency injecting current would not have the capacity to go through the intracellular resistance. So, the deviation in muscle composition and architecture due to this anomaly can be easily assessed by measuring the
impedance of biological tissue and comparing it to the normal value (Rutkove, 2009). This is the fundamental theory behind the application of EIM.

In case of EIM, the muscles internal electrical property, conductivity and relative permittivity depend on conductance, capacitance and geometric factor. The conductivity, $\sigma = K \cdot G$ and the relative permittivity, $\epsilon_r = \frac{K \cdot C}{\epsilon}$ comprises the conductance and capacitance of body tissues according to the relationships

$$G = \frac{R}{R^2 + X^2} \quad \text{and} \quad C = \frac{X}{(R^2 + X^2) \omega}$$

(4)

Where, K is the geometry factor defined by $K = \frac{d}{A}$; $d$ being the distance between the voltage electrodes, $A$ being the cross sectional area of the muscle and $R,X, \epsilon$ is the measured resistance, reactance and relative permittivity at a given frequency $\omega$.

Now, skeletal muscle also possesses a unique feature of anisotropy or directional dependence to current flow (Rush, 1962). Because of the cylindrical shapes of the myocytes constructing skeletal muscle, current flows much more easily along the muscle fibres (Rutkove, 2009). That’s why conductivities for muscle are reported both longitudinally to (parallel to) and transversely to (perpendicular to) the major muscle fiber direction (Gabriel, Lau, & Gabriel, 1996).

EIM was first introduced as a potential tool for neuromuscular evaluation by Seward B Rutkove in 2002. Numerous studies have been conducted afterwards to determine its vulnerability to associated factors, its applicability as a clinical tool and its advantage as well as its shortcoming. EIM experiment can be done in both single frequency and multi frequency domains depending on the objective of the study. In case of single frequency experimentation, most of the studies were done in 50 kHz because the degree of reproducibility of 50 kHz linear-EIM substantially exceeds what is found in other electrophysiological tests of muscle and nerve (Rutkove, Lee, Shiffman, & Aaron, 2006). Multi frequency experiments were conducted in certain range from couple of Hz to
2 to 4 MHz. In most of the cases, the parameters obtained in the MHz range was not considered since at this range of frequency the intracellular matrices gets superconductive and produce erroneous results. EIM have been proved to be successful in detecting numerous neuromuscular disease like ALS, axonal loss injury, radiculopathy and myositis. Early experiments were conducted on animal subjects i.e. rats (Ahad & Rutkove, 2010). Because of unavailability of human tissue, we will be incorporating the muscle conductivity and permittivity from rat study (Wang, et al., 2011) in this study.

Table 1: Conductivity and Permittivity values of rat muscle at selected frequencies (Wang, et al., 2011)

<table>
<thead>
<tr>
<th>Normal</th>
<th>Freq.</th>
<th>$\sigma_L$ (S/m)</th>
<th>$\sigma_T$ (S/m)</th>
<th>$\varepsilon_L$</th>
<th>$\varepsilon_T$</th>
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<td>10000</td>
<td>0.39±0.06</td>
<td>0.12±0.01</td>
<td>146877±16526</td>
<td>94162±8158</td>
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<td>25000</td>
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<td>96105±13335</td>
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<td>50000</td>
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<td>100000</td>
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<td>300000</td>
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<table>
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<th>Acute Crush</th>
<th>Freq.</th>
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<th>$\sigma_T$ (S/m)</th>
<th>$\varepsilon_L$</th>
<th>$\varepsilon_T$</th>
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<td>41997±2814</td>
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100000  0.64±0.041  0.27±0.013  2639±1715  3612±849
150000  0.68±0.043  0.33±0.013  1987±1469  2867±722
300000  0.75±0.046  0.46±0.014  1227±1072  1751±564

<table>
<thead>
<tr>
<th>Freq.</th>
<th>$\sigma_L$ (S/m)</th>
<th>$\sigma_T$ (S/m)</th>
<th>$\epsilon_L$</th>
<th>$\epsilon_T$</th>
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<td>300000</td>
<td>0.82±0.076</td>
<td>0.53±0.047</td>
<td>143±16</td>
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Freq. is the frequency measured in Hz; $\sigma_L$ the longitudinal conductivity; $\sigma_T$ the transverse conductivity; $\epsilon_L$ the longitudinal relative permittivity; and $\epsilon_T$ the transverse permittivity.

Acute and chronic crush conditions are both sciatic crush surgery performed under anesthesia to regenerate the muscular atrophy. Due to unavailability of disease data this study incorporates the acute and chronic crush data for diseased muscle as shown in table 1 which enlists measured mean ± SEM 50 kHz longitudinal and transverse conductivities and permittivities for the three groups of animals.

### 2.2 Finite Element Method

Finite Element Method is a numerical approach for finding solutions to physical reality - formulated in a mathematical model, by subdividing the boundaries into smaller, simpler parts.
The solution to the partial differential equation of each of these small parts are then assembled to reach the final solution for the entire geometry (Süli, 2007). To determine the relationship between EIM measurements and underlying state of muscle, most ideal result would be obtained from actual tissue collection via biopsy, which is not possible for obvious ethical reason. Thus, finite element method serves as a useful tool to study EIM dependency on muscle geometry/volume (Ahad & Rutkove, 2010). The governing equation for FEM study is expressed mathematically by equation 5 which is gradient of the total current density $J_{total}$ along the model is zero.

$$\nabla J_{total} = \nabla \left( \sigma E + j\omega\varepsilon_0\varepsilon E \right) = 0$$  \hspace{1cm} (5)

Where, $\sigma$ is the conductivity and $E$ is the applied electric field.

Preliminary studies used MRI (Magnetic Resonance Imaging) to develop accurate finite element model in simulation domain and experiment was done on rat on primary basis. Average girth of three groups of subject was used as the backbone to construct the model. In two inflicted atrophied condition (acute and chronic crush), the reduced muscle girth was translated into a reduced volume of the muscle compartments, while keeping the other layers (i.e. skin, subcutaneous fat and bone) unchanged (Wang, et al., 2011). The model extended from the knee to ankle joint and consisted of a skin/subcutaneous fat layer, a fascia layer, two bones (tibia and fibula), and several regions of muscle: the biceps femoris (depicted in Fig. 3), the gastrocnemius–soleus complex, and the tibialis anterior (Wang, et al., 2011).
The electrode measurements were 3.5 mm x 18 mm individually with 4 mm inter-electrode separation. No inter-electrode capacitance or conductive impedance was considered for the simulation. The electrode surface on skin layer was assigned as a perfect conductor of electricity (Wang, et al., 2011). The normal component between two adjoining layers was assumed to be continuous and no loss of energy was considered in form of current flow through any exterior boundaries. Dielectric properties of skin, subcutaneous fat and bone were assumed to be isotropic and were obtained from Gabriel’s study (Gabriel, Lau, & Gabriel, 1996). Muscle dielectric properties were obtained by biopsy.

Obtained values from this simulation was compared with the EIM measurements for 50 kHz for establishing the viability of FEM study as shown in table 2. Here \( R, X \) and \( \Theta \) is the resistance, reactance and phase of EIM measurements at 50 kHz.
Table 2: Comparison between EIM measured values and FEM model prediction at 50 kHz

**EIM**

<table>
<thead>
<tr>
<th>Value</th>
<th>Normal</th>
<th>Acute</th>
<th>Chronic</th>
</tr>
</thead>
<tbody>
<tr>
<td>$R_{50}$</td>
<td>73.3±0.69</td>
<td>74.2±1.9</td>
<td>72.7±3.2</td>
</tr>
<tr>
<td>$X_{50}$</td>
<td>22.8±0.02</td>
<td>18.2±0.6</td>
<td>21.2±1.2</td>
</tr>
<tr>
<td>$\Theta_{50}$</td>
<td>17.3±0.3</td>
<td>13.8±0.7</td>
<td>16.1±0.5</td>
</tr>
</tbody>
</table>

**FEM**

<table>
<thead>
<tr>
<th>Value</th>
<th>Normal</th>
<th>Acute</th>
<th>Chronic</th>
</tr>
</thead>
<tbody>
<tr>
<td>$R_{50}$</td>
<td>86.7</td>
<td>87.8</td>
<td>78.3</td>
</tr>
<tr>
<td>$X_{50}$</td>
<td>36.7</td>
<td>29.9</td>
<td>30.3</td>
</tr>
<tr>
<td>$\Theta_{50}$</td>
<td>24.3</td>
<td>18.8</td>
<td>21.2</td>
</tr>
</tbody>
</table>

Multi-frequency spectrum is also plotted using FEM and the dielectric data obtained from the experiment. The frequency range selected for this experiment was from 10 kHz to 4 MHz. Frequency analysis is performed over all the three cases: normal, acute, and chronic. Here the surface impedance values of the chronic group and the normal group are almost similar (Wang L. L., et al., 2011). To summarize the spectrum nature, collapsed parameters (logRslope, the
reactance slope and the phase slope) is also utilized (Wang L. L., et al., 2011). Here predicted values are shown just below the measured value for all three normal, acute and chronic conditions. It is clear from this figure that FEM predicted outcomes are parallel to EIM measurements, except for the case of 50 kHz reactance for the chronic crush and phase slope for the acute crush animals (Wang L. L., et al., 2011). As prominent from Fig. 4 and table 2, by using approximated muscle geometry and immediately postmortem dielectric values, finite-element analysis predicts to
reasonable extent changes in the surface-acquired EIM data. The actual values differed by about 15% for resistance, 60% for reactance and 30% for phase at 50 kHz (Wang L. L., et al., 2011). Inspired by the previous study, a more concentrated study was done by using only FEM to figure the dependency of EIM parameters on subcutaneous fat thickness, muscle thickness and electrode separation variation. Figure 5 depicts the FEM model used in the study (Jafarpoor M., Li, White, & Rutkove, 2013).

The study concluded on three major discoveries in the field of EIM. Firstly, of the three major parameters considered, resistance, reactance and phase; reactance appears to be least affected by alterations in geometric factors. Secondly, keeping the sense electrodes 30mm apart from each other, a separation of 80 mm between the excitation electrodes provides minimum variation in reactance with respect to variation in muscle size and subcutaneous fat thickness (Jafarpoor M., Li, White, & Rutkove, 2013). Finally, muscle conductivity is not affected significantly by the inter-electrode separation. The effect of subcutaneous fat thickness, muscle size and inter-electrode

Figure 5: FEM model of the human arm using Comsol Multiphysics 4.2a based on anatomic data. Inter-electrode spacing 15-30-15 mm (Jafarpoor M., Li, White, & Rutkove, 2013)
distance variation in EIM measurements is illustrated in Figure 6, 7 and 8 correspondingly as assessed by the study.

Figure 6: The effect of altering the thickness of muscle on the measured impedance parameters using the electrode configuration and spacing 15mm-30mm-15mm (Jafarpoor M., Li, White, & Rutkove, 2013).

Figure 7: The effect of altering the baseline 4.4 mm subcutaneous fat thickness on the resistance, reactance and phase. Thicknesses were input in a range from 2.2 mm to 17.6 mm using the electrode arrangement and spacing 15mm-30mm-15mm (Jafarpoor M., Li, White, & Rutkove, 2013).

Figure 8: The effect of inter-electrode distance on the measured 35 kHz resistance, reactance and phase values for several different thicknesses of muscles relative to baseline (1.0) (Jafarpoor M., Li, White, & Rutkove, 2013).
CHAPTER 3

3 Methodology

The objective of this study is to propose the optimized electrode configuration for which the effect of any geometric feature change in EIM measurement can be minimized. Previous discussion establishes that, FEM can be utilized to predict practical EIM results to some extent of accuracy. For convenience, the first part of this study was to design and analyze an FEM model of human upper arm that can reproduce the practical EIM measurements and the findings of the previous study to check its viability. Secondly, the study concentrates on the application of genetic algorithm to propose the desired optimized electrode configuration. For a better implementation of EIM, the study also focuses on how to ease the EIM diagnosis procedure by the application of artificial neural network.

3.1 Finite Element Method

There are several preferable software to do FEM study. However, for this study the model was created and simulated using COMSOL Multiphysics 4.3a. The FEM model was design based on the cross sectional view of human upper arm as shown in figure 9. The model was assumed to be symmetrical on both side of the bone marrow. Incorporating bicep brachii, triceps brachii, brachialis was considered a single domain since all skeletal muscle constitute the same dielectric properties. Also, the skin and fat thickness was distributed homogenously along the model. Considering only the major contributing body tissues, the model consists of four distinct layers-skin, subcutaneous fat, muscle and bone. Figure 10 depicts the 3D view of the basic model used
in this study. The model dimension was changed time to time to meet up the requirements of proposed hypothesis.

Figure 9: Cross Section through Middle Upper Arm (Cross Section Through Middle Upper Arm | ClipArt ETC, n.d.)

Figure 10: FEM model of the human upper arm using Comsol Multiphysics 4.2a (elbow to axilla) based on anatomic data. The inter-electrode spacing was 15 mm-30 mm-15 mm (60 mm in total)
For this study we have only considered the variation in muscle thickness and fat thickness. The top layer of the model skin layer was considered 3mm throughout the study as suggested as the average skin thickness of normal human. All the inter-electrode distances considered in this study was measured from edge to edge. No inter-electrode capacitances or contact impedances were included (Jafarpoor M., Li, White, & Rutkove, 2013). The electrodes conductivity and relative permittivity value were set to 5.0e5 S/m and 1.0 respectively to depict the situation of a perfect conductor. The muscle and tissue material properties were homogeneous throughout the model. For non-electrode boundaries, the normal component of the electric current was assumed to be continuous (Ahad & Rutkove, 2009). Electrodes were modelled as potential surfaces the boundaries of which had either the excitation or zero current, except for the ground electrode, the potential of which was fixed at zero volts (Jafarpoor M., Li, White, & Rutkove, 2013). The discretization mesh was generated automatically with the Comsol software. At each measured frequency, longitudinal and transverse conductivities and permittivities were obtained from the rat studies and were incorporated into the model with rat data substituting for the normal human muscle. Fat, cortical bone and marrow were obtained over the frequency spectrum from Gabriel’s dielectric survey (Gabriel, Lau, & Gabriel, 1996). The skin-subcutaneous fat, cortical bone and marrow were all assumed to be isotropic (Jafarpoor M., Li, White, & Rutkove, 2013). For multi-frequency measurements, this study only incorporates four major frequencies.
<table>
<thead>
<tr>
<th>Frequency</th>
<th>Tissue</th>
<th>Conductivity (S)</th>
<th>Relative Permittivity</th>
<th>Conductivity (S)</th>
<th>Relative Permittivity</th>
<th>Conductivity (S)</th>
<th>Relative Permittivity</th>
</tr>
</thead>
<tbody>
<tr>
<td>10k</td>
<td>Skin</td>
<td>0.0002</td>
<td>1150</td>
<td>0.0002</td>
<td>1150</td>
<td>0.0002</td>
<td>1150</td>
</tr>
<tr>
<td></td>
<td>Fat</td>
<td>0.025</td>
<td>1000</td>
<td>0.025</td>
<td>1000</td>
<td>0.025</td>
<td>1000</td>
</tr>
<tr>
<td></td>
<td>Muscle</td>
<td>(.4, .17, .17)</td>
<td>86e3</td>
<td>(.5, .14, .14)</td>
<td>104e3, 80e3</td>
<td>(.52, .15, .15)</td>
<td>126e3, 102e3</td>
</tr>
<tr>
<td></td>
<td>Bone</td>
<td>0.002</td>
<td>675</td>
<td>0.002</td>
<td>675</td>
<td>0.002</td>
<td>675</td>
</tr>
<tr>
<td>50k</td>
<td>Skin</td>
<td>0.0002</td>
<td>1150</td>
<td>0.0002</td>
<td>1150</td>
<td>0.0002</td>
<td>1150</td>
</tr>
<tr>
<td></td>
<td>Fat</td>
<td>0.03</td>
<td>500</td>
<td>0.03</td>
<td>500</td>
<td>0.03</td>
<td>500</td>
</tr>
<tr>
<td></td>
<td>Muscle</td>
<td>(.45, .2, .2)</td>
<td>55e3</td>
<td>(.58, .2, .2)</td>
<td>41e3, 48e3</td>
<td>(.65, .23, .23)</td>
<td>55e3, 59e3, 59e3</td>
</tr>
<tr>
<td></td>
<td>Bone</td>
<td>0.0035</td>
<td>300</td>
<td>0.0035</td>
<td>300</td>
<td>0.0035</td>
<td>300</td>
</tr>
<tr>
<td>100k</td>
<td>Skin</td>
<td>0.0002</td>
<td>1150</td>
<td>0.0002</td>
<td>1150</td>
<td>0.0002</td>
<td>1150</td>
</tr>
<tr>
<td></td>
<td>Fat</td>
<td>0.03</td>
<td>300</td>
<td>0.03</td>
<td>300</td>
<td>0.03</td>
<td>300</td>
</tr>
<tr>
<td></td>
<td>Muscle</td>
<td>(.55, .3, .3)</td>
<td>36e3</td>
<td>(.65, .2, .2)</td>
<td>26e3, 36e3</td>
<td>(.73, .32, .32)</td>
<td>35e3, 44e3, 44e3</td>
</tr>
<tr>
<td></td>
<td>Bone</td>
<td>0.0035</td>
<td>110</td>
<td>0.0035</td>
<td>110</td>
<td>0.0035</td>
<td>110</td>
</tr>
<tr>
<td>1M</td>
<td>Skin</td>
<td>0.02</td>
<td>990</td>
<td>0.02</td>
<td>990</td>
<td>0.02</td>
<td>990</td>
</tr>
<tr>
<td></td>
<td>Fat</td>
<td>0.05</td>
<td>150</td>
<td>0.05</td>
<td>150</td>
<td>0.05</td>
<td>150</td>
</tr>
<tr>
<td></td>
<td>Muscle</td>
<td>(.65, .4, .4)</td>
<td>4e3, 3e3, 3e3</td>
<td>(.8, .3, .3)</td>
<td>2e3, 1e3, 1e3</td>
<td>(.92, .44, .44)</td>
<td>3e3, 4e3, 4e3</td>
</tr>
<tr>
<td></td>
<td>Bone</td>
<td>0.004</td>
<td>40</td>
<td>0.004</td>
<td>40</td>
<td>0.004</td>
<td>40</td>
</tr>
</tbody>
</table>
3.2 Genetic Algorithm

To make EIM a more established tool for neuromuscular disease diagnosis, an optimized electrode configuration must be proposed that has the least variance in accordance to other anatomic or non-anatomic factors other than muscle electrical properties. In this study, we used the genetic algorithm as our optimization tool since it is one of the most effective means to find good solutions to the problems that are computationally intractable (Goldberg). Genetic algorithm imitates the selection process found in nature by creating a random population of samples at the beginning. Then it delivers a successor population by completing a process of fitness-based choice and recombination. During recombination, first generation samples are chosen and their genetic material is recombined to produce the second generation. This then goes into the next generation. In this process, a set of successive population evolves and the average fitness of the samples tends to converge to an optimized solution (McCall, 2005).

The fitness function used in this study was the slope of the linear regression equation obtained from the reactance at 50 kHz as a function of muscle thickness in normal condition. The population size was set 100, a number obtained from trial and error procedure so that, the solution space is more thoroughly searched and the algorithm doesn’t run too slow. Selection function for this specific study was chosen stochastic uniform since it samples all the solutions at evenly spaced intervals thus minimizing the probability to pick up a local minima rather than the global minima. The reproduction elite count was set to 2 and the crossover fraction was 0.8. The fitness function value differs slightly from the previous one with the same condition due to different correlation coefficient at times. Adaptive feasible is the appropriate type of mutation in this condition. Crossover function was set as arithmetic in which the next generation populations are as the weighted arithmetic mean of two parents.
In this study, the following options of the GA algorithm were specified:

<table>
<thead>
<tr>
<th><strong>Option</strong></th>
<th><strong>Parameter</strong></th>
<th><strong>Description</strong></th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Population</strong></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Population type</td>
<td>Double vector</td>
<td>Individuals in the population have type <em>double</em>.</td>
</tr>
<tr>
<td>Population size</td>
<td>20, 50, 100</td>
<td>Specifies how many individuals there are in each generation. Large population size searches more thoroughly but takes more time.</td>
</tr>
<tr>
<td>Creation function</td>
<td>Constraint dependent</td>
<td>creates the initial population for ga.</td>
</tr>
<tr>
<td>Initial population</td>
<td>Default</td>
<td>specifies an initial population for the genetic algorithm</td>
</tr>
<tr>
<td>Initial score</td>
<td>Default</td>
<td>specifies initial scores for the initial population</td>
</tr>
<tr>
<td>Initial range</td>
<td>Default</td>
<td>specifies the range of the vectors in the initial population</td>
</tr>
<tr>
<td><strong>Fitness Scaling</strong></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Scaling function</td>
<td>Rank</td>
<td>scales the raw scores based on the rank of each individual instead of its score</td>
</tr>
<tr>
<td><strong>Selection</strong></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Selection Function</td>
<td>Stochastic uniform</td>
<td>The algorithm moves along the line in steps of equal size. At each step, the</td>
</tr>
</tbody>
</table>
algorithm allocates a parent from the section it lands on

<table>
<thead>
<tr>
<th>Reproduction</th>
<th>Default</th>
<th>specifies how the genetic algorithm creates children for the next generation</th>
</tr>
</thead>
<tbody>
<tr>
<td>Mutation Function</td>
<td>Adaptive feasible</td>
<td>Randomly generates directions that are adaptive with respect to the last successful or unsuccessful generation. The mutation chooses a direction and step length that satisfies bounds and linear constraints.</td>
</tr>
<tr>
<td>Crossover</td>
<td>Scattered</td>
<td>specify how the genetic algorithm combines two individuals, or parents, to form a crossover child for the next generation</td>
</tr>
<tr>
<td>Stopping Criteria</td>
<td></td>
<td>The algorithm stops if the average relative change in the best fitness function value over Stall generations is less than to the Function tolerance.</td>
</tr>
</tbody>
</table>

Stall generations 10
CHAPTER 4

4 Results and Discussion

Figure 11 illustrates that the FEM model used in this study can predict the experimental EIM outcome to a certain degree. Figure 2 illustrates that, the FEM model predicts the practical results to some extents. Here, the model was designed to have 5mm skin fat and 51mm muscle thickness. The electrode separation between the sense electrodes were 30mm and between the excitation electrodes were 60mm. Only reactance is shown here, because as would be evident a bit later, that reactance is the parameter that can detect the neuromuscular disease in a more convincing manner.

At first, the variation in EIM parameters due to the affect other anatomic and non-anatomic factors. Initially, we have performed the study on varying subcutaneous fat and muscle thickness to observe the dependency of EIM parameters on these anatomic factors. The effect of subcutaneous fat thickness variation on EIM parameters is depicted in figure 12 and 13. As can be summarized
from the figures, increasing the subcutaneous fat thickness has a more prominent effect on resistance than reactance. This is because, the reactance depends on the muscle properties in a large scale. The isotropic skin and subcutaneous fat thickness comprises most of the contribution to resistance measurement. Since, base muscle thickness is fixed on both the case, the change in reactance is not that prominent as the change in resistance.

In case of large subcutaneous fat thickness, the reactance profile shows deviation from normal condition, particularly in very high frequency range. The explanation remains within the simplified circuit model of human body tissue. In case of very high frequency, both the extracellular and intracellular resistance become highly conductive. So, in case of small fat thickness the isotropic SF resistance is not that prominent as in case of larger SF thickness which also contribute to larger reactance value in high frequencies.
Figure 14 and 15 illustrates the variation in above mentioned EIM parameters with respect to muscle thickness alteration. Since, reactance depends on the muscle properties as stated earlier, the change in muscle thickness appears to have more prominent effect on percentage change of reactance than the effect of subcutaneous fat.

Figure 13: Variation in Reactance with change in fat thickness for (15mm-30mm-15mm) electrode spacing and 65mm x7mm rectangular electrode

Figure 14: Variation in resistance with change in muscle thickness for (15mm-30mm-15mm) electrode spacing and 65mm x7mm rectangular electrode
At first, the study was concentrated to explore the effect of SF thickness variation on different EIM parameters and propose a parameter which is least affected and also can detect muscle conditions.

To explore which EIM parameter is least affected by SF thickness variation, we have considered eight different parameters as listed in table 4. Plotting four major parameters results in figure 16.
Table 4: Variation in different parameters with subcutaneous fat thickness alteration

<table>
<thead>
<tr>
<th>SF Thickness (mm)</th>
<th>$R_{50kHz}$</th>
<th>$X_{50kHz}$</th>
<th>$\phi_{50kHz}$</th>
<th>$\phi \left( \frac{Z_{100kHz}}{Z_{50kHz}} \right)$</th>
<th>$\Re \left( \frac{Z_{100kHz}}{Z_{50kHz}} \right)$</th>
<th>$\Im \left( \frac{Z_{100kHz}}{Z_{50kHz}} \right)$</th>
<th>log $R$ Slope</th>
</tr>
</thead>
<tbody>
<tr>
<td>5</td>
<td>35.6</td>
<td>-17.52</td>
<td>-26.20</td>
<td>5.0594</td>
<td>0.7791</td>
<td>0.776</td>
<td>0.0687</td>
</tr>
<tr>
<td>6</td>
<td>39.98</td>
<td>-17.7</td>
<td>-23.88</td>
<td>5.7153</td>
<td>0.7975</td>
<td>0.7935</td>
<td>0.0794</td>
</tr>
<tr>
<td>7</td>
<td>45.7</td>
<td>-17.92</td>
<td>-21.41</td>
<td>5.9863</td>
<td>0.8195</td>
<td>0.8151</td>
<td>0.0855</td>
</tr>
<tr>
<td>8</td>
<td>52.87</td>
<td>-18.17</td>
<td>-18.96</td>
<td>5.847</td>
<td>0.8432</td>
<td>0.8388</td>
<td>0.0859</td>
</tr>
<tr>
<td>9</td>
<td>60.92</td>
<td>-18.44</td>
<td>-16.84</td>
<td>5.4776</td>
<td>0.8644</td>
<td>0.8604</td>
<td>0.0825</td>
</tr>
<tr>
<td>10</td>
<td>69.74</td>
<td>-18.7</td>
<td>-15.01</td>
<td>4.9938</td>
<td>0.8831</td>
<td>0.8797</td>
<td>0.0769</td>
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<td>11</td>
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<td>-13.46</td>
<td>4.4241</td>
<td>0.8988</td>
<td>0.8961</td>
<td>0.0693</td>
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<td>89.24</td>
<td>-19.28</td>
<td>-12.19</td>
<td>3.881</td>
<td>0.9117</td>
<td>0.9096</td>
<td>0.0617</td>
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<td>99.47</td>
<td>-19.57</td>
<td>-11.13</td>
<td>3.3738</td>
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<td>0.9208</td>
<td>0.0543</td>
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<tr>
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<td>109.67</td>
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<td>-10.26</td>
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<tr>
<td>15</td>
<td>119.99</td>
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<td>-9.532</td>
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<td>0.9374</td>
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<td>16</td>
<td>130.15</td>
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<td>159.01</td>
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<td>-7.565</td>
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<td>0.0216</td>
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<tr>
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<td>-7.228</td>
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<td>21</td>
<td>176.84</td>
<td>-21.52</td>
<td>-6.938</td>
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<td>-6.677</td>
<td>0.7048</td>
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<td>0.0119</td>
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<tr>
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<td>193.49</td>
<td>-21.88</td>
<td>-6.451</td>
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<td>0.9683</td>
<td>0.9683</td>
<td>0.0094</td>
</tr>
<tr>
<td>24</td>
<td>201.18</td>
<td>-22.04</td>
<td>-6.252</td>
<td>0.4201</td>
<td>0.9701</td>
<td>0.9701</td>
<td>0.0071</td>
</tr>
</tbody>
</table>
To determine the minimum dependency on subcutaneous fat thickness alternation, we have considered the linear regression technique and determine the slope of each parameter with respect to SF thickness variation. The minimum slope was obtained from the parameter LogRSlope. But it doesn’t have significant distinguishing feature between normal and abnormal muscle conditions as depicted in figure 17 and table 5.

**Table 5: EIM parameters percentage deviation per millimeter skin fat thickness**

<table>
<thead>
<tr>
<th></th>
<th>Deviation/mm</th>
<th>% deviation</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Normal Muscle</td>
<td>Acute Crush</td>
</tr>
<tr>
<td>Resistance at 50kHz (Ohm)</td>
<td>8.715</td>
<td>8.696</td>
</tr>
<tr>
<td>Reactance at 50KHz (Ohm)</td>
<td>0.238</td>
<td>0.295</td>
</tr>
<tr>
<td>Phase at 50kHz</td>
<td>-1.050</td>
<td>-0.724</td>
</tr>
<tr>
<td>Slope of $\frac{\log(R)}{\log(f)}$</td>
<td>-0.014</td>
<td>-0.011</td>
</tr>
</tbody>
</table>

*Figure 16: Linear regression analysis of the EIM parameters considered*
Table 5 enlists the percentage deviation of EIM parameters per millimeter of SF thickness from which we need to consider which parameter has the least deviating pattern out of reactance and phase since other parameters failed to meet the first criterion of disease detection. Reactance at 50 kHz varies 1.36% per millimeter of SF thickness for normal muscle whereas phase has a percentage deviation of 4.53% per millimeters thickness under same condition. Again, it can be seen from figure 17 that reactance is a good parameter to distinguish between different neuromuscular conditions. So, to conclude, reactance is the desired parameter which should be observed for neuromuscular disease detection since it possesses a very small factor of change due to SF thickness. For further study, we will be utilizing the findings of this particular study and use
reactance at 50 kHz as the desired parameter for further development. Though reactance proves to be the minimally affected parameter due to subcutaneous fat thickness alteration, the variation in the reactance due to muscle geometry variation must be considered. Observation of figure 18 and 19 gives an indication that, this variation can be dealt with by optimizing the electrode setup. The electrode separation used for figure 2 was 75mm between the excitation electrodes and 30mm between the sense electrodes and the electrodes were 65mm long and 7mm in width.

Figure 18: Variation in reactance due to different inter-electrode distance between the sense electrodes

Figure 19: Variation in reactance due to different inter-electrode distance between the excitation electrodes
As can be summarized from figure 20, reactance decreases in a rate of around 3 ohms per unit increase of muscle thickness. This creates more complexity to distinguish the normal muscle conditions from the abnormal one based on the value shown in figure 20 when the study is performed in a single frequency scale. The regression co-efficient obtained was the variable that was used as input to the optimization problem in this study. Minimization of this co-efficient means minimum effect on reactance due to the variation of muscle thickness. With a goal to minimize the effect of muscle thickness variation over reactance, inter-electrode distance was the parameter that was chosen as input of the optimization problem. This study considered two inter-electrode distance as named by distance between the sense electrode and distance between the source electrodes. Both distances had a limit from 0.5 cm to 3.5 cm based on the FEM model on which muscle length was 16 cm.

![Figure 20: Trend line of Reactance at 50 kHz for different muscle thickness](image)
As illustrated in figure 21 the fitness value lies around 1.079 which means 1.08 ohm change of reactance per centimeter change of muscle thickness. The optimized configuration is 10mm separation between the sense electrodes and 95mm separation between the source electrodes. The result from the simulation gives a clear indication that, placing the sense electrodes as close as possible and the source electrodes at the far end of the muscle group results in the minimum percentage change of reactance with respect to muscle thickness variation.

To serve the purpose of a diagnosing tool, our proposed configuration must have the capability to detect abnormal muscle conditions. To demonstrate this scenario, we have applied the optimized configuration in our model incorporated with the acute crush and chronic crush data of rat study found in (Wang L. L., et al., 2011). Table 6 illustrates the basic EIM parameters at
different muscle condition with our proposed configuration employed and it is evident from the
results that, the optimized configuration derived in this study still can detect muscular
abnormalities based on EIM parameters namely reactance and phase.

Table 6: A comparison of the EIM parameters at 50 kHz frequency with optimized electrode configuration

<table>
<thead>
<tr>
<th>EIM Parameter at 50 kHz</th>
<th>Normal</th>
<th>Acute</th>
<th>Chronic</th>
</tr>
</thead>
<tbody>
<tr>
<td>Resistance (R) ohm</td>
<td>45</td>
<td>44</td>
<td>47</td>
</tr>
<tr>
<td>Reactance (X) ohm</td>
<td>15</td>
<td>10</td>
<td>9.5</td>
</tr>
<tr>
<td>Phase (P) degree</td>
<td>18</td>
<td>12</td>
<td>11</td>
</tr>
</tbody>
</table>

For further analysis on impact of electrode configuration, this study extends its interest to electrode
surface area. Considering the near circular appearance of arm model, the area covered by an
electrode was defined by angle of rotation. The study was performed for different fat thicknesses
ranging from 5mm to 21 mm and considered the slope of the linear trend line as to be the factor
which should be minimized to maintain a consistent result for normal muscle condition. As we can
see from the figure below increasing the area covered by electrodes results in smaller slope which
means less percentage of deviation per millimeter fat thickness change.

![Figure 22: Resistance at 50 kHz for different fat thickness](image-url)
The variance of resistance for alteration in SF thickness is pretty significant for conventional electrode setup. The results of this study shows that covering more area with the surface electrodes eliminates the variation in both the EIM parameters w.r.t. SF thickness alteration to a significant extent. This phenomenon is described by the current distribution in the model. As can be summarized from the figure below, the model with larger electrodes has a more distributed current density among the surface which results in nullifying the effect of subcutaneous fat thickness.

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This experimentation gave the idea to extend the optimization electrode study by adding electrode surface area as a variable in the genetic algorithm. The goal was to minimize the alteration of EIM parameters with respect to muscle or fat thickness variation. Considering the total length of the model the range of the inter-electrode spacing was set to 3mm to 33mm. The electrode surface area was designed a variable by considering its angular coverage over the model and the range was from 90 to 3 degrees on both side of the symmetry. As the best fitness and best individual plots depicts, the optimized electrode spacing is 87mm between the excitation electrodes and 7mm between the sense electrodes. The best individual score for angular coverage is 6 degree. To simplify, the solution converges when the excitation electrodes are at their maximum limit and the sense electrodes are at their minimum limit. The variation in EIM parameters also depend significantly on the area covered by the electrodes. Best individual score for surface area covered by the electrodes is also at its minimum limit. The difference between the first two variables in the best individual plot of figure 25 are the angular coverage of the optimized electrode configuration. The third variable is the distance between sense electrodes and the fourth variable is the distance
the excitation electrode should be located apart from the sense electrode from optimized configuration.

![Figure 25: Best individual plot from MATLAB GA tool](image)

Figure 26 describes the results in a more convincing way. Here, the conventional configuration is 15mm-30mm-15mm spacing between the electrodes with 65mm x 7mm surface area. And the optimized configuration is 7mm x 7mm surface electrodes with 33mm-7mm-33mm spacing between them. Only reactance at 50 kHz has been highlighted in this figure because it has been stated in previous studies that reactance at 50 kHz is the parameter which is least affected by SF thickness alteration and can diagnose the disease effectively.
Figure 26: Variation in reactance at 50 kHz for alteration in muscle thickness with linear regression line

Figure 27: Variation in resistance at 50 kHz for alteration in muscle thickness with linear regression line
As can be depicted from the figure, the parameters are less prone to change in case of optimized configuration than the conventional configuration. Besides having least dependency over other anatomic and non-anatomic factor, our major goal is to diagnose abnormal muscle condition. As can be seen from figure 28, the proposed optimized configuration can successfully distinguish the normal muscle from the atrophied one.

![Graph showing Reactance in frequency spectrum for 46mm muscle thickness](image)

**Figure 28: Reactance in frequency spectrum for 46mm muscle thickness**

Based on the finding of the optimization problem, we carried out a more concentrated study on the electrode separation and surface area. The profile of the regression line shown in figure 7 shows, keeping the excitation electrodes apart from each other for a more larger distance than the optimized configuration and making the electrode surface area more smaller makes the desired EIM parameter even less variant with muscle thickness alteration with the capability to diagnose the atrophied muscle condition. The slope of the linear regression line is 0.17, which is even smaller than the slope we got from the optimized configuration 0.48. The electrodes shape used here was 1mm x 1mm.
Another interesting finding of this study is the electrode shape. Based on the results of this study, it can be suggested that the less the surface area covered by the electrode is the more the stability of EIM parameters. Though the amount of injected current is same, decreasing the surface area of electrodes results in a higher current density and larger penetration depth through body tissues especially through the subcutaneous fat thickness layer. To prove this hypothesis, we have performed another simulation with 1mm x 1mm electrodes placed in the same separation as the conventional electrode placement.
The comparison is well in favor of the proposed hypothesis as depicted in figure 34. Minimizing the surface area yields to a smaller slope of linear regression line for reactance vs. muscle thickness plot which means EIM parameters shows less variability with geometric change in case of point electrodes. The use of 1mm x 1mm electrode minimizes the changes in reactance with muscle shape variation by 58.26% in comparison to conventional electrode shape of 65mm x 7mm both having the identical electrode separation.
CHAPTER 5

5 Conclusion

EIM technique has been studied more thoroughly over the last decade and can be a perfect and preferred replacement of EMG technique according to the neurophysiologists because of its non-invasive nature and less effort to read the result. But, because of its dependency over other anatomic and non-anatomic factors, it’s yet to be deployed as a clinical tool. The finding of this study suggests that, placing the excitation electrodes at the far end of the muscle group and the sense electrodes closer to each other up to a practical limit and more importantly, keeping the surface area of the electrodes as small as possible can eliminate the variation caused by different fat or muscle thickness for different individuals. The electrode placement and shape was considered symmetrical over the whole study. From theoretical point of view, the surface area of sense electrodes was not supposed to have any impact on the EIM parameters. But, there was deviation from the expected value as we tried to keep the sense electrodes area same as conventional configuration. The finding of this study with proper practical experiment verification can eliminate these dependencies to some extent. The muscle or muscle group that the experiment is interested in can be considered as infinitesimal group of impedance block distributed around the body tissue. Like the transmission model, the smaller the length is covered by the sense electrodes, the smaller the potential difference between them would result in. Placing the current electrodes at the far end of the muscle group forces the alternating current to be distributed among the whole muscle fiber. Our further plan is to extend this study by implementing it in practical experiment.
References


Cross Section Through Middle Upper Arm | ClipArt ETC. (n.d.). Retrieved from Etc.usf.edu: http://etc.usf.edu/clipart/52800/52811/52811_arm.htm


Appendix

Matlab code for Livellink interface with COMSOL

```matlab
function [pot1,pot2] = model_str_test(SFat,BCrntSrc,angl1,angl2,et,ied1,ied2,MT)
close all;
import com.comsol.model.*
import com.comsol.model.util.*

model = ModelUtil.create('Model');
model.geom.create('geom1', 3);
model.geom('geom1').physics.create('ec', 'ConductiveMedia', 'geom1');
model.geom('geom1').lengthUnit('cm');
model.geom('geom1').feature.create('hell', 'Helix');
model.geom('geom1').feature('hell').set('turns', '1.1');
model.geom('geom1').feature('hell').set('rma', '1.1');
model.geom('geom1').feature('hell').set('rmin', '1.2');
model.geom('geom1').feature('hell').set('axialpitch', '15');
model.geom('geom1').feature('hell').setIndex('ax3', [1, 0], 0);
model.geom('geom1').feature('hell').setIndex('ax3', [0, 1], 1);
model.geom('geom1').feature('hell').setIndex('ax3', [0, 0], 2);
model.geom('geom1').feature('hell').set('endcaps', 'perpspine');
model.geom('geom1').feature.create('elp1', 'Ellipsoid');
model.geom('geom1').feature('elp1').setIndex('semiaxes', MT, 0);
model.geom('geom1').feature('elp1').setIndex('semiaxes', MT+0.01, 1);
model.geom('geom1').feature('elp1').setIndex('semiaxes', '15.5', 2);
model.geom('geom1').feature('elp1').set('pos', [8.3, 0], 0);
model.geom('geom1').feature('elp1').setIndex('pos', [0, 0], 1);
model.geom('geom1').feature('elp1').setIndex('pos', [0, 0], 2);
model.geom('geom1').feature('elp1').setIndex('pos', [0, 0], 1);
model.geom('geom1').feature('elp1').setIndex('pos', [0, 0], 2);
model.geom('geom1').feature('elp1').setIndex('ax3', [1, 0], 0);
model.geom('geom1').feature('elp1').setIndex('ax3', [0, 0], 1);
model.geom('geom1').feature('elp1').setIndex('ax3', [0, 0], 2);
model.geom('geom1').feature.copy('elp2', 'geom1/elp1');
model.geom('geom1').feature('elp2').setIndex('semiaxes', MT+SFat, 0);
model.geom('geom1').feature('elp2').setIndex('semiaxes', MT+SFat, 1);
model.geom('geom1').feature('elp2').setIndex('semiaxes', '16', 2);
model.geom('geom1').feature.copy('elp3', 'geom1/elp1');
model.geom('geom1').feature('elp3').setIndex('semiaxes', MT+SFat+0.3, 0);
model.geom('geom1').feature('elp3').setIndex('semiaxes', MT+SFat+0.3, 1);
model.geom('geom1').feature('elp3').setIndex('semiaxes', '16.5', 2);
model.geom('geom1').feature.create('cyl1', 'Cylinder');
model.geom('geom1').feature('cyl1').set('r', 9);
model.geom('geom1').feature('cyl1').set('h', 25);
model.geom('geom1').feature('cyl1').setIndex('pos', [-24.9, 0], 0);
model.geom('geom1').feature('cyl1').setIndex('ax3', [1, 0], 0);
model.geom('geom1').feature('cyl1').setIndex('ax3', [0, 2], 0);
modelgeom(' geom1').feature.create('dif1', 'Difference');
model.geom('geom1').feature('dif1').selection('input').set({'elp1' 'elp2' 'elp3' 'hell'});
model.geom('geom1').feature('dif1').selection('input2').set({'cyl1'});
model.geom('geom1').feature.create('cyl2', 'Cylinder');
model.geom('geom1').feature('cyl2').set('r', 9);
model.geom('geom1').feature('cyl2').set('h', 25);
model.geom('geom1').feature('cyl2').setIndex('pos', '16.3', 0);
```
model.geom('geom1').feature('cyl2').setIndex('ax3', '1', 0);
model.geom('geom1').feature('cyl2').setIndex('ax3', '0', 2);
model.geom('geom1').feature.create('dif2', 'Difference');
model.geom('geom1').feature('dif2').selection('input').set({'dif1'});
model.geom('geom1').feature('dif2').selection('input2').set({'cyl1'});

```plaintext
%%%OK till this point%%%  
poly1=[2,2,MT+SPat+.9,MT+SPat+.9,2];  
% poly2=[2,2,MT+SPat+.9,MT+SPat+.9,2];  
poly1_x=[8.3-2*et-ied1-ied2,8.3-et-ied1-ied2,8.3-et-ied1-ied2,8.3-2*et-ied1-ied2];  
% poly1_x=[8.3-et-7-ied1-ied2,8.3-7-ied1-ied2,8.3-7-ied1-ied2,8.3-7-ied1-ied2];  
poly2_x=[8.3-et-ied1,8.3-ied1,8.3-ied1,8.3-ied1,8.3-ied1];  
% poly2_x=[8.3-7-ied1,8.3-ied1,8.3-ied1,8.3-ied1,8.3-7-ied1];
```

model.geom('geom1').feature.create('wp1', 'WorkPlane');
model.geom('geom1').feature('wp1').geom.feature.create('pol1', 'Polygon');
model.geom('geom1').feature('wp1').geom.feature('pol1').set('x', poly1_x);
model.geom('geom1').feature('wp1').geom.feature('pol1').set('y', poly1);
model.geom('geom1').feature('wp1').geom.feature.create('pol2', 'Polygon');
model.geom('geom1').feature('wp1').geom.feature('pol2').set('x', poly2_x);
model.geom('geom1').feature('wp1').geom.feature.create('rev1', 'Revolve');
model.geom('geom1').feature('rev1').selection('input').set({'wp1.pol1', 'wp1.pol2'});
model.geom('geom1').feature('rev1').setIndex('pos', '8.3', 0);
model.geom('geom1').feature('rev1').setIndex('pos', '0.1', 1);
model.geom('geom1').feature('rev1').setIndex('axis', '1', 0);
model.geom('geom1').feature('rev1').setIndex('axis', '0', 1);
model.geom('geom1').feature('rev1').set('angle1', angl1);  %starting revolution angle for current electrode  
model.geom('geom1').feature('rev1').set('angle2', angl2);  %ending revolution angle for current electrode

model.geom('geom1').feature.create('mir1', 'Mirror');
model.geom('geom1').feature('mir1').selection('input').set({'rev1'});
model.geom('geom1').feature('mir1').setIndex('pos', '8.3', 0);
model.geom('geom1').feature('mir1').setIndex('pos', '0', 1);
model.geom('geom1').feature('mir1').setIndex('pos', '0', 2);
model.geom('geom1').feature('mir1').setIndex('axis', '1', 0);
model.geom('geom1').feature('mir1').setIndex('axis', '0', 1);
model.geom('geom1').feature('mir1').setIndex('axis', '0', 2);
model.geom('geom1').feature('mir1').set('keep', 'on');
model.geom('geom1').run;
model.geom('geom1').feature.create('dif3', 'Difference');
model.geom('geom1').feature('dif3').selection('input').set({'mir1', 'rev1'});
model.geom('geom1').feature('dif3').setIndex('input2').set({'dif2'});
model.geom('geom1').feature('dif3').set('keep', 'on');
model.geom('geom1').run;

model.geom('geom1').feature.create('del1', 'Delete');
model.geom('geom1').feature('del1').selection('input').init(3);
model.geom('geom1').feature('del1').selection('input').set('rev1(1)',1);
model.geom('geom1').feature('del1').selection('input').set('rev1(2)',1);
model.geom('geom1').feature('del1').selection('input').set('mir1(2)',1);
model.geom('geom1').feature('del1').selection('input').set('mir1(1)',1);

a=mphgetadj(model,'geom1','boundary','domain',1);
u=mphgetadj(model,'geom1','boundary','domain',5);
v=mphgetadj(model,'geom1','boundary','domain',8);
crntFace=intersect(a,u);
grndFace=intersect(a,v);

%%% Geometry done %%%

model.material.create('mat1');
model.material.create('mat2');
model.material.create('mat3');
model.material.create('mat4');
model.material.create('mat5');
model.material('mat1').selection.set(1);
model.material('mat2').selection.set(2);
model.material('mat3').selection.set(3);
model.material('mat4').selection.set(4);
model.material('mat5').selection.set([5 6 7 8]);

model.material('mat1').propertyGroup('def').set('electricconductivity', {'0.0002'});
model.material('mat1').propertyGroup('def').set('relpermittivity', ['1150']);
model.material('mat2').propertyGroup('def').set('electricconductivity', {'0.03'});
model.material('mat3').propertyGroup('def').set('relpermittivity', {'500'});
model.material('mat3').propertyGroup('def').set('electricconductivity', {'45' '0' '0' '0' '2' '0' '0' '2'});
model.material('mat4').propertyGroup('def').set('relpermittivity', {'70e3' '0' '0' '0' '55e3' '0' '0' '55e3'});
model.material('mat4').propertyGroup('def').set('electricconductivity', {'0.0035'});
model.material('mat5').propertyGroup('def').set('relpermittivity', {'300'});
model.material('mat5').propertyGroup('def').set('electricconductivity', {'5e5'});
model.material('mat5').propertyGroup('def').set('relpermittivity', ['1']);

model.physics('ec').feature('init1').set('V', 1, '0.01');
model.physics('ec').feature.create('bcs1', 'BoundaryCurrentSource', 2);
model.physics('ec').feature('bcs1').set('Qjs', 1, BCrntSrc); %Boundary current source
model.physics('ec').feature('bcs1').selection.set(crntFace);
model.physics('ec').feature.create('gnd1', 'Ground', 2);
model.physics('ec').feature('gnd1').selection.set(grndFace);

%%Study
model.study.create('std1');
model.study('std1').feature.create('freq', 'Frequency');
model.study('std1').feature('freq').activate('ec', true);
model.study('std1').feature('freq').set('pllist', '50000*1^range(1,1)');

model.mesh.create('mesh1', 'geom1');
model.mesh('mesh1').autoMeshSize(3);
model.mesh('mesh1').run;

model.sol.create('sol1');
model.sol('sol1').study('std1');
model.sol('sol1').feature.create('stl1', 'StudyStep');
model.sol('sol1').feature('stl1').set('study', 'std1');
model.sol('sol1').feature('stl1').set('studystep', 'freq');

model.sol('sol1').feature.create('v1', 'Variables');
model.sol('sol1').feature('v1').set('control', 'freq');
model.sol('sol1').feature('v1').set('control', 'freq');
model.sol('sol1').feature('v1').feature.create('pl1', 'Parametric');
model.sol('sol1').feature('v1').feature.create('pl1', 'Parametric');
model.sol('sol1').feature('pl1').feature.remove('pDef');
model.sol('sol1').feature('pl1').feature('pl1').set('pname', 'freq');
model.sol('sol1').feature('v1').feature('pl1').set('pllistarr',
('50000*1^range(1,1)'));
model.sol('sol1').feature('v1').feature('pl1').set('plot', 'off');
model.sol('sol1').feature('v1').feature('pl1').set('plot', 'off');
model.sol('sol1').feature('v1').feature('pl1').set('probesel', 'all');
model.sol('sol1').feature('v1').feature('pl1').set('probes', {});
model.sol('sol1').feature('v1').feature('pl1').set('control', 'freq');
model.sol('sol1').feature('v1').feature('pl1').set('control', 'freq');
model.sol('sol1').feature('v1').feature.create('fc1', 'FullyCoupled');
model.sol('sol1').feature('v1').feature.create('fc1', 'FullyCoupled');
model.sol('sol1').feature('v1').feature.create('il1', 'Iterative');
model.sol('sol1').feature('v1').feature.create('il1', 'Iterative');
model.sol('sol1').feature('v1').feature('il1').set('linsolver', 'bicgstab');
model.sol('sol1').feature('v1').feature('il1').set('linsolver', 'bicgstab');
model.sol('sol1').feature('v1').feature('il1').set('linsolver', 'bicgstab');
model.sol('sol1').feature('v1').feature('il1').feature.create('mgl1', 'Multigrid');
model.sol('sol1').feature('v1').feature('il1').feature('mgl1').set('prefun', 'gmres');
model.sol('sol1').feature('v1').feature('il1').feature('mgl1').set('iter', 2);
model.sol('sol1').feature('v1').feature('il1').feature('mgl1').set('mgcycle', 'v');
model.sol('sol1').feature('v1').feature('il1').feature('mgl1').set('mcasegen', 'any');
model.sol('sol1').feature('v1').feature('il1').feature('mgl1').set('gmglevels', 1);
model.sol('sol1').feature('v1').feature('il1').feature('mgl1').set('scale', 2);
model.sol('sol1').feature('v1').feature('il1').feature('mgl1').set('massem', 'true');
model.sol('sol1').feature('v1').feature('il1').feature('mgl1').set('mkeep', false);
model.sol('sol1').feature('v1').feature('il1').feature('mgl1').set('rmethod', 'longest');
model.sol('sol1').feature('v1').feature('il1').feature('mgl1').set('mlevels', 5);
model.sol('sol1').feature('v1').feature('il1').feature('mgl1').set('maxcoarsedof', 5000);
model.sol('sol1').feature('s1').feature('i1').feature('mg1').set('amgauto', 3);
model.sol('sol1').feature('s1').feature.remove('fcDef');
model.sol('sol1').attach('std1');

model.sol('sol1').runAll;
p=mphgetcoords(model,'geom1', 'domain', 6);
q=mphgetcoords(model,'geom1', 'domain', 7);
pot1=mphinterp(model,'V','coord',[p(1);p(2);p(3)],'complexout','on');
pot2=mphinterp(model,'V','coord',[q(1);q(2);q(3)],'complexout','on');
clear p;
end