

A Wireless Wearable ECG Sensor for Long-Term Applications

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ABSTRACT

Ubiquitous vital signs sensing using wireless medical sensors are promising alternatives to conventional, in-hospital healthcare systems. In this work, a wearable ECG sensor is proposed. This sensor system combined an appropriate wireless protocol for data communication with capacitive ECG signal sensing and processing. The ANT protocol was used as a low-data-rate wireless module to reduce the power consumption and size of the sensor. Furthermore, capacitive ECG sensing is a simple technique that avoids direct contact with the skin and provides maximum convenience to the user. In our work, small capacitive electrodes were integrated into a cotton T-shirt together with a signal processing and transmitting board on a two-layer standard printed circuit board design technology. The entire system has small size, is thin, and has low power consumption compared to recent ECG monitoring systems. In addition, appropriate signal conditioning and processing were implemented to remove motion artifacts. The acquired ECG signals are comparable to ones obtained using conventional glued-on electrodes, and are easily read and interpreted by a cardiologist.

INTRODUCTION

Electrocardiography (ECG) is one of the most widely used vital sign sensing and health monitoring methods and provides useful diagnostic information about the cardiovascular system. It can be used as powerful indicator of some specific physiological and pathological conditions of humans. With the increase in coronary diseases during the past few decades, continuous monitoring of the ECG signal of high-risk patients can play an important role in immediately detecting pathological signatures and arrhythmias. Using this concept, any deviation of the health status of an individual from their norm can be detected and sent to a health center for early and further analysis and preventative actions. Studies have proven that this type of ECG monitoring, if it does not interfere with daily activities, can improve the diagnosis and therapy of some of the most prevalent cardiac diseases [1–3].

Currently, ECG can be performed using many

methods. The conventional clinical ECG system employs 12 or 15 Ag-AgCl electrodes (wet ECG), which are affixed to specific parts of the chest, arms, or hands and legs. This often requires cleaning the attachment site and, if necessary, shaving the hair off some parts of the body. In this way, the electrodes, which consist of gel in the middle of a pad, can be used to provide a conducting medium for charge transfer between the electrodes and the body. To keep the electrodes in place, extra adhesive tape is also applied. Although this type of ECG provides good signal quality, it is inconvenient and may cause skin irritation, allergic reactions, and inflammation due to toxicological issues of the gels in long-term treatments [1–4]. In addition, the quality of the signal will be reduced as the gel dehydrates during prolonged use, and replacement of the gel is typically infeasible. Furthermore, since it is difficult to keep the adhesives entirely separate from each other over the long term, cross-coupling between neighboring electrode sites can occur through leakage current [5]. Therefore, the wet ECG electrode system may be unsuitable for long-term ECG monitoring.

One of the main goals of this project is to provide maximum convenience to the user or patient during ECG measurements, especially for prolonged use. Therefore, special consideration is taken regarding two interfaces: the patient-sensor and sensor-cardiologist interfaces. A convenient interface between the body and the sensor can be realized using a non-contact ECG sensing method. An efficient wireless protocol plus well designed software is also needed to assure convenience in the sensor-cardiologist interface. Alternatives for wet ECG that potentially provide comfortable patient-sensor interface are dry-electrode or capacitively-coupled ECG (CC-ECG) methods. In the dry-electrode ECG, a metal plate is placed on the skin instead of wet electrodes, so the problems of using gel are eliminated. However, it still has direct contact with the body, so it may cause skin irritation and allergies after prolonged use.

On the other hand, in the capacitive method used in this work, there is no direct contact with the body. Instead, a thin layer of insulator is placed between the body and the metal electrode. In one method, the electrodes can be unnoticeably applied to a cloth that can be worn

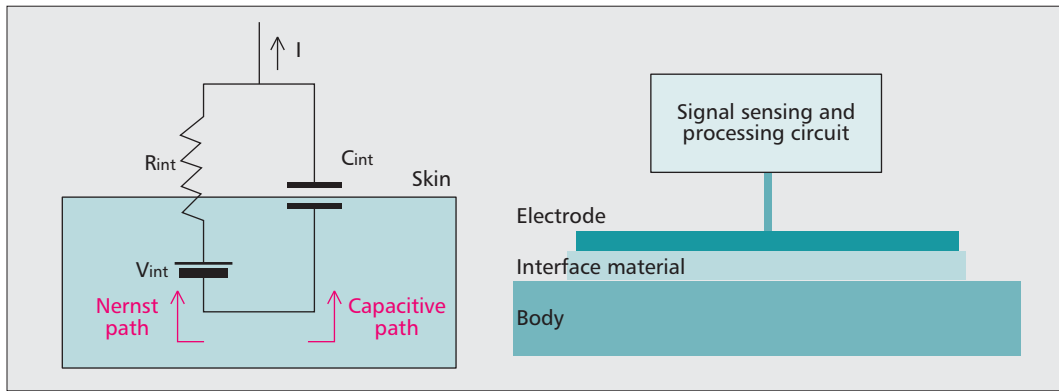


Figure 1. Body equivalent circuit model (left) and electrode-body interface (right).

by the patient to provide ubiquitous (U) health-care ECG sensing. Following this idea, a wireless, portable ECG sensor was designed in which the following key considerations were targeted: convenience, portability, and small size. Convenience and portability were achieved by using the capacitive ECG sensing approach, utilizing a small wireless module, and also by employing inexpensive standard off-the-shelf (OTS) integrated circuits (ICs) in small packages. In addition, we used a standard two-layer printed circuit board (PCB) design for the signal processing board, which resulted in a thin, small-area sensor. To obtain long battery lifetime, all components of the sensor were chosen from low-power ones, and some battery-saving techniques such as idle mode signal sampling were employed. In particular, for data communication, which is always one of the most power-hungry parts in similar sensors, an extremely low-power wireless communication protocol was used. These special considerations resulted in a low-power, small form factor, accurate and easy-to-use wireless ECG sensing system.

In this article, we describe the capacitive sensing method, the system designed including the signal processing and wireless data transmission circuits, and the personal computer (PC) interface and software for further signal processing. We also describe some sample measurement results under different situations, and finally, we present the main findings of our study.

CAPACITIVELY COUPLED ELECTROCARDIOGRAPHY

Among the various kinds of ECG methods, capacitively coupled ECG (CC-ECG) is a very viable method that provides non-contact ECG, which is required for long-term convenience in the user-sensor interface. In this method, a thin layer of insulator is placed between the human body and a metal-plate sensing electrode. The electrode, together with the skin and insulator, form a capacitance that conveys the signal from the body to the sensor. Figure 1 shows the body-electrode interface and its equivalent circuit model [1]. As shown, there are two different paths between the ECG signal source and the electrodes. The path, which consists of a resistor and an interface potential, is called the Nernst

path and is used in resistive ECG (wet method) in which the electrolyte between the skin and the electrode provides the charge-transfer environment. In the capacitive approach, charge transfer occurs through the other path, which is a capacitor that couples the signal onto the electrode.

This CC method has several advantages over conventional ECG. In addition to providing convenience for the patient, contactless ECG sensors can be installed in domestic environments and be configured to communicate wirelessly. Recent examples are CC-ECG sensors on a chair [4, 6], bed [7] or clothing [3, 8, 9]. Also, these CC-ECG systems provide unbiased and accurate data as the person under measurement can be unaware of the sensors or that vital signs monitoring is occurring [1]. These advantages make capacitive ECG an excellent choice for continuous and convenient vital signs monitoring.

It should be noted that CC-ECG sensors do suffer from some shortcomings that should be carefully considered. Because of high impedance between ECG signal source and the sensor, the signal quality is not as good as the signal coming from wet ECG systems. In addition, because of this high impedance, the electrode acts as an antenna for environmental noise [1]. Moreover, capacitive ECG sensors are more sensitive to motion noise than fixed wet electrodes, since any change in the placement of the electrode changes the coupling capacitance and hence the ECG signal acquired. As a result, an undesired displacement current will be generated that may overwhelm the main ECG signal. We discuss more about these shortcomings and their solutions in the following sections.

SYSTEM DESCRIPTION OF THE DESIGNED SENSOR

Our designed ECG sensor is a two-lead system consisting of two basic parts: three identical-sized PCBs as electrodes for signal acquisition and common mode attenuation, and one main board for signal conditioning and transmission. Two of these electrodes are attached to the chest for signal acquisition from the body, and one of them is attached far from the chest on the right hip as a reference electrode (Fig. 2).

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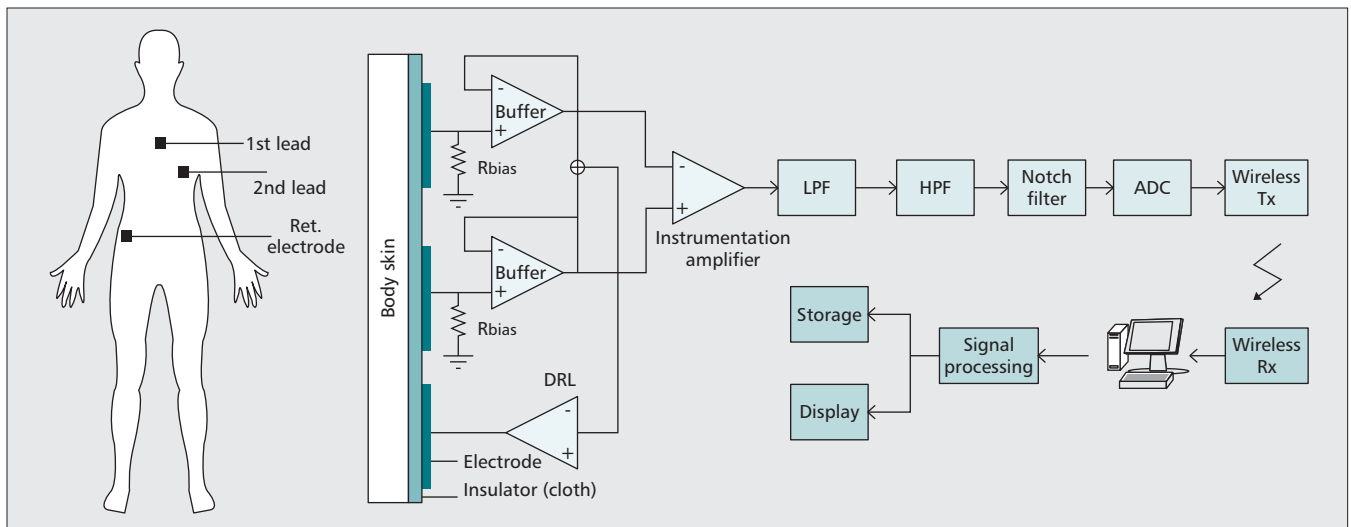


Figure 2. Schematic representation of the ECG system architecture and electrode placement on the human body.

The signal acquired from the electrodes then goes to the main board to be amplified, filtered, digitized, and transmitted. The differential amplifier differentiates the signals coming from the two lead electrodes and amplifies it to fit it to the analog input range of an analog-to-digital converter (ADC). Unwanted frequency components are then removed using a low-pass filter (LPF), a high-pass filter (HPF) and a 60-Hz-notch filter to improve the signal-to-noise ratio (SNR). The processed signal is then digitized with an ADC embedded in a tiny microcontroller (mC) and sent with a wireless module to the main station, which could be a PC in the home or a hospital network. On the PC side, this signal is then received by a commercial USB ANT stick, and goes into the PC for further processing and display on the screen using a software interface, which also has the capability to store the signal for further future examination by a health care professional.

Like many conventional ECG measurement systems, the driven right leg (DRL) circuit is used to reduce common-mode noise [10]. We discuss the DRL circuit in the next section. From the blocks shown in Fig. 2, buffers and their bias circuits were mounted on the back of the lead electrodes' PCB, the DRL circuit was installed on the reference electrode, and other electronic components were mounted on the main PCB.

For powering up the system, as the sensor will be used for long-term monitoring, it is important that a battery with high capacity be used to ensure that it does not need to be changed or recharged very often. On the other hand, as the capacity of the battery goes up, its size also increases. However, as one of our main goals is to keep the total size of the sensor as small as possible, a compromise on the size and battery life was made. A 3 V thin and relatively small lithium battery with a reasonable capacity of 256 mAh was chosen. A low-power DC/DC voltage converter (TPS61040), which was available in extremely small SOT23-5 packages, was used to provide +5 and -5 V biases to supply the amplifiers.

CAPACITIVE ELECTRODES

To achieve excellent capacitive coupling, the capacitance value must be quite high. Increasing the value of the capacitance can be achieved by either increasing the contact area, reducing the insulator thickness, or using an insulator with a higher dielectric constant. Reference [11] considered effects of electrode area and insulator thickness on the coupling capacitor and quality of the ECG signal. In this work, we did not consider increasing the areas of the sensing electrodes as large electrodes are inconvenient to the patient. In addition, although many researchers have used rigid insulators with high dielectric constant for better coupling, we are using normal cloth as the insulator to improve the patient's convenience. However, to increase the coupling capacitance, the cloth was chosen from materials with high dielectric constant and to be as thin as possible. To examine the effect of cloth material on the recorded signal, an experiment was performed; the results are described later. For other experiments, a cotton cloth with dielectric constant of 1.6 and thickness of 0.35 mm was chosen. The square-shaped electrodes with rounded edges and with electrode area of $\sim 33 \text{ mm} \times \sim 33 \text{ mm}$ were used for all the experiments, which resulted in a measured capacitance value of 110 pF at 5 Hz.

The high input impedance (in the range of 10 G Ω) of CC-ECG sensors means that a voltage buffer with high input and low output impedance should be used for impedance matching to the low impedance of the next stage amplifier. Several commercial low input bias current instrumentation amplifiers were used in many recent publications, especially INA116 [5, 12]. However, in this work, the LMC6001 was chosen because it has an input bias current in the same range as INA116, but offers much lower noise and input voltage offset, which helps in reducing the common-mode (CM) noise, the main noise source in CC-ECG sensors.

Before introducing the ECG signal to the instrumentation amplifier, a resistor (R_{Bias} in Fig. 2) was used to discharge the static charges

generated on the cloth, and to filter the DC and low-frequency noises such as motion noise. The CM noise, which is the main source of noise in CC-ECG, is attenuated using the DRL circuit on the back of the third (reference) electrode. Since the body can act as an antenna and pick up electromagnetic interference (EMI) from commercial power line (60 Hz noise) and electromagnetic interferers, there is always a CM signal on the body that normally is larger than biological ECG or electroencephalography (EEG) signals. Any mismatch between the two paths of the lead signals coming from the electrodes helps this CM signal to appear in the output and distort the ECG signal. Normally this distortion occurs because of the mismatch between the values of the electrodes' capacitors, biasing components, and different positions of the electrodes. The capacitance of the electrodes has the most effect on the signal parameters, but as the area and the insulator for both electrodes are the same, we should only be concerned with the drift of the electrode-skin separation. But this may not be a significant issue as both electrodes may be displaced equally during sensing. Also, it was shown in [11] that drift in capacitor values with time improve the signal-to-noise ratio (SNR) due to coupling enhancement. But even if there is some mismatch between the circuit parameters, the high common-mode rejection ratio (CMRR) of the differential instrumentation amplifier causes this CM signal to be attenuated. Also, the DRL circuit was employed to actively attenuate this CM noise even more [10]. Unlike other similar works that have mounted the DRL circuit together with signal processing components in the main board, we mounted the DRL circuit on the back of the reference electrode's PCB to reduce the total size of the system. Therefore, the third electrode also plays the role of driving the patient's body to attenuate the CM noise.

ECG SIGNAL PROCESSING CIRCUITS

Signals coming from capacitive electrodes must be amplified and filtered in order to generate an appropriate low-noise signal that fits within the input range of the ADC and data rate of the transceiver. A differential instrumentation amplifier first differentiates and amplifies the signals coming from two lead electrodes. As CM noise is the most important noise source in CC ECG, the INA106 was chosen for the differential amplifier due to its relatively high CMRR, reasonable gain (60 dB), and low noise. In addition, as the most important frequency components of the ECG signal are approximately in the range of 0.1–100 Hz, a low-pass Butterworth active filter with a corner frequency of 85 Hz and a passive HPF with corner frequency of 0.5 Hz were used to filter the unnecessary frequency components of the signal. A 60-Hz notch filter is also added to attenuate 60-Hz power line noise. To wirelessly send the signal to the main station, it must first be digitized. A low-power microcontroller (μC) with a low operating voltage range (PIC24FJ64GA) and an embedded 10-bit ADC samples the analog ECG signal at 500 samples/s and then sends it to the wireless transceiver. A

28-pin quad flat no-leads (QFN) package of size 6 mm \times 6 mm was chosen for maximum space saving on the PCB. The ADC as well as the wireless module were chosen to be capable of sampling and sending data even in idle mode to increase the battery life.

WIRELESS DATA TRANSMISSION

The ECG capacitively coupled sensor with signal conditioning and processing circuits is capable of providing useful vital signs information on the heart's functioning. However, for long-term bio-signal monitoring, it is more convenient to the patient to have the capabilities of a wireless transceiver to transmit acquired and processed bio-signals to an off-body host computer and control the functioning of the sensor system. However, adding a transceiver circuit to the ECG monitoring system will increase its size and power, and small size and low power are very important for on-body monitoring systems. Therefore, the key factors in choosing the wireless system are the size and power consumption of the module.

Low-power Bluetooth and Zigbee are two candidate wireless modules frequently used in related recent works. For example, [4] used the Zebra module, which consumes about 15 mA and measures 16 mm \times 33 mm. The CC2420 RF from Texas Instruments (TI) was used as another Zigbee module in [9]. It is fairly small, but it consumes about 19 mA in receive mode and 17 mA in transmit mode. BluesenseAD from Corscience, used in [12] as a Bluetooth module, also consumes \sim 30 mW and measures 37 mm \times 21 mm. Reference [8] provided a good review on wireless technologies and modules used in bio-signal measurements in recent years. The modules described in that review mostly used 802.15.4 radio and were fairly large for a portable sensor. However, the new module that was introduced in [8] was compact, measuring 13 mm \times 11 mm \times 7 mm. This module is reasonably small; however, it still consumes 10 mA in transmit mode and 22 mA in receive mode. In our case, an appropriate compromise on size and power consumption of the module led to the choice of the ANT protocol for wireless communication.

The ANT system is a proprietary protocol from Dynastream Innovations that combines a proven silicon solution with a low-level communication protocol that easily handles peer-to-peer, star, tree, and practical mesh topologies. Actually, ANT is an adaptive isochronous ad hoc wireless protocol that operates at 2.4 GHz and can communicate over distances up to 30 m. The isochronous feature relates to how slave and master are made aware of each other. After the master starts sending data, the slave will look for it; and as collisions between its time windows and the master's are noticed, they will start to synchronize with each other. This means that the master node is capable of determining when to transmit based on the activity of its slave neighbors and to remain off in other periods. This adaptivity, together with the low-data-rate transmission of ANT, makes it an ultra-low-power wireless protocol, which is well suited to low-data-rate applications like ECG sensing.

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Each node in an ANT network consists of an ANT engine and a host microcontroller unit (MCU). The ANT engine is responsible for establishing and maintaining connection and channel operation. However, the MCU handles the specifics of a particular application. A typical message between the host and the ANT engine consists of a SYNC byte to start, a CHECKSUM byte at the end, and MSG length, MSG ID, and data bytes in between. There are three main messages conveyed between the host and the ANT engine: configure messages, control messages, and data messages. Channel identification (ID), period, and radio frequency (RF), network settings, and some extra enabling will be set in the configuration message. The control message provides the channel opening and closing information and open receive (Rx) scan channel control. Also, the data message in its extended version consists of 13 bytes, channel number, device number (2 bytes), and device type ID, transmission type, and 8 data bytes.

For our sensor, because of the many attractive features described above, the ANT AP2 RF transceiver module was used. This module has a fairly small size of 20 mm × 20 mm and low power consumption of less than 1 mA at sampling rates suitable for our purposes. Under worst case conditions such as burst continuous mode with maximum sampling and baud rate, it consumes 6.3 mA with 1.9–3.6 V supply range. This module has a relatively low data rate compared to others, but for sampling of the filtered ECG signal in our sensor, it was adequate. Filtering removes some part of frequency informa-

tion of the ECG signal to satisfy the low-data-rate transmission; however, this might decrease the signal quality a little. After sending the signal, the ANT USB stick (Fig. 3) on the PC side, which is a receiver and microcontroller integrated together, provides a bridge between the ANT network and the PC at 38,400 b/s through a virtual COM port.

Utilizing the advantages of the ANT module, the total size of the system, which is mostly due to this module and the battery, now becomes 45 mm × 60 mm × 9 mm. This size is still smaller than similar previous works such as in [2, 8, 9] that implemented portable ECG sensors on shirts. In addition to its small size, low-cost, standard two-layer PCB technology was used for the main board and electrodes, resulting in low overall weight and cost of the system. This PCB also helped achieve ultra-thin electrodes (2 mm), which made them even more unnoticeable to the user. The designed PCB for the electrodes and the main PCB are shown in Fig. 3.

PC INTERFACE AND SOFTWARE ENGINEERING

As mentioned above, the main board does initial signal processing by filtering low- and high-frequency noise sources and attenuating 60 Hz power line noise. The DRL circuit also helps to reduce the CM noise and improve the SNR of the sensor. This signal processing in the main board is usually enough to have a well-shaped ECG signal in the output node that could easily be read by a cardiologist. However, in some cases where monitoring occurs in a high-noise environment, further signal processing and noise cancellation techniques may be required. This additional signal processing in our work was done in the PC, and included 60 Hz and motion noise cancellation. This noise filtering was done using MATLAB and employing the IRRNotch command to remove the 60 Hz noise. In addition, in cases where there are lots of motions like running, the low-frequency components of the signal can be removed by enabling an additional HPF with a corner frequency of 5 Hz, as the motion noise frequency components are usually lower than 5 Hz.

TEST SETUP AND MEASUREMENT RESULTS

To test the operation of the ECG sensor system, frequency domain and time domain responses were observed. To acquire the frequency spectrum of the sensor, the two lead electrodes were placed on a support, and then a 0.35-mm-thick layer of cotton was inserted on top of them as cloth; then two metal plates that model the body were attached to the insulator layer on top of the lead electrodes. Then a small pressure that models the force of a stretchable belt used for time domain measurements was applied by putting some weights on top. The frequency spectrum was measured by applying the signal to the plates and sweeping frequency from 0 to 10 kHz. Figure 4 shows the measurement results compared to simulations.

As shown in Fig. 4, the two data sets agree with each other quite well. Also, as expected, the

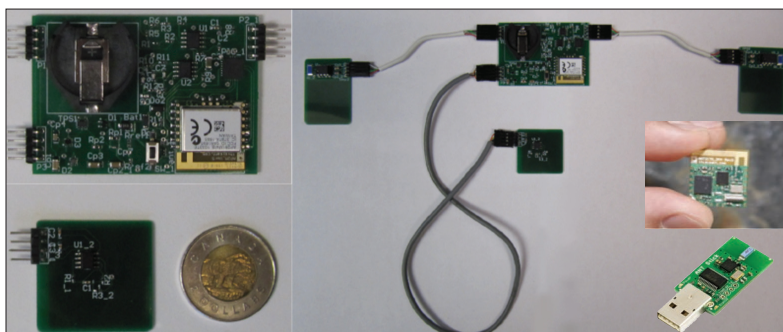


Figure 3. Photograph of the designed PCBs: signal processing PCB (top left), electrode PCB and a two-dollar Canadian coin (bottom left), the entire system (right), and the ANT AP2 transceiver module and USB stick (lower right).

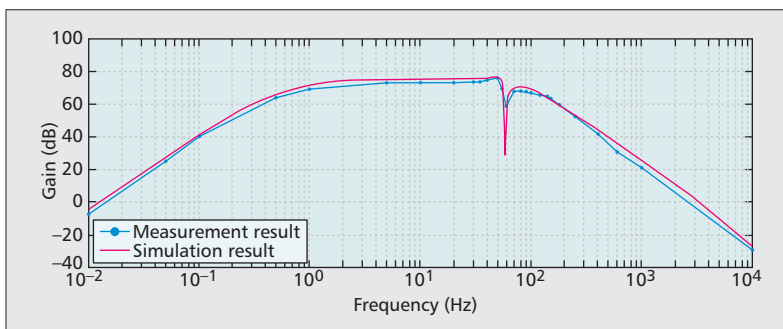


Figure 4. Frequency spectrum of the sensor. Both measurement and simulation results are shown.

corner frequencies of the spectrum are between 0.5 and 85 Hz, and there is 20 dB attenuation of the 60 Hz frequency. Even though this attenuation helps to reduce the CM noise, it is followed by another 60 Hz noise reduction block in the PC, as described earlier. In this way, almost complete cancellation of the noise is feasible.

To test the time domain response of the sensor, the electrodes were woven under a stretchable belt (to attenuate motion noise effects) that was fastened to the body of a healthy 24-year-old male on top of a 350- μm -thick cotton T-shirt. The main PCB was also attached to the T-shirt. The two lead electrodes were attached to the chest and the reference electrode far from them on the hip of the test person, as previously shown in Fig. 2. He sat on a chair and was not deliberately moving during the measurements. Figure 5 shows the results before and after software signal processing from the capacitive approach together with the signal from 2-lead standard wet-electrode (Ag-AgCl) ECG on the same person.

The raw signal (bottom trace in Fig. 5) indicates that there is noticeable 60 Hz noise, which was removed via 60 Hz notch filtering using MATLAB (middle trace in Fig. 5). Also, the processed signal from the capacitive method is a bit weaker and has a little more noise than the signal coming from wet ECG. In addition, results from the CC method show some variation from pulse to pulse, which might be because of change in the coupling capacitance due to body movements and perspiration. However, in the wet ECG system, the electrodes are glued to the body, which makes them less susceptible to motion artifacts, and therefore, such pulse-to-pulse variations are less. Despite these small changes in the CC ECG signal, it is qualitatively comparable to the wet ECG trace and the QRS complex, and important peaks like P and T waves are easily detected. The total current consumed from the battery during the measurements was less than 25 mA. So, considering total capacity of the chosen battery and idle mode working of the sensor, it should acquire the ECG signal and send it to the base station for at least 15 h continuously, which is reasonable for such a small-sized lithium battery.

Another experiment to investigate the effects of the insulator material on the measured ECG signal was performed. The signal was measured when the electrodes were in direct contact with the skin without an insulator (dry-electrode ECG) and compared with CC-ECG results from cotton and wool insulators, each 0.35 mm thick. The results are shown in Fig. 6. As shown in this figure, the signal from the electrodes with the wool insulator is weaker than the other two signals and has lower quality due to the lower dielectric constant of wool (1.4). The trace using wool also has some undesired peaks that may be due to unwanted charge transfer between the electrode's plate and the wool. These unwanted peaks make it difficult for cardiologists to clearly detect the P and T waves. On the other hand, the CC-ECG from cotton cloth and dry-electrode ECG resulted in good-quality signals that are comparable to the signals from the wet ECG method in which QRS complex and P and T waves can easily be detected.

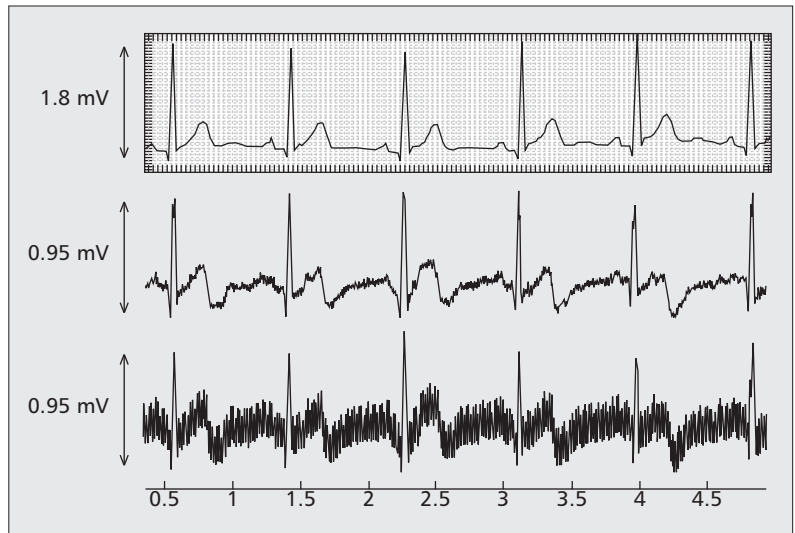


Figure 5. Time domain measurements of a conventional wet ECG (top), CC-ECG from the sensor after signal processing (middle), and CC-ECG from the sensor before signal processing (bottom).

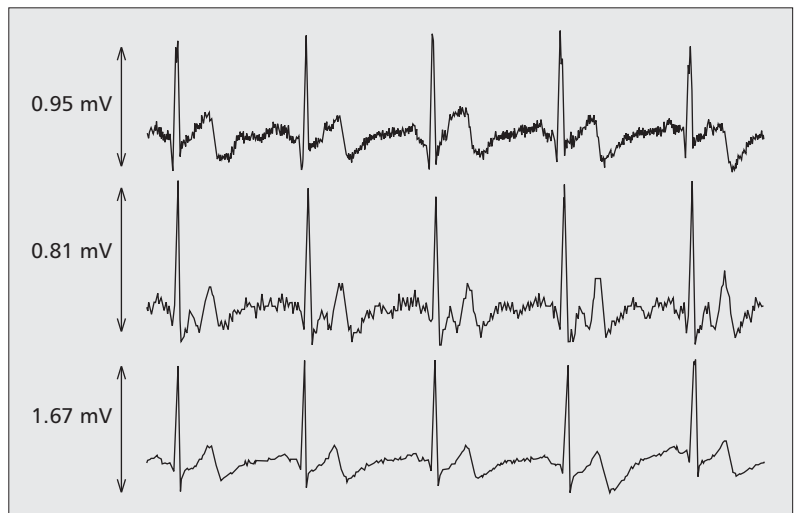


Figure 6. Measurement results from CC-ECG with cotton cloth (top), wool cloth (middle), and dry-electrode ECG (bottom).

Many previous publications that employed the contactless CC-ECG approach did not provide a portable wireless sensor [6, 7, 11]. Since there are no strict limitations on the size of the electrode in these type of sensors, the signal quality can be improved as the size of electrodes and coupling quality increases. Compared to those publications that used wireless communication, our proposed sensor system is smaller and thus more convenient to use. Some previous publications implemented their sensors on a chair or bed [4, 6, 7, 11], but these systems are not very suitable for ambulatory applications. However, compared to most of the cases in which continuous ECG monitoring were possible [2, 8, 9, 12], our proposed sensor is smaller in total size and weight, and has lower power consumption because of the low-power components and the extremely low-power wireless module used. Our proposed wireless ECG sensor system could be made more sensitive and compact if we

Reference	Technologies used	Implementation	Advantages	Limitations	Wireless technology
[2]	Capacitive sensing	Integrated on cloth	Flexible electrodes — 24 mm × 41.7 mm conductive fabric with conductive acrylic adhesive	Large inconvenient electrode, large system	FM transmitter at 315 MHz — high power
[3]	Capacitive sensing	Integrated on a 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 cm ²	Bluetooth
[4]	Capacitive sensing	Integrated on chair	Un-noticeable on-chair electrodes – 4 cm × 8 cm	Not for ambulatory applications, No motion artifact suppressing method, Ref electrode — 34 cm × 23 cm	Zebra Zigbee module — 15 mA, 16 mm × 33 mm
[6]	Capacitive sensing, Active shielding	Integrated on chair	Integrated on-chair electrodes –4 cm × 4 cm	Not for ambulatory applications, Thick electrodes — 12 mm (due to shielding plates), Large ground (Ref.) plane — 45cm × 30cm	—
[8]	Capacitive sensing	Integrated on cloth and belt	Fairly small electrodes (15 mm diameter) and wearable sensor	Extremely low-capacity battery (maximum 12 h usage), No motion artifact suppressing method	Eco wireless module 2.4 GHz — 10 mA transmit and 22 mA receive modes, 13 mm × 11 mm
[12]	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
Proposed sensor	Capacitive sensing	Integrated on a stretchable cloth	Lowest power consumption compared to other systems, relatively small sensor size, ultra-thin electrodes (2 mm) using standard 2-layer PCBs	Rigid Electrodes	ANT module, extremely low power — 1 mA (lowest compared to others), 20 mm × 20 mm

Table 1. Comparison between our proposed sensor and other recent similar works.

use integrated circuits (ICs) fabricated from a standard IC manufacturer using some of our previous high-performance ICs such as oscillators and amplifiers [13, 14] and more efficient energy sources. Finally, Table 1 provides a summary of recent ECG sensor systems and their advantages and limitations compared to our proposed sensor.

CONCLUSIONS

In recent years, many efforts were made to develop contactless, portable ECG sensors for continuous vital signs monitoring. But as of now, there are no standards for the system's size, architecture or performance. Based on our experience, for a contactless, easy-to-use, ubiquitous ECG sensing system, three guidelines can be provided. First, direct contact with the skin should be avoided. Second, for the sensor to be

carried conveniently without disturbance of users' normal lives, small size, low power consumption, and wireless communication should be employed. Third, deploying the sensing electrode on a T-shirt or undergarment is a good option as it can be widely and easily used under many physical conditions of the user. In this work, a small portable ECG sensor was designed using a two-lead capacitive coupling approach. Inexpensive, standard two-layer PCB design was employed to achieve thin electrodes that were easily mounted on a T-shirt. Using several size miniaturization techniques such as selecting small commercial components and distributing them to four PCBs, the final size of the electrodes and main PCB was smaller than many similar recent efforts. A small wireless module was employed to send the processed ECG signal from the main board to a PC using the low-power ANT protocol. In addition to signal pro-

cessing in the main board, software engineering was used to provide additional signal processing and noise cancellation and for displaying the ECG signal on the PC's screen. Due to its small size, low weight, and good quality of the measured signal, we believe that this sensor is a good candidate for long-term ubiquitous ECG monitoring. Future improvements can include making the capacitive electrodes even more convenient by employing flexible electrode PCBs or removing the battery and providing power wirelessly to further decrease the size of the sensor. Also, further signal processing in the main board that detects major abnormalities could also help in reducing power consumption and increasing the battery life to 24 to 48 hours of continuous monitoring.

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BIOGRAPHIES

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Future improvements can include making the capacitive electrodes even more convenient by employing flexible electrode PCBs or removing the battery and providing power wirelessly to further decrease the size of the sensor.