Measuring Skin-electrode Impedance Variation of Conductive Textile Electrodes under Pressure

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***Abstract*— Electrocardiogram (ECG) is the first bio-signal physicians use to diagnose cardiovascular diseases, since any kind of heart abnormality reflects on it. ECG electrodes play an important role in collecting this signal. One type of ECG electrodes which is recently getting more popular, especially due to its user friendly features compared to traditional Ag/AgCl electrodes, is conductive textile. Similar to any other type of electrode, conductive textile is associated with skin-electrode interface impedance located between the body (source of signal) and ECG monitoring device, thus affecting the recorded ECG quality. In this paper, we measure the skin-electrode impedance of conductive textile electrodes under various pressure levels, because in some applications ECG electrode is located under a blood pressure cuff and therefore it is under pressure when the cuff is inflated. We show that in fact under pressure the impedance decreases, resulting in a higher quality ECG measurement.**

Keywords—ECG electrodes; Skin-electrode Impedance; pressure

I. INTRODUCTION

ECG has always been considered an informative vital sign carrying enough information about one's heart situation, and even a human identifier in some new approaches [1]. Measuring ECG with accuracy is thereby an important issue that researchers are concerned with. To achieve the best quality in ECG measurement, many parameters should be considered. One of the parameters that can affect ECG quality is the ECG electrode acquiring ECG signal and its impedance [2].

At the same time, nowadays many home healthcare devices are commercially available allowing cardiac patients to measure their heart condition continuously. Such devices are improving by becoming more accurate, recording or monitoring multi vital signs [3], calling the emergency ward in case of emergency [4], detecting the presence of pathology occurrences [5], adding some diagnostic properties [4], and increasing the user friendliness level of the device. For devices that measure ECG as a vital sign, it would be best if they perform tasks in a non-intrusive manner. In order to achieve this, different ECG electrodes are entering the market [6]. One recent ECG electrode which is getting more popular is conductive textile. Since it is as soft and flexible as an ordinary fabric and there is no need for skin preparation, they are a good choice for wearable home health care devices. On the other, hand home health care devices which are capable of measuring more than one vital sign, e.g. ECG and blood pressure, are more useful. In addition to that, there are some blood pressure measurement algorithms which rely on ECG as a more consistent signal to measure blood pressure more accurately, such as [7] and [8] which show that conductive textile is a suitable choice to be utilized in devices using this algorithm. In such applications, conductive textile as an ECG electrode can be simply embedded inside a cuff, and when the cuff inflates to measure blood pressure, it applies pressure to the conductive fabric ECG electrode. Considering that any ECG electrode is associated with skin-electrode impedance that may affect the ECG measured at the electrode, we need to know what happens to the skin-electrode impedance under this pressure, since it will have implications in the ECG measured by the electrode.

In this paper, we measure the skin-electrode impedance of conductive textile electrodes when they are under pressure. We place electrodes under a regular Omron blood pressure cuff and apply pressure by inflating it. This particular measurement set up is chosen for our experiments, because this is how ECG is obtained practically, in many portable home health care devices that measure blood pressure based on some information from ECG. We repeat the measurements for four pressure values and all measurements are done for three subjects. We also extract all components of skin-electrode impedance, based on its electrical equivalent model and measure their variation rate in response to applying pressure.

The rest of the paper is organized as follows. Section II represents the skin-electrode impedance and its electrical equivalent models and measurement methods. It is followed by our impedance measurement set up in section III and also presenting how we apply pressure to the skin-electrode impedance. In section IV we show the experimental results and the paper concludes in section V.

II. RELATED WORK

A. Skin-electrode impedance

Skin-electrode impedance is the impedance between the body and the electrode; this plays a major role in the quality of the signal sensed by the electrode. If it is too high, and thus associated with a small signal-to-noise ratio, then it will negatively affect the quality of the signal [9], [10].

B. Electrical circuit equivalent to the skin-electrodes impedance

To better understand and analyze skin electrode impedance behaviour, an equivalent electrical circuit is helpful. Such a circuit can be extracted by studying the electrical characteristics of both electrode and the skin. Warburg [11] was the first to propose an equivalent circuit model for the electrode-electrolyte interface. Moreover, Feate et al. [12] identified the components of electrode circuit model, analyzing the electrical properties and conductive nature of biological tissues [12]. Their study helped in estimating the values of capacitors and resistors in the electrode-skin model. Basically two electrical models are suggested for skin electrode interface, explained next.

1) Single–time constant model

Swanson and Webster [13] suggested a model for skin-electrode impedance, which is a combination of a resistor in series with a paralleled resistor and capacitor as shown in Fig. 1. E_{hc} in this model is mostly applicable for wet electrodes since it shows the voltage between the skin and electrolytes, for example, gel. C_d stands for the electrical charge between the electrode and skin [14], [9], and R_d shows the resistance occurs between the skin and electrode during charge transfer [13]. Another component of the model, R_s, represents the total resistance of electrolyte gel (if any), sweat and also the underlying skin tissue [10], [15]. This model is known as the single–time constant model [16].

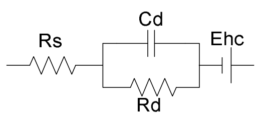

Fig. 1. Single–Time Constant Skin Electrode Interface Model

2) Double–time constant mode

Neuman's [16] suggestion for the skin-electrode impedance electrical equivalent model is more complicated. As illustrated in Fig. 2, it is actually composed of two stages of the single–time constant model and known as double–time constant model. One stage only represents the skin and the other one only represents the electrode.

The authors in [17] explored both models in terms of ECG measurement and frequency response. Their study shows that the double–time constant model exhibits more accurate results. In this study, due to some restrictions in our measurement devices, we apply the single–time constant model.

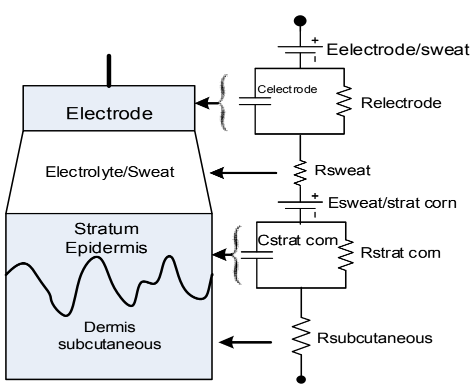

Fig. 2. Double–Time Constant Model [17]

C. Skin-electrode impedance measurement

Skin-electrode impedance measurement has been always of interest because it influences the reliability of the collected signal. Thus, many papers in the literature have discussed methods of measuring it accurately. The authors in [18] proposed a method with a measurement setup shown in Fig. 3.

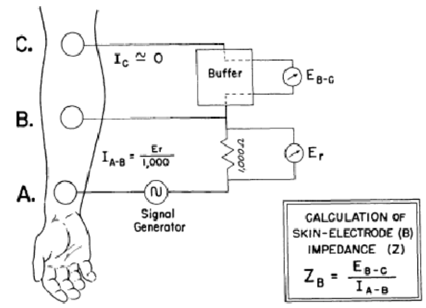

Fig. 3. Skin Electrode Impedance Measurement Setup [18]

In this method, three electrodes are applied on the arm and the skin-electrode impedance of the middle one is calculated. The current flowing in A and B can be calculated by measuring the voltage drop in E_r. A buffer unit is used to prevent any significant current flow between B and C. Furthermore, a known sine wave is applied to electrode B and the voltage between B and C is measured. Thus, impedance of B can be calculated using the following equation

$$Z_B = \frac{E_{B-C}}{I_{A-B}}$$

Another method is presented in [19] which uses a double electrode configuration, as shown in Fig. 4.

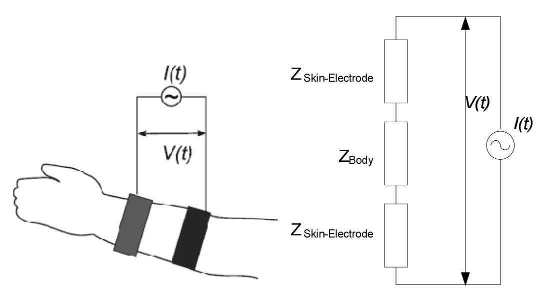

Fig. 4. Two-Electrode Configuration for Skin-Electrode Impedance Measurement [19]

All available studies in the literature measure the skin-electrode impedance for traditional Ag/AgCl electrodes. Since conductive textile as an ECG electrode is becoming popular for medical devices, in this study we measure the skin-electrode impedance of conductive textile. Furthermore, we do the measurements while we are applying pressure to the interface.

III. MEASUREMENT SYSTEM AND APPLYING PRESSURE

We utilize commercially available state-of-the-art devices that are designed specifically for this purpose in order to measure skin electrode impedance variation over a particular frequency range. A block diagram of the experimental setup is shown in Fig. 5. This represents a two-electrode configuration that works based on injecting a current sine wave signal from one electrode and collecting the same signal from the other electrode, as well as measuring the voltage drop between two electrodes.

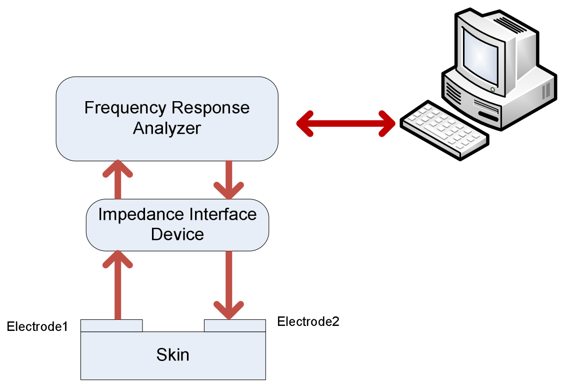

Fig. 5. Experimental Setup for Skin-Electrode Interface Measurement

This system includes a frequency response analyzer (FRA) (Model # 1255B, Solartron Analytical, UK) and an impedance interface device (Model # 1294A, Solartron Analytical, UK), as well as a personal computer.

Skin-electrode impedance was measured by injecting a 10 µA AC current sweeping frequency from 1 Hz to 1 MHz to one electrode located on the left bicep and collecting it from the other electrode on the same bicep at 10 cm distance from the first one. Our measurement configuration is depicted schematically in Fig. 6. Based on it, our measurement shows the total value of both electrodes plus tissue impedance.

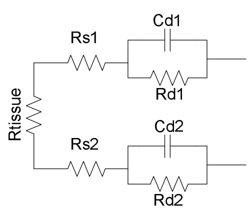

Fig. 6. Schematic View of Skin-Electrode Impedance Measurement

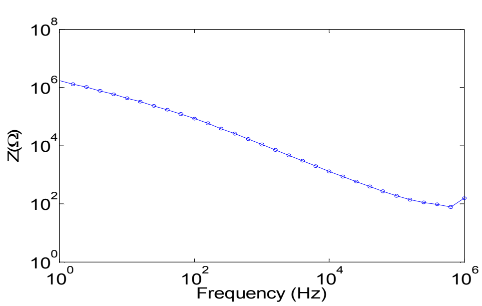

Fig. 7. Skin-Electrode Impedance Frequency Response

Since the electrodes we apply are identical in terms of type, material and size, we can assume that they have equal skin-electrode interface impedance. R_{tissue} represents the resistance of tissues between two electrodes. When we place the electrodes on a healthy human's middle upper arm, this value is found to be approximately 150 Ω [20]. On the other hand, the electrode skin impedance is larger than 1 MΩ. According to this, in this study, R_{tissue} is assumed to be negligible and its contribution is considered in R_s. Therefore, we determine the skin-electrode interface for a single electrode by dividing the total measured value by two. Fig. 7 shows the results of our experiments in logarithmic scale.

A. Estimating skin-electrode impedance model components and verification

We have already measured the magnitude of impedance over a range of frequency from 0 to 1 MHz. In order to estimate our model parameters, specifically R_s , R_d and C_d, as depicted in Fig. 1, such that our measured $|Z(\omega)|$ shown in Fig. 7 matches the magnitude of the obtained model. Applying the least squares nonlinear curve fitting method, we extract three components of the simplified single-time constant model. This fitting method minimizes the summed squared of the error between the measured data point and the fitted model and provides the best fit impedance model from the measured impedance values. In this study, we implemented this method in MATLAB (R2008a) by applying the "lsqcurvefit" function. MATLAB code based on the experimental data returned the following values for the three components:

$$R_d = (9.87\,e + 5)\Omega,\ C_d = (1.58\,e - 8)F,\ R_s = 16.69\ \Omega$$

In order to compare the obtained data and observed one, we substituted the values of the components in Equation (1), which shows the mathematical relation between |Z(ω)| and its three components. Fig. 8 illustrates the result graph of obtained |Z(ω)| and observed graph data on the same plane for comparative purposes.

$$|Z(\omega)| = |Re(\omega) + jIm(\omega)| = \left| R_s + \frac{R_d}{1+\omega^2 C_d^2 R_d^2} - j\frac{\omega C_d R_d^2}{1+\omega^2 C_d^2 R_d^2} \right| \tag{1}$$

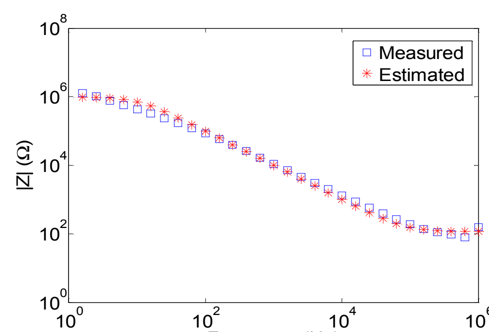

Fig. 8. Measured Data and the Estimated Model

B. Applying pressure to the ECG electrodes

For our measurements, to investigate the effects of pressure on conductive textile skin-electrode interface impedance, we set up the system as explained above; the only difference was that, we applied pressure with the cuff, to the electrodes located underneath, while doing the measurements. We kept the pressure of the cuff constant during each measurement and applied pressure from 0 mmHg up to 120 mmHg in 40 mmHg steps. For consistency with our previous measurements, we used the single–time constant model for the interface and the least squares nonlinear curve fitting method for component value estimation. The variation trend line of each component, that is, C_d, R_d and R_s, was tracked and drawn in the following graphs. We only chase the variation of impedance amplitude under the pressure, because the phase part, only cause time delay. This impedance affects the quality of collected ECG, and ECG is a periodic signal and time delay has no effect on it [21].

IV. EXPERIMENTAL RESULTS

In this study, we expand our preliminary measurement which was done only for one subject [21] and do the measurements for more subjects. Here, measurements were carried out on three healthy subjects (2 females, 1 male). None of the volunteers was taking any medication or had a known history of cardiovascular disease.

Our results including magnitude of skin-electrode impedance in various pressure values and also variation of all three components of its model will be illustrated using tables and graphs. The entire measurement procedure was carried out as follows:

a. Both electrodes were placed on the subject's bicep at a 10 cm distance from each other. The electrodes were also connected to Solartron devices.
b. The Omron cuff was wrapped around subject's bicep on top of the electrodes.
c. The cuff was inflated and its pressure was read continuously until it reached a certain level.
d. The measurement procedure with Solartron devices began and the result was saved in the PC as .xlsx files.
e. The components of the model were determined by applying the "lsqcurvefit" function in MATLAB.
f. All components from the single–time constant model were extracted.

TABLE 1 shows the subjects' gender, height, weight and body mass index (BMI), which is defined as follows:

$$\text{Body Mass Index (BMI)} = \frac{\text{Weight}}{(\text{Height in m})^2}$$

TABLE 1. SUBJECTS' CHARACTERISTICS

	Gender	Height	Weight	Body Mass Index
Subject1	Female	172 cm	68 kg	23.25
Subject2	Male	181 cm	59 kg	18
Subject3	Female	162 cm	56 kg	21.34

Fig. 9, Fig. 10 and Fig. 11 show the |Z| variation per frequency for the first, second and third subjects, respectively. The frequency varied between 0 Hz and 1 MHz. The |Z| values for various pressures are shown on the same graph in logarithmic scale. The value of |Z| corresponding to the lowest pressure (P = 0 mmHg) and that corresponding to the maximum pressure we apply, (P = 120 mmHg) represent the highest and lowest curves, respectively.

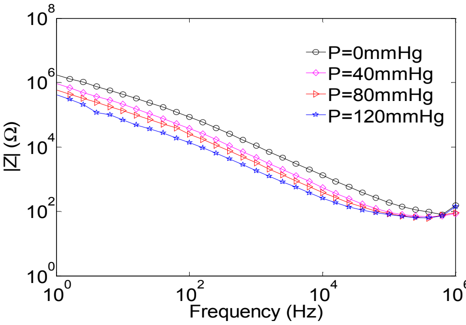

Fig. 9. Skin-Electrode Impedance Variation for Subject 1

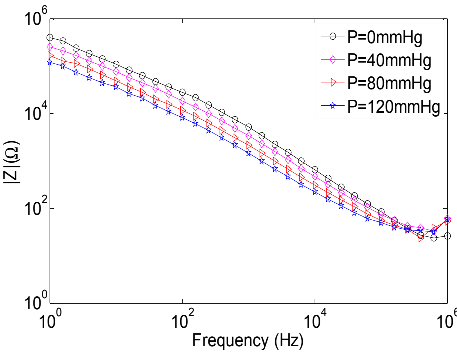

Fig. 10. Skin-Electrode Impedance Variation for Subject 2

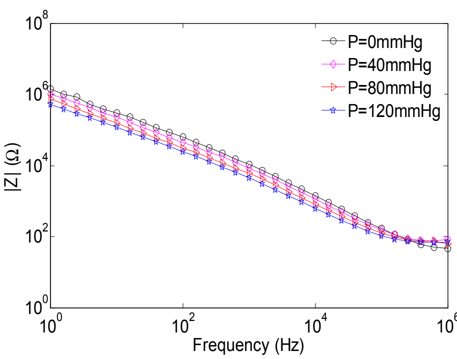

Fig. 11. Skin-Electrode Impedance Variation for Subject 3

Measurements were followed by parameter extraction for each curve. Fig. 12 to 14 show the model components, including R_d, C_d and R_s variations for subject 1 to 3 respectively. The first graph shows the variation in R_d, which in the middle shows variation in the capacitor of the model, C_d, and at the bottom shows the variation in R_s.

V. CONCLUSION AND FUTURE WORK

Conductive textile as an ECG electrode, appropriate for wearable home health care devices, is investigated in terms of its skin-electrode interface impedance when under pressure. Our observations show that this impedance decreases under pressure resulting in a better ECG quality. Therefore, we can conclude that conductive textile is a suitable choice for blood pressure measurement algorithms using ECG as an auxiliary consistent signal.

For future work we will repeat the same measurements for more people and classify the results based on different parameters such as gender, age and BMI. In addition to that, we will acquire ECG at the same time as we apply pressure to the electrode and compare ECG signals collected under different pressure levels. Another idea we have to expand this work, is to apply pressure to other types of ECG electrodes and verify how components of skin-electrode model vary as time passes.

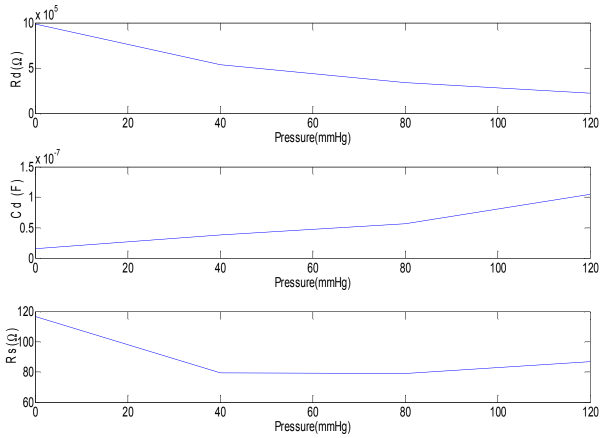

Fig. 12. Component Variation Rate Versus Pressure for Subject 1

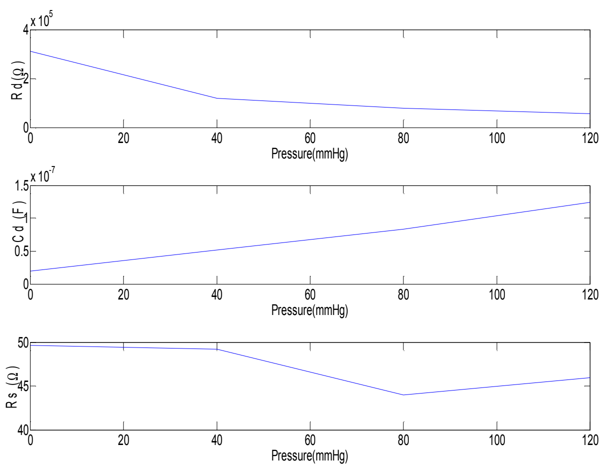

Fig. 13. Components Variation Rate Versus Pressure for Subject 2

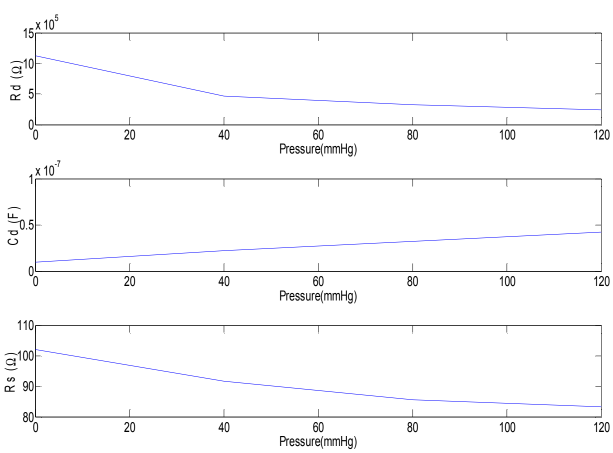

Fig. 14. Components Variation Rate Versus Pressure for Subject 3

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