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Analysis of Skin-Electrode Impedance Using Concentric Ring Electrode

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Abstract:-**A significant contributor to artefact generation in surface electromyography (sEMG) and, functional electrical stimulation (FES) intensity is the skin-to-electrode impedance ($Z_{S/E}$). While using electrolytic gels may initially lower $Z_{S/E}$, the impedance may not remain stable. It can vary over time due to changes in underlying structures such as sweat glands and physical deformations due to movements. An experiment seeking to identify major factors in the reduction of $Z_{S/E}$, and therefore mitigate these artefacts, was performed by varying a series of control factors on the concentric ring electrode (CRE). Unlike conventional disc electrodes, CREs have small surface areas which may exacerbate $Z_{S/E}$ changes. The factors tested were electrode material, electrode size, skin preparation, and surface pressure. This work analyzes how these factors in their various combinations effect changes in $Z_{S/E}$ and suggests protocols for improving recording or stimulation with CREs via lowered and consistent $Z_{S/E}$.**

Index Words: - **Surface Electromyography, sEMG, Functional Electrical stimulation, FES, skin-to-electrode impedance and Concentric ring electrode (CRE).**

I. Introduction

Loss or shift of skin contact with electrodes can cause severe motion artefacts, lowering the recorded signal fidelity. Since concentric ring electrodes (CRE) have smaller surface areas than conventional disc electrodes, variations in the skin topography due to muscle movement may alter the $Z_{S/E}$ causing motion artefacts. These motion artefacts may cause the signal processing amplifiers to saturate, with resulting data loss [1]. Skin abrasion, electrolytic gel interfaces and pressure on the electrode surface have all been individually observed to help in reducing $Z_{S/E}$ and stabilizing the electrode, thus minimizing motion artefact [2]. Skin preparation methods using electrolytic gels have been shown to initially lower $Z_{S/E}$ and hold the electrode in place. Other work has found a correlation between electrode contact surface area and $Z_{S/E}$ [3].

Our work was performed in an effort to formalize recommendations for obtaining high signal fidelity, low artefact, and low interface impedance via the appropriate combination of these factors for a given experimental process. The experiments performed consisted of three main functional trial sets: 1) computational modeling and analysis 2) impedance meter construction and validation for use with CRE trials 3) agarose validation of computational model and impedance meter, and 4) human trials. A computational model of the CRE interfaced with a skin functional unit, was generated and used to simulate the various treatment combinations to be tested in the agarose and human experiments.

While there are a wide variety of commercially available impedance meters, the precision, resolution, and frequency range required for these CRE experiments made most of these available meters either unsuitable or too expensive. An impedance meter was designed to work specifically with the CRE that could be easily interfaced to a computer for data acquisition.

Agarose provides a safe, repeatable, and controlled medium for testing both the impedance meter functionality and to provide a controlled set of responses for the various CRE trials to be performed in the human experiments. These results were used as a validation set for the computational model. Human trials were performed to provide final validation for the computational model and a basis for the development of the suggested CRE utilization protocols.

II. BACKGROUND

A) Skin-Electrode Interface Model

By analyzing the $Z_{S/E}$, it might be possible to determine how to help ascertain whether a CRE, such as in Fig 1, is making proper contact with the body surface. In his book, *Medical Instrumentation-Application and Design*, [2] John G. Webster gives the electrical equivalent model for the skin, electrodes, and gel, which has been adapted to the model as shown in Fig 2, where, Vh is the half cell potential of the electrode/gel interface, the parallel combination of resistive (Re) and capacitive (Ce) components make up the impedance associated with the electrode-gel interface and polarization effects (Ze), and Zg is the series impedance associated with interface effects and due to resistance in the gel.

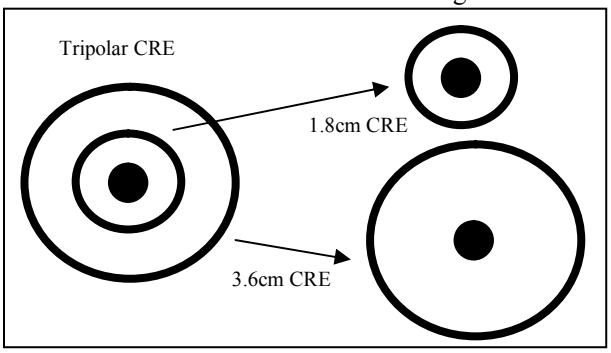

Fig 1 CRE Configuration showing how one Tripolar CRE can be used as two separate electrodes of 1.8cm and 3.6cm diameter

The potential difference across the epidermis is given by Vep. The electrical impedances in the epidermis (Zp) are modeled as a parallel combination of resistive (Rp) and capacitive (Cp) components. The total subdermal impedance is modeled as Zt [2]. See Fig. 2 for a circuit diagram showing these impedances incorporated into the overall impedance $Z_{S/E}$.

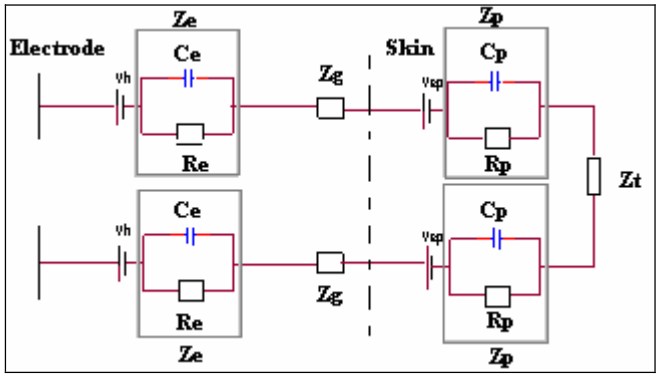

Fig 2 Electrical equivalent model for the total impedance in a two electrode system

B) Calculation of Skin-to-Electrode Impedance

The impedance is measured between two electrodes, all the impedances given in the Webster model must be doubled and result in a series combination of Ze, Zg, Zp, and Zt related to the first electrode and Zt, Zp, Zg, and Ze related to the second electrode, as shown in Figure 2.

C) Measurement of Skin-to-Electrode Interface Impedance:

The impedance measurement instrumentation used consisted of a 2.0 VAC signal generator, as shown in Fig.3. Due to skin impedance approaching higher values at lower frequencies, [2] a high input impedance, R1, was included. R1 was included to prevent damaging currents from flowing into the test subject. The returning current from the recording electrode impressed a voltage across resistor R2. An interface circuit for Nyquist filtering was developed and custom signal processing software for this experiment were written in LabVIEW™ & MATLAB™ and were used to acquire and process the signals.

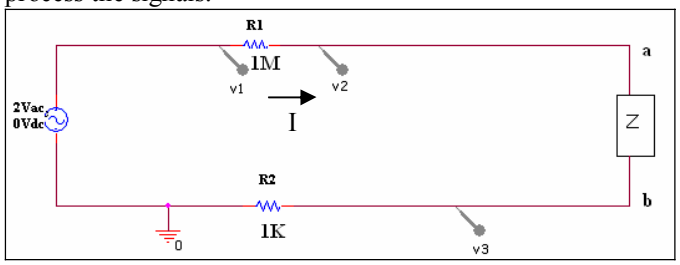

Fig 3 Schematic of impedance measuring instrumentation

D) Measurement of Current, I:

The current flowing through the circuit, *I*, in Fig 3 was measured to calculate the skin-to-electrode impedance. This was performed by measuring V_1 and V_2 to determine the voltage across the resistor R_1 and dividing it by R_1.

$$I = (V_2 - V_1) / R_1 \qquad \text{(Eq. 1)}$$

Where, V_1 = input voltage and V_2 = CRE center disc voltage.

E) Measurement of Unknown Impedance $Z_{S/E}$:

The differences in voltages V_2 and V_3 recorded results in the voltage dropped across the unknown skin-to-electrode impedance Z in Fig 3. By dividing this resulting voltage by the current *I*, the impedance value $Z_{S/E}$ is determined. The magnitude of *Z* is given by,

$$Z_{S/E} = (V_2 - V_3) / I \qquad \text{(Eq. 2)}$$

III. Methodology

A) Computer Modeling:

A computational model was developed using FEMLAB (Comsol, Stockholm, Sweden). The model consisted of the following components: 1) a CRE, 2) electrode-skin interface material (simulating gel preparations), and 3) a bilayer skin model. The controlled simulation factors were 1) CRE material, 2) CRE size, 3) CRE conductivity, and 4) paste, and 5) modeled skin layer thickness were individually varied and the results recorded for future comparison to the physical trials.

The finite element method (FEM) model generated in FEMLAB was used to estimate the current density distribution in the homogeneous conductive medium that was used to represent the skin. This skin bilayer model consisted of two separate homogeneous layers representing the epidermis (conductivity 0.4 S/m), and dermis/subdermal (0.06S/m) structures. Two additional layers represented the CRE (Copper (Cu), gold (Au), silver (Ag), and tin (Sn), and the interface material between the CRE and the skin (NuPrep and Ten-20). All structures were modeled as individually homogeneous distributive resistance units. Constant voltage of 2.0 V was applied to the outer electrode during simulations and center disc was connected to the ground (0.0 V).

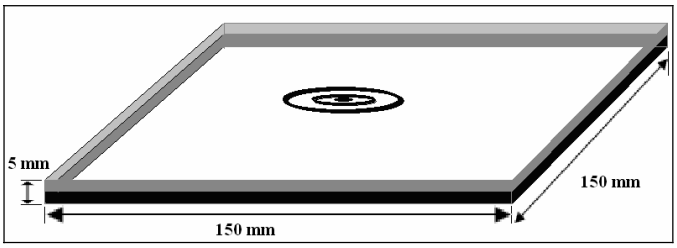

Fig 4 Agarose Skin Analog Bilayer Model, with CRE resting on top layer

B) Agarose analog of skin:

A physical model of the multilayered human skin was developed using agarose gel and is displayed in Fig 4. The conductivities can be controlled to be close to that of the epidermis and dermis/sub dermal structures by varying the quantity of Tris 1M solution (pH 6.1) which is the solvent of the agarose and the quantity of agarose itself. Agarose (5.6gm) was dissolved in Tris (120 ml at 6.1pH). This mixture was stirred continuously and boiled to 100°C. At approximately this point, agarose boils, breaking its links and dissolving completely in the Tris to form a dense colorless liquid which further solidifies into a gel at room temperature. The agarose is considered as a distributed resistance with a combined conductivity of 0.44 S/m for the bilayer. [4].The epidermis layer consisted of 2.0 mm thick agarose with a conductivity of 0.44 S/m, while the dermis/subdermal

structures were modeled with a 3.0 mm-thick Agarose layer with conductivity 0.06 S/m [5].

A sinusoidal input with amplitude of 2.0 V at a frequency of 30Hz was applied to the outer ring of the CRE, and the center disc was kept at ground. The current drawn through the CRE circuit was used within the Energy method (integrating energy drawn through the electrode over the range of currents, obtained by measuring the current drop across R1 and R2, see Fig 3) to calculate the appropriate impedances. The $Z_{S/E}$ was measured from CREs of 3.6 cm and 1.8 cm diameters. The variations in the impedance patterns were plotted and a comparison was made with the results obtained using a commercial impedance meter (BIOPAC^{TM}).

C) Electrode Material:

$Z_{S/E}$ Skin-to-electrode impedance can change due to different materials used for the electrodes. Au, Ag, Cu, and Sn were tested for their effect on measured $Z_{S/E}$.

D) Human Subjects - Measurement of Skin-to-Electrode Impedance:

To test the skin-to-electrode impedance on humans, twenty healthy subjects ages 21-32 were recruited for testing. Each subject was given a full disclosure of the experimental procedure which had been previously approved by the University IRB. All impedance recordings were made with the subjects resting in the supine position on their back. An inflatable jacket (Cutting-edge Hal key-Roberts Pro-1F CO2 Automatic Inflator, oral tube backup system). strapped to the subjects was used to generate the surface pressures on the CRE's during testing. For all combinations of treatments, the electrodes were placed on the false ribs (last two ribs that are not connected to sternum). Incremental changes in pressure, roughly none, 1/3, 2/3, and fully inflated jacket, were applied to the electrodes using the inflatable jacket. For the individual treatment variations, the skin sites were treated with different preparations: a) no preparation, b) Nuprep (Weaver Co), c) TEN-20, and d) EC2 conductive pastes. Impedance readings for each treatment were taken with the various test pressures on the CRE surface. This procedure was repeated for the different electrode sizes and again for the different electrode metals.

Pressure to the electrode was applied by filling an inflatable jacket. The pressure was not calibrated, but rather was monitored for relative correlation of pressure increase to $Z_{S/E}$. Sampling was performed at 1,000 Hz for 10 seconds for a total of 10,000 samples per trial. A band pass filter from 1 to 100 Hz was used for hardware filtering. Data were acquired from the CREs via a 6036E National Instruments Daq Card. The 6036E was controlled using a custom LabVIEW^{TM} program. Post-processing was performed with a custom MATLAB^{TM} program to determine $Z_{S/E}$. Each experiment was repeated fifteen times with out disturbing the position of electrode sites.

IV. RESULTS

A) Validation of FEMLAB and Impedance Meter with Agarose Test:

Figure 5 shows the high correlation between the computational model and the agarose testing from a typical subject trial. The data are arranged by electrode material type, and shown in two major categories for each of the electrodes: no preparation (skin only), and Ten-20 (electrolytic gel with skin abrasion) for each of the two sizes tested per material.

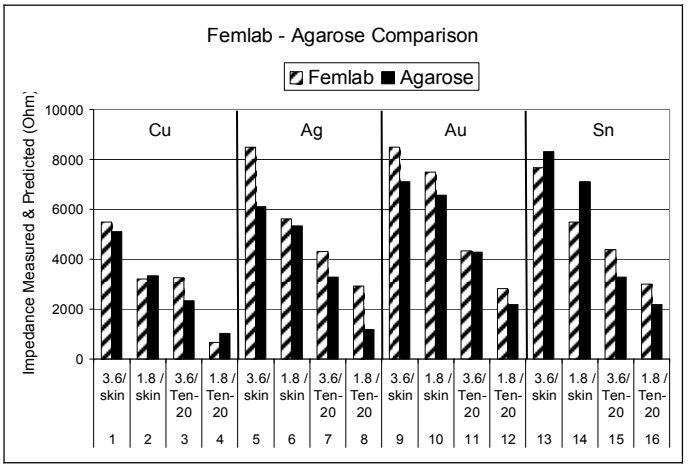

Fig 5 Femlab computational model and agarose validation trial comparison

Figure 6 shows the results for the Cu electrodes. The results for the other materials tested followed similar patterns, with only minimal variations. Due to space limitations, this was selected as the representative set, as Cu is the electrode material that showed the greatest benefit.

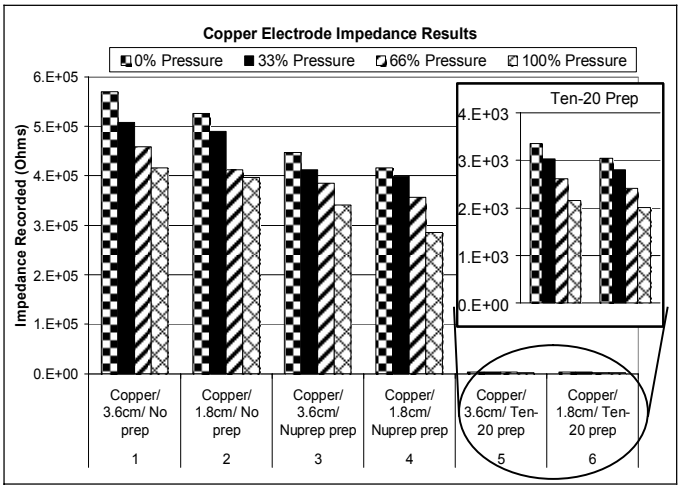

Fig 6. Human trial impedance measured for Cu electrodes

V. DISCUSSION

For resistors and capacitors, the impedance measuring instrumentation was accurate to the tolerance of the components. The impedance didn't vary over time. When impedance was measured from human subjects, it was found

to vary over time. In most subjects, after wearing the electrodes continuously for more than 15 minutes the impedance began decreasing without changing the pressure or preparation. After further investigation there was evidence that moisture was forming between the skin and electrode due to sweat, which lowered the impedance. A breathable material should be used as a substrate for the electrodes. The computational model was validated based on these results and can be used to reasonably predict ranges of operating parameters prior to experimentation.

The highlights of the items to include in any experimental protocol are suggested in the following section. These, however, should not be considered a definitive, final, or complete set of protocols for CRE experiments, but rather a companion guide of protocols for defining the sub-components of an experiment related to minimizing artefact and maximizing useable data collected from any CRE measurement point.

Suggested Decision-Making Hierarchies for CRE Protocols: Currently the preffered combination: 1.8cm diameter, copper plated CRE.

(1) *Surface Preparation:* Choice of appropriate surface preparation method, which can be a combination of abrasion and/or electrolytic gel, is the primary driver in the decision making process. The experiments suggested the most effective method was the surface abrasion with electrolytic gel.

(2) *Electrode metal*: As was expected, within the scope of this experiment Cu, with the highest conductivity of the metals tested, was found to be the most suitable material in reducing $Z_{S/E}$. However, Cu corrodes quickly and isn't suitable for most electrode applications, therefore Ag would be the most appropriate in a wide number of cases.

(3) *Electrode Size*: From computational analysis agarose, modeling, and the human experiments conducted, consistent results were obtained from smaller diameter CRE versus larger sized CRE. This is an encouraging result as smaller sized electrodes allow for a greater versatility of experiment design. This leads to greater diversity of sites where it can be used.

(4) Pressure: The greater the pressure applied to the surface of the electrode, regardless of material size and surface preparation, will aid in reducing $Z_{S/E}$ and increasing electrode location stability, decreasing the chance of motion artefact. We didn't measure the absolute pressure on the electrodes, so we only know that as pressure increased, the $Z_{S/E}$ decreased, as one would expect.

(5) Modeling: A validated computational model, within the constraints of the modeled material's applicability to the new desired function, can be used to predict functionality of a CRE size, material, and interface material prior to any experiments being actually performed.

VI. CONCLUSIONS

If the $Z_{S/E}$ impedance is less than 10kΩ it can be concluded the electrode makes good contact with the skin. The $Z_{S/E}$ for CRE after skin preparation is within the common range reported in the literature for typical surface potential electrodes. Applying pressure also helps to improve the $Z_{S/E}$.

These protocols that were generated are a good guide for a researcher attempting to minimize impedance and artefact in experimentation using CREs performed.

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REFERENCES

[1] Rosell,J.;Colominas,J.;Riu,P.;Pallas-Areny,R.;Webster,J.G.; "Skin impedance from 1Hz to 1Mhz." IEEE Trans.Biomed Eng., vol.BME-35, pp.649-651.1988.

[2] Webster,J.G.; "Medical Instrumentation. Application and design" Ch. 5.4, 5.5, pp.194-200, Ed., 3rd ed., Newyork:Wiley, 1998.

[3] Patrick,K.M.; A Thesis presented to the College of Graduate studies, Lamar University-Beaumont, "Skin impedance measuring device" 1998.UMI Pro Quest Digital Dissertations. Publication No. AAT 1391191.

[4] Yamanoto,T.;Yamanoto,Y.;1997. "Electrical properties of epidermal stratum corneum" Med. Biol. Eng. 3:151-158

[5] Peters, M. J.;Stinstra,G.;Hendriks,M.; "Estimation of the Electrical Conductivity of Human Tissue", *Electromagnetics*; Vol. 21 Issue 7/8, p545-557, Oct-Dec2001