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New dry electrodes based on iridium oxide (IrO) for non-invasive biopotential recordings and stimulation

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ABSTRACT

This paper presents a new type of dry electrodes for acquisition of biopotentials and stimulation. These dry electrodes are composed by 16 microtip structures (by forming an array of 4 × 4 microtips), which were fabricated through bulk micromachining of a ⟨1 0 0⟩-type silicon substrate in a potassium hydroxide (KOH) solution. The fabrication process was trimmed in a way that each microtip presents solid angles of 54.7°, a width in the range 150–200 μm, a height of 100–200 μm, and an inter-microtip spacing of 2 mm. The electrodes have a thin layer (obtained by reactive DC-sputtering) of iridium oxide (IrO) to improve the contact with the skin. These dry electrodes penetrate the outer skin layer (i.e. *stratum corneum* that is 10 μm thick) to allow a direct contact with the electrolyte fluids of the inner skin layers. The new electrode avoid the use of conductive gels and reduce the skin preparation time in EEG experiments, which may take about 45 min for a set of 32 standard silver/silver chloride (Ag/AgCl) electrodes. The electrode–electrolyte impedance spectrometry (IS) of the IrO thin-films was performed in a saline solution, 0.9% concentration by weight, to mimic the electrode–tissue interface. The IS measured results for the IrO coatings were comparable to the results observed for the standard Ag/AgCl electrodes. The new dry microtips array constitutes an inexpensive, low resistance and mechanically robust alternative electrode for non-invasive biopotential recording/stimulation with fast application on skin.

1. Introduction

Recent advances in the biomedical field related with medicine and biology have been demanding more sophisticated electrode fabrication technologies [1]. Electrical activity occurs between neurons as well as in the muscles (e.g. heart) and nerves. The biopotential electrodes, jointly with acquisition systems, sense the electrical activity and make it accessible for clinical and research trials. Electrodes may also be employed on stimulation of excitable tissue.

Biopotential recording and excitable tissue stimulation have been accomplished by recurring to invasive and non-invasive electrodes. The neuroscience field has been demanding invasive electrodes that are implanted for single and multiple recording sites, deep brain stimulation (DBS) and alleviate symptoms of Parkinson's disease [2]. Non-invasive electrodes are used for biopotential recordings like ElectroEncephaloGram (EEG) [3], ElectroOculoGram (EOG), ElectroCardioGram (ECG), ElectroMyoGram (EMG), among several other signals from the skin surface. The surface functional electrical stimulation (FES) [4] and electrotactile stimulation [5] is also accomplished by non-invasive electrodes.

Different electrode materials have been tested on biopotential recording systems: silver/silver chloride (Ag/AgCl) [6], titanium nitride (TiN) [7], aluminum (Al) [8], platinum (Pt) [9] and iridium oxide (IrO) [10]. Standard sintered Ag/AgCl ring electrodes are frequently used for clinical and biomedical applications (e.g. ElectroCardioGraphy and ElectroEncephaloGraphy) and they usually present very low skin-contact impedances and reasonable stability over the required frequency range. Hereafter, the Ag/AgCl electrodes were used as the control electrodes. Nevertheless, IrO has been proved to be one of the most promising stimulation materials (high charge delivery capacity and low constant impedance over the entire frequency range for neural stimulation and biocompatibility) [11]. Among other deposition processes, reactive sputtering may be used in iridium oxide thin-films deposition, whether the target is DC-powered [12] or RF-powered [11] in a plasma environment.

It has been stated that skin impedance is determined mainly by the *stratum corneum* at frequencies below 10 kHz [13] (see Fig. 1). Furthermore, ECG and EEG applications fall entirely into this frequency range. This outer skin layer has high-impedance characteristics since it is mainly constituted by dead skin cells and has very small water content. Consequently, biopotential electrodes require skin preparation (e.g. skin abrasion) and use of electrolytic gel to bypass the *stratum corneum* isolation properties and reduce interface impedance—see Fig. 1(a). Therefore and as depicted in Fig. 1(b),

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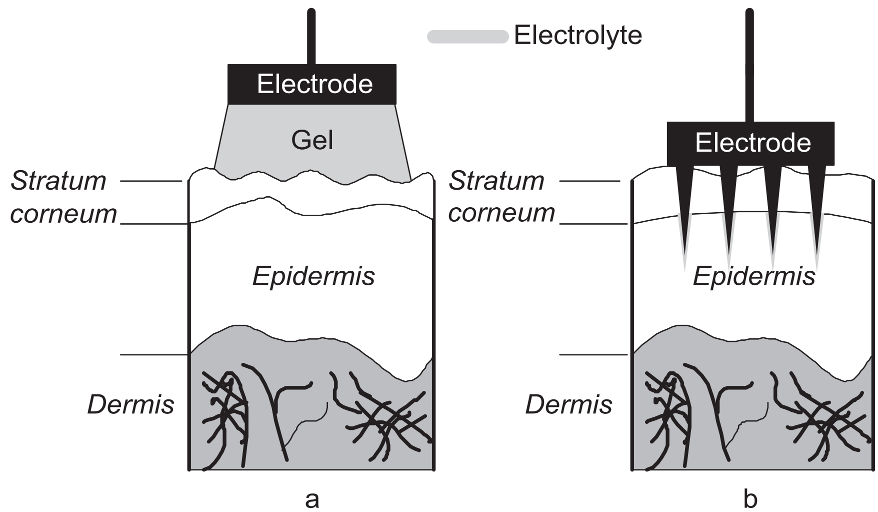

Fig. 1. Application of biopotential electrodes: (a) standard EEG electrode; (b) EEG electrode with microtips.

biopotential electrodes able to penetrate the outer skin layer and interface directly to the electrolytic fluids (abundance of some chlorides like NaCl) of deeper tissues (e.g. living dermis) are of great interest. Ag/AgCl has been proposed as a dry electrode coating with promising results [14]. However, the silver chloride showed to be toxic and has an associated infection risk since it dissolves on skin [15].

Each dry electrode presented in this paper is an array with 4×4 microtips covered with a thin layer of iridium oxide (IrO). The IrO coating layer was deposited by DC-sputtering technique after the microtips array bulk micromachining through a wet-etching process with undercut in a potassium hydroxide (KOH) solution.

The design of the microtips consists in a pyramidal structure. The penetration of the microtips in the skin requires a specific pressure on the array (axial load is applied) and is given by:

$$P = \frac{F}{A} \tag{1}$$

where P is the pressure resulted in the solid structure (i.e. microtip), F, the perpendicular force applied, and A, the section where the force is applied [16]. The force necessary to insert and remove the electrode is about 10 N [16]. Fig. 2(a) shows the pairs of action-reaction forces, established in the electrode when inserted on the skin considering, as the worst case scenarium, that skin and silicon have the same mechanical resistance. Assuming a uniform distribution of the force over the 4×4 microtips array (0.625 N per microtip), the base of each microtip structure is subjected to a pressure of 15.6 MPa. Therefore, every microtip achieves a maximum pressure endured by silicon at 165 μm from the base—red section in Fig. 2(a). Since the average height of the pyramidal structures is about 150 μm and the mechanical resistance of the skin is lower than silicon's, few (or none) microtips are expected to break during skin penetration. Additionally, only 5% of the microtip electrode arrays with aspect ratios higher than the proposed electrode are expected to break [16]. As depicted in Fig. 2(b), each microtip in the array is at least 100 μm high in order to pass through the *stratum corneum*. The fabricated dry electrodes were compared with the standard Ag/AgCl biopotential electrodes—see Fig. 3 for comparison.

2. Fabrication

A wet-etch process using KOH was applied in the bulk micromachining [15,17,18,19] of the silicon microtips. The tip shape was defined by the undercut effect in the etch process, where locally fastest-etching planes are revealed. A silicon wafer with ⟨1 0 0⟩ orientation was used with a silicon nitride layer as mask for the

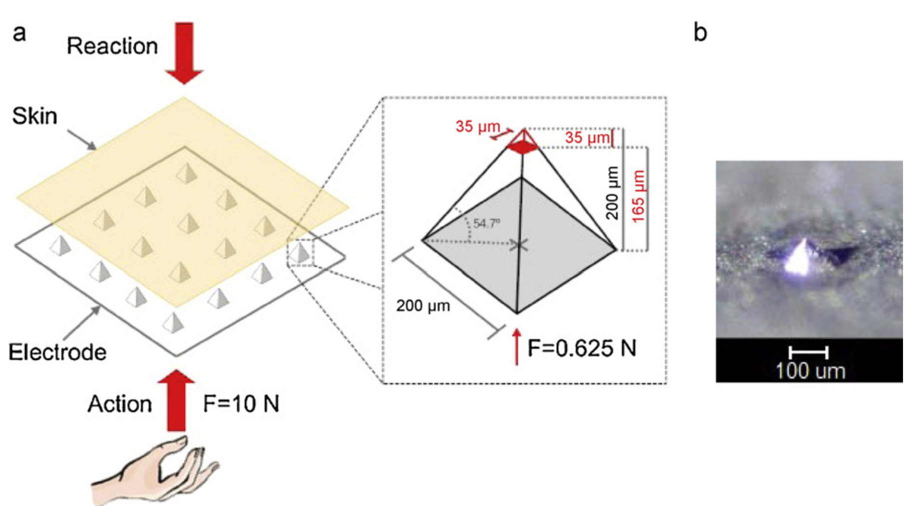

Fig. 2. (a) Pairs of action–reaction forces on a electrode with an array of 4×4 microtips; (b) magnified photograph of a microtip with pyramidal shape.