ORIGINAL ARTICLE

Bared Microtip Arrays for ECG Measurement

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Abstract: Barbed dry microtips arrays are fabricated using simple and low-cost silicon wet etching techniques for ECG recording. Compared with standard wet electrodes, the proposed dry electrodes do not require skin preparation or electrolytic gel during the ECG measurement process. We also discuss parameters affecting skin-electrode impedance, such as the thickness of the conductive layer and the driving frequencies. We successfully demonstrate that the proposed dry electrodes can be used to measure ECG signals.

Keywords: pyramidal microtip, barbed microtip, wet silicon etching, microtip array, contact impedance, ECG recording

Introduction

Rapid progress in the development of biomedical devices for biopotential recording have increased demand for novel and sophisticated sensing electrodes [1]. Non-invasive electrodes are widely used for biopotential recording applications, such as ElectroCardioGram (ECG), ElectroMyoGram (EMG), ElectroEncephaloGram (EEG), ElectroOculoGram (EOG), and so on [2-3]. In general, conductive gel is required for such non-invasive applications.

Silicon-based microstructures with sharp tip arrays used for recording biopotential have recently received increased attention [4-6]. The tiny tips on the electrodes can pierce the layer of the stratum germinativum (without penetrating to the dermis), which in turn reduces skin impedance. Therefore, conductive gel is generally not required for this type of electrode. Silicon microtips can be fabricated using various etching techniques, such as anisotropic wet etching and dry isotropic/anisotropic etching methods. Different types of tip structures have been successfully realized by dry etching methods with appropriate masks and process designs [7-9]. However, dry etching techniques are generally considerably more expensive than wet etching. Since the etching rate is strongly dependent on the crystal orientation, it is possible to realize microtip structures which are defined by the slow etching planes [10-11].

This work presents a simple fabrication process for barbed microtip arrays using low-cost wet etching techniques. Impedance and ECG measurement results are discussed. The proposed microtip arrays were used as dry electrocardiogram (ECG) electrodes. The barbed tips on the proposed electrode should result in stronger attachment than found in typical commercially-available electrodes. A performance comparison between commercial available ECG electrodes (wet electrodes) and the proposed tip-array electrodes (dry electrodes) is provided.

Theory and Fabrication

A. Theory

Figure 1 shows a schematic illustration of human skin. The outermost layer is the stratum corneum, a layer with a thickness ranging from 10–20μm. This layer consists mainly of dead skin cells, resulting in a high degree of resistance which typically degrades the quality of signals for biopotential recording. Electrodes with microtips arrays can be used to pierce the stratum corneum, putting the electrodes into direct contact with relatively conductive epidermal layers, thus effectively reducing impedance and improving signal quality without the need for conductive gel or skin preparation prior to attaching the electrodes to the skin [12-13]. In addition, barbed tips should result in sturdier electrode attachment.

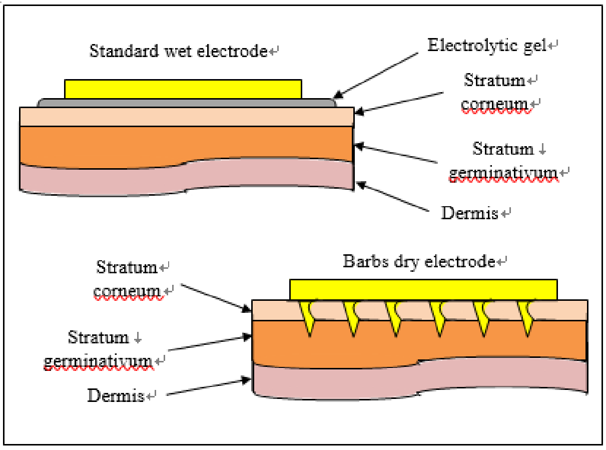

Figure 1. Cross-sectional schematic and simplified electrical model of standard wet (top) and microtips-based dry electrodes (bottom).

B. Fabrication

Figure 2 illustrates the fabrication process. The starting substrate is a standard P-type silicon wafer with a (100) orientation. As shown in Fig. 2(a), a dielectric layer consisting of silicon nitride (Si3N4, 4500 Å) and silicon oxide (SiO2, 500 Å) layers was deposited on the wafer using low pressure chemical vapor deposition (LPCVD).

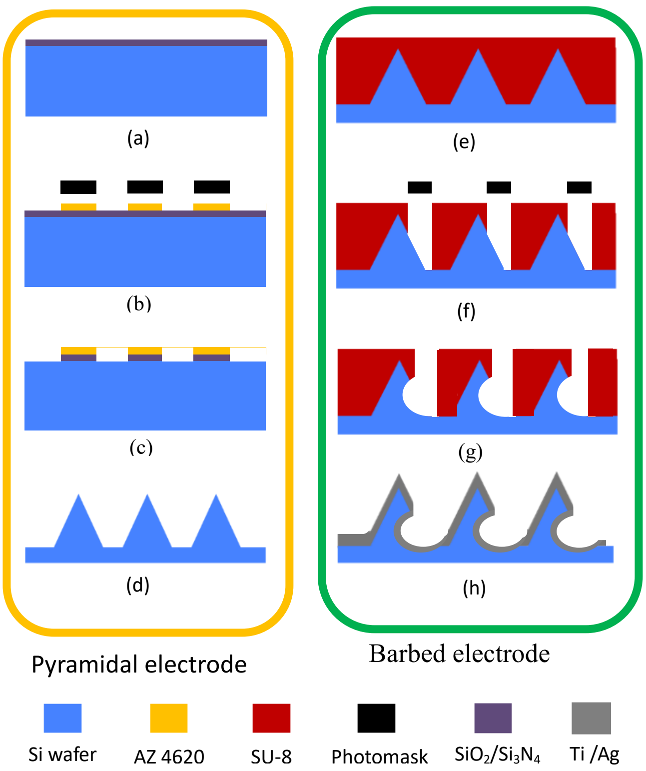

Figure 2. Fabrication and process flow for barbed microtips array.

The dielectric layer was patterned into square arrays using the RIE process to serve as a mask during KOH wet etching (Figs. 2(b) and 2(c)). The pyramidal microtip arrays were fabricated by KOH etchant mixed with IPA, as shown in Fig. 2(d). Note that IPA was used to reduce the roughness of the etched surface. Then, SU-8 photoresist was spun as the mask during HF/HNO3 etching (Fig. 2(e)), and was patterned into barbed etching holes (Fig. 2(f)). The barbs on the pyramidal microtips were formed by isotropic silicon etching using HF/HNO3 (Fig. 2(g)). After HF/HNO3 etching, the SU-8 was removed using the piranha solution (a 3:1 mixture of concentrated sulfuric acid (H2SO4) with hydrogen peroxide (H2O2)). Finally, the conductive layer of Ti/Ag was deposited by sputtering (Turbo Sputter Coater K575X) (Fig 2(h)). Note that Ti was used as the adhesion layer.

Figure 3 shows an SEM image of the tip array after wet etching. The width at the base of the microtips is about 86μm, and the height of microtips is about 155μm, thus they were not long enough to stimulate nerve endings within the skin, thus their application is painless.

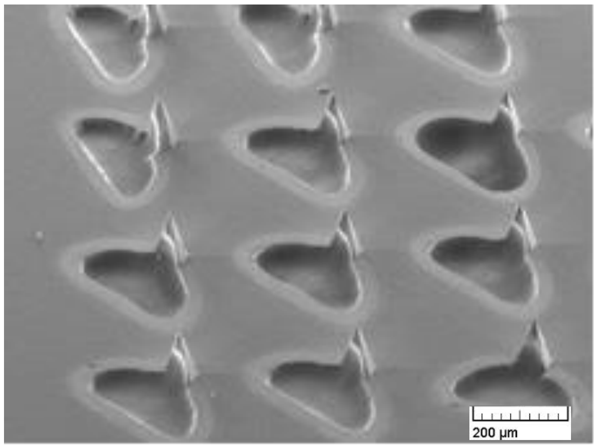

Figure 3. SEM image of the bared microtips array.

The fabricated microtip-array electrode was bonded using conductive adhesive onto a standard commercially-available ECG electrode, from which the conductive gel was removed. Figure 4 shows the assembled electrodes

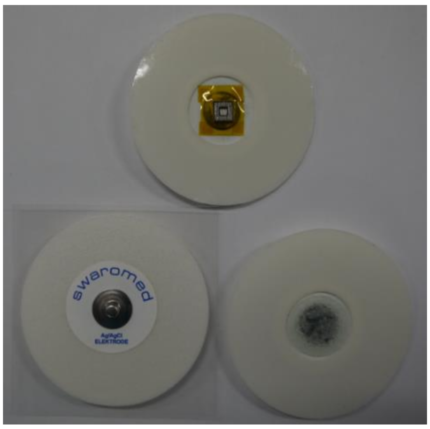

Figure 4. Assembled microtip-array electrodes.

Measurement

A. Skin-Electrodes Contact Impedance Measurement

The total skin-electrode impedance is measured by an LCR meter (6440A, Wayne Kerr Electronics Ltd., U.K). The measurement data were transferred to a computer with LabVIEW. During the measurement, two electrodes were placed on human skin, at a distance (center-to-center) of about 3 cm. The driving signal of the LCR meter is set to 1 V and the frequency sweeps from 120 Hz to 10020 Hz [14-15].

Figure 5 shows the results of skin-electrode impedance measurement. As the thickness of the deposited silver film increases, the total impedance decreases. The contact impedance of the commercial wet electrodes at a frequency of 1,000Hz was 19,214 Ohm.

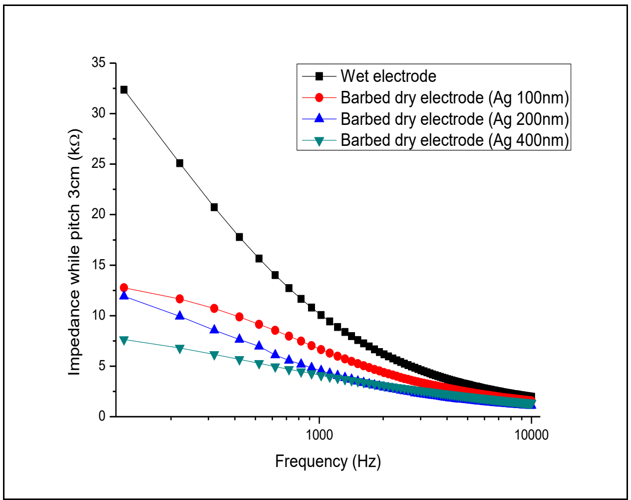

Figure 5. Skin-electrode contact impedance in frequency domain.

B. ECG Recording

To confirm the usefulness of the proposed electrodes, we also performed ECG testing using a commercially-available ECG recording device. Figure 6 shows the ECG measurement results. Figure 6(a) was recorded with wet electrodes. Figure 6(b) was recorded with barbed dry electrodes. Figure 6(c) shows the results of adding water to barbed dry electrodes, and Fig. 6(d) shows the results for barbed dry electrodes with electrolytic gel. The barbed dry electrodes can record the ECG characteristic peaks reasonably well due to their low contact impedance as compared with commercial wet electrodes. Figure 7 shows the electrode position for the ECG recording.

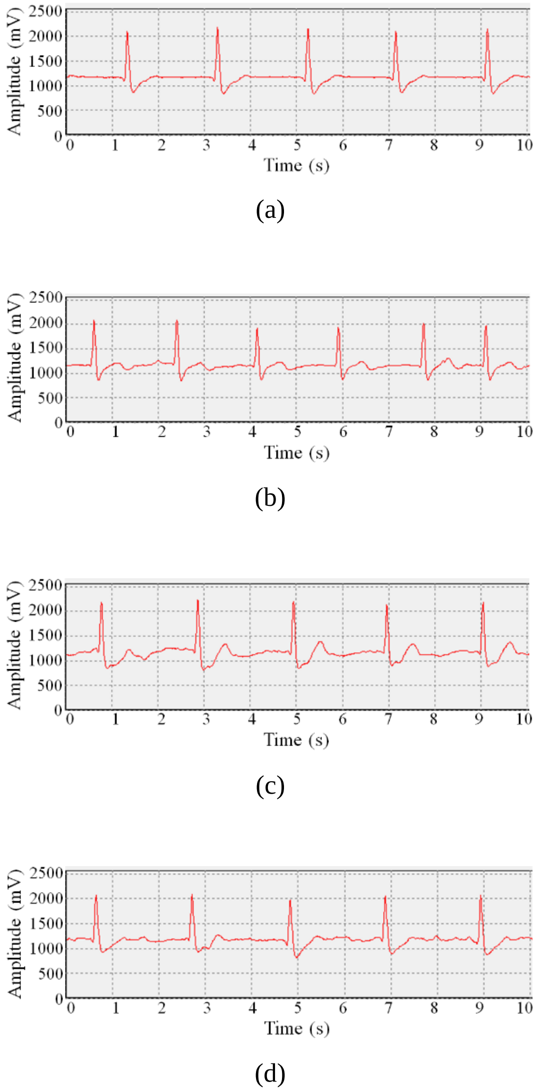

Figure 6. ECG recording results using (a) wet electrodes, (b) by barbed dry electrodes, (c) by barbed dry electrodes with water, (d) by barbed dry electrodes with electrolytic gel.

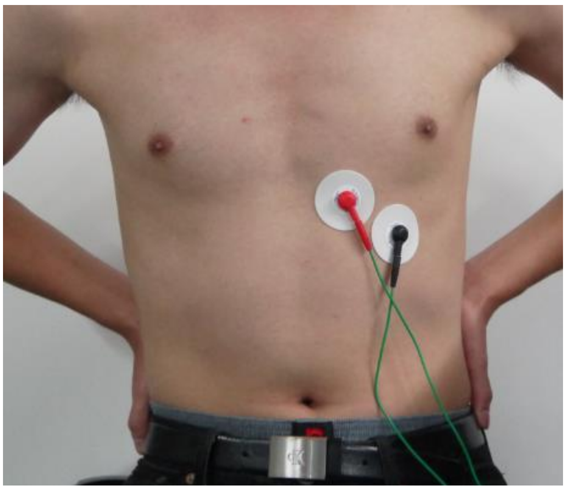

Figure. 7. ECG recording.

Conclusion

We presented a low-cost process for fabricating barbed dry electrodes for ECG recording and measured the total impedance between the electrodes and skin. Furthermore, a commercially-available ECG recording device was used to demonstrate ECG recording with the barbed dry electrodes and found that the performance of the proposed barbed dry electrodes was almost identical to that of standard conventionally-available wet electrodes.

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