A NON-INVASIVE BIOPOTENTIAL ELECTRODE FOR THE CORRECT DETECTION OF BIOELECTRICAL CURRENTS

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ABSTRACT

An advanced current measure probe is described for biological applications. Singular points on the body skin emit a semi-constant current that could be potentially useful for a medical diagnosis of the body organs to which any singular point is connected. In order to achieve an accurate and reliable analysis of the emitted current, all the error sources have been studied in order to design a ready to use probe.

KEY WORDS

Biosensors, Non-invasive measurements, Biopotential Electrodes, Measurement and Instrumentation.

1. Introduction

Significant effort has been devoted to evaluate the electrical properties of paths on the human body described by Chinese Medicine (i.e., meridians) [1]. The specific points of interest (acupuncture points) are described as being one millimeter in diameter. The electrical characteristics of the acupuncture meridians have been extensively studied since the 1950s to establish their existence by scientific methods [2]-[3]. To date, the bioelectric properties of acupuncture points are not clearly understood. For example, many discrepancies and inconsistencies appear in the literature concerning skin impedance on the acupuncture points [4]. The change in impedance at acupuncture points is reportedly anywhere from 1/2 to 1/20 of the impedance of the surrounding skin. These low impedance points are thought to be the result of sensory and motor nerves emerging from deep tissue to superficial layers of the skin [5]. It is by now generally accepted that both meridian and acupuncture points have lower electrical resistance or impedance than nearby surrounding points. Small electrical potentials can be recorded on the skin at the terminal points of acupuncture meridians and the results are replicable under controlled experimental conditions [6]-[7]. As a results of these potentials, the current produced ranges from a few to several hundred nanoamperes, which can be recorded by an external measuring device. The electromotive force producing these endogenous currents results from electrical potentials at the acupuncture points on the skin and is a function of the internal resistance of the point.

Figure 1 An acupuncture point seen as the result of sensory and motor nerves emerging from deep tissue to superficial layers of the skin.

Figure 2 shows a typical behaviour of the electric resistance on an acupuncture point. The resistance is a function of the cutaneous region, skin humidity and pressure of the measurement electrode. The arrow points out the strong decrease of resistance in the point of acupuncture. The ordinate reflects the resistance in ohms, the abscissa the run of the electrode on the cutaneous surface. Most of research on electroacupuncture has been based on skin resistance and feeding in external currents [8]-[9]. All of these instruments use an external voltage source to measure the skin conductance and the measurement can be easily influenced and disrupted by many factors, making it difficult to achieve reliable and replicable results. Moreover, feeding the point with an
external current in order to measure the internal resistance constitutes an interference from a physiological point of view and the exact effect on the human organism of applying external invasive currents of even a few μA to acupuncture points are not known. Thus, any method permitting readings to be taken in nanoamperes at acupuncture points without feeding in additional external current would be diagnostically useful. Figure 3 shows the schematic representation of current measurement without use of an external generator[10].

2. Main error components

For the correct evaluation of the physiological parameters of medical interest it is necessary that the measures are repeatable and reliable. We performed an analysis of the main error components that may influence the measurements:

1. It is necessary to avoid the formation of spurious potentials that would contaminate our measurements. In particular, the measurement technique must not depend on the skin ph and on the internal resistance of the characteristic point. This can be achieved using electrodes that minimize the overpotentials, such as the silver/silver chloride ones.

2. The tip of the probe must have a proper shape in order to minimize pain for the patient and to keep a constant contact area on the skin. The contact area will vary as a function of the pressure exerted by the operator whereas the measured current will be a function of pressure. Therefore a pressure compensation system must be introduced such that measurements are virtually insensitive to fluctuations in pressure over a wide range of values.

3. The current emitted by the characteristic point, when measured in a short circuit, decreases asymptotically from a peak value to a lower steady state value as a function of many physiological parameters and conditions. Measured data validation is therefore demanded to a software procedure to calculate the standard deviation of the datum in a given number of samples. When the standard deviation is lower of a preset threshold, the current value is accepted and displayed.

3. The silver/silver-chloride electrode

To measure and record potentials and currents in the body an electrode can be used as biomedical sensor [11]. This seems to be a very simple function, but in fact an electrode recording biopotentials is actually a transducer, converting ionic currents in the body into electronic currents in the electrode. This transduction function greatly complicates electrode design. Figure 4 shows an electrode-electrolyte interface. The electrode only has one type of charge carrier (electron), whereas the electrolyte has two types of charge carriers (cation and anion). The electrolyte is an aqueous solution containing cations of the electrode metal C⁺ and anions A⁻. The electrode consists of metallic atoms C and the current crosses the interface from left to right. At the interface, charge is exchanged through chemical reactions, which can be generally represented as:
where \( n \) is the valence of \( C \) and \( m \) is the valence of \( A \).

\[ C \leftrightarrow C^{n+} + ne^- \quad (1) \]

\[ A^{m-} \leftrightarrow A + me^- \]

Figure 4 Electrode-electrolyte interface

A potential difference known as the half-cell potential is determined by the metal involved, the concentration of its ions in solution, and the temperature. The standard half cell potential, \( E_0 \), is the potential for 1M concentration solution at 25°C when no current flows across the interface. When a circuit is constructed to allow current to flow across an electrode-electrolyte interface, the observed half-cell potential is often altered. The difference between the observed half-cell potential for a particular circuit and the standard half cell potential is known as the overpotential. Three basic mechanisms contribute to the overpotential: ohmic, concentration, and activation.

1. **The ohmic overpotential** is the voltage drop across the electrolyte itself due to the finite resistivity of the solution. Overall, this is usually not a big voltage in high concentration solutions.
2. **The concentration overpotential** results from changes in ionic concentration near the electrode-electrolyte interface when current flows. Oxidation-reduction reaction rates at the interface change with excess charge due to a finite current. This modifies the equilibrium concentration of ions changing the half-cell potential.
3. **Charge transfer** in the oxidation-reduction reaction at the interface is not entirely reversible. For metal ions to be oxidized, they must overcome an energy barrier. If the direction of current flow is one way, then either oxidation or reduction dominates, and the height of the barrier changes. This energy difference produces a voltage between the electrode and the electrolyte, known as the activation overpotential.

The overpotential of an electrode is then given by the sum of these three polarization mechanisms:

\[ V_p = V_r + V_c + V_a \quad (2) \]

where \( V_r \) is the ohmic overpotential, \( V_c \) is the concentration overpotential, and \( V_a \) is the activation overpotential. Note that overpotentials impede current flow across the interface. A way to minimize \( V_p \) is to use nonpolarizable electrodes. These allow conduction current to flow across the interface with no energy exchange and there are no overpotentials for this type of electrode. The best electrode to use for all possibilities for biological electrode system is the silver/silver chloride (Ag/AgCl) electrode. This is made of a silver metal base with attached insulated lead wire coated with a layer of the ionic compound AgCl. The electrode is then immersed in an electrolyte bath in which the principle anion of the electrolyte is Cl-. For the best results, the electrolyte solution should also be saturated with AgCl so there is little chance for any of the surface film on the electrode to dissolve. Cl- is an attractive anion for electrode applications with mammals since these animals (including humans) have an excess of chloride ions in solution. The electrode-electrolyte interaction is described by the reaction

\[ Ag^+ + e^- \leftrightarrow Ag + \quad (3) \]

And the Nerst equation for the reaction can be written as:

\[ E = E_0 + \frac{RT}{nF} \ln \left( \frac{[Ag]}{[Ag^+]} \right) \quad (4) \]

The first two terms on the right side of this last expression are constants - only the third is related to ionic activity. In biological systems, the large chlorine ion concentration makes its activity fairly constant. This means that the half-cell potential for this electrode is quite stable for biological systems. In this article we will only consider low current densities, and consequently the electrode-electrolyte interface can be modeled as a linear system with an equivalent circuit composed exclusively of linear components (i.e., voltage/current sources, resistors, capacitors and inductors). The terminal characteristics of an electrode have both resistive and reactive components.

Figure 5 Equivalent circuit for a biopotential electrode in contact with an electrolyte. \( E_{hc} \) is the half-cell potential, \( R_d \) and \( C_d \) make up the impedance associated with the electrode-electrolyte interface and polarization effects, and \( R_s \) is the series resistance associated with interface effects and due to resistance in the electrolyte.
Fig. 5 shows the equivalent circuit of the electrode-electrolyte interface. In this circuit Rd and Cd represent the resistance (i.e., conduction currents) and the capacitance (i.e., displacement currents) respectively resulting from the double-layer of ionic charge at the electrode-electrolyte interface. The resistance Rs is the series resistance associated with equivalent losses in the electrolyte itself.

There are three primary layers in the skin. The outermost layer, or epidermis, plays the most important role in the electrode-skin interface. It is a constantly changing layer, the outer surface of which consists of dead material on the skin’s surface with different electrical characteristics from live tissue. The deeper layers of skin contain the vascular and nervous components of the skin as well as the sweat glands, ducts, and hair follicles. These layers are similar to others in the body, and with the exception of the sweat glands, can be modeled as equivalent to the electrical characteristics of the rest of the viscera. Given this anatomy, a general equivalent circuit describing the characteristics of both the electrode-electrolyte interaction and the connection to the skin can be developed, as illustrated in figure 6.

Figure 6 Total electrical equivalent circuit obtained for a body-surface electrode placed against skin

The epidermis can be considered a semipermeable membrane to ions, so a potential given by the Nernst equation, can be developed if there is a difference in ionic concentrations across this membrane. The dermis and subcutaneous layer under it behave in general as pure resistances. They generate negligible DC potentials. Finally, the electrical characteristics of the sweat glands must also be taken into account for a complete model of a skin electrode. The fluid secreted by sweat glands contains Na+, K+, and Cl- ions, the concentrations of which differ from those in extracellular fluid. This produces a potential between the lumen of the sweat duct and the dermis and subcutaneous layers. There is also a parallel RpCp combination with this potential representing the wall of the sweat gland and duct. This equivalent model has been used by many authors to simulate the electrical behaviour of the electrode-skin interaction.

4. The Non-invasive Probe and the pressure compensation system

To realize the measurements, a noninvasive probe has been designed. The probe is illustrated in Fig. 7 and its electrode is a rod 2 mm in diameter made of casting of Ag-AgCl powder. The rod must minimize the skin pain to the patient during the measurements; to achieve this goal it has been decided to use a semisphere shaped rod instead of a cylindrical one because the last one could injure the skin at some probe-skin angles.

Figure 7 The realized probe

The contact surface will increase or decrease according to the pressure and pressure unbalance will introduce an error at a rate proportional to the unbalance itself. Also the contact resistance will be influenced by the pressure. It can be easily seen that an error of more than 100% of the measured current can be introduced for a pressure variation of about 300 grams. Assuming that the reliability of the measure must be as high as possible, the above mentioned error cannot be tolerated, hence a pressure compensation system must be introduced to make the measure almost insensitive to pressure changes for a wide span of values. It was then studied a pressure compensation system to allow the operator to push the probe on the measure point to a well defined and repetitive pressure.

Figure 8 Probe dimensions
The presence of a pressure sensor allows the operator to work in a linear range of pressures, namely between 100 and 250 grams, in order to get the contact surface variation in a linear range. The probe tip holder can slide in its guide for about 1mm to transmit the force on the pressure gauge, but at the same time, a perfect guide sealing must be provided in order to avoid the saline solution to penetrate the inner space of the housing of the probe. A sealing grease should have in any case very low friction, in order to assure a complete repositioning [reset] of the tip after measures. The residual pressure on the sensor must be at any rate checked by the application software before any current measure. Figure 8 shows the probe dimensions.

5. Data validation

As seen in thousands of measures, the current emitted by acupuncture points of the body, when measured in a short circuit, decreases asymptotically from a peak value to a lower steady state value, as a function of many physiological parameters and conditions. Measured data validation is therefore demanded to a software procedure. A software has been then developed to calculate the standard deviation of the data in a given number of samples. When the standard deviation is lower of a preset threshold, the current value is accepted and displayed. Many characteristics points on different subjects have been tested and some results are reported in the following paragraph.

As an example, figure 9 shows some results obtained measuring for ten times the current emitted by two characteristics points (point A and point B) of a male subject. Measurement were taken every 30 seconds with a nominal pressure of 200 gr. Point A is the point LU11(Shao Shang) of the Lung Channel and is located on the radial side of the thenar eminence, 0.15 cm posterior to the nailbed. Point B is the point HE9 (Shao Chong) of the Heart Channel and is located on the radial side of the fifth digit, 0.15 cm from the corner of the nail bed. Figure 10 shows the locations of the two points on the hand of the subject and the locations of the two channels along the arm.

6. Results

The probe was used to test some chosen characteristics points. The currents were recorded using BE 100-04 apparatus (patented by Biophysics Research srl, Rome, Italy), set for the appropriate measuring mode. All the measurements were done holding the probe in an upright position, at an angle of about 80 degrees with the skin, with the tip in contact with the acupuncture point to be tested. The majority of tests took place during the day between 9am and 6pm. Table 1 and Figure 11 show the results of measuring for 8 times the currents emitted by Point A at different moments during the day. All the indicated currents are in nA and the table also provides the standar deviation (SD) and the mean value (MV). Table II shows the results of measuring the currents emitted by Point A in 12 different subjects for 10 times with a time step of 1 minute. Figure 12 illustrates the measurements on the 12 subjects.

![Figure 9 Current deviation](image)

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![Figure 10 Location of point A and B](image)

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![Table 1 Different current measurements during the day](image)

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![Figure 11 Different current measurements during the day](image)
Table 2 Different current measurements in 12 subjects

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References


Figure 12 Measurements of current emitted by Point A in 12 different subjects.

7. Conclusion

This study shows that small electric currents can be measured at characteristic points on the skin in a replicable manner with a non-invasive probe. A correct evaluation of the magnitude of these currents could be significant for a diagnosis of the organs correlated with the points. More investigation need to be done in order to understand the correlation between any measure point and the activity of the body organs.