MEMS Capacitive Sensor for Wound Monitoring Applications

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Abstract. This paper presents development of a capacitive microsensor to be used in monitoring of wound healing process. Wounds will heal more quickly if they are kept under a bandage and presently bandages must be removed to track the healing process. This could potentially delay the healing process. A sensor can be used to detect the presence of blood in the wound by measuring the capacitance. Blood has a higher permittivity than air or any other substance that may be in the wound and will significantly increase the total measured capacitance in the device. The presence of blood in the wound under the bandage and so in the device would signify that the wound is not fully healed. The sensor was fabricated using standard MEMS (Microelectromechanical Systems) techniques in cleanroom. The sensor was tested in present of no solution, DI water and saline solution, and capacitance was measured to be 1 pF, 35 pF and 53.3 pF, respectively.

1. Introduction

There have been very few breakthroughs in wound care over the past century. During the 1960’s, it was discovered that a wound would heal at a faster rate when it was kept humid and moist [1]. Therefore, when a wound is bandaged, it will heal quicker than when it is un-bandaged. The concept behind wound monitoring sensors is that the bandaging will not need to be removed, but the wound can still be monitored for healing progress.

Currently, a wound can be monitored in a variety of ways. The most common method is to measure pH in the wound area [2-3]. Healthy skin has a pH value in the range of 5.5 to 6.5, while infected wounds can exhibit pH levels of 7.4 or more [4]. Since early 2000, wearable bio-sensing devices have been used for health management and allow the caretaker to avoid unnecessary removal of wound dressing [5-6]. They are made of a material which integrates sensors as a wearable part of the bandage. The sensors are used to detect infections by measuring the pH in the wound. The pH can be measured using optical signals which measure the refractive index of the surface of the skin using a built in LED and spectrometer. A potential issue with measuring pH as a method of wound monitoring is that pH can change drastically with temperature, thus making the temperature of the skin and ambient air temperature uncontrolled variables.

Another method for measuring wound healing progress is by use of wound mapping technology, which measures the electrical properties of the wound tissue throughout the healing process [7]. Specifically, the impedance of the skin will be measured, allowing for the wound progress to be determined. This method may not be able to determine whether the wound has healed completely, as
many factors can affect impedance, including blood flow in the skin and potentially skin temperature [8].

As another alternative, capacitance can be used to simply detect the presence of blood, which will allow for the track of a healed wound. This would simply act as a notification that bandages may be removed without disrupting the healing process. The sensor will consist of a small enclosed channel, which will attract blood due to the capillary effect when it is present. The blood will flow through the channel, which has electrodes on both sides. The blood will change the capacitance between the electrodes due to different permittivity, thus signalling that the wound has not healed fully.

This paper covers the design, fabrication and experimentation of the capacitive sensor for wound healing. The microsensor consists of glass and silicon wafers bonded to each other. It works based on the detection of saline as a substitute for blood around the wound by measuring the capacitance around an open wound. The capacitance will change when the particles travel between aluminum electrodes through the microfluidic channel. The microsensor is able to read whether the fluid is adequate for the wound and if it is bleeding.

2. Design

Schematic perspective and top views of the layers in the sensor are shown in Figure 1. The working principle of the capacitive sensor for wound monitoring relies on a gap between electrodes on an etched area in a silicon chip and glass substrate. This will allow for the capillary effect to take place, drawing the blood into the device. To allow for a reading of capacitance to be taken, aluminum is deposited on the bottom of the etched groove in the silicon chip and in the corresponding area on the glass substrate. Aluminum pads are connected to the aluminum electrodes in order to create the capacitor. The electrodes are then passivated with insulting material in order to avoid electrical current passing between them and enable capacitance readings.

The aluminum pad on the silicon extends to one side, allowing for wire to be bonded to the pad without interfering in the glass layer. The aluminum pad on the glass is offset to the other side, allowing wire bonding without interfering with the bonding of the glass and silicon as shown in Figure 1. In addition the aluminum electrode on the glass is slightly smaller than the electrode on the silicon to allow for slight misalignment without affecting the capacitance value. The dimensions of key features on the designed capacitance microsensor are presented in Table 1.

![Figure 1](image.png)

**Figure 1.** a) Perspective view of the sensor, showing different layers and features of the device. b) Top view of the entire design.
Table 1. Dimensions of key features on the designed capacitance microsensor

| Feature                              | Value     |
|--------------------------------------|-----------|
| Overall Area of Microsensor          | 1 cm²     |
| Depth of Groove                      | 60 μm     |
| Length of Main Channel               | 8700 μm   |
| Width of Main Channel                | 4000 μm   |
| Width of Exit Channel                | 1000 μm   |
| Effective Length of Al Electrodes    | 8500 μm   |
| Effective Width of Al Electrodes     | 3000 μm   |

A capacitance will increase significantly when a liquid, in this case blood, is present in the gap between the aluminum electrodes on the glass and the silicon. There is a small groove at the end of the etched area on the silicon chip to allow air to escape. This will allow blood to be evenly dispersed through the gap, maximizing the capacitance outputs.

The microsensor was designed to measure the capacitance between the aluminum electrodes. The equation used to calculate capacitance of the sensor is:

\[ C = \varepsilon_0 \varepsilon_r \frac{A}{d} \]

where \( C \) is capacitance, \( \varepsilon_0 \) is the permittivity of free space (constant), \( \varepsilon_r \) is the relative permittivity of the medium between the electrodes, \( A \) is the projected area of the electrodes and \( d \) is the distance between the electrodes. Both the area of the electrodes and the distance between them are fixed in this design. The only parameter that changes the capacitance is the relative permittivity of the medium which depends on the type of fluid present in the groove. The relative permittivity of air that is supposed to be present in the gap when the wound is completely healed is 1 while the relative permittivity of blood which enters the gap when the wound is not healed is over 500, resulting in a significant drop in capacitance after healing. In order to make the experiments easier, a 0.01 M saline can be used as a substitute for the blood. The saline has a relative permittivity of about 90. It has a slightly lower electrical conductivity than blood at this concentration [9-10].

3. Fabrication

Fabrication of the device was performed based on common MEMS techniques in the cleanroom. Figure 2 shows the fabrication steps of the sensor, including silicon wafer fabrication, glass wafer fabrication and bonding them. The silicon wafer was a double side polished n-type wafer with initial resistivity of 1-10 Ω.cm. The wafer was first oxidized in an oxidation furnace to form a 0.5 μm thick oxide layer on both sides of the wafer.

Defining the groove was done by depositing photoresist on each side of the wafer and soft baking. After developing the photoresist on the front side of the wafer, the photoresist was hard baked on the hotplate. In order to pattern the oxide layer as an etch mask, the wafer was dipped in hydrofluoric acid (HF) which opened up the places where bulk etching will take place. To finalize the groove, a wet etch of TMAH heated to roughly 90°C was used to create a 60 μm deep grooves. The remnants of oxide layer was then stripped by dipping the wafer in HF.

After cleaning the wafer with piranha and RCA clean techniques, another 0.5 μm thick oxide layer as an electrical passivation film was grown on all surfaces in the oxidation furnace. The wafer was then loaded into the physical vapor deposition (PVD) thermal evaporator to deposit an aluminum layer over the entire wafer. The lithography process was used to pattern the photoresist on the aluminum layer. The wafer was then submerged into a commercial aluminum etchant to form aluminum electrodes, conductors and pads. After rinsing and drying the wafer, the wafer is covered with acetone to remove the remaining photoresist.

A double sided polished borosilicate glass wafer was used as the mating part of the device. The processes on the glass wafer were similar to those on the silicon wafer, however only an aluminum
layer was deposited onto the glass. The glass was simply cleaned first with soap. It was then placed into the PVD thermal evaporator to deposit an aluminum layer on it. The aluminum layer was then patterned using lithography process and subsequent aluminum etch in order to form electrodes, conductors and pads on the glass wafer. Both silicon and glass wafers were cleaved to small individual pieces for further processing.

The aluminum electrodes on both silicon and glass wafers required to be insulated from the medium. A dual purpose process was designed to passivate the electrodes and bond the chips in a single step. This step was performed by spinning and patterning a 0.7 µm thick photoresist on the chips, covering electrodes and conductors. The same wet photoresist layers were used as a bonding interface between the two chips. In order to bond silicon and glass chips, they were aligned and pushed against each other on a 135°C hotplate for five minutes. The photoresist was baked in this process and made a bond between two chips while the aluminum electrodes were still passivated by the thin photoresist layer.

![Silicon Wafer Fabrication](image1)

![Glass Wafer Fabrication](image2)

**Figure 2.** Fabrication steps of the capacitive microsensor

### 4. Results and Discussion

The capacitive sensor was characterized in present of air, deionized (DI) water and 0.01 M saline solution. The saline solution was used to represent blood, as blood is a biohazard and unsafe to be used in this experiment. The sensor electrodes were connected to a capacitance meter for measurement. A syringe was used to deposit each solution into the channel on the sensor.

The completed experiment yielded results as expected. Air yielded a capacitance reading across the electrodes of 1 pF as a baseline. This is because the permittivity of air is as low as ~1. The capacitance reading in present of the DI water was measured significantly higher at 35 pF. The saline solution yielded the highest reading at 53.3 pF. These results are summarized in Figure 3. The capacitance reading when the saline solution was present in the channel was much higher than that of the air, meaning that the presence of a small amount of blood can be easily detected. Although the readings detected during experimentation did not completely match the design values, the design was still theoretically successful. The inconsistent values are likely due to misalignment of the electrodes which result in a drastic area difference of the electrodes. Since the permittivity value of blood is significantly higher (5-10 times) than that of saline solution, it is reasonable to think that the presence of blood in a wound would result in a significantly higher reading that could easily contrast that of a bloodless environment.
Figure 3: Comparison of capacitance readings of each fluid tested; Air, DI Water and 0.01 M Saline solution.

5. Conclusions
This paper described the development steps of a highly sensitive microsensor for wound monitoring applications. The microsensor design is highly plausible as a way to monitor a wound healing progress. The sensor could be used to detect the presence of blood under a bandage, which would therefore eliminate the need to remove a bandage to check on the healing process. Since the bandage could stay in place during the monitoring process, wounds would be able to heal quicker. This sensor can be integrated into bandages if needed or used as a disposable sensor. The results showed that the device can accurately detect the present of blood in the wound, monitoring the healing process.

References
[1] Winter G D 1962 Formation of the scab and the rate of epithelization of superficial wounds in the skin of the young domestic pig Nature 193(4812)293
[2] McColl D, Cartlidge B and Connolly P 2007 Real-time monitoring of moisture levels in wound dressings in vitro: an experimental study International Journal of Surgery 5(5)316-22
[3] Guinovart T, Valdés - Ramírez G, Windmiller J R, Andrade F J and Wang J 2014 Bandage - based wearable potentiometric sensor for monitoring wound pH Electroanalysis 26(6)1345-53
[4] Rahimi R, Ochoa M, Parupudi T, Zhao X, Yazdi I K, Dokmeci M R, Tamayol A, Khademhosseini A and Ziaie B 2016 A low-cost flexible pH sensor array for wound assessment Sensors and Actuators B: Chemical 28(229)609-17
[5] Pasche S, Angeloni S, Ischer R, Liley M, Luprano J and Voirin G 2008 Wearable biosensors for monitoring wound healing Advances in Science and Technology (57)80-87
[6] Stoppa M and Chioriero A 2014 Wearable electronics and smart textiles: a critical review Sensors 14(7)11957-92
[7] Weber S A, Watermann N, Jossinet J, Byrne J A, Chantrey J, Alam S, So K, Bush J, O’Kane S and McAdams E T 2010 Remote wound monitoring of chronic ulcers IEEE Transactions on Information Technology in Biomedicine 14(2)371-7
[8] Buono M J, Burke S, Endemann S, Graham H, Gressard C, Griswold L and Michalewicz B 2004 The effect of ambient air temperature on whole-body bioelectrical impedance Physiological Measurement 25(1)119
[9] Stiles D K and Oakley B 2003 Simulated characterization of atherosclerotic lesions in the coronary arteries by measurement of bioimpedance IEEE Transaction of Biomedical Engineering (50)916-21
[10] Peyman A, Gabriel E H 2007 Complex permittivity of sodium chloride solutions at microwave frequencies Bioelectromagnetic (28)264-74