Analyzing Dry Electrodes for Wearable Bioelectrical Impedance Analyzers

M. Usman[1], A.K. Gupta[2], W. Xue[1]

1. Mechanical Engineering, Rowan University, Glassboro, New Jersey, USA
2. Osteopathic Medicine, Rowan University, Stratford, New Jersey, USA
usmanm9@students.rowan.edu, guptaad@rowan.edu, xuew@rowan.edu

Abstract— Dry electrodes are gaining popularity in the area of electronic health for biosignal measurements due to their reusability and comfort as compared to traditional gel-based wet Ag/AgCl electrodes. This paper presents a performance comparison of dry and wet electrodes for medical devices, in particular, for bioelectrical impedance analysis (BIA). BIA is an emerging technology widely used for body composition analysis by computing the impedance of the human body. The designed system for BIA consists of a wearable silicone ring with four copper electrodes. The experiment is conducted on 40 healthy human subjects using both the ring and the Ag/AgCl electrodes. The linear regression demonstrates a high correlation between both electrodes (r = 0.96 for resistance and r = 0.93 for reactance). The measurement of root mean square noise is determined for both electrodes. The dry electrodes demonstrate a higher noise level (1.96 mV) as compared to the wet electrodes (0.282 mV), mainly due to the absence of conductive gel. Moreover, fast Fourier transform is performed to determine and filter out unwanted signals and to reduce the noise level in the dry electrodes. The results demonstrate that the designed ring electrodes have a comparable performance with commercial Ag/AgCl electrodes and can be used in mobile wearable medical devices.

I. Introduction

In this work, we report a comparative analysis of dry and wet electrodes in wearable medical devices for bioelectrical impedance analysis (BIA). Recent advancements in sensor technology, VLSI design, and wireless communication offer compact systems with high computation powers at low costs. This facilitates the accurate acquisition of physiological signals in wearable medical devices. An important factor that affects the signal accuracy from the human body is the characteristics of the electrodes used in the acquisition. Medical devices typically use conventional gel-based wet Ag/AgCl electrodes for bio-potential signal recording. The conductive gel on the electrodes can reduce the noise and the electrode-skin impedance, enhancing the quality of the signal. However, these gel-based electrodes can cause discomfort, skin irritation, and sometimes even rashes on patient's skin. Furthermore, these wet electrodes are not designed for long-term use because the gel tends to dry out and loses its adhesive capability over time, leading to an increased electrode-skin impedance. As a result, frequent replacements of the electrode are required. In contrast, dry electrodes have been studied as a potential replacement for Ag/AgCl electrodes as they are durable and their application or separation causes little skin discomfort [1, 2].

BIA is an emerging technology widely used for body composition analysis, nutritional assessment, and clinical research in medical electronics. In BIA, the electrical impedance of the human body is measured by passing a small amount of electricity through the body and, at the same time, recording the response signals [3]. In this paper, dry electrodes are analyzed using a wearable bioelectrical impedance analyzer for body fat estimation based on our prior work [4]. The dry electrodes are made of copper due to its several advantages: copper electrodes are sturdy and reusable; the material has a low electrical resistivity (1.68×10^{-8} Ωm); the metal surface provides a uniform and smooth contact with the skin; copper is inexpensive and commercially available in the form of flexible, one-sided adhesive tapes which can be easily applied to a variety of substrates. The analyzer designed to measure the bioelectrical impedance is calibrated and tested using discrete passive components (resistor and capacitor). The novel ring-based bioelectrical impedance analyzer provides new opportunities for future wearable health monitors.

II. System Design

Fig. 1 shows the block diagram of the designed bioelectrical impedance measurement system. The circuit generates a sine wave of 50 kHz using a voltage-controlled oscillator (VCO). This frequency is widely used in BIA as it enables electrical current to flow through extracellular as well as intracellular paths [3]. The output voltage signal of the VCO is converted into a current signal by a voltage-controlled current source (VCCS). The VCCS delivers the current through the body via source electrodes (I^+, I^-). The voltage drop across the body is measured using two sense electrodes (V^+, V^-). The current also flows through a known reference resistor (R_{ref} = 100 Ω) connected in series with the body. The measurement of the voltages across the body and the reference resistor is performed using two separate amplifiers. The gain and phase detector then acquire both amplified voltage signals and computes the ratio of amplitude and difference of phase between them.

$$\mathrm{K} = \frac{A_1 . V_{body}}{A_2 . V_{ref}} \tag{1}$$

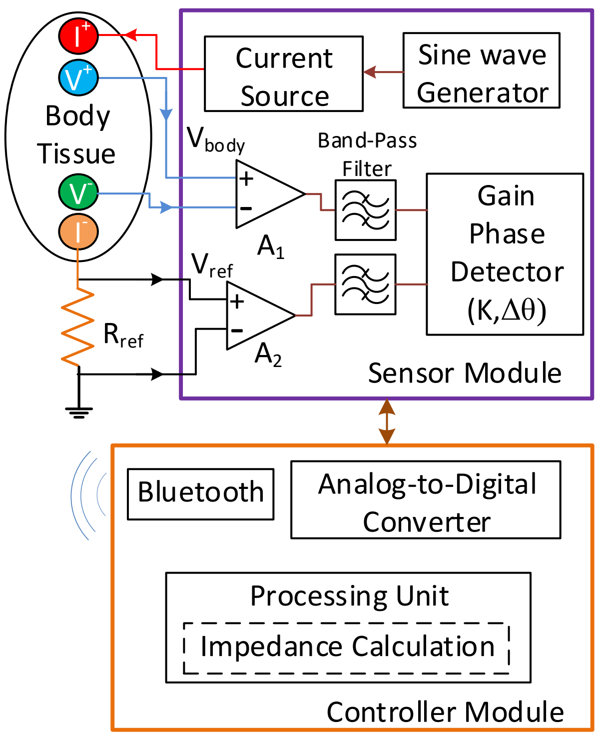


Figure 1. Simplified block diagram of the designed bioelectrical impedance analyzer.

$$\Delta\theta = |\theta_{body} - \theta_{ref}| \tag{2}$$

where V_{body} and θ_{body} are the voltage and the phase across the body while V_{ref} and θ_{ref} are the voltage and the phase across the reference resistor.

These gain and phase signals are processed using Adafruit Feather 32u4 development board which contains a 10-bit analog to digital converter and an 8 MHz clocked ATmega microcontroller. The microcontroller computes the body impedance from the signals acquired from the sensor module.

$$|Z_{body}| = \frac{A_1.V_{body}.R_{ref}}{A_2.V_{ref}} = \mathrm{K}.R_{ref} \tag{3}$$

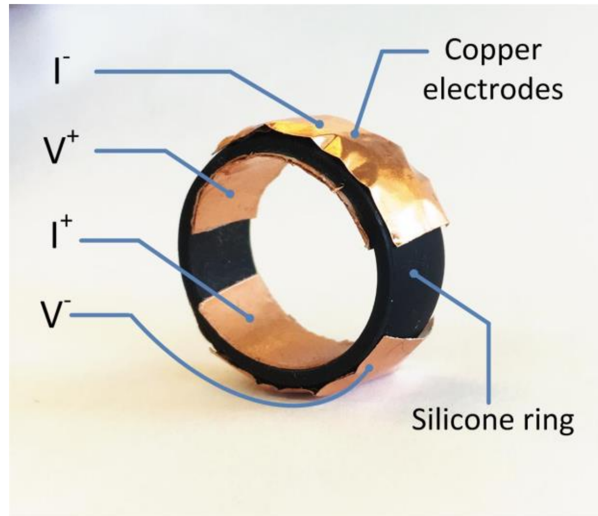
$$Z_{body} = |Z_{body}|.e^{-j.\Delta\theta} \tag{4}$$

$$Z_{body} = R_{body} - \mathrm{j}.X_{body} \tag{5}$$

where K is the gain obtained from the sensor module, R_{body} and X_{body} are the resistance and reactance components of the body tissue, respectively.

An onboard low-power 2.5 GHz Bluetooth module provides wireless communication to control and acquire data from the impedance analyzer using a connected smartphone.

Fig. 2 shows a wearable ring with attached dry electrodes, along with a wrist band enclosure that contains the electronic hardware. A four-electrode method is used for BIA to reduce the errors caused by electrode-skin contact impedance [5]. The electrode-skin impedance is important for acquiring biopotential signals in BIA as a higher impedance can lead to a lower signal-to-noise ratio (SNR), a decreased amplitude of the bio-impedance signals, and eventually increased errors from the measurements [6]. Although the four-electrode method is less sensitive to contact impedance, problems still do arise [7]. In order to understand and compensate for the remaining errors from the measurements, the dry electrodes were quantitatively analyzed and the results were compared with those from the Ag/AgCl electrodes.



(a)

(b)

Figure 2. (a) Dry electrodes integrated with a silicone ring. Four-electrode measurement method is used, in which current is passed through the body using two source electrodes (I^+, I^-) and the voltage drop is measured across the body using two sense electrodes (V^+, V^-). (b) Wrist wearable electronics enclosure interfacing electrodes on the ring. Measurement is performed by the subject wearing the ring on one hand and placing the fingers of the other hand on the outer electrodes.

III. METHODOLOGY

Forty healthy volunteers participated in the study. The study was approved by the Institutional Review Board (IRB) at Rowan University (ID: PRO 2018002232). The subjects were asked to restrain from eating, drinking, and exercising for at least 3 hours before the experiment. All the measurements were recorded in a single sitting because a number of factors such as body temperature, humidity, caffeine level, and post-activity effects can greatly affect the bio-impedance signal. Bio-impedance measurements were performed on each subject using both the dry and the Ag/AgCl electrodes. For the dry electrodes, neither conductive gel nor abrasive cream was used. The subjects were asked to wear the ring on the index finger of their left hand and place the right-hand fingers on the outer electrodes as shown in fig. 2b. The electrodes on the ring deliver a high frequency (50 kHz) current signal from the left hand to the right hand through the upper body to measure the body impedance. Further details regarding the measurement procedure using the designed analyzer are provided in our previous report [4]. Fast Fourier transform (FFT) and noise levels were also examined to provide a more thorough understanding of the signal acquired using dry electrodes.

IV. RESULTS AND DISCUSSION

A. System performance evaluation

In order to verify the accuracy, the designed impedance analyzer is tested on an RC series circuit (tissue-equivalent model) by changing the values of resistor and capacitor [8]. The measured versus the calculated resistance and reactance at 50 kHz are shown in fig. 3. The system demonstrated errors of 4.3 %, 3.5 %, 3.7 %, 2.8 %, and 1.7 % when tested on circuits with calculated resistance of 104 Ω, 240 Ω, 379 Ω, 534 Ω, and 914 Ω, respectively. Similarly the system demonstrated errors of 2.4 %, 3.8 %, 4.47 %, 4%, and 6.25 % when tested on circuits with calculated reactance of -212 Ω, -144 Ω, -96 Ω, -67 Ω, and -31.8 Ω, respectively. According to the results, the measurement of resistance and reactance

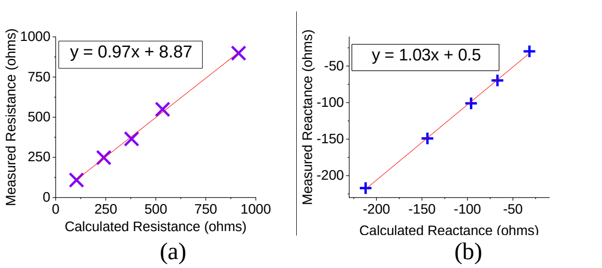
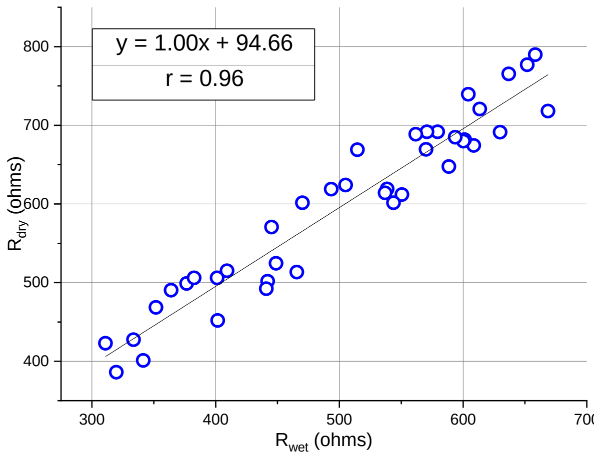


Figure 3. Impedance measurement using the designed system for a tissue equivalent model (RC series circuit) using frequency of 50 kHz. (a) Resistance measurement. (b) Reactance measurement.

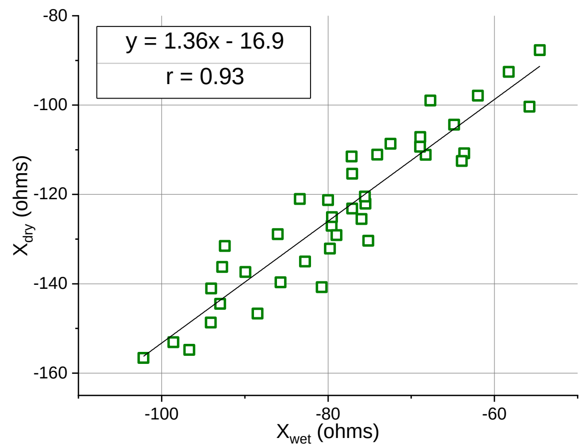
demonstrates high consistency and linearity with r = 0.998 for resistance and r = 0.997 for reactance. The power consumption of the entire system is measured to be 95.46 mW in active mode when supplied with a voltage of 3.3 V. The system has an impedance magnitude measurement range of $0\ \Omega < |Z| < 3.7\ k\Omega$ and a phase angle measurement range of $0° < |\theta| < 90°$.

B. Electrode analysis for bio-impedance measurement

After circuit verification the analyzer is used to measure the bioelectrical impedance using the dry ring electrodes as well as the gel-based Ag/AgCl electrodes. The major challenge while working with small-sized dry electrodes is the electrode-skin contact impedance. The ring electrodes are small with a dry surface, which imposes a larger contact impedance [9]. Although the four-electrode method is used to eliminate the contact impedance, this method cannot remove the parasitic effects imposed due to larger contact impedance. The mean parasitic impedance of dry electrodes is measured to be R_{par} = 23.73 Ω, X_{par} = -13.82 Ω. These values are acquired through experiments in which BIA is performed by placing both dry and wet electrodes at the same locations. The dry electrodes have a surface area of 260 mm^2 (26 mm × 10 mm) while the wet electrodes are more than 5 times larger at 1320 mm^2 (40 mm × 33 mm). The small size constricts the injected electrical current, leading to a higher measured impedance [10]. The absence of conductive gel in dry electrodes also contributes to the contact impedance. This contact impedance can be reduced by increasing the size of the electrodes [11]. However, it will be difficult to implement larger electrodes on wearable electronic devices due to their tight size constraints. Another issue is that the hands and fingers have a very high proportion of bone tissues. Although they only represent a very small part of the body, a large amount of impedance occurs. This is a source of error when computing the body composition. The average impedance caused due to hands (wrist-to-finger) for our subject group is measured to be R_H = 79.02 Ω and X_H = -34.42 Ω. All these factors require compensation to reduce errors in the final measurement. In order to compensate for these problems, the impedance of the upper body measured using the dry ring electrodes (finger-to-finger) is compared with that from the Ag/AgCl electrodes (wrist-to-wrist). Fig. 4a shows the obtained resistance of the 40 subjects using the wet Ag/AgCl electrodes (R_{wet}) plotted against the measured results using the dry electrodes (R_{dry}). The correlation coefficient for the linear regression is 0.96. The plot for the body reactance of dry versus wet electrodes is shown in fig. 4b. The correlation coefficient for the reactance plot is 0.93. These high correlation coefficients show that the measurement obtained from the Ag/AgCl electrodes can also be reliably obtained



(a)



(b)

Figure 4. (a) Linear regression for body resistance using dry electrodes (R_{dry}) and Ag/AgCl electrodes (R_{wet}). (b) Linear regression for body reactance using both electrodes X_{dry} and X_{wet}.

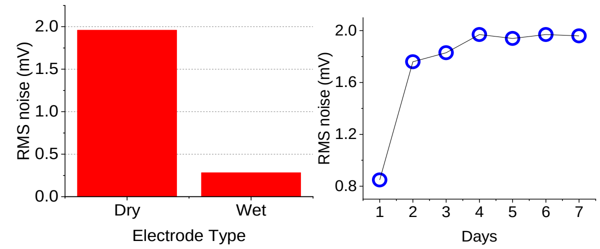
from the dry electrodes. The difference in the measured impedance can be obtained by subtracting the absolute average of the dry electrodes from that of the wet electrodes These values are then used for compensating for the errors caused by the hands and the parasitic effects caused by dry electrodes.

$$R_{diff} = R_{dry} - R_{wet} = 102.75\,\Omega \quad (6)$$

$$X_{diff} = |X_{dry}| - |X_{wet}| = 48.24\,\Omega \quad (7)$$

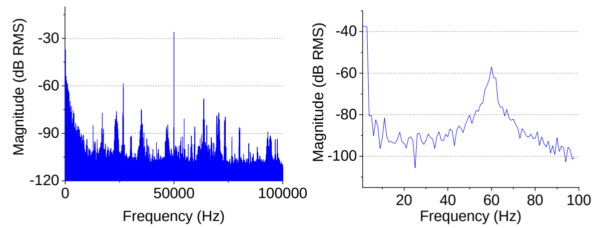
where R_{diff} is the difference of body resistance measured by dry electrodes (R_{dry}) and wet electrodes (R_{wet}). X_{diff} is the difference of body reactance measured by dry electrodes (X_{dry}) and wet electrodes (X_{wet}).

Fig. 5a demonstrates a comparison of root mean square (RMS) noise levels of dry and wet electrodes. The RMS noise is a measure of the absolute difference of a voltage



(a) (b)

Figure 5. (a) RMS noise level comparison of Ag/AgCl (wet) with designed ring electrodes (dry). (b) RMS noise of dry electrodes as a function of time.



(a) (b)

Figure 6. FFT of dry electrodes. (a) 0 Hz to 100 kHz. (b) 0 Hz to 100 Hz

from its baseline. The dry electrodes impose a higher noise level (1.96 mV) than the wet electrodes (0.282 mV). The dry electrodes have reduced contact areas with the skin due to the micro gaps between the two surfaces; while the conductive gel can fill the gaps and increase the contact areas for the wet electrodes. Fig. 5b shows the RMS noise measurements of the dry electrodes recorded for 7 consecutive days. It is observed that the noise level increases rapidly within the first 24 hours, mainly caused by the surface oxidation of copper, and becomes saturated on the third day. Other non-oxidative materials such as gold or platinum can be used as the electrode materials, but they are expensive and are not suited for wearable products for consumers

Fig. 6a shows the FFT spectrum of the signal recorded from the dry electrodes. The 50 kHz bio-impedance signal along with its harmonics can be clearly seen in the spectrum. Fig. 6b highlights the high amplitude noise at low frequencies. The power line noise at 60 Hz is often coupled in bio-potential measurements, along with other physiological signals such as electrocardiography (ECG), electromyography (EMG), and electroencephalography (EEG) at frequencies less than 10 Hz. All these noise signals can be easily removed using a narrow band-pass filter with a center frequency of 50 kHz.

The dry electrodes are lighter in weight, smaller in size, and at a lower cost when compared to the commercial Ag/AgCl electrodes. The body impedance of the 40 subjects measured by the dry electrodes is higher, due to the reduced electrode-skin contact areas, additional wrist-to-finger distance, and extra bone tissue impedance in hands. However, these errors can be quantified and then compensated for.

V. Conclusion

This paper presented a comparative analysis of dry and wet electrodes in wearable medical devices for bioelectrical impedance analysis. The designed electrodes were compared with the traditional Ag/AgCl electrodes and demonstrated a comparable performance. Analysis of the dry electrodes demonstrated a higher impedance and a lower signal-to-noise ratio. However, these factors were quantified and then compensated for. The impedance of 40 volunteer subjects was measured and an average difference was calculated to compensate for the errors caused by the dry electrodes. The bio-impedance signal was filtered out from the noise by using a narrow band-pass filter. Overall, the dry electrodes can provide an inexpensive solution for long term health monitoring in mobile biomedical devices by fulfilling wearability and user comfort. Future work includes developing a more accurate system by using other advanced techniques in data sciences such as fuzz logic and artificial intelligence to detect the physiological signals.

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