A Transcutaneous Electrode Measurement System

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Abstract

Arrays of transcutaneous electrode elements have been proposed as a method to improve selective muscle stimulation and functional control. Measuring the current distribution across the surface of these electrode elements is important for understanding how the electric field penetrates into human tissue.

A new measurement system has been developed that simulates human skin at the electrode interface. Small structures were incorporated, allowing localised changes in skin impedance to be simulated. Current flow through the measurement system was recorded and used to produce current distribution maps at the simulated electrode skin interface.

First measurements indicated that coupling gel helped to uniformly distribute stimulation current across the surface of the electrode. Simulating localised changes in skin impedance produced large current densities and non-uniform current distributions, which is in agreement with published literature.

1. INTRODUCTION

Transcutaneous electrode arrays have the potential to improve functional movements through selective stimulation of different muscle groups [1,2]. Assessment of the current distribution from multiple active elements at the electrode-skin interface is important when defining new electrode array properties. A pseudo-skin layer was developed by [3] to mimic the impedance of skin. Transcutaneous electrodes were placed on the pseudo-skin, which covered a saline filled tank. Current distributions were then calculated by manipulating a differential voltage probe through the tank. Direct measurements of the current distribution at the electrode skin interface were not possible.

A measurement surface that simulates the impedance of skin can be constructed through discrete networks of electronic components [4, 5, 6]. By measuring the current flow through the components it is possible to estimate the current distribution at the electrode interface. Spatial resolution of the system then depends upon the size of the measurement surface(s). Both [4, 5] used pad sizes of 1cm^2, which are insufficient to analyse current distribution between closely packed (< 5mm) electrode elements [1].

Local changes in skin impedance due to 'voltage breakdown' or underlying structures (e.g. sweat glands) are often cited as ways for current flow into the body [6,7]. Further experiments [6:pp23] would imply that current can be considered as being conducted through discrete channels of ~1mm². Incorporating such structures into an in-vitro measurement system would enable their effects to be quantifiably assessed.

A new in-vitro measurement system is herein described which is able to measure the current density at sub mm^2 resolution, as well as simulate localised changes in skin impedance.

2. METHODS

2.1. Measurement Circuit

A gold plated measurement surface (53×53mm) comprising of 100 small pads (0.81mm^2) surrounded by 24 larger pads (112.33mm^2) was designed (fig 1). To accommodate larger electrodes, a gold plated pad of uniform impedance surrounds the measurement surface.

A schematic of the measurement circuit is shown in fig 2. A constant current stimulator (CCS) [8] delivers electrical current to the electrode under test. The current flow through each pad is calculated from the voltage V measured across each of the impedances Z_M and Z_S. The voltage across Z_C is used to record the actual current flow through the whole circuit.

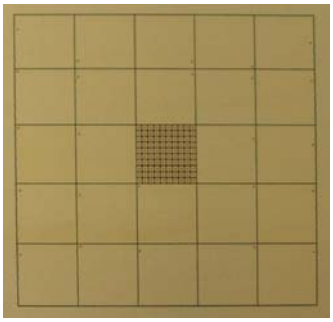


Fig 1: Measurement array of 100 small (0.81mm^2) and 24 larger pads (112.33mm^2). Each pad is isolated with a 0.15mm gap.

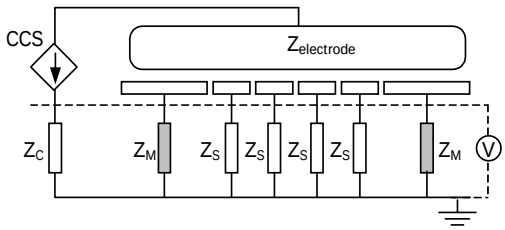


Fig 2: Schematic of the electrode measurement circuit. The current distribution is calculated by measuring the current flow through each pad impedance (24×Z_M and 100×Z_S). The supplied current can be measured from the voltage across Z_C..

The impedance for the small (Z_S=6.8MΩ) and medium (Z_M=54.4kΩ) pads were calculated using $Z = \rho d / A$, assuming a uniform skin resistivity of ρ=1kΩm (~550Ωcm^2) for a thickness of d=6mm.

2.2. Equipment

The measurement surface was manufactured on a 6 layer PCB board, with gold plated pads placed in a two dimensional structure (fig 1). The voltage through each resistive element (1% tolerance) was recorded using a 64-channel acquisition card (NI 6071E) running at 620kHz with custom LabVIEW (National Instruments, Texas USA) software.

2.3. Validation

An electrode uniformly distributing a current of 20mA across its surface area (20cm^2) should produce a current density of J=10μAmm^{-2}. This would generate a voltage of ~55V between each of the measurement points and ground. This is consistent with simulations (Protel DXP, Altium) and measurements of electrode-skin interface voltages undertaken by the authors.

The 6071E limits the range of measured voltages to ±10V; to allow large currents (>3mA) to be analysed the circuit was scaled by a factor of 0.1 (i.e. Z_M=5.4kΩ, Z_S= 680kΩ) simulating a surface resistivity of 100Ωm. Mixed mode simulations (Protel DXP) and measurements of electrode current densities and distributions before and after scaling indicated there would be no detrimental effects.

2.3. Experiments

A constant current stimulator [8] was set to produce monophasic pulses (width 1ms, 25Hz repetition frequency). Measurements were repeated with and without electrode coupling gel (Compex Medical SA). A 500g weight was used to apply constant pressure across the electrode surface.

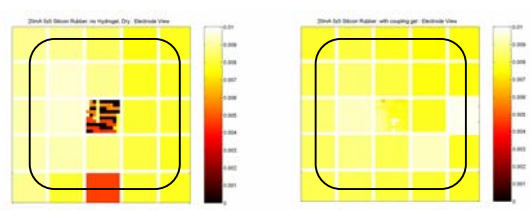
3. RESULTS

A train of 20 constant current pulses was recorded for each data set. The average amplitude of each pulse was extracted using custom algorithms written in Matlab 7.1 (Mathworks). Standard deviations were less than 5% of the measured value.

3.1. Uniform surface resistivity (100Ωm)

Current distributions were recorded for a 5×5cm metal plate, a 5×5cm dry silicon rubber electrode without hydrogel (fig 3), and normal 5×5cm self-adhesive hydrogel electrodes (Compex Medical SA) (fig 4). To observe edge effects, a 2×4cm metal plate was placed across half of the measurement surface and coupled using electrode gel (fig 5).

There was no significant difference between



a) b)

Fig 3: 5×5cm Silicon rubber electrode at 20mA; a) without coupling gel and b) with coupling gel. [Scale mAmm^{-2}]

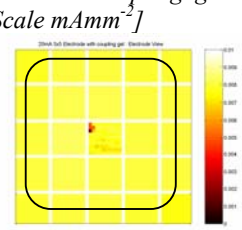


Fig 4: 5×5cm self adhesive electrode at 20mA with coupling gel. [Scale mAmm^{-2}]

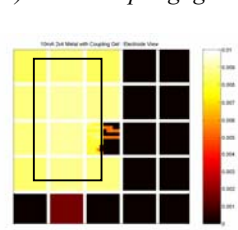


Fig 5: 2×4cm metal plate at 10mA with coupling gel.

a) b) c)

Fig 6: Current distribution maps for a 5×5 cm self adhesive electrode at 20mA with coupling gel [Scale mAmm^{-2}]. Non-uniform surface impedances (ρ=29mΩm) were generated for one (a) and two (b,c) small electrode pads. Peak current densities were a) 6mAmm^{-2}, b) 5.2mAmm^{-2} & 4.8mAmm^{-2} and c) 2.6mAmm^{-2} & 2.4 mAmm^{-2}

using electrodes with and without hydrogel (fig 3a, vs. fig 4). The use of coupling gel helped improve the uniformity of the current distribution by providing better contact between the electrode and measurement surface. The coupling gel also enables current to spread beyond the extent of the electrode surface as observed in fig 5. However there is no current flow in pads that are not in contact with the metal plate or coupling gel.

3.2. Non-uniform surface impedance

To simulate non-uniform surface impedances (e.g. sweat pores) the resistance of one and two elements (fig 6) were reduced to Z_S=200Ω (ρ=29mΩm). Significant changes in current density were measured at these points, which affect the uniformity of the localised current distribution. The total current density across the entire electrode is also reduced when compared to uniform surface impedances (fig 4). The spatial distribution of the low impedance elements also appears to affect the current density across the electrode surface.

4. DISCUSSION AND CONCLUSIONS

A new 124-element measurement surface that simulates non-uniform skin impedances with a resistivity (ρ) of up to 1kΩm has been developed. The system can be used to measure current distributions in existing and new transcutaneous electrode structures. Scaling the impedance measurements (i.e., ρ=100Ωm) did not appear to effect measured current distributions.

Localised changes in skin impedance did effect the current distribution as predicted by [7]. Very high current densities that could lead to tissue burning could be clearly measured without the need to incorporate networks of resistors [5] or capacitive and non-linear components [7].

Coupling gel helped to uniformly dissipate current across large surface areas but did not reduce the effect of localised impedance changes.

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