

# Impact of Skin–Electrode Interface on Electrocardiogram Measurements Using Conductive Textile Electrodes

Bahareh Taji, Shervin Shirmohammadi, *Senior Member, IEEE*, Voicu Groza, *Fellow, IEEE*, and Izmail Batkin

**Abstract**—Physicians’ understanding of biosignals as measured with medical instruments becomes the foundation of their decisions and diagnoses of patients, as they rely strongly on what the instruments show. Thus, it is critical and very important to ensure that the instruments’ recordings exactly reflect what is happening in the patient’s body so that the acquired signal is the real one or at least as close to the real in-body signal as possible. This is such an important issue that sometimes physicians use invasive measurements to obtain the real biosignal. Generating an in-body signal from what a measurement device shows is called “signal purification” or “reconstruction” and can be done only when we have adequate information about the interface between the body and the monitoring device. In this paper, first, we present a device that we developed for electrocardiogram (ECG) acquisition and transfer to PC. To evaluate the performance of the device, we use it to measure ECG and apply conductive textile as our ECG electrode. Then, we evaluate ECG signals captured by different electrodes, specifically traditional gel Ag/AgCl and dry golden plate electrodes, and compare the results, allowing us to investigate if ECG measured with the device is proper for applications where no skin preparation is allowed, such as ECG-assisted blood pressure monitoring devices. Next, we propose a method to reconstruct the ECG signal from the signal acquired by our device, with respect to the interface characteristics and their relation to the ECG. The interface in this paper is skin–electrode interface for conductive textiles. In the last stage of this paper, we explore the effects of pressure on skin–electrode interface impedance and its parametrical variation.

**Index Terms**—Bioimpedance, biomedical electrodes, electrocardiography, impedance measurement, pressure effects.

## I. INTRODUCTION

**A**MONG all vital signs that track an individual’s health, electrocardiogram (ECG) has a significant importance, since it includes meaningful information about the person’s heart and performance. It is the primary diagnostic for people with cardiac diseases and is a standard procedure in current cardiac medicine. Much effort has been spent lately to make ECG monitoring an easy and anywhere–anytime available procedure for people with cardiac problems and; especially, for those at risk of heart attack or stroke. Cardiovascular diseases

are some of the most prevalent and serious life-threatening health problems in the world, and represent the main cause of death for people between 44 and 64 years old, and the second most frequent cause of death for people between 24 and 44 years old [1]. Previous works on biosignals monitoring have come up with different devices commercially available. Such systems are mainly composed of three subsystems: the first subsystem is a special electrode as interface between the patient’s body and the device. The second subsystem is the signal conditioning and processing unit which is the core of the system. Since almost all acquired biosignals are of small amplitudes ( $\mu\text{V}$  to  $\text{mV}$  level) and low frequencies (0.1 Hz to 1 kHz), they generally include conditioning circuits that amplify the signals and filter external noises. The last subsystem is the communication unit which allows connecting the device to a host node that can be a computer or Smartphone or in another level transferring data using a web-based architecture to a hospital or emergency ward. Devices introduced in [2] and [3] have such performance that can diagnose some abnormalities and call the emergency service.

However, the electrodes used in current ECG acquisition devices may cause several problems. First, since they are in direct contact with the patient’s body, they can induce skin irritation. Second, they could make monitoring a cumbersome process if they are not convenient for people using them. Finally, the time duration that they can deliver constant signal quality without degrading is limited.

The aforementioned problems require new types of ECG electrodes and new systems based on them. Wearable electrodes are one solution used in bioelectric monitoring [4]. In [5], the authors present a flexible electrode based on conductive nanoparticles, applicable for long-term use, and embeddable in a belt. We will see more about this in Section II.

In addition to the electrodes themselves, another issue is that the ECG signals obtained with the electrodes have some distortions [6]. Various ways are proposed in the literature to mitigate distortion; e.g., authors in [6] suggest to maximize the input impedance of the amplifier and [7] proposes an algorithm to separate the muscular component of ECG signals to detect a rectified representation of cardiac activity. Furthermore, in [8] and [9], ECG purification is done by removing motion artifact as a significant source of noise. In [10], we presented a device able to measure ECG signals and also we did some experiments with conductive textile as our ECG electrodes. Besides, we used gel Ag/AgCl electrodes as reference for

Manuscript received July 6, 2013; revised September 16, 2013; accepted October 7, 2013. The Associate Editor coordinating the review process was Dr. Domenico Grimaldi.

The authors are with the School of Electrical Engineering and Computer Science, University of Ottawa, Ottawa, ON K1N 6N5, Canada (e-mail: btaji016@uottawa.ca; shervin@eecs.uottawa.ca; groza@eecs.uottawa.ca; vbatkin@rogers.com).

Color versions of one or more of the figures in this paper are available online at <http://ieeexplore.ieee.org>.

Digital Object Identifier 10.1109/TIM.2013.2289072

comparison. Electrodes were also tested on different parts of the body to find out whether conductive textile is a suitable choice for our device to obtain acceptable ECG to be used in blood pressure (BP) algorithms. In this paper, we extend our previous paper and we explore how skin–electrode interface impedance affects the ECG signal measurement and we also propose an algorithm to generate pure ECG signal free from skin–electrode impedance impacts. We also enhance our device to make it an all-inclusive platform for both ECG and BP measurement. ECG electrodes for our device are conductive textile strips embedded in the inner side of a cuff. We measure skin–electrode impedance under external pressure applied by a cuff included in the device, to study skin–electrode impedance variation under pressure especially because in ECG-assisted BP measurement algorithms, ECG is taken at the same time that the cuff is inflating and thus automatically applying pressure to the skin–electrode interface.

The rest of this paper is organized as follows. Section II gives a brief review of existing ECG electrodes and monitoring devices. Architecture of our proposed ECG measurement system is described in Section III. The results obtained by our system are presented in Section IV. This is followed by looking into influences of impedance of the electrode and its contact interface with the skin on the quality of the acquired ECG, as presented in Sections V and VI. In Section VII, we measure the interface impedance when it is under external pressure and follow its variation rate with increasing pressure. Finally, this paper concludes in Section VIII.

## II. RELATED WORK

### A. ECG Monitoring Systems in the Literature

Nowadays cardiac patients have the opportunity for their ECG signal to be monitored continuously using devices specific for this purpose. Portable biosignal monitoring devices are now commercially available and used in telemedicine and home health-care scenarios for disease prevention and management, early diagnoses, and home rehabilitation using monitoring devices [4]. In addition to commercially available devices, there are many prototypes presented in the literature. In this subsection, we briefly explain some of them and their features.

The authors in [3] present an ECG measuring device which sends patients' ECG data via a web-service-oriented architecture. It is not only able to acquire ECG, but can also provide diagnosis using "patients" clinical history and call emergency services if necessary. In this paper, standard Ag/AgCl electrodes are used. Another device presented in [2] is an ambulatory ECG device using disposable electrodes and equipped with a built-in warning system. It can diagnose some prevalent types of heart abnormality, such as atrial fibrillation, and an alert is activated once an abnormal ECG is detected. In [11], a wearable ECG monitoring device is presented that claims to be most convenient for users, particularly in long-term applications [11] since it uses dry capacitive electrodes embedded in a cotton T-shirt. In another approach, authors in [12] explain a different perspective of ECG monitoring by introducing a system which generates ECG signals as opposed

to monitoring ECG signals. This system allows researchers and ECG monitoring device developers to assess and analyze their device performance in an empirical and realistic way in addition to numerical and theoretical methods [12].

### B. ECG Electrode

Irrespective of the monitoring system, there is an interface between the body as the signal source and the monitoring system that collects signals from the body. This interface is in direct contact with the body and it is called an "electrode." Thus, ECG electrodes sense the ECG biosignal that comes from a biological source, such as the body. Because the main role of an ECG electrode is to change the voltage from its ionic form in the body to its electron form in the wires, all electrodes are polarizable or nonpolarizable, also known as dry or wet electrodes, respectively [13]. Below, we explain each category.

1) *Nonpolarizable Electrodes:* The ECG electrodes clinically in use are adhesive gel Ag/AgCl electrodes which have been in use for a long time. In this group of electrodes, an electrochemical process between the gel and the biological tissue yields a conductive path between the patient's skin and the electrode [13], and a real electrical current flows along this path. The advantages and disadvantages of nonpolarizable electrodes are as follows.

- 1) They are adhesive, thus can easily be fixed on the skin, preventing motion artifacts.
- 2) They show very clean and reliable ECG signals.
- 3) The adhesive part and the gel inside them can cause skin irritation and contact skin dermatitis if they are in use for long time [13].
- 4) They normally require skin preparation in advance, such as shaving, using alcohol to clean the contact area and even using sand paper to remove the dead layer of the skin [13].
- 5) Technically, the most significant drawback of nonpolarizable ECG electrodes is that the signal they detect will degrade when the gel inside them dries out. Therefore, they are not appropriate for long-term use and are considered disposable electrodes. It is inconvenient to replace electrodes, making them problematic for the users in home healthcare and telemedicine.

2) *Polarizable Electrodes:* In contrast to nonpolarizable electrodes, in polarizable electrodes no actual electrical current flows. Instead, a displacement current occurs as a result of a change in ionic concentration in the electrode–skin interface. These electrodes work based on the capacitive coupling between a conductive material and skin. They do not need any kind of gel; instead, they operate by moisture and sweat on the skin [13]. Dry electrodes vary from simple stainless steel plates to new flexible textiles capable of conducting electrical potential. Dry electrodes can be categorized as contact or noncontact, as explained below.

a) *Dry contact electrodes:* The simplest polarizable electrode is a metal disc in direct contact to the skin. Although the performance of such electrodes is comparable to that of gel electrodes, due to sweat and humidity of the skin, their

use is still limited because of stiffness and the fact that they can cause skin irritation. Another disadvantage is that they are highly sensitive to motion artifacts, since they are not fixed on the body unless they are embedded in a belt, in which case the belt should maintain tight contact between the metal disc and the skin.

One of the newest electrodes is conductive textile, with the advantage that it is just like an ordinary fabric, soft, and flexible and it does not cause skin irritation; moreover, it is appropriate for chronic applications. Conductive textile can be made of various types of conductive yarn, including silver-coated nylon (AgNy), stainless steel yarn and silver-coated copper (AgCu) [14]. Conductive textiles are increasing in popularity because they are easy to use, and unlike Ag/AgCl electrodes do not need to be changed in long-term applications. Moreover, they are not stiff and can be shaped like an ordinary wristband or embedded inside garments that the user can wear [15]. Hence, they are good options for long-term applications or home healthcare monitoring devices. In [16], authors compare textile electrodes made with different techniques of attaching the conductive yarns to the fabric, including weaving, knitting, and embroidering, showing the latter to have better results in terms of accuracy.

*b) Dry noncontact electrodes:* The difference between this category of electrodes and the previous one is that there is an insulating layer between the skin and the electrode. Dry noncontact electrodes work based on the capacitive coupling between the skin and the conductive disc which leads to the polarization of the electrode, causing displacement current to flow for a while. To achieve strong capacitive coupling, we can increase the area of contact, use a thin insulator layer or include an insulator with a high dielectric constant [11]. Similar to dry contact electrodes, dry noncontact electrodes are also sensitive to motion artifacts, but show better behavior in terms of decreased skin irritation.

In this paper, we opt for conductive textile made of AgNy yarns as our ECG electrode, as shown in Fig. 1, because AgNy textile shows bactericidal effect [17]. We will show that conductive textile is capable of delivering good quality ECG signal. Furthermore, this type of electrodes does not need skin preparation; thus, it is a good choice for applications, such as home healthcare where the user would follow only a regular procedure. Our device can also be used for BP measurement using the algorithm presented in [18].

To detect ECG with conductive textile electrode, first we need to develop a hardware foundation capable of acquiring ECG. The architecture of such system is described in the next section.

### III. PROPOSED ECG MEASUREMENT SYSTEM

#### A. Data Acquisition

Fig. 2 shows the block diagram of our device. After the first part, which is the conductive textile ECG electrode made of silver-coated nylon filaments, there is an ECG board consisting of amplifiers, three stages of filters (high pass, low pass, and notch), analog to digital convertor (ADC), and a master processor unit (MPU). The signal collected by the



Fig. 1. AgNy conductive textile [10].

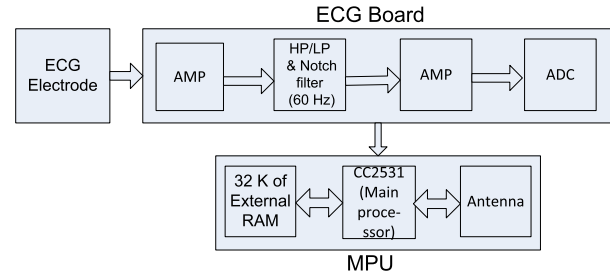


Fig. 2. Device block diagram [10].



Fig. 3. MPU [10].

ECG electrodes is fed to the ECG board amplifiers and the amplified signals are filtered by high pass, low pass, and notch filters. A 60-Hz notch filter is specifically applied to remove power line noise. The output signal of the filter stage is fed to the ADC and is digitized. Our ADC is ADS8320, Texas Instruments, with 16 bits of resolution and maximum sampling rate of 100 KS/s. The output of the ADC is then fed to the MPU, which is shown in Fig. 3 and consists of a CC2531F256 microprocessor from Texas Instruments, 32 kB serial RAM memory and an 802.15.4-based wireless control node [10]. CC2531F256 has 256 kB of built-in programmable flash memory that hosts the embedded software controlling the system. Furthermore, the MPU has an on-board chip antenna and also a ribbon cable connector implemented on it, the latter used to connect the device to embedded software programming resources.

#### B. ECG Capturing Procedure

To capture ECG, we use IAR embedded workbench to upload a control code sequence in the built-in flash memory of the CPU. The code consequence is written in a way that starts data acquisition (DAQ) after a push button is pressed by the user, and then digitized data are sent to a PC via the antenna of the MPU. The PC receives the data and another software, written in MATLAB, processes the received samples, and shows the signal on the screen [10].

To capture data, we apply two wristbands made of conductive textile on the subject's bicep with 10 cm of distance between them. Since we use conductive textile, there is no need for skin preparation and it causes no skin irritation. For our prototype, we have not implemented an accelerometer yet, so we ask the user to avoid moving during measurements.

In some commercial ECG monitoring systems, an accelerometer is included; therefore, if the user is running or walking, his/her ECG signal is adjusted accordingly.

#### IV. EVALUATION OF THE ACQUIRED ECG SIGNAL

##### A. ECG-Assisted BP Methods

BP indicates the pressure of blood against the walls of the arteries as the blood is circulating in them. Since it is an important signal, many BP devices are designed specifically to allow people to monitor their BP. One of the most popular noninvasive methods of measuring BP is the oscillometric method [19]. This method is not only used for measuring systolic and diastolic BP, but also has other applications in heart rate variability measurement [20]. This method requires a BP cuff wrapped around the patient's bicep or wrist. Steadily inflating the cuff gradually stops blood flow in arteries underlying it. A pressure transducer is used to record the pressure. On the other hand, oscillations in the cuff pressure are made by arterial pulsation and when they have their maximum amplitude, cuff pressure is equal to mean arterial pressure. In this method, systolic and diastolic BP are estimated from the mean BP and oscillation pattern [21]. Users have to ensure that the cuff is at the heart level; otherwise, its measurement will not be accurate. Moreover, in cases, such as obesity, heart arrhythmia, or even posture change, arterial amplitude that the cuff senses is not distinct [18]. The inaccuracies evident in this method have caused researchers to try to find a more reliable method for BP measurement. Some extract and remove the effects of other signals, such as respiratory signal on BP, thereby resulting in a higher level of accuracy [22]. Some other use more consistent signals, such as ECG [18], [19], as described below.

The authors in [19] apply ECG in an improved oscillometric BP measurement algorithm. In the first step, they apply the pulse transit time, the time delay between R-peaks and photoplethysmographic pulses, for estimating the standard deviation of both systolic and diastolic BP. In the second step, they use signal-to-noise ratio of the ECG signal to ensure that the patient was in rest. They found that the measured BP was valid if this ratio was below a given threshold. In [18], the authors propose a new method of BP estimation concerning the R-peaks of ECG to improve the oscillometric pulse peak detection which leads to better estimation of BP. R-peak information was also applied to determine the maximum amplitude of oscillometric pulses interleaved between two consecutive R-peaks. Eventually, the maximum amplitude algorithm was used to estimate systolic and diastolic BP. The authors also developed a prototype to implement their ECG-assisted BP measurement algorithm. In their prototype, two ECG electrodes made of AgNy conductive textile are used to acquire the ECG signal. One electrode was embedded in the cuff maintaining contact with the patient's bicep when the cuff was wrapped around the arm. The other electrode was in the shape of a wristband, and was worn on the patient's wrist.

Next, we describe experiments to find out how our device and electrodes behave in terms of detecting ECG signals for algorithms similar to the above two.

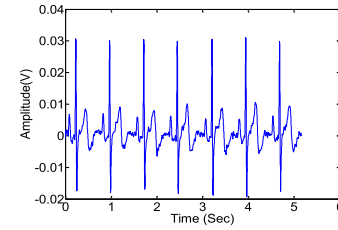


Fig. 4. ECG( $t$ ) captured from wrists using Ag/AgCl electrodes.

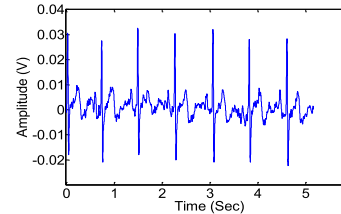


Fig. 5. ECG( $t$ ) captured from wrists using textile electrodes.

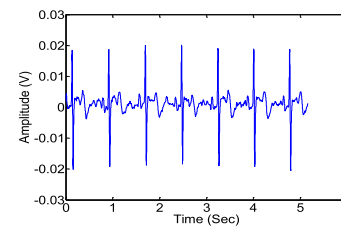
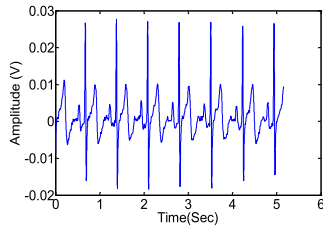
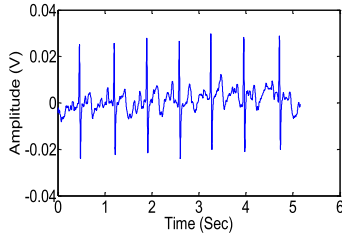
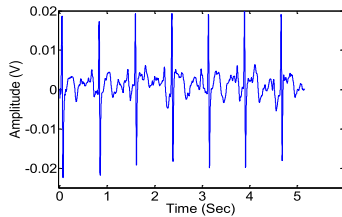
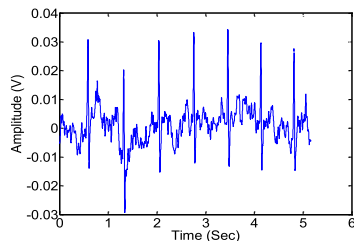
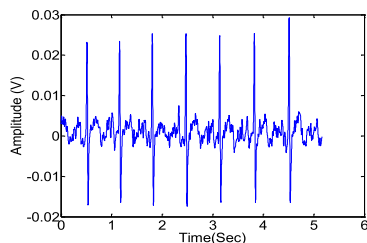


Fig. 6. ECG( $t$ ) captured from wrists using one Ag/AgCl and one textile electrode.

##### B. Experimental Results

In this part of this paper, we evaluate by what type of electrode and in what position on the body we will obtain a proper ECG signal to be used in ECG-assisted BP measurement algorithms. A proper ECG signal means a signal without any missing R-peaks and also without any falsely detected R-peaks due to severe distortion or noise. To do so, we collect ECG signals from one person's wrists and biceps using both the conductive textile electrode and regular gel Ag/AgCl (RedDot) electrode and compare the results. Next, we acquire the ECG signal using two golden plates touched to the fingertips. The golden plates are dry contact electrodes with a size of  $12 \times 18 \text{ mm}^2$ . In the next step, we acquire the ECG signal using one golden plate touched by the index fingertip of one hand and conductive textile on the bicep of the opposite hand as the second electrode. The results of all these measurement configurations are shown in Figs. 4–11. It should be mentioned that we did not use any kind of skin preparation before DAQ. Figs. 4–6 show the results of our experiments for wrists with various electrodes. Figs. 7–9 show the results for ECG detection from biceps with various electrodes.

As can be seen from the figures, the ECGs taken with Ag/AgCl electrodes have stable P-, R-, and T-wave amplitudes, while signals taken with textile electrodes exhibit some variation in amplitudes and are more contaminated by noise; but, their main parts specifically; P-, QRS-, and T-waves are still recognizable. Thus, the acquired signals are acceptable for

Fig. 7. ECG( $t$ ) captured from biceps using Ag/AgCl electrodes.Fig. 8. ECG( $t$ ) captured from biceps using textile electrodes.Fig. 9. ECG( $t$ ) captured from biceps using one Ag/AgCl and one textile electrode.Fig. 10. ECG( $t$ ) captured from fingertips using gold plate electrodes.Fig. 11. ECG( $t$ ) captured from fingertips using one gold plate and one textile electrode.

our purpose. We can also see that when we use two different electrode types, we obtain different R-peak amplitudes. These observations were valid for both wrists and biceps.

Fig. 10 shows the ECG signal acquired from two index fingertips touching golden plates, while Fig. 11 depicts result

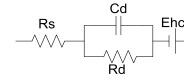


Fig. 12. Skin-electrode impedance equivalent circuit [13].

of collecting ECG from one index fingertip touching one golden plate and one conductive textile on the bicep of the opposite arm.

As the graphs imply, ECG acquired by two golden plates and from fingertips shows severe fluctuation. This is because when the user touches the golden plate, the external pressure applied to the electrode varies as time passes; therefore, the contact area between fingertip and electrode is not constant. Contact area affects the skin–electrode interface and its impedance which strongly affects the acquired ECG signal. In contrast, the ECG taken with one golden plate and one conductive textile has more stability in R-peak amplitudes. However lots of noise in T- and P-waves is observed which is the impact of using golden plates and due to aforementioned reasons.

To sum-up, in all tested configurations, the acquired ECG signals are suitable for ECG-assisted BP measurement algorithms, since we did not detect any extra R-peaks, nor did we miss any of them. However, some other factors still need to be investigated. Skin–electrode impedance and its effects on the acquired ECG is our first concern, followed by determining how this impedance varies under pressure. This is because in our device, one ECG electrode made of conductive textile, will be embedded in a cuff, and ECG will be measured when the cuff is inflating; thereby, applying pressure to the skin–electrode interface. Therefore, we first explore the effects of the skin–electrode interface impedance of conductive textile on ECG (Section V describes the model, and Section VI presents the results), and then study the effects of applying pressure on skin–electrode interface impedance (Section VII).

## V. SKIN-ELECTRODE IMPEDANCE MODELING AND MEASUREMENT

### A. Skin–Electrode Interface Models

To better understand the electrical behavior of the skin–electrode interface, an equivalent electrical circuit can help. This circuit can be modeled in single or double-time constant, as explained next.

1) *Single-Time Constant Model*: Swanson and Webster [23] have developed a model for skin–electrode impedance, as shown in Fig. 12. Their model is a combination of a resistor in series with a paralleled resistor and capacitor.  $E_{hc}$  is the voltage between the skin and electrolytes, such as gel. Thus, the model is mostly applicable for wet electrodes. The capacitance  $C_d$  represents the electrical charge between the electrode and skin [13], [24], while  $R_d$  stands for the resistance that occurs between the skin and electrode during charge transfer [23].  $R_s$  represents the electrolyte gel (if any), sweat, and the underlying skin tissue [25], [26].

Equation (1) represents the total impedance of this interface as a function of frequency. Equations (2) and (3) represent the

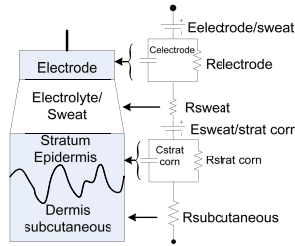


Fig. 13. Skin-electrode interface and electrical equivalent [27].

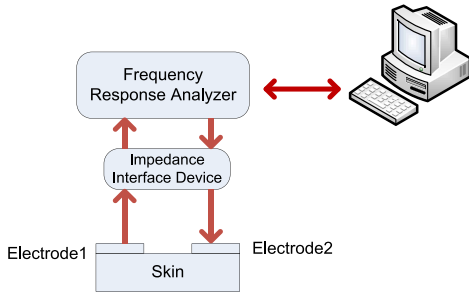


Fig. 14. Block diagram of experimental set-up applied in this paper.

real and imaginary parts, respectively; hence, (4) can be used to determine magnitude of impedance

$$Z(\omega) = R_s + \frac{R_d / jC_d\omega}{R_d + 1/jC_d\omega} = R_s + \frac{R_d}{1 + j\omega C_d R_d}$$

$$= \frac{R_s + R_d + j\omega C_d R_s R_d}{1 + j\omega C_d R_d}, \quad \omega = 2\pi f \quad (1)$$

$$\text{Re}(Z(\omega)) = R_s + \frac{R_d}{1 + \omega^2 C_d^2 R_d^2} \quad (2)$$

$$\text{Im}(Z(\omega)) = -\frac{\omega C_d^2 R_d^2}{1 + \omega^2 C_d^2 R_d^2} \quad (3)$$

$$|Z(\omega)| = |\text{Re}(\omega) + j\text{Im}(\omega)|$$

$$= \left| R_s + \frac{R_d}{1 + \omega^2 C_d^2 R_d^2} - j \frac{\omega C_d^2 R_d^2}{1 + \omega^2 C_d^2 R_d^2} \right|. \quad (4)$$

2) *Double-Time Constant Model*: Fig. 12 proposes a general model for all electrodes, and is known as the single-time constant model [27]; however, there are some more complicated models for skin–electrode impedance, such as Neuman’s [27] model, which uses two stages of the single-time constant model. The first stage only represents skin, while the second only represents the electrode. This is called double-time constant model, and is illustrated in Fig. 13. The authors in [28] studied both models in terms of ECG measurement and frequency response and showed that double-time constant model exhibits more accurate results. However, in this paper, we use the simplified single-time constant model due to restrictions with our measurement devices.

### B. Set-Up and Measurement Configurations

A block diagram of the experimental set-up of the model is shown in Fig. 14. We use 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

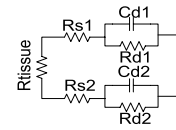


Fig. 15. Schematic diagram of skin-electrode impedance measurement system.

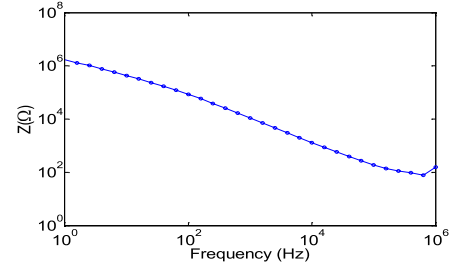


Fig. 16. Skin-electrode impedance frequency response.

electrode, as well as measuring the voltage drop between two electrodes.

1) *Devices and Measurement Configuration*: Our measurement configuration is shown in Fig. 15, showing two electrodes and their interface with user’s tissue.  $R_{tissue}$  represents the resistance of tissues between two electrodes.

We need to determine the impedance of a single skin–electrode interface, while our measurements show the total value of both electrodes plus tissue impedance. If both electrodes are identical in terms of type, material and size, we can assume that they have the same skin–electrode interface. In the experiments, we applied two identical electrodes made of medical-grade silver plated 92% nylon and 8% Dorlastan stretchable conductive textile. Each electrode was 0.50-mm thick and  $3 \times 18 \text{ cm}^2$  in size. They were, therefore, identical and we could divide the total amount by two to determine the skin–electrode interface for a single electrode. For a healthy human’s middle upper arm,  $R_{tissue}$  is known to be almost  $150 \Omega$  [29], while the electrode skin impedance is larger than  $1 \text{ M}\Omega$ . According to this, in our study  $R_{tissue}$  is assumed to be negligible and its contribution is considered in  $R_s$ .

The bioimpedance measurement system we used includes a frequency response analyzer (Model# 1255B, Solartron Analytical, U.K.), an impedance interface device (Model# 1294A, Solartron Analytical, U.K.) and a personal computer. Fig. 16 shows our impedance measurement results using this system.

### C. Estimating Skin–Electrode Impedance Model Components and Verification

Applying the least squares nonlinear curve fitting method, we tried to estimate three components of the skin–electrode impedance model specifically,  $R_s$ ,  $R_d$ , and  $C_d$ . We implemented this method in MATLAB (R2008a) by applying the “lsqcurvefit” function.

We have already measured the magnitude of impedance against frequency in Fig. 16. We now need to estimate our model parameters such that  $|Z(\omega)|$  of the model matches

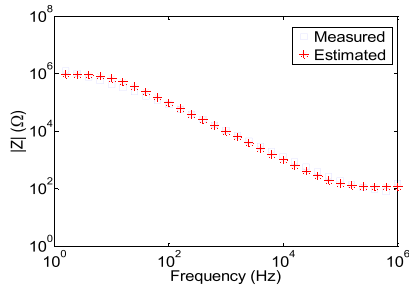


Fig. 17. Measured data and estimated model.

our observation and the curve in Fig. 16. The least squares nonlinear curve fitting method provides the best-fit impedance model from the measured impedance values; this is done by minimizing the summed square of the error between the data point and the fitted model. The MATLAB code based on the experimental data returned the following values for the three components:

$$R_d = (9.8726e + 5) \Omega, \quad C_d = (1.5884e - 8) F$$

$$R_s = 116.6971 \Omega.$$

To verify whether our model matches the observation or not, we substituted the above values in their respective components in (4). We then drew the graph of  $|Z(\omega)|$  as a function of frequency and observed both the measured and estimated curves on the same plane, as shown in Fig. 17, where we can see the validity of our model.

## VI. EFFECTS OF THE SKIN-ELECTRODE INTERFACE IMPEDANCE ON ECG

### A. Effects of $|Z(\omega)|$ on ECG Acquiring

Considering the general circuits for ECG detection and knowing the value of  $|Z(\omega)|$ , a basic question comes up: Do we miss any information in the ECG waveform because of the ECG monitoring device is fed by a signal affected by  $|Z(\omega)|$ ? If the answer is yes, then the next question is how is ECG affected? Besides, we need to find a way to compensate this effect. In [30], the authors clearly answer the first question and show that there is an error ratio in ECG signal associated with the skin-electrode impedance. Their suggestions for improvement, include decreasing the skin-electrode impedance, increasing the input resistance of the amplifier or compensating skin-electrode impedance effects in acquired ECG. In another paper, presented in [31], the electrode is a gel electrode and both its model and its impact on ECG registered on skin-electrode interface are investigated by applying a square stimulus signal and recording its response.

In this paper, we propose another approach to address this issue and determine the effects of skin-electrode interface on recorded ECG. Our approach is based on skin-electrode impedance variation per frequency. We also consider input impedance of the amplifier. Fig. 18 gives a typical view of an ECG acquiring circuit.  $R_{i1}$  and  $R_{i2}$  are the input impedances of amplifiers, while  $Z_1(\omega)$  and  $Z_2(\omega)$  present the skin-electrode impedance of electrodes collecting the signal.

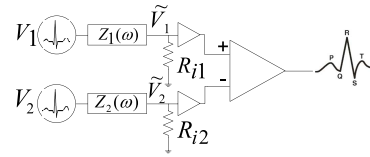
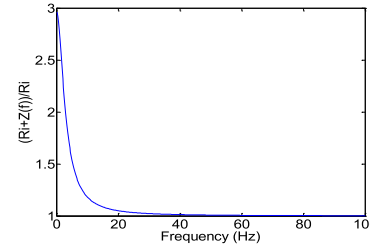


Fig. 18. General circuit for ECG acquiring.


 Fig. 19. Magnitude of  $(R_i + Z)/R_i$  per frequency.

The relations are shown in (5)–(7)

$$\tilde{V}_1(\omega) = V_1(\omega) \times \frac{R_{i1}}{R_{i1} + Z_1(\omega)} \quad (5)$$

$$\tilde{V}_2(\omega) = V_2(\omega) \times \frac{R_{i2}}{R_{i2} + Z_2(\omega)} \quad (6)$$

$$\text{Acquired ECG}(\omega) = \tilde{V}_1(\omega) - \tilde{V}_2(\omega). \quad (7)$$

In our case  $Z_1(\omega)$  and  $Z_2(\omega)$  are equal and amplifiers used for both electrodes are also identical ( $R_{i1} = R_{i2}$ ). Thus, ECG can be presented as follows:

$$Z_1(\omega) = Z_2(\omega) = Z(\omega), \quad R_{i1} = R_{i2} = R_i$$

$$\text{Acquired ECG}(\omega) = \frac{R_i}{R_i + Z(\omega)} (V_1(\omega) - V_2(\omega)). \quad (8)$$

However, the real ECG signal before it is affected by skin-electrode impedance, called In\_body ECG, is

$$\text{In\_body ECG}(\omega) = V_1(\omega) - V_2(\omega). \quad (9)$$

From (8) and (9)

$$\text{In\_body ECG} = \text{Acquired ECG}(\omega) \times \frac{R_i + Z(\omega)}{R_i}. \quad (10)$$

As can be seen, all equations are in the frequency domain, because  $Z$  varies by frequency. However, acquired ECG signals shown in Figs. 4–11 are in the time domain. Therefore, to calculate the In\_body ECG in the time domain, we first need to find the acquired ECG in the frequency domain, calculate the In\_body ECG in the frequency domain, then translate this from the frequency domain to the time domain. Eventually, we need to compare acquired ECG (t) with In\_body ECG and discuss about distortions caused by skin-electrode interface, as well as how this interface can influence the interpretation of ECGs from a medical perspective.

We first have to calculate  $R_i + Z(\omega)/R_i$  using the values of all components, including  $R_d$ ,  $C_d$ , and  $R_s$ . The input resistance of the amplifier we used was measured and found to be 1 MΩ. By substituting all values in equation  $R_i + Z(\omega)/R_i$  and plotting its graph per frequency in linear scale, the graph in Fig. 19 is obtained.

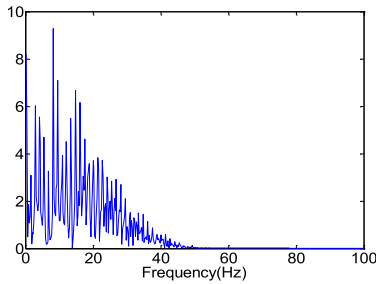


Fig. 20. Acquired ECG in frequency domain.

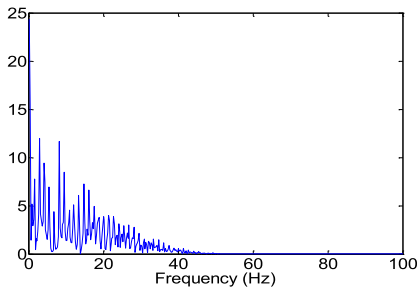


Fig. 21. In-body ECG in frequency domain.

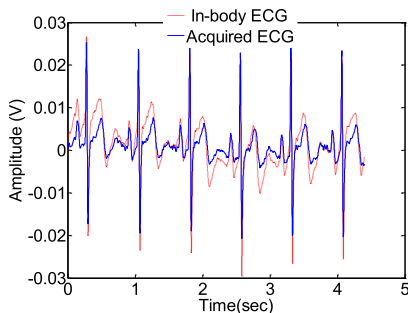


Fig. 22. Acquired and in-body ECG waveforms.

Consequently, we needed to find acquired ECG in the frequency domain. To obtain that, we apply Fourier transform in MATLAB by applying the “fft” function on samples we obtained when we took ECG measurements in the time domain. Fig. 20 depicts the FFT of the acquired ECG.

To obtain in In\_body ECG the frequency domain, according to (10), we need to multiply the data shown in Fig. 19 by the data shown in Fig. 20. The result is shown in Fig. 21.

What we have obtained so far is In\_body ECG in the frequency domain. By integrating this, we convert it to time domain. Integration is carried out by applying the “ifft” function in MATLAB. Fig. 22 presents the In\_body ECG( $t$ ) and acquired ECG( $t$ ) on the same graph to make the differences easy to observe. We must also mention that, since ECG is a periodic signal, we eliminated the phase shift effects resulting from the imaginary part of impedance. This also makes comparison of the primary ECG signal and the obtained one easier. We repeated this process for several acquired ECG( $t$ ) samples and they all show the same results. Only one sample is presented in this paper due to shortage of space.

In next section, we will discuss the differences between these two signals and any possible missed information.

It should also be noted that temperature is known to affect skin impedance and thus skin–electrode impedance. Authors in [32] show that resistance and reactance of skin impedance decrease when temperature increases. Based on their research skin impedance is highly dependent on the total body water. Higher temperature results in body dehydration and causes skin impedance variation. To avoid skin–electrode impedance variation due to change of temperature, we performed all our experiments in the same ambient temperature of 22 °C.

### B. ECG Morphological Changes and Interpretation

As expected, In\_body ECG( $t$ ) and acquired ECG( $t$ ) have differences in some parts of the ECG signal, specifically in lower frequency contents, such as T-wave. The typical frequency range of T-peaks is between 4 and 7 Hz [33]. Therefore, they are considered low frequency contents of ECG as compared to R-peaks and P-peaks being in the frequency range of 10–17 Hz [33]. Hence, the question is what risks are associated with this type of information missing in ECG and what misinterpretation may occur according to this error from medical point of view? Since T-wave is affected, any information it carries can be missed. Since it is the most unstable part of the ECG recording [34], the T-wave changes with many ECG abnormalities. For instance, it has a critical role in diagnosing ST elevation, which is a marker of acute coronary occlusion [35].

Another parameter that is affected by the T-wave hump amplitude is the  $R/T$  ratio. This ratio is used clinically to monitor sympathetic tone during isoflurane anesthesia [36]. In addition to the aforementioned parameters, the  $Q-T$  interval, which indicates specific form of ventricular tachycardia [34], is also affected by the skin–electrode. Thus, if T-wave is not detected accurately,  $Q-T$  prolongation may be missed.

In the next step, we try to find a way to minimize these effects. One possible solution for this is to minimize the skin–electrode interface impedance. The technical method we selected is to apply pressure to make a better contact between the electrode and the skin that leads to smaller skin–electrode impedance. In [37], the authors showed that applying pressure will decrease the impedance. We apply pressure to our electrode; i.e., the AgNy conductive textile, and investigate how its skin–electrode impedance changes. This is described next.

## VII. APPLYING EXTERNAL PRESSURE

### A. Enhancing the Device

For our device to be an appropriate platform for implementing ECG-assisted BP measurement, a cuff is needed. Hence, to have a fully automated device, we used 5 V motor and an air pump to the device. The motor is driven by the MPU and powered up by batteries. It pumps air into the bladder inside the cuff and inflates it. To measure the cuff’s pressure, we add a pressure sensor (26PC series, Honeywell) to the device, which converts pressure to voltage. Its output is read by the ADC unit. Hence, we can watch the pressure of the cuff and control it by MPU which is a part of BP measurement procedure.



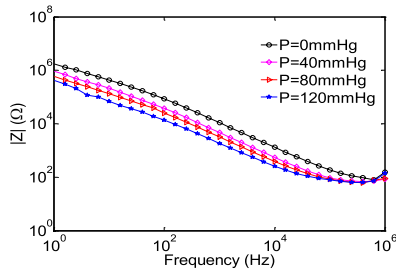
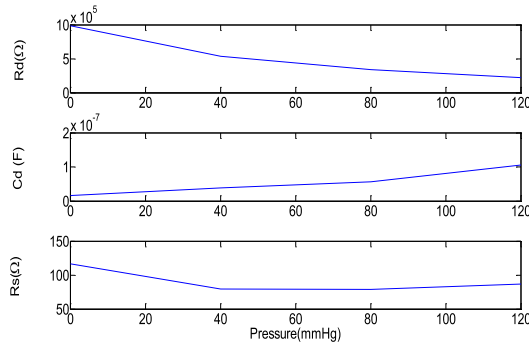
Fig. 23. Measured  $|Z|$  under various pressure values.

Fig. 24. Component variation under pressure.

### B. Measurement Set-Up

We setup the same system as explained in Section VI; the only difference being that when we want to measure the impedance, we apply pressure on the cuff, while electrodes are located under the cuff. The pressure of the cuff remains constant during each measurement. We applied pressure from 0 to 120 mmHg in 40 mmHg steps. For consistency with our previous measurements, we used the singletime constant model for the interface and the least squares nonlinear curve fitting method for component value estimation. The variation trend line of each parameter  $C_d$ ,  $R_d$ , and  $R_s$ , was tracked and drawn in the following graphs.

### C. Results and Discussion

Fig. 23 shows  $|Z(\omega)|$  variation per frequency. The frequency varied between 0 Hz and 1 MHz. Applying MATLABs “lsqcurvefit” function, all components were determined for each graph. Fig. 24 shows the rate of variation for each component.

As can be seen from the graphs,  $R_d$  continued to decrease while external pressure increased. This is due to  $C_d$ , which is the electrical charge between the skin and the electrode. The value of any capacitor can be expressed by

$$C = \epsilon_0 \epsilon_r \frac{A}{d} \quad (11)$$

where  $C$  is the capacitance in Farad (F),  $\epsilon_r$  shows the dielectric constant of the material between two parallel plates,  $\epsilon_0$  stands for the electric constant (approximately  $8.85 \times 10^{-12} \text{ Fm}^{-1}$ ),  $A$  is the effective common area between two parallel plates ( $\text{m}^2$ ) and  $d$  (m) is the distance between the parallel plates [38].

This equation clarifies  $C_d$ 's response under pressure. Applying pressure makes the effective area between the two parallel plates larger, while making the distance between them smaller. Since  $C$  is directly proportional to the effective area  $A$  and inversely proportional to distance  $d$ , pressing the electrode toward the skin results in a larger value of  $C_d$ .

We were also able to anticipate the behavior of  $R_d$  under pressure.  $R_d$  is the resistance occurring in the charge transfer path between the skin and the electrode. Resistance of any material can be formulated as in

$$R = \rho \frac{L}{A} \quad (12)$$

where  $R$  is the resistance in ohm ( $\Omega$ );  $\rho$  is the resistivity of the material with the dimension of ( $\Omega \text{ m}$ );  $L$  is the length of the material with the dimension of (m);  $A$  is the cross sectional area of the conductor material with the dimension of ( $\text{m}^2$ ) [38].

When applying pressure, the contact area between the skin and electrode increases, resulting in smaller values of  $R_d$ .

For analyzing the behavior of  $R_s$ , we can again refer to (12). Applying pressure creates more contact area between sweat and the electrode, thereby lowering the value of  $R_s$ . In our case, we let the sweat evaporate after each measurement; thus,  $R_s$  remained approximately constant after 40 mmHg of pressure.

## VIII. CONCLUSION

In recent years, conductive textile has become increasingly popular in medical applications particularly for home health-care devices. This is because it is a user-friendly type of electrode, bringing users convenience and comfort as compared to gel electrodes, which are disposable and cause skin irritation. In this paper, we developed a hardware-based configuration appropriate for acquiring ECG signals with conductive textile, and we acquired ECG using various types of electrodes from different locations on the body. This was done to evaluate the overall quality of acquired ECG by textile electrodes. In the next step of this paper, we looked into the problems that skin–electrode impedance can cause for the quality of ECG signal registered at the electrodes. A new method of reconstructing ECG was also proposed and we showed that the skin–electrode interface influences the quality and shape of the ECG signal registered at the electrode. skin–electrode impedance impact on ECG was shown and the procedure for detecting the differences between affected ECG and in-body ECG was explained. Moreover, we proposed a method to optimize the ECG acquired by our electrodes. This method was based on minimizing skin–electrode interface impedance by applying pressure.

For our future work we will expand our research to do under pressure skin–electrode impedance measurement for more subjects to analyze how skin–electrode interface vary according to age, gender, and body mass index. We will also take ECG under different pressure values to evaluate the differences between acquired ECG and in-body ECG under pressure.

## REFERENCES

- [1] A. Karilainen, S. Hansen, and J. Muller, "Dry and capacitive electrodes for long-term ECG-monitoring," in *Proc. 8th Annu. Workshop Semicond. Adv.*, 2005, p. 156.
- [2] C. T. Lin, K. C. Chang, C. L. Lin, C. C. Chiang, S. W. Lu, S. S. Chang, et al., "An intelligent telecardiology system using a wearable and wireless ECG to detect atrial fibrillation," *IEEE Trans. Inf. Technol. Biomed.*, vol. 14, no. 3, pp. 726–733, 2010.
- [3] C. De Capua, A. Meduri, and R. Morello, "A smart ECG measurement system based on web-service-oriented architecture for telemedicine applications," *IEEE Trans. Instrum. Meas.*, vol. 59, no. 10, pp. 2530–2538, 2010.
- [4] A. Lymberis, "Advanced wearable sensors and systems enabling personal applications," in *Wearable and Autonomous Biomedical Devices and Systems for Smart Environment* (Lecture Notes in Electrical Engineering), A. Lay-Ekuakille and S. C. Mukhopadhyay, Eds. New York, NY, USA: Springer-Verlag, 2010, pp. 237–257.
- [5] K.-P. Hoffmann and R. Roman, "Flexible dry surface-electrodes for ECG long-term monitoring," in *Proc. 29th Annu. Int. Conf. IEEE EMBS*, Jun. 2007, pp. 5740–5743.
- [6] A. S. Berson and H. V. Pipberger, "Skin-electrode impedance problems in electrocardiography," *Amer. Heart J.*, vol. 76, no. 4, pp. 514–525, Oct. 1968.
- [7] P. Augustyniak, "Separating cardiac and muscular ECG components using adaptive modelling in time-frequency domain," in *Proc. WACBE World Congr. Bioeng.*, 2007, pp. 1–4.
- [8] P. S. Hamilton and M. G. Curley, "Adaptive removal of motion artifact," in *Proc. IEEE 19th Annu. Conf. EMBS*, vol. 1, Nov. 1997, pp. 297–299.
- [9] B. Ko, T. Lee, C. Choi, Y. Kim, G. Park, K. Kang, et al., "Motion artifact reduction in electrocardiogram using adaptive filtering based on half cell potential monitoring," in *Proc. Annu. Int. Conf. IEEE EMBC*, Apr. 2012, pp. 1590–1593.
- [10] B. Tajji, S. Shirmohammadi, V. Groza, and M. Bolic, "An ECG monitoring system using conductive fabric," in *Proc. IEEE Int. Symp. MeMeA*, May 2013, pp. 309–314.
- [11] E. Nemati, M. Deen, and T. Mondal, "A wireless wearable ECG sensor for long-term applications," *IEEE Commun. Mag.*, vol. 50, no. 1, pp. 36–43, Jan. 2012.
- [12] G. Lamarque, P. Ravier, and C. D. Viou, "A new concept of virtual patient for real-time ECG analyzers," *IEEE Trans. Instrum. Meas.*, vol. 60, no. 3, pp. 939–946, Mar. 2011.
- [13] J. Webster, *Medical Instrumentation: Application and Design*, 4th ed. New York, NY, USA: Wiley, 2009, pp. 196–198.
- [14] M. A. Mestrovic, R. J. Helmer, L. Kyrtatzis, and D. Kumar, "Preliminary study of dry knitted fabric electrodes for physiological monitoring," in *Proc. 3rd ISSNIP*, Dec. 2007, pp. 601–606.
- [15] I. Lee, S. Shin, Y. Jang, Y. Song, J.-W. Jeong, and S. Kim, "Comparison of conductive fabric sensor and Ag-AgCl sensor under motion artifacts," in *Proc. IEEE 30th Annu. Int. Conf. EMBS*, Aug. 2008, pp. 1300–1303.
- [16] T. Pola and J. Vanhala, "Textile electrodes in ECG measurement," in *Proc. 3rd Int. Conf. ISSNIP*, Dec. 2007, pp. 635–639.
- [17] M. Ishijima, "Cardiopulmonary monitoring by textile electrodes without subject awareness of being monitored," *Med. Biol. Eng. Comput.*, vol. 35, no. 6, pp. 685–690, Nov. 1997.
- [18] S. Ahmad, S. Chen, K. Soueidan, I. Batkin, M. Bolic, H. Dajani, et al., "Electrocardiogram-assisted blood pressure estimation," *IEEE Trans. Biomed. Eng.*, vol. 59, no. 3, pp. 608–618, Mar. 2012.
- [19] P. Csordás, A. Mersich, and A. Jobbágy, "Advanced indirect method for measuring blood pressure," *Electr. Eng. Comput. Sci.*, vol. 53, nos. 3–4, pp. 115–121, 2009.
- [20] S. Ahmad, M. Bolic, H. Dajani, V. Groza, I. Batkin, and S. Rajan, "Measurement of heart rate variability using an oscillometric blood pressure monitor," *IEEE Trans. Instrum. Meas.*, vol. 59, no. 10, pp. 2575–2590, Oct. 2010.
- [21] J. Jilek and T. Fukushima, "Oscillometric blood pressure measurement: The methodology some observations and suggestions," *Biomed. Instrum. Technol.*, vol. 39, no. 3, pp. 237–241, May/June. 2005.
- [22] S. Chen, M. Bolic, V. Groza, H. Dajani, I. Batkin, and S. Rajan, "Extraction of breathing signal and suppression of its effects in oscillometric blood pressure measurement," *IEEE Trans. Instrum. Meas.*, vol. 60, no. 5, pp. 1741–1750, May 2011.
- [23] D. Swanson and J. Webster, *A Model for Skin-Electrode Impedance*. New York, NY, USA: Academic, 1974, pp. 117–128.
- [24] B. Eegins, "Skin contact electrodes for medical applications," *Analyst*, vol. 188, no. 4, pp. 439–442, Apr. 1993.
- [25] S. Grimnes and O. Martinsen, *Bioimpedance and Bioelectricity Basics*, 2nd ed. Amsterdam, The Netherlands: Academic, 2008.
- [26] L. Geddes and M. Valentinuzzi, "Temporal changes in electrode impedance while recording the electrocardiogram with 'Dry' electrodes," *Annu. Biomed. Eng.*, vol. 1, no. 3, pp. 356–367, 1973.
- [27] M. Neuman, "Biopotential Electrodes," in *Medical Instrumentation Application and Design*, 3rd ed. New York, NY, USA: Wiley, 1988, pp. 183–232.
- [28] C. Assambo, A. Baba, R. Dozio, and M. Burke, "Determination of the parameters of the skin-electrode impedance model for ECG measurement," in *Proc. 6th WSEAS Int. Conf. Electron., Hardw., Wireless Opt. Commun.*, 2007, pp. 90–95.
- [29] S. Grimnes, "Impedance measurement of individual skin surface electrodes," *Med. Biol. Eng. Comput.*, vol. 21, no. 6, pp. 750–755, Nov. 1983.
- [30] J. J. Almasi and O. H. Schmitt, "System and random variations of ECG electrode system impedance," *Ann. New York Acad. Sci.*, vol. 170, no. 2, pp. 509–519, Jul. 1970.
- [31] K. Tomczyk, "Procedure for correction of the ECG signal error introduced by skin-electrode interface," *Metrol. Meas. Syst.*, vol. 18, no. 3, pp. 461–470, Oct. 2011.
- [32] B. Cornish, B. Thomas, and L. C. Ward, "Effect of temperature and sweating on bioimpedance measurements," *Appl. Radiat. Isot.*, vol. 49, no. 5, pp. 475–476, May/June. 1998.
- [33] R. Shepov'nikov, A. Nemirko, A. Kalinichenko, and V. Abramchenko, "Investigation of time, amplitude, and frequency parameters of a direct fetal ECG signal during labor and delivery," *Pattern Recognit. Image Anal.*, vol. 16, no. 1, pp. 74–76, Jan. 2006.
- [34] E. Khan, "Clinical skills: The physiological basis and interpretation of the ECG," *British J. Nursing*, vol. 13, no. 8, pp. 440–446, May 2004.
- [35] I. Rokos, W. French, A. Mattu, G. Nichol, M. Farkouh, J. Reiffel, et al., "Appropriate cardiac cath lab activation: Optimizing electrocardiogram interpretation and clinical decision-making for acute ST-elevation myocardial infarction," *Amer. Heart J.*, vol. 160, no. 6, pp. 995–1003, Dec. 2010.
- [36] P. Annila, V. Jantti, L. Lindgren, and A. Y. Hankala, "Changes in the T-wave amplitude of ECG during isoflurane anaesthesia," *Acta Anaesthesiol. Scand.*, vol. 37, no. 6, pp. 611–615, Aug. 1993.
- [37] A. Albulbul and A. Chan, "Electrode-skin impedance changes due to an externally applied force," in *Proc. IEEE MeMeA*, May 2012, pp. 1–4.
- [38] D. Halliday and R. Resnick, *Fundamentals of Physics*, 2nd ed. New York, NY, USA: Wiley, 1970, pp. 452–453.



**Bahareh Tajji** received the B.Sc. degree in electrical engineering from Isfahan University of Technology, Isfahan, Iran, and the MA.Sc. degree in biomedical engineering from the University of Ottawa, Ottawa, ON, Canada, in 2013. Her current research interests include the ECG measurement and medical devices.

She was a Hardware Designer in the biomedical and also automation and monitoring industries for many years. She is now a PhD student at the University of Ottawa.



**Shervin Shirmohammadi** (SM'04) received the Ph.D. degree in electrical engineering from the University of Ottawa, Ottawa, ON, Canada.

He is currently a Full Professor with the School of Electrical Engineering and Computer Science, University of Ottawa, Ottawa, ON, Canada. He is Co-Director of both the Distributed and Collaborative Virtual Environment Research Laboratory, and Multimedia Communications Research Laboratory, conducting research in multimedia systems and networking, specifically in gaming systems and virtual environments, video systems, and multimedia-assisted biomedical engineering. The results of his research have led to more than 200 publications, over a dozen patents and technology transfers to the private sector, and a number of awards and prizes. He is Associate Editor-in-Chief of IEEE TRANSACTIONS ON INSTRUMENTATION AND MEASUREMENT and Associate Editor of ACM Transactions on Multimedia Computing, Communications, and Applications, and was Associate Editor of Springer's Journal of Multimedia Tools and Applications from 2004 to 2012.

Dr. Shirmohammadi is a University of Ottawa Gold Medalist, a licensed Professional Engineer in Ontario and a Lifetime Professional Member of the ACM.



**Voicu Groza** (M'97–SM'02–F'11) received the Dipl. Eng. degree in computer engineering and Dr.Eng. degree in electrical engineering from the Polytechnic Institute of Timisoara, Timisoara, Romania, in 1972 and 1985, respectively.

He joined the university of Ottawa, Ottawa, ON, Canada, in 1996, where he is currently a Professor with the School of Electrical Engineering and Computer Science. Prior to this, he was a Professor with the Polytechnic University of Timisoara. He has published more than 250 technical papers, authored

two books, and received two patents. His current research interests include biomedical instrumentation and measurements, high speed data acquisition systems and reconfigurable computers.

Dr. Groza has held leadership roles on the organization and technical program committees of numerous international conferences, such as IEEE International Symposium on Medical Measurement and Applications (MeMeA) from 2008 to 2013, IEEE International Conference on Instrumentation and Measurement (I2MTC), and IEEE Canadian Conference on Electrical and Computer Engineering (CCECE). He has been volunteering in the frame of the IEEE Instrumentation and Measurement Society both at the Ottawa chapter and at the worldwide society administration level. Currently he serves as chair of the IEEE Working Group on Standard for Objective Measurement of Systemic Arterial Blood Pressure in Humans. He is a Fellow of the Engineering Institute of Canada.



**Izmail Batkin** received the M.S. and Ph.D. degrees in theoretical physics from the Voronezh State University, Voronezh, Russia, and the Dr.Sci. degree in nuclear physics from the Leningrad State University, Saint Petersburg, Russia, in 1965, 1969, and 1982, respectively.

He is currently a Research Consultant with the School of Electrical Engineering and Computer Science, University of Ottawa, Ottawa, ON, Canada, and Ottawa General Hospital, Ottawa. He was also the Chief Scientist with the Advanced Bioelectric and Biopeak Corporations, Ottawa. He was a Full Professor with the Voronezh State University and an Adjunct Professor with Carleton University, Ottawa. He has been involved in the successful development and testing of a new generation of wearable physiological electrodes and monitors for the home and clinical environments. He has obtained one patent and published more than 150 refereed papers, many in top notch journals such as *Physical Review*, the *Journal of Physics*, and the *Soviet Journal of Nuclear Physics*. His current research interests include the noninvasive monitoring of physiological parameters, theoretical and nuclear physics, and medical physics.