SPIKED BIOPOTENTIAL ELECTRODES

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ABSTRACT

We describe the microfabrication, packaging and testing of a dry biopotential electrode (*i.e.* electrolytic gel is not required). The electrode consists of an array of micro-dimensioned, very sharp spikes (*i.e.* needles) designed for penetration of human skin which circumvents high impedance problems associated with layers of the outer skin. Deep reactive ion etching (DRIE) technology was used to fabricate the spikes.

The main advantages of these electrodes include a fast and uncomplicated application procedure, low electrode-skin-electrode impedance (lower than standard electrodes), and comfortable use. The spiked electrode offers a promising alternative to standard electrodes in biomedical applications (*i.e.* monitoring EEG signals) and is of interest in research of new biomedical methods.

INTRODUCTION

Electrodes applied to the skin measuring biopotentials (*i.e.* biopotential electrodes) are extensively used in modern clinical and biomedical applications (*e.g.* electrocardiography, electromyography, electroencephalography (EEG), peripheral nerve compound action potentials and evoked potentials [2]). The importance of the electrode should not be neglected since measurement electronics equipment is likely to display misleading artefacts if an inappropriate electrode is used.

Important drawbacks of standard electrodes are related to skin preparation and the use of electrolytic gel (*i.e.* wet use). Long application times, long stabilisation times (diffusion of the electrolytic gel into the skin), long cleansing times (removal of the gel after use), low comfort and large electrode size complicate the use of standard electrodes. Research into eliminating or reducing the inconveniences of standard electrodes has lead to several new approaches to measuring biopotentials including NASICON ceramic electrodes [4], on chip amplified dry electrodes [5, 6] and the Zipprep™ electrode.

In this paper an entirely new approach is presented. The micromachined electrode includes an array of micro-dimensioned spikes designed to pierce the outer skin layer (*Stratum Corneum SC*). The high impedance characteristics of the *SC* is thereby circumvented and skin preparation and electrolytic gel is not needed. This results in significantly shorter and easier application- and cleansing procedures as well as shorter stabilisation times. Since spiked electrodes can be applied to the skin without gel or additional conducting media, they are also attractive for long-term measurements.

The electrode consists of a micromachined Si chip (electrode chip) and a simple package which improves the attachment of the lead wire to the electrode chip as well as the attachment of the electrode to the skin.

BIOMEDICAL BACKGROUND

Biopotentials result from the electrochemical activity of a certain class of cells, known as excitable or active cells, which are part of nervous, muscular or glandular tissue. Within the human body, active cells are surrounded by body fluids having a high Cl^- concentration. An active cell acts as a constant current source when stimulated and creates an ionic current within the body fluid. This current induces electrical potentials within the human body. These biopotentials decrease in amplitude with increasing distance from the active cell. The detection of biopotentials is therefore closely related to the detection of the ionic current created by the active cells. As modern electronics (*i.e.* EEG-Amplifier) require electronic current, a biopotential electrode is a transducer transforming ionic current into electronic current. This current transformation is possible when an electrochemical electrode-electrolyte interface can be established (*i.e.* the human body fluids act as the electrolyte and the metallic biopotential electrode as the chemical electrode [1]).

Skin anatomy is another important topic to understand the basics of biopotential measurements. The skin shows a layered architecture. The outer skin layer

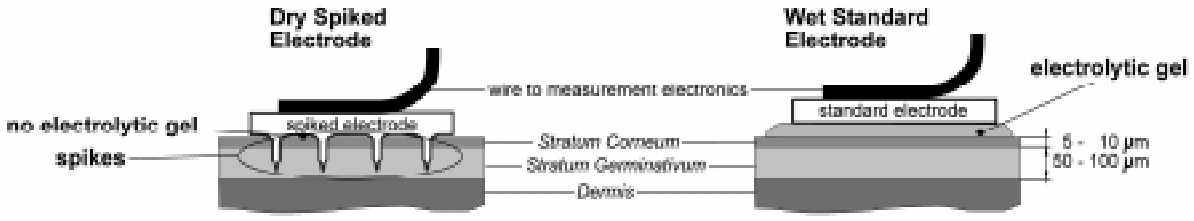

Figure 1: Comparison of the use of the presented spiked electrode a) with a standard electrode b). In b), electrolytic gel is needed to measure biopotentials (decrease of the isolating characteristic of the *Stratum Corneum* by diffusion of the gel). In a), the spikes penetrate into the electrically conducting *Stratum Germinativum* to sense biopotentials. The *Dermis* contains nerves and blood vessels. Note that the spiked electrode does not require electrolytic gel (*i.e.* dry use) which is in contrast to the standard electrode (*i.e.* wet use).

(*Stratum Corneum SC*) acts as a fluid barrier and therefore has electrical isolation characteristics. This layer is constantly renewing itself and consists of dead cells (Figure 1). The *Stratum Germinativum* (*SG*) is the area where the cells divide, grow and are displaced outward to the *SC*. Since the *SG* is composed of living cells which predominantly consist of liquid, this layer of the skin is an electrically conducting tissue comparable to an electrolyte (Figure 1). The *Dermis*, which is below the *SG*, contains vascular and nervous components as well as sweat glands and hair follicles and is also electrically conducting (Figure 1). It is in the *Dermis* where pain has its origins.

When a metal electrode is applied on an unprepared surface of the skin, very high Electrode-Skin-Electrode Impedances (ESEI) result since no direct electrochemical electrode-electrolyte interface with the body fluids can be established. Therefore, standard biopotential electrodes, not having micro-spikes, demand special skin preparation prior to the application of the electrode. The two most common preparation methods are 1) abrasion of the *SC* and 2) the use of electrolytic gel. The goal of abrasion is to reduce the thickness of the *SC*. However, complete removal of the *SC* is painful and not recommended. Alternatively, an electrolytic gel having a high concentration of conductive ions can be applied to the *SC*. The gel diffuses into the *SC* and improves its conductivity.
The combination of the above stated preparing methods gives the possibility of reducing the ESEI by creating an electrode-electrolyte interface. Important drawbacks of these methods are obvious: long application times, long stabilisation times (diffusion of the electrolytic gel into the skin), long cleansing times (removal of the gel after use), and discomfort.
It is now easy to understand that an electrode containing micro-spikes (or needles) penetrating into the *SG* (but not the *Dermis*) has good sensing performances since a direct electrode-electrolyte interface can be established in the *SG* (Figure 1). Furthermore, such an electrode would not have the drawback of time consuming skin preparation and would not be painful.
Biopotential electrodes that are not based on an electrochemical electrode-electrolyte interface have been published [5]. These biopotential electrodes are active (*i.e.* signal amplification on the electrode) and use capacitive detecting principles. The main drawback of active electrodes are their high cost and complexity.

CONCEPT

The goal of the spiked electrode is to pierce the *Stratum Corneum* (*SC*) and to penetrate the electrically conducting *Stratum Germinativum* (*SG*) in order to circumvent the high impedance characteristics of the *SC*. Skin preparation and electrolytic gel is therefore not needed (Figure 1).
Direct batch fabrication of the spikes using silicon (Si) has been chosen. The length of the spikes must take into account the layered architecture of the skin. The spikes must not reach the tissue layers below the *SG* (containing nerves and blood vessels) so as to avoid pain or bleeding. The thickness of the *SC* is approximately 10 to 15 µm. The thickness of the *SG* is about 50 to 100 µm [3]. Thus, spikes penetrating the skin more than 10-15 µm but less than 50-100 µm produce a pain-free electrode-electrolyte interface at the *SG* and transform the ionic current induced by active cells into an electronic current.
To reduce the most important noise source in an electrode-electrolyte interface, *i.e.* polarisability [8], the spikes are coated with a silver-silverchloride (Ag-AgCl) double layer where only the AgCl is in contact with the electrolyte. The advantages of Ag are low electrical resistivity and biomedical compatibility.

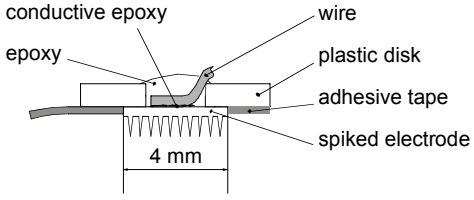

Figure 2: Concept of the spiked biopotential electrode

The spiked electrode is connected to the analysing electronics by a lead wire. As the wire diameter is larger than the length of the spikes, it must be attached to the back side of the electrode to avoid any influence

on the proper penetration of the spikes into the skin. This implies that an electrical interconnection of the front side and back side of the electrode chip must be established. Therefore, through-holes were fabricated and both sides were coated with Ag, paying special attention that the front side and back side Ag coating overlap. This was found to be the simplest and most effective way to do electrically interconnect front and backside.

Dry use of the spiked electrode without any skin preparation presupposes that the electrode chip is securely attached to the skin allowing the spikes to pierce the *SC* and penetrate the *SG*. A simple package fulfilling this requirement consists of a thin, circular disk glued to the back side of the electrode chip. The diameter of the disk exceeds the diagonal dimension of the electrode chip. A ring-shaped adhesive tape is fitted onto the disk and firmly adheres the disk and the electrode chip to the skin. The lead wire is connected to the back side of the electrode chip through a hole in the centre of the disk.

FABRICATION

The fabrication of spikes standing out of the wafer plane has been reported before [3, 7, 9]. Our spikes were manufactured at wafer level using a new three step deep reactive ion etching (DRIE) process. First, a circular oxide mask, patterned using standard photolithography, was isotropically underetched (Figure 3a). Second, an anisotropic process defines the spike-length (Figure 3b). Third, a second isotropic etch, which was stopped before the mask was completely underetched, smoothens the spikes and gives them a shape allowing easy penetration of the *SC* (Figure 3c). As the underetching is not perfectly uniform, this step was stopped before the oxide masks of early finished spikes are completely underetched. If they would fall down and stick to the sidewalls of the spikes, the sharp tip would be severely damaged. This process allows good control of the spike shape, length and diameter. For one wafer, dimensions vary less than ±2%. The length and diameter of the fabricated spikes range from 100 to 210 µm and 30 to 50 µm, respectively.

Through-holes were KOH-etched using a thermally grown oxide as a stop layer (Figure 3d) and a (111) wafer glued with black wax as a front side protection. In addition to acting as a stop layer, the oxide was used for a controlled removal of the circular oxide masks and for sharpening the spikes.

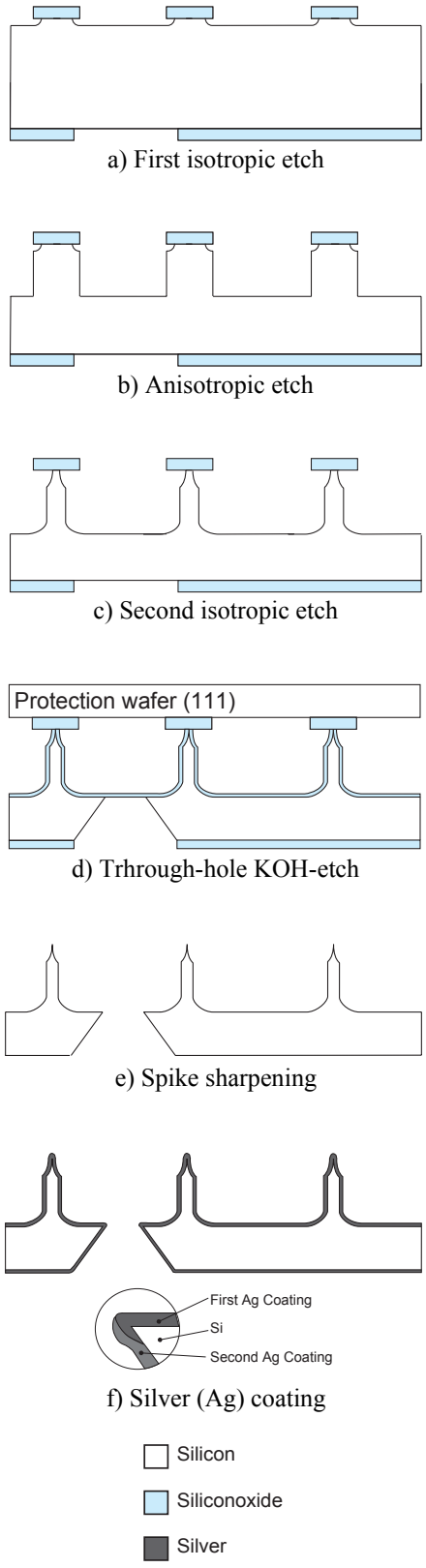

Figure 3: Process flow performed on wafer level. a)-c) is a three step DRIE process. The detail of f) shows the electrical interconnection of front and backside obtained by two successive evaporation steps. The two silver layers overlap at through-hole level.

By etching the oxide with HF, extremely sharp tip radii of less than 0,5 µm can be obtained (Figure 3e).

A double-sided silver evaporation process using a mechanical system which tilts and rotates the wafer (Figure 4) was developed to coat the front- and back side of the electrode chip. This convenient process allows overlapping of front- and back side coating at the through-hole while maintaining a uniform coating of the spike sidewalls. Hence, electrical interconnection of front side and back side as well as complete homogenous spike coating is obtained. The detail of Figure 3f demonstrates the overlapping of the two Ag layers at the through-hole level and Figure 5 shows a SEM picture of a silver coated spiked electrode chip front side.

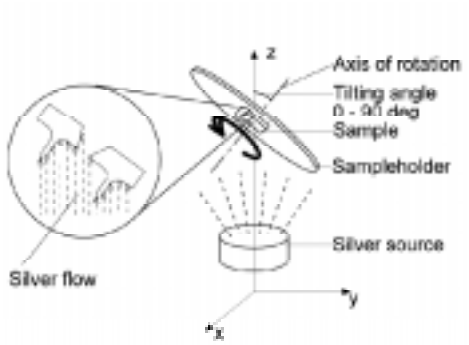

Figure 4: Silver evaporation process using a tilting and rotating mechanism. Very uniform coating of the entire spike as well as overlapping of the two successive evaporation steps at the through-holes can be obtained with this process.

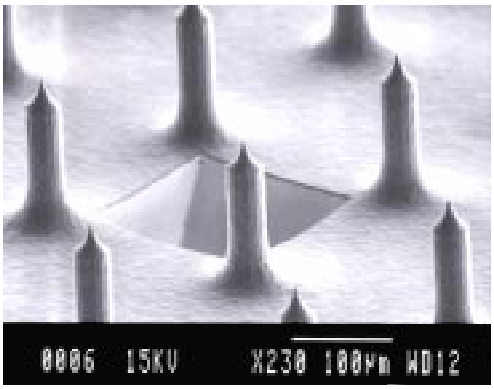

Figure 5: SEM picture of silver coated spikes. The length of the spikes is 160 µm, the diameter is 40 µm. Note the through hole in the center of the picture. This device is ready to be packaged

After dice sawing the wafer, the electrode chips were glued onto a plastic ring with standard epoxy. The lead wire was attached with conducting epoxy and secured with standard epoxy.

Subsequently, AgCl is grown on the Ag via an uncomplicated electrochemical cell process [8]. It required a 0,9 molar Cl^- solution in which a pure Ag reference electrode and the spiked electrode chip to be chlorided were dipped. By applying a constant current, the spiked electrode chip is chlorided. The control of current-density and process time controls the thickness and quality of the AgCl layer. Finally, the double side self adhesive tape was added to the plastic ring. At this point, the spiked electrode is ready for biopotential recordings (Figure 6).

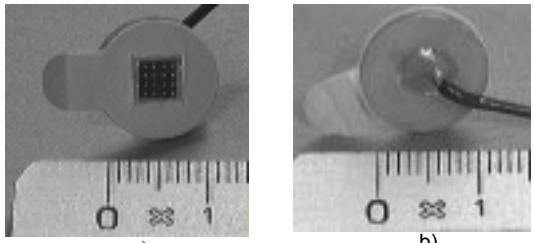

Figure 6: Photographs a) and b) show both sides of the spiked electrode ready to be applied to the skin. As the silver is covered by silverchloride, the electrode chip is dark coloured. The packaging measures 10x10x2.5 mm.

EXPERIMENTAL

The most important characteristic of high performance biopotential electrodes is low electrode-skin-electrode impedance (ESEI). Measurements were conducted to prove that dry spiked electrodes show ESEI as low as wet standard electrodes, are less complicated to use, faster to apply and more comfortable. Furthermore, EEG measurements were recorded to check the usefulness of spiked electrodes in a clinical environment.

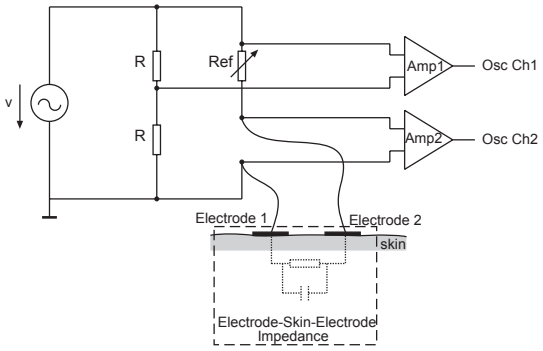

Figure 7: Measurement set-up for evaluating the ESEI of biopotential electrodes

The measurement set-up for evaluating the ESEI is showed in Figure 7 and should not be confused with biopotential recording. Signal v is a sinusoidal alternating voltage. R_{ref} is manually adjusted so the peak to peak value of the output voltage of Amp1 and Amp2 are equal. Knowing the temporal phase-shift of Amp1 and Amp2 and the value of R_{ref}, the ESEI (real and imaginary part) can be determined. The ESEI is dependent on the measurement frequency. In the presented measurements, the frequency of v was chosen to range from 0,5 to 500 Hz which includes the interval of interest for biomedical applications. The RMS value of v is below 60mV so that it does not interfere with the membrane potential of active cells, so that there is no risk of harm for the test person and so that current densities can be kept on a level as low as it is comparable to currents during biopotential recordings. This is important since the electrical conductivity of the skin is dependend on the current density. Such low voltage levels imply that the measurements must be conducted in an environment devoid of extrinsic noise (*e.g.* in a Faraday Cage). For all ESEI measurements, the spiked electrode chips have a size of 4x4mm^2.

Standard EEG measurement recordings were conducted with spiked electrode sizes of 4x4 mm^2 and 2x2 mm^2. Electroencephalographic measurements exhibit the lowest biopotential levels and therefor need low ESEI to obtain reliable recordings of the brain activity.

RESULTS AND DISCUSSION

Figure 8 clearly demonstrates the reduction of the ESEI by the action of the spikes by a factor of at least 13 when comparing a *spiked electrode with a standard one, both applied without gel and skin preparation*.

In Figure 9, the ESEI of *dry spiked electrodes* is reduced by a factor of at least 1,3 *compared to standard electrodes using gel.*

Figure 10 represents recorded raw EEG data comparing a wet standard electrode and a dry spiked electrode. Both recordings are very similar and show that the spiked electrode is suitable for EEG measurements. An EEG recording with eye movements was used to better show the similarity of both signals. The size of the spiked electrode chip can be reduced from 4x4 mm^2 to 2x2 mm^2 while maintaining very satisfactory results for these EEG measurements.

Captured electrical signals are stable with low noise levels even at low frequencies.

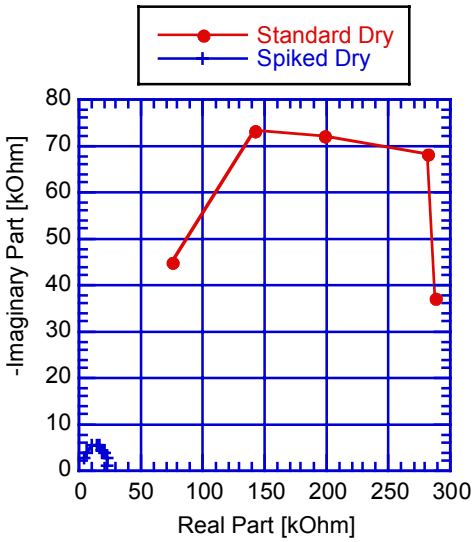

Figure 8: Nyquist plot which shows the influence of the spikes on the measured ESEI. The spiked electrode is compared with a standard electrode, both without gel (*i.e.* dry) and without skin preparation. The decreasing of impedance by the action of the spikes (both real and imaginary part) demonstrates clearly the advantages of spiked electrodes.

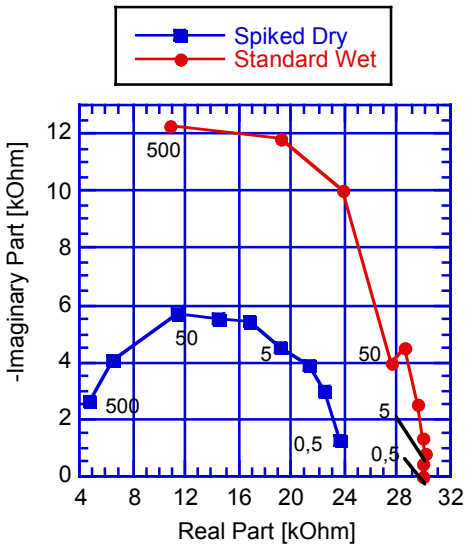

Figure 9: Nyquist plot of the ESEI of a standard electrode using gel (standard wet) and of a spiked electrode without gel (spiked dry). The numbers beside the curves indicate the measurement frequency. The spiked electrode shows lower impedance than standard ones.

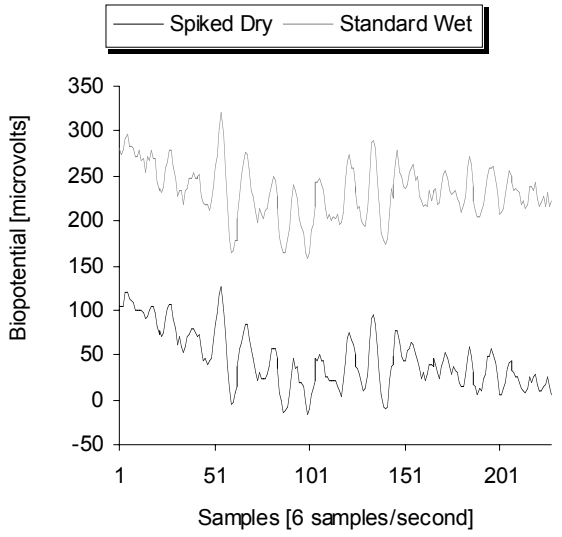

Figure 10: Comparison of EEG raw data recorded by a dry spiked electrode and a wet standard electrode. Note that the signals are very similar.

The packaging showed very good performance during testing. The electrode was firmly attached to the skin and allowed the spikes to penetrate the skin. The spikes can withstand even severe handling, therefore, experimental as well as clinical use of the spiked electrode is possible. The application procedure of the micromachined electrode is easy and fast while no pain is induced and stabilisation times are shorter compared to standard electrodes.

Low ESEI, short stabilisation times and uncomplicated use without skin preparation are important characteristics of the spiked biopotential electrode and were confirmed during the experiments.
EEG recording is possible even with very small electrode sizes (2x2 mm^2). The fact that low signals of EEG can be sensed indicates, that the spiked electrode can be used for other biomedical applications since, in general, they have stronger signals. No standard biopotential electrodes reliably measuring biopotentials on the surface of the skin and showing dimensions as small as 2x2 mm^2 have been found by the authors. ESEI measurements for 2x2 mm^2 electrodes are not meaningful since current densities are higher than for actual biopotential recording.

CONCLUSION

Silicon based spiked biopotential electrodes have been successfully manufactured. The electrodes contain an array of micro-dimensioned, very sharp spikes (*i.e.* needles). Electrode-Skin-Electrode Impedance measurements and EEG recordings confirmed that spiked electrodes do not need any skin preparation or electrolytic gel to achieve better performances than standard electrodes. This results in great advantages of spiked electrodes, *i.e.* uncomplicated and fast application on and removal from the skin, short stabilisation times and comfort during use. Furthermore, the size of the spiked electrode can significantly be reduced compared to standard electrodes.

The spiked electrode is a promising device in various biomedical applications.

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