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2-Scale Topography Dry Electrode for Biopotential Measurements

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Abstract—**The design and fabrication of a novel 2-scale topography dry electrode using macro and micro needles is presented. The macro needles enable biopotential measurements on hairy skin, the function of the micro needles is to decrease the electrode impedance even further by penetrating the outer skin layer. Also, a fast and reliable impedance characterization protocol is described. Based on this impedance measurement protocol, a comparison study is made between our dry electrode, 3 other commercial dry electrodes and a standard wet gel electrode. Promising results are already obtained with our electrodes which do not have skin piercing micro needles. For the proposed electrodes, three different conductive coatings (Ag/AgCl/Au) are compared. AgCl is found to be slightly better than Ag as coating material, while our Au coated electrodes have the highest impedance.**

I. Introduction

CURRENT medical practice uses Ag/AgCl coated electrodes with a conductive gel for ECG/EEG biopotential recordings. The gel improves the signal quality by reducing the equivalent impedance of the electrode-skin interface and by suppressing the stretch-related skin motion artifacts (gel acts as a kind of buffer). Such wet gel electrodes unfortunately have several issues. First, the subject's comfort is limited by the gel: electrode/gel placement takes time, the wet electrodes are not very comfortable to wear on the skull (in particular on hairy skin) and after electrode removal the (dried) gel needs to be washed away, which is often not easy. Furthermore, smearing of the conductive gel may lead to shorts when electrodes are placed nearby each other. Finally, frequent gel replacement due to gel drying is required to ensure high quality biopotential recording, and allergic skin reactions are possible. Especially for ambulatory monitoring or for ECG/EEG monitoring over long periods of time, such wet gel electrodes are not ideal at all.

Dry electrode technology is an interesting alternative. A good review of dry electrodes and their evaluation is provided in [1]. A first category of dry electrode is non-contact sensors, which rely on a capacitive coupling to the body rather than on a resistive contact. In [2] Sullivan et al. reported a non-contact, capacitive, EEG sensor that uses active shielding to lower the noise. This capacitive sensor was further improved in the work by Chi et al. [3][4] demonstrating good performances on EEG and ECG signals. Their approach focused on compensating the noise by active electronics, not on optimizing the design of the electrodes. Matthews et al [5] looked at the electrode design, and proposed a hybrid electrode combining resistive and capacitive coupling. The electrodes consist in a set of macroscopic pins (so-called "fingers"), that enable contact through hair. Contact impedance of $10^7\Omega$ was reported.

Another approach to dry electrodes is to use micro-machining techniques to lower the contact impedance of the electrode-skin interface. Special (micro) machining treatments are applied to the contact surface of the electrode, in order to create a needle-like substrate. Macro needles open up possibilities of recording biopotentials even on hairy areas. Micro needles at the skin-electrode interface provide a means to reduce the contact impedance by penetrating the thin stratum corneum (upper layer of skin), which is responsible for the high skin impedance. Needles of different size, shape and material have been reported in literature [6] – [9]. Most papers however are limited to the electrode fabrication process and to a very basic, qualitative performance comparison with the standard wet gel electrode. Moreover, often expensive clean room fabrication processes are used for electrode fabrication. Finally, in literature, there is an important non-uniformity in electrical testing procedures which makes it almost impossible to compare the performance of electrodes from different papers.

This paper describes how micro precision milling and inexpensive molding steps can be used to create a 2-scale dry electrode in a cost-effective way. Further, a fast and accurate impedance measuring protocol is developed to compare and evaluate the contact impedance of various types of electrodes. Finally, our own electrodes are characterized and compared to other available dry and wet electrodes.

II. Materials and Methods

A. Electrode Design

By designing a dry electrode with 2-scale topography *(Fig. 1)* the dry electrode impedance can be optimized even when recording an EEG on hairy places like the skull. Millimeter structuring provides a way to penetrate through hair while the micro needles on top penetrate the outer skin

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layer. It is the outer skin layer or stratum corneum that is known to act as a high impedance barrier. This layer is 10 to 20 µm thick depending on the body area so penetration can be painless if the needle height is well designed.

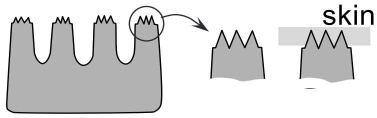

Fig. 1. Proposed design for a 2 scale-topography dry electrode

B. Fabrication procedure

In order to develop a cost effective fabrication technique for this disposable electrode, a vacuum casting technology [7] is used to create the proposed 2-scale design. This fabrication process consists of four steps *(Fig. 2)*.

Step 1: first of all computer numerical control (CNC) micromilling is used to create a metal master pattern. Aluminum was chosen because this soft metal keeps the milling process fairly easy but still provides a strong and hard master. The aluminum master makes it possible to perform subsequent molding without damaging the initial master profile.

Step 2: A molding step is performed in order to vacuum cast a silicone rubber (Sylgard 184) negative mold from the master.

Step 3: In a second molding step the epoxy resin SU-8 (Microchem SU-8 2002) is vacuum cast in the silicone mold to create a replica of the initial master pattern. The use of a flexible Polydimethylsiloxane (PDMS) mold facilitates the extraction of the replica structure, no demolding agents are necessary. The replica material SU-8 is selected because of its biocompatibility [10]. SU-8 is an insulator, a conductive coating is essential to create a functional electrode.

Step 4: Au and/or Ag coatings with a total thickness of 0.9 µm were created by sputtering. AgCl interfaces were created by applying to the Ag coated electrodes a subsequent chloridization step. A 450 mQ/cm2 chloride deposit is applied using a 5 mA/cm2 current density in a 0.9% NaCl solution for a period of 100s [11].

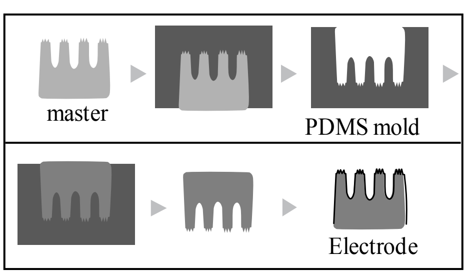

Fig. 2. Schematic representation of the 4 step fabrication procedure

Using the double molding technique electrodes with and without micro needles are fabricated *(Fig. 3)*. The electrode has a circular shape and total diameter of 3 mm. Nine big pillars having a diameter and height of 0.5 mm are pitched 0.75 mm apart. The micro needles on top of the bigger pillars *(Fig. 3b)* are ~80µm high and ~25µm in diameter, the pitch is 155µm. The inner tip angle is 35°.

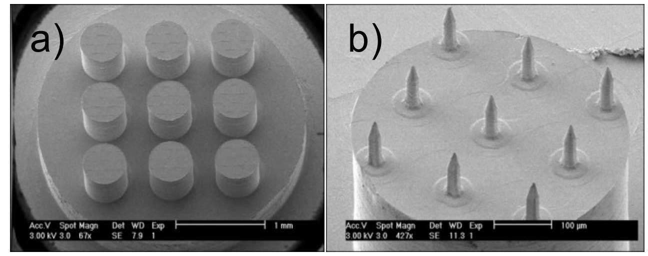

Fig. 3. a) SEM picture of a dry electrode without any micro needles on top of the macro pillars b) Zoomed-in SEM picture of one macro pillar counting 9 micro needles on top

C. Impedance measuring protocol

The Association for the Advancement of Medical Instrumentation (AAMI) provides guidelines and specifications for disposable biopotential electrodes [12], primarily on electrode-to-electrode impedance. To compare different dry electrodes however, the key parameter to consider is the skin-electrode contact impedance. This will directly affect the quality of the recording and its susceptibility to noise. In general, the coupling between skin and electrode can be described as a layered conductive and capacitive structure, with series combinations of parallel RC elements [1]. For dry (contact) electrodes, this model can be reduced to a single RC element.

The electrode characterization is thus performed by measuring the impedance of the skin-electrode contact over a range of frequencies from 0.1Hz to 1MHz. Measurements at very low and high frequencies are required for circuit modeling purposes. It is imperative that the low and high frequency resistive plateaus are known to unambiguously fit the data to a circuit simulation.

A three electrode configuration *(Fig. 4)* is chosen as described in [13]. Electrodes are placed on the forearm. Two out of three electrodes are standard 24mm wet gel electrodes (Z2) from TYCO Healthcare, the third one is the electrode under testing (Z1). Electrodes are placed on the forearm without any skin pretreatment and impedances are measured after 30 min of in situ settling time.

The IVIUM Technologies 'CompactStat: Electrochemical interface and impedance analyzer' is used for the measuring setup. Advantages of this tool are its versatility (it can either be used as a potentiostat or a galvanostat), and its bandwidth, allowing measurement over the frequency range of interest. Using a three electrode setup, a voltage can be applied over the reference electrode (RE) and the working (WE) electrode, respectively connected to Z1 and the non-current carrying Z2 electrode. By connecting the counter electrode (CE) to the current carrying Z2 electrode a closed loop is formed with the three impedances in a star configuration as in *(Fig. 4b)*.

Impedance spectra and values are given in absolute and normalized values, where normalization is done by the area of the electrodes. The absolute value gives an indication of the impedance for a particular electrode configuration. The normalized impedance is however more appropriate to compare different electrode designs, as it removes any geometrical dependencies and is an intrinsic property of the electrode design. For each electrode under test, the geometrical area which is used by the electrodes on the skin has to be considered ('active skin area'). The electrical conducting area is used for normalization purposes. For the wet electrode for instance, only the gel area is considered, not the insulating bandage next to the gel used to stick the electrode to the skin. For all electrodes under test, the 'active skin area' is listed in table I.

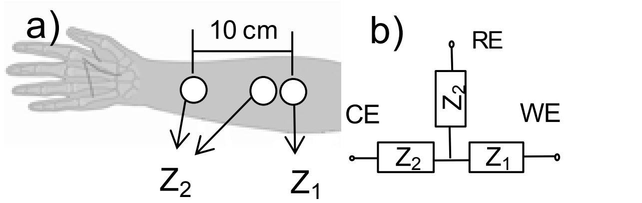

Fig. 4. a) Illustration of the electrode placement. Z2 being a standard wet gel electrodes and Z1 the electrode under testing; b) equivalent electrical circuit used for electrode impedance measurements.

III. Results

A. Electrode Impedance comparison

The impedance of the proposed dry electrodes is compared to commercial off-the-shelf electrodes. Five different electrode types are considered in total: a wet gel electrode, 3 commercial dry electrodes and prototypes of our electrodes. An Ag coated dry electrode without micro needles was used for the comparison *(Fig. 3a)*. Measurement is performed 30 min after electrode placement and an average of 3 measurements is plotted.

When comparing the absolute impedances (*Fig. 5*), the wet gel electrode has the lowest impedance. There is roughly one order of difference between the wet and dry electrodes.

The normalized impedance results plotted in Fig. 6 show that the proposed dry electrode outperforms other dry electrodes, and even achieves lower impedance at low frequencies than a stabilized wet gel electrode (measurement taken 3h after electrode placing). The results shown in Fig. 6 are for our dry electrode with only one scale topography, so improvements are even still possible.

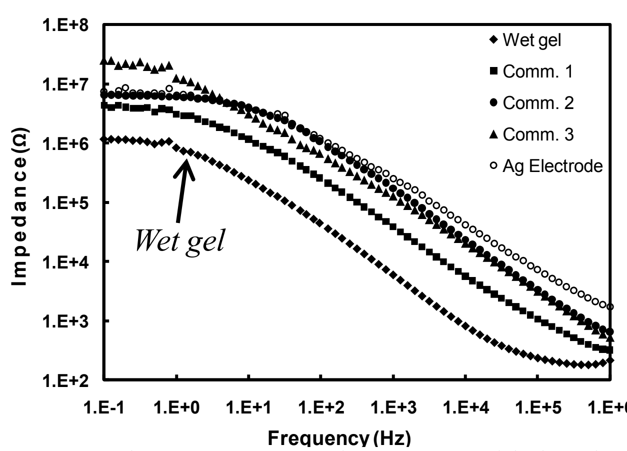

Fig. 5. Impedance spectrum comparison: 3 commercial electrodes, a standard gel electrode & imec's one scale dry electrode with Ag coating

TABLE I

ELECTRODE ACTIVE SKIN AREA

Electrode	Diameter [cm]	Area [cm^2]
Wet gel	1.6	2.0
Comm. 1	1.4	1.5
Comm. 2	1.0	0.8
Comm. 3	1.0	0.8
Ag Electrode	0.3	0.07

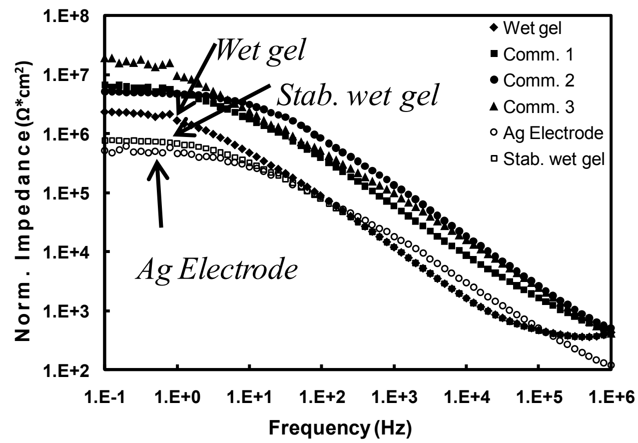

Fig. 6. Normalized electrode impedance comparison. The results from Fig. 5 are multiplied with the respective electrode areas from Table I

B. Comparison of conductive coatings

Three different coating materials are also investigated. An impedance comparison is made between one scale topography coated electrodes (*Fig. 3a*). Au and Ag coatings were created by sputtering. AgCl interfaces were created by applying to the Ag coated electrodes a subsequent chloridization step. Figure 8 shows the impedance comparison for the Au, Ag and AgCl finished electrodes. The Au coating has the highest impedance while the AgCl slightly outperforms Ag for frequencies from 1Hz to 1kHz.

Equivalent electrical circuit fitting is performed using a small adaptation of a basic electrode model (*Fig. 7*). Instead of an ideal capacitance in the parallel RC-element a Constant Phase Element (CPE) is used. This CPE element accounts for distributed surface reactivity and surface inhomogeneity, it is extensively used in fitting of electrochemical impedance data [14]. Figure 9 shows the fitting results on a Cole-Cole plot using the Levenberg–Marquardt algorithm for the 3 different coatings, corresponding parameters are summarized in Table II.

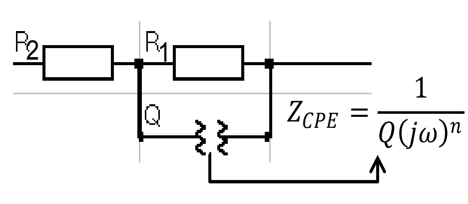

Fig. 7. Equivalent electrical circuit used to fit the data in figure 9

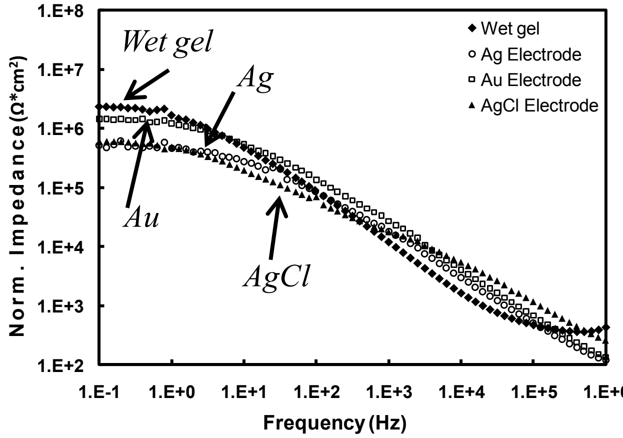

Fig. 8. Impedance comparison of Au, Ag and AgCl coatings for imec's one scale dry electrode

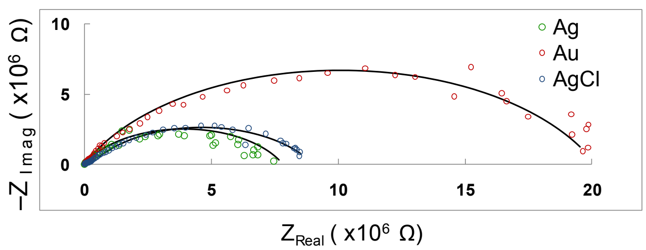

Fig. 9. Cole-Cole plot and fitting of coating impedance

TABLE II
EQUIVALENT ELECTRICAL CIRCUIT FITTING PARAMETERS

	Ag	Au	AgCl
R1	7.80E+06	2.01E+07	8.81E+06
R2	4.00E+02	4.00E+02	2.30E+05
Q	6.77E-12	7.11E-12	5.38E-12
n	7.27E-01	7.455E-01	6.88E-01

C. *Dry electrode evaluation for ECG*

With imec's two lead ECG device and two Au coated dry electrodes (*Fig 3a*), ECG signals were successfully recorded (*Fig. 10*). In the recorded signal the characteristic PQRST complex is clearly visible and the different features can be easily distinguished.

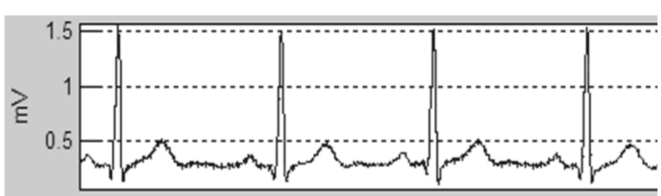

Fig. 10. ECG signal recorded using 2 Au coated dry electrodes

IV. CONCLUSION

A 2-scale electrode design is proposed for recording of bio-potential using dry electrodes. The dry electrodes are fabricated using high precision micro milling and cheap molding techniques. The design addresses the problem of contacting hairy body areas by providing space between the macro pillars. On top of this, each macro pillar can be equipped with multiple micro needles aiming to penetrate the thin stratum corneum, in order to reduce the contact impedance.

An impedance measuring protocol is described to measure and compare electrode impedances in a relevant and reproducible way. The importance of calculating a 'normalized impedance', by taking into account the active skin area for each electrode under test, is illustrated.

Measurements show that the proposed dry electrodes have lower impedance than three commercial available dry electrodes. Impedance is also lower than that of wet electrodes for most relevant frequencies considering ECG and EEG applications (<400Hz). Furthermore, experiments revealed that Ag en AgCl coatings have lower impedances than Au for these frequencies (<400Hz).

Using imec's two lead ECG device and two imec Ø3 mm Au coated dry electrodes without micro needles we successfully recorded good quality ECG signals. The clear PQRST complex in the recordings is a promising result for real ECG applications.

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