A Bioelectrical Sensor for the Detection of Small Biological Currents.

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Abstract
A novel instrumentation system has been designed for the correct detection of small dc currents emitted by characteristic points of the human skin using a non invasive probe. The probe presents an integrated pressure compensation system based on a force sensor allowing to obtain reliable and repeatable measurements that are virtually insensitive to pressure fluctuations. The current acquisition circuit and the pressure compensation system are illustrated together with experimental results.

Keywords: Biosensors, Non-invasive measurements, Biopotential Electrodes, Bioelectrical Currents, Small Localized Biosources.

1 Introduction
The objective of this work was to design and build a non-invasive probe and an amplifier for biopotential recording from characteristic points located on the body surface, considered here as small localized biosources (SLB). This non-invasive sensor aims to obtain its information without a physical penetration of the skin. To eliminate the possibility that the operator may influence the testing process through unconscious variations in pressure, the probe has an integrated pressure compensation system based on a force sensor. This system allows the operator to push the electrode of the probe on the measure point to a well defined and repetitive pressure obtaining reliable and repeatable measurements. These are virtually insensitive to fluctuations in pressure over a wide range of values and operating at all time under specified conditions. The probe can be used to detect small dc currents emitted by points of the human skin. The specific points of interest (acupuncture points) are described in literature as being one millimeter in diameter and their electrical characteristics have been extensively studied since the 1950s to establish their existence by scientific methods \[1\]-\[4\]. To date, the bioelectric properties of acupuncture points are not clearly understood but they are frequently described as having distinct electrical properties. \[5\]-\[7\]. These properties include increased conductance, reduced impedance and resistance, increased capacitance, and elevated electrical potential compared to adjacent points \[8\]-\[9\]. This assertion has been used as a means to localize and analyze acupuncture points for diagnostic purposes \[10\]. Yet, the electrical characterization of acupuncture points is associated with important technical issues that are often overlooked. Electrode polarizability, stratum corneum impedance, presence of sweat glands, choice of contact medium, electrode geometry, and other factors contribute to the final electrodermal reading and may cause doubts about the validity of electrodiagnostic devices \[11\].

2 Generality on Bioelectrical Sensors
Electrical signals produced by various bodily activities are commonly used in monitoring and diagnosis \[12\]. Measured signals are generally compared with normal reference signals and variations in signals can be useful to provide accurate diagnosis. The most common devices include the electrocardiography (ECG) – electrical activity of different sections of the heart, the electroencephalography (EEG) – electrical activity of the brain and the electromyography (EMG) – electrical activity of the muscles. In general biomedical sensors take signals representing biomedical variables and convert them into what is usually an electrical signal. As such, the biomedical sensor serves as the interface between a biologic and...
an electronic system and must function in such a way as to not adversely affect either of these systems. In considering biomedical sensors, it is necessary to consider both sides of the interface: the biologic and the electronic, since both biologic and electronic factors play an important role in sensor performance. Sensor convert signals of one type of quantity such as hydrostatic fluid pressure into an equivalent signal of another type of quantity, for example, an electrical signal. Sensors of electrical phenomena in the body, usually known as electrodes, play a special role as a result of their diagnostic and therapeutic applications. There are, however, special problems that are encountered by biomedical sensors that are unique to them. These problems relate to the interface between the sensor and the biologic system being measured.

3 The non-invasive probe

The main parts of the realized instrumentation system are the probe, the electronics interface and the computation unit. The whole system can be modelled using the simple diagram illustrated in Figure 1.

![Figure 1: Block diagram of the system.](image)

The instrument must interface with the skin tissue through the direct contact of the non-invasive probe. Main purpose of the probe is the detection of biochemical currents providing at the same time a safe interface with the skin. The electronics interface matches the electrical characteristics of the measurement with the computation unit, preserving the efficiency of the probe and providing secondary signal processing functions. Finally, the computation unit provides primary user interface and control for the overall system thanks to a Graphical User Interface and to specific software. It also provides data storage, primary signal processing functions and maintains safe operation of the instrument. The software implementation combines a variety of graphical techniques to create a powerful system that will enable users to perform an accurate and reliable analysis of the emitted currents and to easily go on to further applications and research. Figure 2 shows the non-invasive probe that has been designed in order to realize the measurements. The electrode of the probe is made of casting of Ag-AgCl powder and it is actually a transducer, converting ionic currents in the body into electronic currents in the probe. This need to be done avoiding the formation of spurious potentials that would contaminate the measurements (overpotentials). In particular, the measurement technique must not depend on the skin ph and on the internal resistance of the characteristic point.

![Figure 2: The non invasive probe.](image)

The overpotential of the electrode is given by the sum of three polarization mechanisms: ohmic, concentration, and activation overpotentials:

\[
V_p = V_r + V_c + V_a
\]

where \( V_r \) is the ohmic overpotential, \( V_c \) is the concentration overpotential, and \( V_a \) is the activation overpotential [13]. These overpotentials impede current flow across the interface and need to be minimised. A way to minimise \( V_p \) 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. Two different shapes have been analyzed for the electrode; initial experiments have been performed using a 2mm diameter cylindrical tip. It has been found that this shape could injure the skin during the measurements at some probe-skin angles and a second semispherical tip, also with a 2 mm diameter, has been then used to minimise the pain of the patient. As it can be seen in Figure 3, the contact area between the semispherical tip and the skin changes according to the force exerted by the operator. In a) the tip is leaning on the skin and the contact surface is just a single point; in b) a certain force is exerted on the tip and the contact surface increases; in c) the tip is pushed on the skin with a value of the force that causes the maximum contact surface. In the figure the probe is always kept vertically to the skin; changing the inclination of the probe leads to changing in the contact surfaces. The contact resistance is then function of the pressure and of the contact surface. If the pressure of application exceeds a certain value, traumatisation of test points could occurs through repeated application of the test electrode. Thanks to the pressure compensation system the value of the pressure is always kept to a well defined and safe value.
The current acquisition circuit

The acquisition of small dc current using the realized probe requires some precautions in order to minimize the error on the measured data. The characteristic point is considered as a small localized bio-source emitting a small dc current and it can be modeled as a current generator $I$ and its internal resistance $R_i$ [14]. Figure 4 shows a simplified schematic of the acquisition system. The main components of the scheme are the current generator, the probe and the electronic interface.

The current generator is in series with the skin and connected to a current operational amplifier through the contact tip of the probe $P$, being $v_0$ the output voltage of the amplifier. The component in the red dashed square in figure 4 represents the skin/electrode interface and the biopotential electrode. To describe the electrical behaviour of the skin/electrode interaction it can be used the equivalent model of Figure 5. The right part of the model is the equivalent circuit of the biopotential electrode. $E_{hc}$ is the half-cell potential, $R_e$ and $C_e$ make up the impedance associated with the electrode-electrolyte interface and polarization effects.

![Figure 5: Equivalent model used to describe the electrical behaviour of the skin/electrode interaction.](image)

Figure 6 shows the variation of the impedance of the electrode-electrolyte interface with the frequency.

![Figure 6: Variation of the impedance of the electrode-electrolyte interface with the frequency.](image)

The epidermis is the outermost layer and plays the most important role in the electrode-skin interface. Its equivalent circuit is similar to the one of the electrode (associated impedance $R_{ep}$, $C_{ep}$ and overpotential $E_{ep}$). 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 contain the vascular and nervous components of the skin as well as the sweat glands, ducts, and hair follicles. Those are taken into account also by a parallel $R_gC_g$ combination with a potential $E_g$ representing the wall of the sweat gland and duct. The dermis and subcutaneous layer under it behave in general as pure resistance ($R_{de}$) and generate negligible dc potentials. The characteristic resistance of the skin varies from about 100K to 1 MΩ. Acupuncture points present a lower resistance (10-50 KΩ). Thanks to the semi-spherical shaped tip of the electrode, the probe can directly enter in contact with the skin within a certain pressure range and the equivalent resistance of the contact spot can be considered in series to the current generator. As illustrated in the previous figure 3, different values of pressure on the skin lead to
different equivalent contact surfaces and to a different equivalent contact resistance. 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. 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 and a pressure compensation system must be introduced to make the measure almost insensitive to pressure changes for a wide span of values.

5 The pressure compensation system

As seen in the previous paragraph the reliability of the measure must be the highest as possible, hence to get repetitive measurements, the error introduced by different values of pressure cannot be tolerated. To solve this problem there are two possible ways: the first and simplest solution is to keep the pressure always to a constant value. In manually executed measurements this is not often feasible and the operator may influence the testing process through unconscious variations in pressure. The second solution consists in compensate the measurement making it a function of the exerted pressure. A pressure compensation system has been then studied to allow the operator to push the probe on the measure point to a well defined and repetitive pressure. The block diagram of the compensation system is shown in figure 5. A force transducer S is connected to an operational amplifier that provides an output voltage $V_p$ proportional to the pressure on the tip.

![Figure 5: Schematic of the pressure compensation system.](image)

The force sensor operates on the principle that the resistance of a silicon implanted piezoresistors will increase when the resistors flex under any applied force. The sensor concentrates force from the application, through a stainless steel plunger, directly to the silicon sensing element. The amount of resistance changes in proportion to the amount of force being applied. This change in circuit resistance results in a corresponding mV output level. The pressure transducer provides precise and reliable force sensing performance using a piezoresistive micromachined silicon sensing element. A low power Wheatstone bridge circuit design provides inherently stable mV outputs over the force range. The sensor has a typical sensitivity of 0.24 mV/g (0.20 mV/g min, 0.28 mV/g max) and supports an operating force from 0 to 1500g. The sensor performance has been evaluated and tested using a deadweight or compliance force in order to maintain operation within design specifications.

The voltage $V_p$ coming from the pressure compensation system and the voltage $V_o$ coming from the acquisition circuit are then connected through a multiplexer to an analog to digital converter used as front-end of a microprocessor µP (see Figure 6).

![Figure 6: Schematic of the final stages of the system.](image)

The microprocessor is programmed to modify the value of $V_o$ according to the value of $V_p$. A conversion algorithm is used to keep into account the transfer function of the force sensor in the dynamic of interest. As a first approximation, a linear function has been used but different algorithms can be analyzed. As far as the range of compensable pressures is concerned, two of the essential elements are constituted by the elasticity of the skin and by the diameter of the semi sphere. Usually a span of 50-300gr can be compensated allowing the operator to work in a linear range of pressures and maintaining the contact surface variation in a linear range. The status of the probe and of the pressure sensor can be at any rate checked by the application software. It is also possible to modify the pressure compensation system parameters. Before performing any current acquisition the application software checks any possible residual pressure on the sensor. Figure 7 shows the complete electronic scheme of the electronics interface. This scheme and the current acquisition system are patented by Biophysics Research srl, Rome, Italy.

6 Results

The realized biological sensor and the pressure compensation system have been tested and evaluated by a group of representative users. The probe has been then used to measure currents emitted from different points of the body on various subjects. The measurements were done holding the probe in an upright position, at an angle of 80 degrees with the skin, with the tip in contact with the characteristic point to be tested. All the measurements shown that the pressure compensation system allows the operator to work on the measure point to a well defined and repetitive pressure obtaining reliable and repeatable measures that are virtually insensitive to fluctuations in pressure over a wide range of values.
The pressure compensation system parameters can be checked and modified by the user through the Graphical User Interface. Moreover, the user can configure different kinds of current measurements, in a range of 30 to 3,000 nA, saving and displaying data in graphical or textual formats and generating a report [15]. The computational unit calculates the standard deviation of the data coming from the acquisition circuit in a given number of samples. When the standard deviation is lower than a preset threshold, the current value is accepted and displayed in the results mask. A vocal synthesis system provides also an audio output of the results. The software package manages different kinds of current measurements, in a range of 30 to 3,000 nA, saving and displaying data in graphical or text formats. All data either in text or in graphics form are stored in a database together with all the information related to the test session.

In order to minimize the measurement error, the developed algorithm is able to automatically compensate the measurement as a function of the pressure exerted on the probe. As can be seen in Figure 8, thanks to the action of the compensation algorithm, the current (red line) remains unaffected by the large variations in the pressure exerted on the probe (blue line). One of the main issues has been the replicability of the measurements. To analyse this question the consistency of the currents emitted by different characteristic points over time for 10 minutes has been studied.

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<th>Time (sec)</th>
<th>Standardized Current (p.u.)</th>
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Table 1: Results of measuring for 10 times the currents emitted by point A.

Table 1 shows the results of measuring for 10 times the currents emitted by point A with a time step of 60 seconds. Point A is the point LU11 (Shao Shang) of the Lung Channel and is located on the hand on the radial side of the thenar eminence, 0.15 cm posterior to the nailbed. All the indicated currents are in nA and...
the table also provides the standard deviation (SD) and the mean value (MV). Figure 9 illustrates the standardized values of the current (in p.u.) showing that no significant change in current amplitude was found during the 10 minutes period with regard to the point or its averaged value.

Figure 9: Standardized values of the currents emitted by point A every 60 seconds.

7 Conclusions
The presented non-invasive electrode features a pressure compensation system designed to improve the detection of small electric currents at characteristic points on the skin. The force sensing element allows to operate over a wide range of values and to obtain measurements insensitive to pressure’s fluctuations. The main components of the realized instrument have been described in the paper and results of measurements have been provided showing that small electric currents can be measured at characteristic points on the skin in a replicable manner. The proportionality between these currents and the homeostatic level of the cells of the organ correlated with the selected points needs to be further investigated in order to understand if the correct evaluation of the magnitude of these currents can be meaningful for diagnostic purposes.

8 Acknowledgements
This work was supported in part by the Portuguese Foundation for Science and Technology (FCT) under Project Nº SFRH/BPD/46224/2008 and Project Nº SFRH/BSAB/950/2009.

9 References