[[1]](#footnote-1)

A Non-Contact Wearable Sensing System Towards Long Term ECG Monitoring

Leon Chen, Ognian Marino, Member, IEEE, Chih-Hung Chen, *Senior* Member, IEEE,   
Tapas Mondal, and M. Jamal Deen, *Fellow*, IEEE

*Abstract*—Electrocardiography (ECG) is a commonly used vital sign health monitoring method used extensively in the modern healthcare systems. Different designs of ECG system have been developed as alternatives to the common twelve-lead ECG systems. The approaches in these designs target portability and user convenience. Other issues such as signal integrity for acceptable quality ECG traces, power consumption and new electrodes to replace standard hydrogel ones for wearable systems, were addressed. In this work, an ECG sensing system is developed towards long-term monitoring using capacitive electrodes that can be easily incorporated into clothing. The two electrodes are not in direct contact to the skin, but have an interface clothing material between the skin and electrodes. The electrodes are connected to the ECG system that performs signal acquisition and transmits wirelessly the acquired data to a computer for data storage and processing. Experiments are conducted on different types of clothing materials with various body movements introduced for observations. The results of the detected ECG signals are comparable to other ECG sensing system alternatives.

*Index Terms*—Capacitive coupling electrodes; ECG sensors; wearable ECG sensing system

# INTRODUCTION

E

lectrocardiography (ECG) has proven to be among the most useful diagnostic tests in clinical medicine. The ECG is now routinely used in the evaluation of patients with implanted defibrillators and pacemakers, as well as to detect myocardial injury, ischemia, and the presence of prior infarction. In addition to its usefulness in ischemic coronary disease, the electrocardiogram, in conjunction with ambulatory ECG monitoring, is of particular use in the diagnosis of disorders of the cardiac rhythm and the evaluation of syncope. Other common uses of the ECG include evaluation of metabolic disorders, direct and side effects of pharmacotherapy, and the evaluation of primary and secondary cardiomyopathic processes 0.

|  |
| --- |
|  |
| Fig. 1. Basic features in the waveform of an ECG signal. |

In the early days of ECG, the sensitivity of the signal acquisition was a significant challenge. In the late 1800's, attempts to measure the electric activity in frog's hearts became successful only when the hearts were exposed directly to the measuring equipment. The measuring conditions were quite difficult and invasive, so scientists wanted to be able to record the electric signals of the heart without having to enter inside the body. However, the electrical heart signals attenuate when travelling through body tissues, becoming quite weak on reaching the skin’s surface. This problem was solved later by Willem Einthoven [2]. He managed to improve the sensitivity of the ECG by using a string galvanometer, and this represented a great leap forward in electrocardiography. Einthoven’s improvements were very significant because the now familiar P, Q, R, S, and T waves were apparently defined (Fig. 1), while previously, the scientists had demonstrated only ventricular depolarization and repolarization, e.g., in Waller’s work [3].

Different types of ECG systems (e.g., [4]–[7]) have been introduced to improve the signal quality and to provide convenience to the patients. The conventional ECG method is often known as a wet method because it uses a hydrogel between the skin and the electrodes to increase the conductivity of the signal path. However, the wet method uses conductive gels that contain certain metallic materials that can irritate the skin of the patients. Some patients may even be allergic to metallic materials and other ingredients of the conductive hydrogel [8]–[12], which in turn may affect the quality of ECG signals and subsequent diagnosis. From these references, propylene glycol in the conducting gel is found underneath ECG electrodes, and there is possibility of allergic reactions due to nickel particles or to the acrylic adhesive present in the popular disposable ECG electrodes. Also, there is change of conductivity type from ion-based in living tissues to electron-based in metal electrodes, and the conductive hydrogel modifies the contact interface.

In this work, in order to prevent the above problems associated with using gel, an alternative using dry electrodes has been developed. This dry-electrode method does not rely on the conductive gel that may irritate patients with allergy. The capacitive dry electrode method [4], [5] does not require direct contact with the skin and is suitable for long-term monitoring. Despite the layer of interfacing material between the skin and electrodes, the capacitive coupling is still capable of detecting the ECG signals from the skin’s surface potentials.

The interface material for capacitive electrodes can be a thin layer of cloth material such as cotton that is tightly bonded to the skin for an optimal ECG acquisition. Cotton is a common fabric for clothes. In comparison with other materials, such as wool, silk or nylon, cotton has higher dielectric constant, which translates in better capacitive coupling when the materials are dried [13]. The capacitive electrodes are an alternative to smart textiles [14]–[16]. The smart textiles are dry conductive fabrics with embedded electrodes and electronics. Both the capacitive electrodes and smart textiles are of increasing interest in ECG-based integrated sensors. The development of dry electrodes allows the electrodes to be wearable with an appropriate biocompatible interface and concurrently have reasonable coupling to the electrical ECG signal at the skin' surface.

In this paper, a capacitive-coupled method for the ECG system, based on the concept for active electrodes described in [5], is proposed. The system is capable to obtain ECG traces in the presence of a layer of cloth between the skin and the electrodes. The proposed ECG system is designed for ambulatory applications, and it is based on capacitive electrodes. It is a portable and wearable system for both inpatients and outpatients, and can transmit ECG data wirelessly to a computer for data storage and processing [17].

This paper is organized as follows. In Section II, a description of the major blocks of the proposed ECG system with capacitive electrodes is given. In Section III, we present the test results using different interface materials and with different body movements. Also, a comparison between the proposed method and results from previous publications are discussed. Finally, the conclusions are presented in Section IV.

# proposed method

## Overview

The purpose of using the proposed dry ECG method is to obtain the ECG traces when a layer of cotton or other clothing material is present between the skin and the electrodes. Now, since there is no direct contact with the skin, then skin irritations or possible allergies of the patients can be prevented. The dry ECG method using the capacitive electrodes is implemented using the system setup as shown in Fig. 2. Two capacitive electrodes, one for LA (left arm) and another for RA (right arm), are placed on the human body as sensors with a cloth material between the skin and the electrodes. They are connected by wires to the portable ECG device. Our portable ECG device is small in size, with low power consumption, and transmits the ECG data wirelessly to a personal computer with monitoring and display software. This entire system is designed to allow the patient to carry around the ECG device throughout the day for longer term monitoring.

|  |
| --- |
|  |
| Fig. 2. Block diagram of the ECG sensing system with a photograph of a designed capacitive electrode on the top. |

The wireless ECG device connects several key processing components. LA and RA are the inputs of the electrometric buffers of the capacitive electrodes. The buffers repeat the potentials of the inputs and forward the ECG signals to the inputs of the differential amplifier of the DataLog. The differential amplifier amplifies the ECG signal and an analog-to-digital converter (ADC) converts the analog signal to digital format. The digital codes from ADC are then transmitted wirelessly to the computer system. The wireless link uses the IEEE 802.15.1 protocol. Bluetooth, the low power consuming wireless communication protocol, is used for our ECG system [14].

## Capacitive Electrodes

The capacitive electrodes sense the bio-potential of ECG through the capacitance between the electrode and skin surfaces [4], [5], as shown in Fig. 3. The use of these capacitive electrodes allows for having an interface material between the electrodes and the skin. The interface material is commonly a piece of cotton. In practice, the patient wears the electrodes over their clothing. The cotton material is expected to be a thin layer that is in tight contact to the skin for optimum electrostatic pickup of the ECG signals. Also, as mentioned before, the two electrodes are placed on the two arms (LA and RA).

Figure 4 presents the schematic of the designed capacitive electrode. The sensing area of the electrode is 6 cm², providing about 20pF capacitive coupling to the skin. A 3 V coin battery serves as power source for the operational amplifiers (Op-Amps). The power switches on when the electrode is connected with the cable of the portable ECG device. Thus, the battery charge starts draining only when the wire is plugged into the electrode. The coin battery has a capacity of 40 mAh, allowing for the electrode to last for up to 2 weeks using the micro-power AD8617 CMOS dual Op-Amp. Two-week operation of the electrode is determined to be sufficient time for long-term ECG monitoring.

|  |
| --- |
|  |
| Fig. 3. Electrode-body interface with capacitive coupling of ECG bio-potential. |

|  |
| --- |
|  |
| Fig. 4. Schematic diagram of the capacitive electrode. |

The operational amplifier (op-amp) AD8617 is selected for its low input bias current (< 5 pA) and low current consumption (< 0.1 mA). This op-amp also has low offset voltage, and low input voltage and current noises. The electrometric amplifier AD8617 is very suitable for portable medical devices. One of the op-amps is used for splitting the battery voltage symmetrically around the virtual ground (RET). The other op-amp is used for impedance buffering in the signal path, providing high impedance (50 GΩ) at the input and low impedance (few kΩ) at the output. Shielding and guarding are carefully considered in the layout in order to prevent external interferences being coupled to the extremely sensitive input of the electrode.

Resistors and a capacitor are added at the output of the sensor in order to comply with safety requirements for ECG, initial signal conditioning and lead-off detection. The safety requirements include very small or no current to keep the heart beating. Overall, the advantages of the proposed electrodes are *small size* and *low power consumption*, which provide both *portability* and use in longer term ECG monitoring.

## Portable ECG Device

The Biometrics DataLog [18] is selected for the portable ECG device in these experiments. The size of the device is 104 × 6 × 22 mm3, weight is 129 g. It is powered by two AA batteries for about 24 hours. The device provides real time data transfer from up to 24 programmable channels. In the experiments, only one channel is used. Automatic data backup on a Micro SD card is also available.

A photograph of the capacitive electrodes attached on RA on the cotton material and the portable ECG device is shown in Fig. 5. The electrode shown is a prototype. In future, its packaging and integration will be improved and it will be more securely attached to the arm to minimize position displacement from body motion. As shown earlier in Fig. 2, the size of an electrode is about 3 cm in diameter. The capacitive electrodes are connected to the Biometrics DataLog device with flexible wires for convenience to the user. The flexibility of the wires is essential in preventing the displacement of the electrodes in the case of body motion.

|  |
| --- |
|  |
| Fig. 5. Photograph of a capacitive electrode attached to RA on top of cotton material with portable ECG device (Biometrics DataLog) on the right. |

## Computer Monitoring System

In Fig. 6, the monitoring system with a software application in a laptop computer is shown. A Bluetooth wireless receiver is connected to the USB port of the computer to receive the data from the portable ECG device. The latter had acquired the ECG signals by the sensitive capacitive-coupled electrodes, as explained above. The data received from the portable ECG device are processed by the infinite impulse response (IIR) notch (60 Hz) and band-pass (1 Hz - 30 Hz) filters in the software application to minimize the noise in the ECG traces.

|  |
| --- |
|  |
| Fig. 6. Block diagram of the computer monitoring system for ECG sensing. |

After the filtering, the ECG readings are displayed in real time. The software can also store the ECG readings in CSV files. Each CSV file contains a time frame of 10-second ECG acquisition. Consecutive acquisitions are stored in files with consecutively incrementing filenames. Thus, the stored files contain the whole record of ECG acquisition, until the storing function is stopped by the user. Therefore, the duration of the record is practically unlimited, or until the large free disk space (gigabytes) of the laptop is consumed.

# Performance Evaluation

The next sub-section presents the ECG experiments conducted using the proposed method with the capacitive electrodes described in the previous sections. LA and RA electrodes were placed on the forearm and the arm. The experiment was conducted in a standard room environment on three different healthy subjects, observing similar results, as shown in Fig. 7 and Fig. 8. Formal consent was taken as per the McMaster Research Ethics Board (REB) approved informed consent forms. Simple questionnaires were asked to exclude any known cardiac or significant health problems. Different interface materials and body movements are considered in the ECG experiments. The comparison with existing results is then given in a subsequent subsection.

## Experiments

The ECG acquisition is first tested by placing the sensor on a dry cloth over the skin (sensor not in direct contact with the skin), and without using any gel or liquid to enhance signal coupling to the electrodes. Reasonably good results were obtained. However, as shown in Fig. 7(a), moisturizing the skin with water results in enhanced signal coupling to the electrodes, resulting in better quality ECG traces. Now, the QRS complex and the peaks of the P and T waves are easily distinguished.

|  |
| --- |
|  |
| Fig. 7. ECG acquisitions in the time domain without body movement, with electrodes on (a) moisturized skin and (b) moisturized cotton face towel. |

The following experiments consider different dry and moisturized (by water) interface materials. A moisturized thin layer of cotton, e.g., cotton face towel in Fig. 7(b), preserves the quality of the ECG signal, while lightly moisturized cotton cloth hand wipe also sustains the quality of the ECG signal. Overall, when the interface materials are dry, the electrodes are more unlikely to pick up the ECG signals, while once the interface materials are moisturized by water, then the capacitive electrodes are able to acquire good ECG waveforms. Thus, it becomes clear from Fig. 7 that in order to acquire good ECG traces, the electrostatic coupling between skin and electrode has to be increased, that is, the interface material must be of high dielectric permittivity, e.g., moisturized by water, the latter having high dielectric constant in the order of 80.

|  |
| --- |
|  |
| Fig. 8. ECG acquisitions in time domain with body movements. Electrodes on moisturized cotton towel by (a) slow abduction of left shoulder, (b) slow walking, (c) body rotation, and (d) fast abduction of left shoulder. |

In Table 1, a summary of the results of the electrodes integrated on skin and different types of cotton interface materials is provided. A *good result* denotes waveforms with *QRS complex with a typical period of 1 second*. The amplitude of the waveform varies depending on how close the electrodes are placed to the heart. By placing the electrodes closer to the heart, a stronger signal can be detected, which results in a higher amplitude of the waveform. On the contrary, a *poor result refers to either no signal* (*flat*) or *waveform with significant fluctuations due to noise*. In between the interval of the QRS complex as shown in Fig. 1, the wave should be flat with small pulses that represent the P and T waves right before and after the QRS complex, respectively. In this case, any pulses detected other than the QRS complex, and P and T waves are considered as noise.

|  |
| --- |
| Table 1. Effects of interface materials and moisture on the quality of the ECG acquisition |
| |  |  |  |  | | --- | --- | --- | --- | | Interface Materials and Moisturizing | | Electrode Location | | | **Forearms** | **Arms** | | Electrode Directly on Skin Surface | **Dry** | Good | Good | | **moisturized** | Good | Good | | Thin Cotton Napkin | **Dry** | Bad | Bad | | **moisturized** | Good | Good | | Cotton Face Towel | **Dry** | Bad | Bad | | **Moisturized** | Good | Good | | Cotton Hand Wipe | **Already moisturized** | Good | Good | |

A second set of the experiments was performed to evaluate body movement artifacts in ECG acquisitions. The body movements in this experiment include shoulder abduction, walking and body rotation. The ECG motion artifact experiments used moisturized interface materials - cotton face towel and cotton hand wipe. The ECG results by body motion are shown in Fig. 8. Now, in the ECG traces, it is observed that the body motion superimposes “noise” on the ECG signals. Left shoulder abduction slow movement is conducted in Fig. 8(a). The QRS complex is maintained, however, the P and T waves are distorted by the induced movement noise. Similar P and T waveform distortion applies for slow walking and body rotation in Fig. 8(b) and Fig. 8(c), respectively.

The motion noise is due to the displacement of the electrodes, when not secured properly on the interface material. Fig. 8(d) shows the result of the fast abduction of the left shoulder movement. In comparison with Fig. 8(a), it is observed that more movement noise with increase in amplitude of the ECG traces in Fig. 8(d) are due to faster or larger shoulder abduction movement.

## Comparison with Existing Results

Many types of ECG systems have been developed with different technologies implemented such as the front-end sensors, integration and data acquisition techniques. A summary of recent ECG sensor systems is shown in Table 2, comparing 10 different methods of implementing ECG systems that are representative for the state-of-the-art, including the proposed method. From the literature, some electrodes are developed for different performance goals and constraints. Therefore, there are always trade-offs between different designs of the ECG systems.

Compared to the technologies reported in these publications, the proposed ECG system presents a good balance among the following parameters - signal quality, size and power consumption. Our proposed design might be slightly larger than [6], [14], [19] in the size of the electrodes, but it provides better flexibility in the integration on clothes. Additionally, our proposed method provides good signal filtering as in [5]-[15], [17], [20]–[23], but simultaneously has lower power consumption and smaller size in the acquisition module for better portability. In comparison with [6], [7], [14]–[16], [22], [23], our proposed method has better signal integrity mainly due to the better circuit design of the capacitive electrode, and by optimization of the filters in the system.

Different wireless technologies have been adopted in different publications [6], [14], [17], [19]–[21]. For example, in [6], [14], [19], [20], it was suggested to build mobile ECG monitoring applications using Android cell phones. However, there is significant power consumption in the cell phones during RF communications, which is an issue towards long term ECG monitoring. For long-term monitoring, an application specific transmitting device is preferred so that the battery can last for a minimum of one day of continuous operation of the portable unit.

Currently we tested the ECG acquisition as routinely done in an outpatient clinic. Using our system, we aim to identify Tachy and Brady arrhythmias including atrial fibrillation, ventricular and supraventricular tachycardia (narrow complex and wide complex), sick sinus syndrome, episodic syncope which are particularly intermittent and may not be evident on regular short-term ECG. The low power consumption allowed us using the porotype for prolonged periods, confirming the potential of the capacitive electrodes and ECG method for long term monitoring. Also, apart from long-term ECG we can have regular short-term ambulatory multi-lead ECGs as well.

# Conclusion

An ECG sensing system has been developed in a way to improve the signal detection, portability and power consumption, addressing the trade-offs in ECG system designs among size, power consumption, signal quality and integration. Recent developments have been leaning towards the idea of contactless ECG sensing for long-term monitoring, without direct contact to the skin to minimize the skin irritation during prolonged patient monitoring. The proposed method places the electrodes on biocompatible interface materials such as with tightly contacting cotton. For better portability and convenience, the size of the electrodes must be small. In the proposed system, there is no need for a ground electrode in capacitive method. So two leads are enough, as opposed to minimum three electrodes in the hydrogel method.

The proposed ECG sensing systems must have low power consumption for long-term monitoring. The electrodes are connected to a portable ECG device that is capable of transmitting signals wirelessly through Bluetooth to a personal computer in real time. From the results of our experiments, the proposed ECG sensing system properly acquires QRS complexes with P and T waves when the capacitive sensors are placed on various cotton materials moisturized with water. In our proposed wearable system, we have convincing proof-of-concept of acquiring good quality ECG signals that has strong potential to diagnose abnormal tracings. Additionally, good results are observed with slow body movements. However, fast or larger movements cause motion artifacts in the gathered ECG waveforms.

Several directions for future improvements of the wearable ECG sensing systems can be considered. Among these, the most critical is that the electrodes must be firmly integrated onto the clothes to reduce the noise from sensor displacement. Furthermore, the electrostatic coupling of the electrodes to the skin potential can be improved by employing a dedicated film or membrane materials with increased dielectric permittivity.

Acknowledgment

We gratefully acknowledge the Canada Research Chair program, NSERC of Canada as well as the Canada Foundation for Innovation and the Ministry of Research and Innovation for supporting our work. We also acknowledge our colleagues at Celestica for help in developing another version of the ECG electrode system as a possible commercial product.

|  |  |  |  |  |  |
| --- | --- | --- | --- | --- | --- |
| Table 2. Comparison between the proposed methods and recent alternatives for ECG sensing systems. | | | | | |
| Ref. | **Technologies** | **Implementation** | **Advantages** | **Limitations** | **Wireless Technology** |
| [[20]] | Capacitive sensing | Integrated on chest belt | Portable and integrated to the Android phones | Power consumption issues | Zigbee |
| [[19]] | Flexible PDMS dry electrode sensing | Integrated on wrist band | Small, low power consumption | Limited by the ECG recording systems | N/A |
| [[14]] | Capacitive sensing | Integrated on belt | Flexible electrodes integrated into garment, convenient integration on the body | Poor filtering (relatively small upper-corner frequency, No notch Filter), Poor signal quality, Large Ref electrode - 250 cm2 | Bluetooth |
| [[5]] | Capacitive sensing | Integrated on chest vest | Picks up good signals through one or two layers of cotton | Large size of electrodes and devices gives poor portability | N/A |
| [[6]] | Capacitive sensing | Integrated on thin cloth | Small, low power consumption electrodes | Limited by the phone and monitoring network system | Bluetooth |
| [[21]] | Multiwall carbon nanotubes/cotton-based electrode sensing | Integrated on cloth | Low cost, good conductivity on cotton materials | Limited by the circuits in electrodes and other part of the ECG system | N/A |
| [[16]] | Wet textile, Silver/Silver chloride electrodes sensing | Integrated on belt | Good conductivity | Large in size, not portable | N/A |
| [[22]] | Dry electrode sensing | Integrated on cloth | Low power consumption, good noise filtering | Power consumption issues | Bluetooth/Zigbee |
| [23] | Capacitive sensing using conductive foam | Embedded in chest belt | Flexible electrodes integrated with chest belt | Three electrode, Very high resistor 5G used, Used for 24 hours, depends on clothing material and thickness, contact pressure and humidity | Bluetooth |
| [24] | Dry electrode sensing | Nanowoven fabrics with Ag/AgCl conductive inks | Screen printed dry electrode woven into fabric – comfortable, easy to use and suitable for wearable systems | Only sensing electrode demonstrated. | N/A |
| [[7]] | Dry-electrode sensing, Active shielding | Integrated on cloth | Small electrodes (down to 8 mm diameter), conductive rubber electrodes | Relatively high power consumption, Low signal quality, Large sensor size | BluesenseAD module, high power consumption (33 mA × 4.8 V), 37 mm × 21 mm |
| This work | **Capacitive sensing using flexible PCB** | **Integrated on a stretchable cloth** | **Very low power consumption compared to other systems, relatively small-size, ultra-thin flexible electrodes combined with standard 2-layer PCBs, easy to use** | **Semi-rigid electrodes using thin, flexible PCB** | **Bluetooth** |

References

1. R. C. Schlant, R. J. Adolph, J. P. DiMarco, L. S. Dreifus, M. I. Dunn, C. Fisch, A. Jr. Garson, L. J. Haywood, H. J. Levine, and J. A. Murray, “Guidelines for electrocardiography. A report of the American College of Cardiology/American Heart Association Task Force on Assessment of Diagnostic and Therapeutic Cardiovascular Procedures (Committee on Electrocardiography),” *Circulation*, vol. 85(3), pp. 1221-1228, 1992.
2. W. Einthoven, “Un nouveau galvonometre,” *Nat. Arch Neerl Sci Exactes*, vol. 6, pp. 623-33, 1901.
3. W. Einthoven, “Le telecardiogramme,” *Arch Int Physiol.*, vol. 4, pp. 132-63, 1906.
4. E. Nemati, M. J. Deen, and T. Mondal, “A wireless wearable ECG sensor for long-term applications,” *IEEE Communications Magazine*, vol. 50, pp. 36-43, 2012.
5. A. Arcelus, M. Sardar, and A. Mihailidis, “Design of a capacitive ECG sensor for unobtrusive heart rate measurements,” *2013 IEEE International Instrumentation and Measurement Technology Conference* (*I2MTC*), 2013, pp. 407-410.
6. B. S. Lin, W. Chou, H. Y. Wang, Y. J. Huang, and J. S. Pan, “Development of novel non-contact electrodes for mobile electrocardiogram monitoring system,” *IEEE Translational Engineering in Health and Medicine*, vol. 1, pp. 1-8, 2013.
7. G. Gargiulo, P. Bifulco, M. Cesarelli, M. Ruffo, M. Romano, R. A. Calvo, C. Jin, and A. van Schaik, “An ultra-high input impedance ECG amplifier for long-term monitoring of athletes,” *Med. Dev. Evid*. *Res.*, vol. 3, pp. 1–9, 2010.
8. L. Stingeni, E. Cerulli, A. Spalletti, A. Mazzoli, L. Rigano, L. Bianchi, and K. Hansel, “The role of acrylic acid impurity as a sensitizing component in electrocardiogram electrodes,” *Contact Dermatitis*, vol. 73(1), pp. 44-8, 2015.
9. E. Ozkaya and P. Kavlak Bozkurt, “Allergic contact dermatitis caused by self-adhesive electrocardiography electrodes: a rare case with concomitant roles of nickel and acrylates,” *Contact Dermatitis*, vol. 70(2), pp. 121-123, 2014.
10. N. Sakamoto, N. Sato, M. Goto, M. Kobayashi, N. Takehara, T. Takeuchi, A. K. Talib, E. Sugiyama, A. Minoshima, Y. Tanabe, K. Akasaka, J. Kawabe, Y. Kawamura, A. Doi, and N. Hasebe, “Three cases of corticosteroid therapy triggering ventricular fibrillation in J-wave syndromes,” *Heart Vessels*, vol. 29(6), pp. 867-872, 2014.
11. A. C. Deswysen, E. Zimerson, A. Goossens, M. Bruze, M. Baeck, “Allergic contact dermatitis caused by self-adhesive electrocardiography electrodes in an infant,” *Contact Dermatitis*, vol. 69(6), pp. 379-381, 2013.
12. B. Nunez-Acevedo, M. T. Gonzalez-Fernandez, M. M. Juangorena, and C. Vidal, “Multifunctional acrylates as possible sensitizers in electrocardiogram electrode allergy,” *Ann Allergy Asthma Immunol*, vol. 111(1), pp. 77-78, 2013.
13. M. S. Naidu and V. Kamaraju, *High Voltage Engineering*, 4th ed.; Publisher: New Delhi, India, pp. 107-108, 2009.
14. S. Fuhrhop, S. Lamparth, and S. Heuer, “A textile integrated long-term ECG monitor with capacitively coupled electrodes,” *IEEE Biomedical Circuits & Systems Conference*, 2009,pp.21-24.
15. C. L. Lam, N. N. Z. M. Rajdi, and D. H. B. Wicaksono, “MWCNT/ Cotton-based flexible electrode for electrocardiography,” *Sensors IEEE*, pp. 1-4, 2013.
16. G. Andreoni, A. Fanelli, I. Witkowska, P. Perego, M. Fusca, M. Mazzola, and M. G. Signorini, “Sensor validation for wearable monitoring system in ambulatory monitoring: Application to textile electrodes,” *2013 7th International Conference on* *Pervasive Computing Technologies for Healthcare and Workshops* (*PervasiveHealth*),2013, pp. 169-175.
17. M. J. Deen, “Information and communications technologies for elderly ubiquitous healthcare in a smart home,” *Personal and Ubiquitous Computing*, vol. 19(3-4), pp. 573-599, 2015.
18. Biometrics Ltd. Biometrics DataLog. Available online: http://www.biometricsltd.com/datalog.htm (accessed on 26 January 2014)
19. C. Y. Chen, C. L. Chang, C. W. Chang, S. C. Lai, T. F. Chien, H. Y. Huang, J. C. Chiou, and C. H. Luo, “A low-power bio-potential acquisition system with flexible PDMS dry electrodes for portable ubiquitous healthcare applications,” Sensors, vol. 13, pp. 3077-3091, 2013.
20. P. C. Hii and W. Y. Chung, “A comprehensive ubiquitous healthcare solution on an Android™ mobile device,” Sensors, vol. 11, pp. 6799-6815, 2011.
21. B. Jin, T. H. Thu, E. H. Baek, S. H. Sakong, J. Xiao, T. Mondal, and M. J. Deen, “Walking-age analyzer for healthcare applications,” *IEEE Journal of Biomedical and Health Informatics*, vol. 18(3), pp. 1034- 1042, 2014.
22. M. Magno, L. Benini, C. Spagnol, and E. Popovici, “Wearable low power dry surface wireless sensor node for healthcare monitoring application,” *2013 IEEE 9th International Conference on Wireless and Mobile Computing, Networking and Communications (WiMob)*, 2013, pp. 189-195.
23. J.S. Lee, J. Heo, W. K. Lee, Y. G. Lim, Y. H. Kim and K. S. Park, “Flexible Capacitive Electrodes for Minimizing Motion Artifacts in Ambulatory Electrocardiograms,” *Sensors*, vol. 14, pp. 14732-14743, 2014.
24. M.A. Yokus and J.S. Jur, “Fabric-Based Wearable Dry Electrodes for Body Surface Biopotential Recording,” *IEEE Trans Biomed Eng*, vol. 63(2), pp. 423-430, February 2016.

1. Manuscript received June 29, 2016. This work was supported in part by the National Sciences and Engineering Research Council of Canada (NSERC).

   L. Chen, O. Marinov, C.-H. Chen, and J. M. Deen are with the Department of Electrical and Computer Engineering, McMaster University, Hamilton, Ontario, L8S 4K1, Canada (phone: +1-905-525-9140 ext. 27137; fax: +1-905-523-4407; e-mail: chenl79@mcmaster.ca, omarinov@yahoo.com, [chench@mcmaster.ca](mailto:chench@mcmaster.ca), and jamal@mcmaster.ca).

   T. Mondal is withthe Department of Pediatrics, McMaster University, Hamilton, Ontario, L8S 4K1, Canada (e-mail: mondalt@mcmaster.ca). [↑](#footnote-ref-1)