A Portable Wearable Tele-ECG Monitoring System

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Abstract—This paper introduces a wearable Tele-ECG and heart rate (HR) monitoring system which has a novel architecture including a stretchable singlet redesigned with textile electrodes (TEs), textile threads, snap fasteners, Velcro, sponges, and an ECG circuit. In addition, a Bluetooth low energy (BLE), a smartphone, a server, and a web page have been added to the system for remote monitoring. The TE can be attached to and removed from the singlet by a Velcro, which allows the user to dry-clean the TE easily for long-term use. A new holter-based ECG system has been designed to evaluate the TE-based ECG system and the average correlation between the recorded ECG signals is obtained as 99.23%. A filtered digital signal, with a high signal-to-noise ratio of 45.62 dB, is transmitted to the smartphone via BLE. The ECG signal is plotted, the HR is calculated with 1.83% mean absolute percentage error, and displayed. The data are sent to the server, allowing the patient’s physician to analyze the signals in real time through the web page or the smartphone. If HR reaches beyond the normal range or user presses the “HELP” button on the smartphone screen, the physician is informed automatically by an SMS and a flashing LED. This fast and uninterrupted telemonitoring system has the potential to improve the patient’s life quality by providing a psychological reassurance.

Index Terms—Bluetooth low energy (BLE), ECG telemonitoring, healthcare, Internet of Things (IoT), telehealth care, wearable monitoring system, wearable sensors.

I. INTRODUCTION

PATIENT care demands on hospitals are on the rise in parallel with the growth of the world population and the accompanying rise in chronic diseases. According to the mortality statistics recorded in 2015, 17.7 million people have lost their lives due to cardiovascular diseases in the world. This number corresponds to 31% of the total number of deaths [1]. Instrumentation and measurement (I and M) are a keystone to diagnose each disease [2]. ECG measuring is quite important to examine and monitor a patient with heart problems. In a conventional ECG measuring system, Ag/AgCl electrodes are attached to the body, with conductive gel at the electrode–skin interface, for signal acquisition to derive certain ECG leads. The conductive gel may be toxic and cause skin irritation [3], [4] and the Ag/AgCl electrodes may cause allergic reactions of the skin [5]. Despite their high conductance initially, after prolonged use, Ag/AgCl lose their adherence resulting in signal loss and requiring replacement. Therefore, the use of Ag/AgCl electrodes for extended periods of patient monitoring is not appropriate. As alternatives, dry metal electrodes have been used [6], [7] and some studies have explored textile electrode (TE) that has a minimal effect on patient’s normal lifestyle [8]–[12].

The TE also has high contact conductivity that is essential for a reliable ECG acquisition [13], [14]. TE can be used to pick up ECG, EEG, and electromyogram (EMG) signals and they have been proven to be as reliable as Ag/AgCl electrodes [15]. In normal conditions, the conductance of TE-based ECG monitor is regarded as equivalent to Ag/AgCl. Humid conditions are advantageous rather than detrimental to TE-based monitor due to the electrolytic properties of sweat. The electrolyte includes molecules and ions. Therefore, it increases electrical conductivity [16], [17].

Contactless ECG sensors provide the convenience of monitoring in nonhospital environments [18]. Diverse ECG measurement systems have been reported including systems that can be fixed to chair [19], bed [8] or can be adapted to clothes [20]–[22]. Nemati et al. [3] demonstrated a wireless ECG monitoring system which consisted of a belt stretched inside a T-shirt and equipped with three capacitive electrodes, where the ECG measuring is transmitted to a PC, requiring the user to own a PC. However, once this proposed system is preferred, the patients can use their existing cell phone for monitoring the vital signs. Mahmud et al. [6] collected ECG signals using two electric potential integrated circuit sensors fitted in a phone case and were able to transmit the signal with an Rfduino through Bluetooth low energy (BLE) and displayed the ECG signal on a cell phone. They also use smartphone memory to save ECG data, which is not appropriate for multiple usages due to limited data storage. In this paper, both local memory and server are implemented to allow multiple patient follow-ups without limitation to data storage. Chamadiya et al. [8] performed ECG measurements with electrodes fitted on stretchers, wheelchairs, and patient beds. They installed the conventional Ag/AgCl and TE on the same measurement setup to compare the results of both electrode types and concluded that there was no discrepancy between the measurements of the two types. In their work, the patient had to be stable in the supine position dictated by the measurement system or had to maintain contact with electrodes reclining back in a chair, which restricts the movement of the patient. On the other hand, in this paper, the person does not need to recline against or maintain contact with anything and, hence, favorable since the ECG is acquired under normal motion and living conditions. Pani et al. [23] concluded that ECG devices as wearable devices, using either dry or wet TE, without any
To monitor ECG, Yoo et al. [24] developed a 24-h data acquisition system comprising TE in a smart dress. However, they reported ECG signal without heart rate (HR) or location. Lam-onaca et al. [25] equipped a smartphone with new sensors and used existing sensors on the smartphone to measure the body posture, falling, HR, blood oxygen saturation, eye pathologies, and respiratory system. Their system does not have a wearable part, and the HR was calculated using captured face image providing an added benefit for the user in case of instant HR measurement [25]. However, in case HR must be measured continuously, the system proposed in this paper would be preferred since the system measures ECG and HR continuously in real time. Pandian et al. [26] measured noninvasive physiological parameters and saved ECG data using electrodes placed on a vest. They also indicated photoplethysmogram, body temperature, blood pressure, galvanic skin response, and HR [26]. They used wireless and GPS module with their circuits for data transmitting without smartphone or BLE. As a result, their system consumes more power, and they reported that their system runs for a limited time of 4.5 h, which is not optimal. However, the proposed system in this paper can run approximately 335 h (14 days) continuously. Using a smartphone instead of a device without GPS was found more useful [27]. Arteaga-Falconi et al. [28] utilized the benefits of smartphone and developed a mobile biometric authentication by using measured ECG signal. This could be merged with the wearable ECG monitoring systems. To solve energy consumption, Luo et al. [29] proposed an ECG compression process, using which allowed lowering of data transmission and power consumption. They send the data to a computer with a wireless module. The user is required to use a computer for the initial setup. However, once the proposed system is chosen, the users do not need to have a computer. They can easily use an existing smartphone with the developed custom application. Dionisi et al. [30] proposed a flexible solar panel for their wearable monitoring system, which is a convenience to the user as it mitigates power consumption and frequent battery replacement problems. However, in the evening or midnight, the system stops since it needs daylight to operate. In this paper, the demonstrated system runs uninterrupted for a long time independent of daylight. In the future, a hybrid system could be developed, consisting of both solar panel and a battery.

In this paper, a wearable Tele-ECG monitoring system with a novel architecture consisted of TE, Velcro, textile thread, snap fasteners, sponge, and ECG front end has been designed to monitor ECG of the users while allowing the users to have a comfortable daily life and maintain better hygiene. In the previous studies, TE was generally sewn on a T-shirt or belt [9], [13]. In this case, cleaning of TE is not easy since the TE needs to be dry-cleaned. In addition, the TEs have to be sewn to different sized clothes for different weighted people. On the other hand, in this paper, the novel architecture is designed to achieve good quality ECG signal with TEs that can be easily attached, transferred, or removed from any different sized clothes by the Velcro. The TE can be easily dry-cleaned and its position adjusted on a human body. Also, there is no electrical cable on the system as the ECG signal is carried by the textile thread. The attached snap fasteners receive the ECG signals from the textile thread and transfer it to the electronic ECG circuit. The developed ECG I and M system measures the ECG signal, detect the HR, monitor them on both a cell phone and a web page, and record the signals both in the cell phone memory (if necessary) and custom server. The ECG front-end circuit has been designed and integrated with a CC2650MODA which is a 32-bit ARM Cortex-M5-M0 microcontroller (MCU) in industrial scientific and medical band incorporating a BLE to transmit the data using a radio link. The electronic card prototype has been enclosed in a case produced with a 3-D printer. The circuit card and three TEs have been fitted inside the stretchable singlet. Analog signals picked up by TE are amplified and filtered by the front-end circuit, then digitized and further filtered by a digital filter (DF) in the MCU and ultimately sent to a smartphone using the BLE. In addition to viewing the ECG signal and HR on the physicians’ smartphone screen, the data are retransmitted to the server. Moreover, the physician can be alerted automatically during an emergency HR level or by pressing a specialized “HELP” button for immediate attention, which could provide faster healthcare access to the patients. The medical specialists can view the ECG, HR, medical history, and location information of patients. To evaluate ECG, a holter-based new ECG system has also been designed and the data have been recorded from both TE- and holter-based systems. Their average correlation of ECG signals is 99.23% and maximum signal-to-noise ratio (SNR) value of TE-based system is 45.62 dB. To evaluate HR, the both designed system and a fingertip pulse oximeter have measured the HR values in the same condition and their minimum mean absolute error (MAE) and mean absolute percentage error (MAPE) are 1.1% and 1.83%, respectively.

Merging new technologies provides several novelties to this paper as easily washable TE and BLE usage are more novel than Ag/AgCl electrodes and Bluetooth/cable usage. Battery life of nearly 14 days of the proposed study is another novelty. The automated emergency alert with calculated HR value, manual help request, and battery replacement alert contribute novelties of the system that are aimed to enhance user experience. Additional novelties are that the physician can control the system; can add or remove new patients, follow vital signs in real time with location information, and add patients’ medical history to the server and observe them continuously on the smartphone or web interfaces. The developed system is appraised to be quite beneficial for cardiac patients and their physicians.

II. SYSTEM OVERVIEW

The flowchart of the proposed system is shown in Fig. 1. The system can access multiple patients simultaneously. On the patient side, the developed telemonitoring system comprises four main components as follows.

1) Underwear with three textile ECG electrodes, an analog ECG front end, an MCU with an onboard radio transmitter.
2) A user smartphone which receives patient data via BLE and relays it to a remote IoT server.
3) An IoT server to store the ECG data and transmit it to the physician’s smartphone or the specially designed web interface.
4) User interfaces for physicians to see patients’ ECG signal and HR remotely, by adding or removing patients to the database from both web interface and cell phone interface.
With very large transconductance amplifier gain, the negative feedback forces the inputs to the transconductance amplifier to become nearly equal, that is, \( V_{\text{IN}} - V_{\text{FB}} = V_{\text{REF}} \). The IA produces an output \( V_{A} \) in response to a differential input \( V_{D} \) and the voltage at the reference pin in accordance with: \( V_{A}(s) = (V_{D}(s) + V_{\text{REF}}(s))A(s) \). Consequently, unlike the traditional IA, the feedback to \( V_{\text{REF}} \) pin to remove dc offset in the differential input need not swing as far as the supply rails; \( V_{\text{REF}} \) pin subtracts the same amount of dc offset to remove BD in the ECG signal which is another advantage of using AD8237. Exploiting these relationships, the frequency response at node A shown in Fig. 2 is derived. The voltage gain of the IA is\( A(s) = \left( s + \frac{R_{1A}}{R_{1B}C_{1}} \right) \left( s + \frac{R_{1A}C_{1}}{1} \right)^{-1} \). On the other hand, \( V_{\text{REF}}(s) \), the output of the integrator, is given as \( V_{\text{REF}}(s) = -\left( sR_{2}C_{2} \right)^{-1}V_{A} \).

Combining \( V_{A}(s), V_{\text{REF}}(s) \), and \( A(s) \), the following equation is obtained:

\[
V_{A}(s) = \frac{s(s + R_{1A} / R_{1B}C_{1})^{-1}}{s^{2} + [(R_{1A}C_{1})^{-1} + (R_{2}C_{2})^{-1}]s + (R_{1A}R_{2}) / (C_{1}C_{2})^{-1}}.
\]

This stage is followed by an additional gain stage and filter which adds 20 dB to the overall gain and further attenuates the ECG signal at 125 Hz. The gain stage is a single-pole noninverting OA (AD8609) with gain function

\[
V_{B}(s) = \frac{(R_{2}C_{3})^{-1}}{s + (R_{3}C_{3})^{-1}} \left( 1 + \frac{R_{AA}}{R_{AB}} \right) A(s).
\]

The last stage is a Sallen–Key AAF with a transfer function

\[
V_{\text{ECG}}(s) = \frac{\omega_{0}^{2}}{s^{2} + \left( R_{6}C_{6} \right) \omega_{0} + 1},
\]

where \( \omega_{0} = \left( R_{6}C_{6} \right)^{-1} \) is the square of the natural frequency. Multiplying (1)–(3), the overall transfer function of the ECG front-end circuit is obtained by substituting the component values (4), as shown at the bottom of this page.

Fig. 3 depicts the frequency response of the analog front end between 0.1 and 125 Hz. The 3-dB cutoff frequencies are 0.4 and 24 Hz. The ECG signal is attenuated by 12 dB at 50 Hz and 35 dB at the Nyquist frequency (125 Hz). Group delay can be readily measured from the phase characteristics to be 10 ms approximately in the passband.

Fig. 4 shows the performance of the front-end circuit obtained from a SPICE simulator. A 1-Hz pulse signal which drifts sinusoidally with a frequency of 0.05 Hz has been used to test the circuit. This test aims to show the performance of the high-pass filter of developed front-end circuit to remove BD. Fig. 4(a) shows an input test signal with BD and Fig. 4(b) shows the output signal without the BD.

The designed ECG front end is interfaced to the CC2650MODA MCU. The front-end and MCU circuits are manufactured on the same printed circuit board (PCB) (Fig. 5). The component side of the developed circuit board hosts the ECG front end, CC2650MODA, an on–off switch, a push-button, and an LED. The system can be run or stopped with the on–off switch. The pushbutton-activated LED is used to test whether the system runs and to check if the battery is full manually. LED is not powered continuously to save battery life. In addition, to check the power automatically, the battery voltage is sampled and read by an analog input of the MCU every 6 h. The system stops working when the battery voltage is under 2 V. Therefore, it is compared with 2.05 V (1.025 V

\[
V_{\text{FB}} = \frac{R_{S}}{R_{1}} V_{\text{IN}}
\]

where \( V_{\text{IN}} \) is the input voltage and \( V_{\text{FB}} \) is the feedback voltage. In this configuration, the feedback voltage is equal to the input voltage, \( V_{\text{FB}} = V_{\text{IN}} \), and the voltage gain is

\[
A(s) = \frac{s(s + R_{1}C_{1})^{-1}}{s^{2} + \left( R_{2}C_{2} \right) s + \left( R_{2}C_{2} \right)^{-1}}.
\]

Fig. 1. Stages of the proposed system.

III. SYSTEM HARDWARE DESIGN

A. ECG Front-End Circuit Design

The wearable ECG device contains a CR2450 (620 mAh, 3.0 V) coin battery for the measurement and real-time transmission of ECG signal during daily routines. An analog filtering stage follows the amplification, and there are three main issues concerning it: motion artifacts and related baseline drift (BD) compensation, 50-Hz power line rejection, and antialiasing filtering (AAF) before analog-to-digital converter (ADC). A fifth-order bandpass filter (BPF) made up of cascading three stages of filtering provides a moderate performance on these filtering issues. Higher performance could have been achieved by increasing the analog filter order. However, this would call for additional active filters and would increase the physical size of the circuit as well as the power demand from the battery. Consequently, the filtering performance has been complemented with a DF. At this point, there is a tradeoff as to divide filtering between analog and DFs. Less analog filtering means higher order digital filtering and more computational burden on the MCU.

Fig. 2 shows the designed ECG front end and connected CC2650MODA MCU circuit. The ADC input range is 0–3 V, whereas the ECG signal range is −1.5 to +1.5 V. To ensure that the ECG signal complies with the ADC input range, a virtual ground (VG) is generated from the +3-V battery voltage and used it as the analog ground. Resistive divider made of \( R_{7}, R_{8} \), and the unity-gain buffer U2A produces +1.5 V at the output (labeled VG) with respect to the negative terminal of the battery and behaves as the signal ground.

The ECG analog signal is processed by an instrumentation amplifier (IA) and three operational amplifiers (OAs). This unit serves five purposes in the project as follows.

1) It presents a very high resistance to the electrodes.
2) It generates a VG for analog signals.
3) It amplifies the weak ECG signals.
4) It reduces BD and noise in the ECG signal.
5) It acts as an AAF before sampling and ADC operation.

AD8237 type IA and the sample ECG circuit of it in the datasheet have been chosen for use. An AAF and additional gain have been added to the sample circuit. The overall midband gain of the designed ECG front end is 60 dB.

\[ VA(s) = \frac{\omega_{0}^{2}}{s^{2} + \left( R_{6}C_{6} \right) \omega_{0} + 1} \]

The designed ECG front end is interfaced to the CC2650MODA MCU. The front-end and MCU circuits are manufactured on the same printed circuit board (PCB) (Fig. 5). The component side of the developed circuit board hosts the ECG front end, CC2650MODA, an on–off switch, a push-button, and an LED. The system can be run or stopped with the on–off switch. The pushbutton-activated LED is used to test whether the system runs and to check if the battery is full manually. LED is not powered continuously to save battery life. In addition, to check the power automatically, the battery voltage is sampled and read by an analog input of the MCU every 6 h. The system stops working when the battery voltage is under 2 V. Therefore, it is compared with 2.05 V (1.025 V
This article has been accepted for inclusion in a future issue of this journal. Content is final as presented, with the exception of pagination.

IEEE TRANSACTIONS ON INSTRUMENTATION AND MEASUREMENT

4

**Fig. 2.** Simplified ECG front-end and MCU circuits.

**Fig. 3.** Front-end filter frequency response.

**Fig. 4.** Response of the front-end circuit to a pseudo-ECG signal superimposed on a baseline of 0.05 Hz. (a) Input test signal with BD. (b) Output signal without BD.

from VG output). When less than this threshold, the ECG signal value is reset once to all zeros. Upon receiving an all zeros ECG signal, the smartphone alerts the user about the battery condition.

The solder side of the PCB has a coin cell battery housing and battery. ECG front end has five different stages: IA, OA, VG, AAF, BD compensation and MCU has four different stages: ADC, digital BPF, ARM Cortex, and BLE. The board size is 8 cm \( \times \) 3.75 cm.

**B. System Architecture**

A novel TE-based wearable system architecture has been designed for ECG monitoring to provide a good quality signal, long battery life, an option of cleaning it easily, and without interrupting users’ daily life. The redesigned singlet with the TEs and circuit is shown in Fig. 5 as the first version of ECG monitoring system. Three TEs are attached under the singlet with one each on the left and right side over the breasts and one under the left breast. The electrodes pass the ECG signal to the analog front-end circuit. The front-end circuit is positioned under the right breast via conducting textile threads. The TE under the left breast serves as the reference signal ground with other two serving as exploring electrodes. Although it is possible to acquire three standard ECG limb leads using three electrodes, it is opted to use Lead I, as this lead conveys sufficient information to the remote health personnel.

TEs are 7 \( \times \) 7 cm\(^2\) geometric structure patch of fabric made of silver-based threads (Adafruit Knit Conductive Fabric-Silver (Sn: 91612-0-1002, P1167)) placed on three locations. The fabric, certified to be Restriction of Hazardous Substances-compliant, is highly conductive with a resistance of less than 1 \( \Omega/\text{ft}^2 \) in any direction across the textile. The attachment has been designed to enable electrodes to be fixed onto any place in the singlet underwear and removed easily when necessary to allow adaptability for different sizes and the ease of washing. The three TEs are attached under the singlet by a Velcro with a soft sponge and a conductive textile thread from each electrode is terminated on a circuit box by a snap fastener (Fig. 5). The 3-D printed box houses the ECG and the MCU circuit card and the box can be placed in a pouch on

\[
\frac{V_{ECG}(s)}{V_D(s)} = \frac{1.438 \cdot 10^3 s^2 + 8.15 \cdot 10^{12}}{s^5 + 1111s^4 + 721 \cdot 10^3 s^3 + 9.75 \cdot 10^7 s^2 + 7.658 \cdot 10^9 s + 7.648 \cdot 10^8}.
\] (4)
the lower right corner of the singlet. The TE can be removed for washing and then returned.

To evaluate the performance of the TE-based ECG system, a new holter-based ECG monitoring system that uses Ag/AgCl conventional wet electrodes has been designed (Fig. 5) as a second version. The ECG front end is the same in two versions. The 3-D-printed cover has been modified to connect the cables of the Ag/AgCl electrodes to the circuit with three jacks. The two types of ECG monitoring systems have been operated in standing, walking, going upstairs conditions, and the results are compared.

IV. SYSTEM SOFTWARE DESIGN

System software has been developed in four parts: First, ECG signal digitizing, filtering, and transmission to a smartphone; second, Android-based smartphone implementation; third, IoT server; and fourth, web design. ECG signals acquired by the wearable TE and processed by the ARM MCU are transmitted using BLE protocol to an Android cell phone.

The firmware on the CC2650MODA MCU provides codes to configure the ADC, set up the infinite impulse response (IIR) filter coefficient registers and set up the BLE parameters. The firmware, with MCU timer, generates an interrupt signal to acquire the ECG signal every fourth millisecond. The IIR code processes the digitized signal, and the filtered signal is passed on to the BLE code. Cortex M3 kernel does not support floating point arithmetic. Therefore, provisions have been made to use integer arithmetic instead of floating point. The radio transmits the ECG signal to the smartphone.

On the cell phone, the ECG data are plotted in graphical form, and the calculated HR is displayed. In the meantime, ECG, HR, and the location information are sent to the server by a background code. Data transmitted to the server are passed to the web panel to be displayed by the physician. The doctor can view the patient’s ECG, HR, and location with previously saved patient’s medical histories in real time.

A. ECG Signal Transmission to a Smartphone

Analog signal after being digitized passes through a digital BPF which further reduces the BD and high frequencies. Five samples of the ECG signal are collected to form a packet, and 50 packets with 250 samples are sent to the cell phone in 1 s.

1) Digital Filter Design: A fourth-order Butterworth type BPF has been implemented on the MCU. As mentioned previously, the sampling frequency is 250 samples per second. For the sake of ease, an IIR implementation has been used rather than an infinite impulse response implementation as the latter would necessitate a much heavier computational burden on the MCU. Since the DF is a real-time filter, it is important that the MCU finish filtering and all the other jobs in 4 ms. The DF refines the ECG signal, improving the BD and further suppressing the line frequency interference. Analog filter side attenuates 12 dB at 50 Hz and DF side attenuates an additional 14 dB (Fig. 6); thus the total attenuation reaches 26 dB.

The method of obtaining IIR filter coefficients is by transforming a proper analog Butterworth BPF to a DF through a bilinear transform. A second-order low-pass filter and a second-order high-pass filter have been cascaded to implement a fourth-order BPF (5), as shown at the bottom of this page. The DF cutoff frequencies are set at 1 and 25 Hz with the transfer function.

Bilinear transform warps the frequencies as it maps the jω-axis onto the unit circle in the z-plane. Denoting the discrete frequency by ohm, the relation between analog and discrete frequencies is given by ω = 2f,tan(Ω/2) with Ω1 = 1/250. 2π = 0.0251327 rad/sample and Ω2 = 25/250. 2π = 0.6283185 rad/sample, the analog cutoff frequencies have been found as ω1 = 6.283516 rad/s and ω2 = 162.4598481 rad/s. Plugging these cutoff frequencies into the BPF transfer function (5) results in

\[
H_{\text{BP}}(s) \approx \frac{26393.202s^2}{s^4 + 238.645s^3 + 28474.3s^2 + 243607.4s + 1042071.5}
\]

Bilinear transform gives the discrete-time equivalent of this transfer function in the following equation:

\[
H_{\text{BP}}(z) = \frac{0.0663 - 0.1325z^{-2} + 0.0663z^{-4}}{1 - 3.1074z^{-1} + 3.6232z^{-2} - 1.9140z^{-3} + 0.3984z^{-4}}
\]

This can be translated into discrete-time domain and it involves two shift registers of length four, one each for x and y besides multiplication and addition of floating-point numbers. However, Cortex-M3 MCU has no floating-point unit. The filter input x[n] is a 12-bit unipolar binary number which is output by the on-chip 12-bit ADC. The arithmetic logic unit (ALU) of the MCU is capable of 32-bit signed multiply–add operations in a single instruction cycle. Since the most significant bit of the computation must be reserved for the sign bit, the filter equation mentioned above can use 31 bits for multiply–add operations and must avoid overflow of the ALU result. To perform the filter equation with integer arithmetic, the filter equation has been manipulated by multiplying and dividing through 8192, i.e., 2^13. Choosing a power of 2 is to facilitate the division by shifting the result right 13 times. The translated discrete time-domain filter which has its shifted coefficients is as follows:

\[
\]

Fig. 6(a) shows the frequency response of the DF and Fig. 6(b) depicts a simulation of the digital BPF. A test signal has been created by superimposing 0.2-Hz sinusoidal BD and 50-Hz interference signal both of magnitude of 50 on a clean ECG signal [Fig. 6 (top trace)]. Bottom trace shown in Fig. 6 indicates the response of the DF to the test signal.

In CC2650MODA MCU, the forward and reverse filter coefficients are held as constants while an intermediate signal named u[n] is stored in a four-element shift register. The required arithmetic multiply–add operations are performed between shift register locations and filter coefficients to produce the filtered output.
2) Packing and Transmitting the Data: For the data packet to be transmitted accurately, using BLE communication protocol, maximum byte counts, and interpacket intervals must be configured correctly. Interpacket interval for BLE is specified to be between 7.5 ms and 4 s [31]. In addition, a packet can transfer a maximum of 27 bytes.

In this paper, 250 samples of ECG must be sent in real time. It has been selected to send 50 packets per second with each packet containing five samples. Thus, every packet consists of 15 bytes. Every 20 ms, a data packet is transmitted to the cell phone. With this scheme, the restrictions on maximum byte count per packet and the packet transfer rates are not violated. Twenty three bytes of data can be accommodated per packet as the header uses four bytes. Fifteen bytes of data per packet are sent. Thus, the packet size of this paper is 19 bytes. BLE throughput of the developed system is: 1000 ms × 19 bytes/20 ms = 7.6 kb/s from CC2650MODA to the mobile phone. The ECG data are transmitted by writing a service routine for CC2650MODA using Bluetooth developer studio.

B. Android-Based Smartphone Implementation

An Android mobile phone application has been developed to make it possible for the physician and the patient to view in real time the ECG waveform and the HR. The mobile application, which uses the BLE technology to access the ECG module, sends the ECG data to the server for the physician’s appraisal. This application runs on Android phones with 4.3 Jelly Bean and higher versions that has Bluetooth 4.0 hardware and higher versions. Android interface screens are shown in Fig. 7(a) and (b) for patient and doctor, respectively. The system relies on the username to decide whether the user is a patient or a physician and directs them to the relevant user interface screen.

1) Reading and Displaying the ECG Signal: Wearable ECG circuit employs the CC2650MODA. A connection between the Android application and the CC2650MODA MCU is made using the media access control address. After the connection is achieved, transmit services of the CC2650MODA are determined. Next, the data are read off the service which sends the ECG data. During initial phases of development, Android BLE application program interface has been used, and with a maximum priority assigned to connections, a maximum of 40 packets have been read per second. Fast BLE library has been adopted for reading more packets per second. The designed Android application unpacks the samples and ECG waveform is displayed on the screen using the Spark library, and also the data are sent to the server in the background.

2) Calculating and Displaying HR: HR is the number of detected R waves per minute. To determine the peak value of the R wave, the ECG signal is differentiated. The instant of time where the derivative change from positive to negative is the highest is labeled an R-wave peak. The number of peaks in a minute has been displayed as HR on the cell phone screen.

To transfer the data to the server, message queuing telemetry transport (MQTT) protocol has been preferred and Android version of the Paho MQTT library is used. Data are passed to an HTTP server using MQTT Broker application on a LINUX-based server. Data arriving at the HTTP server are transmitted by the MQTT protocol to the physician’s phone. The latency is approximately 300 ms.

C. Web Design for ECG Monitoring

HTTP server which hosts the physician’s application has been developed using the NodeJS/Express Framework infrastructure. Angular 4 runs on the web panel interface. MongoDB database, which is an open-source and a free-to-use program, has been used for database function. A user registered as a doctor can view the list of patients assigned and their real-time ECG and HR (Fig. 8).

Reactive programming technology reads data arriving at Angular 4 application and is presented in graphics form using Peity Chart and Easy Pie Chart libraries. Titles of entries, which the physician uses, are presented in the left column of the web interface. The physicians can add or remove a patient as well as update their own profile and configure other preferences on this page. The patient’s name, birthdate, gender, and address are on the topmost page are found. ECG and the HR in real time are displayed below these entries. The prerecorded medical history of the patient can be examined on this page as well. A pin for the patient location on the map can be seen adjacent to the medical information. If there is an emergency, the marked location information sent by the doctor makes it easy for the ambulance staff to reach them.
V. EXPERIMENTAL RESULTS

The proposed system has been implemented and operated on both the singlet-with-TE monitor and the holter-type device with Ag/AgCl electrodes. Fig. 8 shows the TE-based system in operation on a 28-year-old male subject with the relevant screen views that are acquired with both cell phone and web interfaces. Real-time ECG signal, HR, location pin on the map, and medical history are clearly discernible on the liquid-crystal display screen.

Thirty volunteers, aged 25–50 years, have worn the singlet and attached the holter-type device on their body. They also had a fingertip pulse oximeter affixed to their forefinger to verify the measured HR. The ECG signals have been recorded for 10 min continuously. Arbitrarily selected 4-s intervals from the recordings of three subjects, aged 25, 33, and 48 years, are shown in Fig. 9. The red and blue ECG signals have been recorded from the holter- and singlet-type devices, respectively. The tests have been performed in standing, walking, and going upstairs conditions. No significant difference has been observed between the measured ECG signals acquired from holter-type device and singlet-type device. There is a slight difference between their SNR values (Table I). As a result, the proposed system can be used in both wearable singlet-type and holter-type ECG measurement and monitoring devices.

HR increases from standing to walking and stair-climbing conditions while the general appearance of the ECG signal is preserved. It has resulted in an HR of 60, 75, and 90 b/s for the first person while he was standing, walking, and going upstairs, respectively. The second person has had 75 b/s while standing and 90 b/s while both walking and going upstairs. The third person has same HR values of 105 b/s while standing, walking, and going upstairs conditions; however, the amplitude of the ECG T wave has varied.

To assess the noise reduction capabilities of the DF of the ECG front end and MCU; ECG signals have been recorded from the first subject with/without DF while standing, walking, and going upstairs conditions. The experiments have been performed using both the singlet-based wearable system and holter-based system. SNR values have been calculated as indicated in Table I. As it can be seen, the designed DF obtains satisfactory SNR levels. Due to the wet Ag/AgCl electrodes in the holter-based system, SNR values are better than the other system; however, the discrepancy is insignificant. The average correlation between the two systems in each condition is 99.23%. BD is seen removed (Fig. 9). Also, P, Q, R, S, T peaks are identified clearly. In the proposed system, TE can be easily washed and, hence, preferred for daily use instead of Ag/AgCl electrodes.

To evaluate the measured HR, MAE, and MAPE have been calculated. During the experiment in each condition, HR has been recorded ten times from both the singlet- and holter-type devices and fingertip pulse oximeter. It is observed that the results of singlet- and holter-type devices are the same. MAE and MAPE have been calculated across the HR results of the singlet-type device and the pulse oximeter as given in Table II. The error rates of the measured HR from the proposed Tele-ECG monitoring system increase when the subject is moving. However, they are still within satisfactory levels.

In the meantime, the HR has been cycled from 60 to 80, 100, and 120 bpm from ECG simulator every 30 min. In 4-day data that have been collected, a total of 80 times on the cell phone screen and saved to the server. For HR data verification, the Bland–Altman plot [30], [32], [33] has been used to show the difference between the measured HR and ground-truth HR against their average (Fig. 10). Low–high limits of agreement are calculated with \((\mu \pm 1.96\sigma)\) and displayed, where \(\sigma\) is the standard deviation and \(\mu\) is the mean of difference. The 96% of all data are between the low and high limit lines. Also, the Pearson correlation between the measured HR and ground-truth HR has been calculated as \(r = 0.999\).

The wearable device is powered by a CR2450 (3.0 V) lithium coin battery since it is thinner, smaller in size, and easy to find. The battery housing can easily be replaced by a larger one to accommodate a new battery with a higher mAh rating. To reduce the power consumption, the unused pins of the MCU are three-stated and sleeping BLE is woken with an

![Fig. 9. Three cases of ECG recording in the conditions of standing, walking, and going upstairs. Red and blue: ECG signals that are recorded holter- and TE-based devices, respectively.](image)

![Fig. 10. Bland–Altman plot of the measured HR.](image)

<table>
<thead>
<tr>
<th>TABLE I</th>
<th>PERSON’S SNR VALUES IN DIFFERENT CONDITIONS</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Using the TE-based system</td>
</tr>
<tr>
<td></td>
<td>Standing</td>
</tr>
<tr>
<td>Without DF</td>
<td>33.21 dB</td>
</tr>
<tr>
<td>With DF</td>
<td>45.62 dB</td>
</tr>
<tr>
<td>Using the holter-based system</td>
<td>33.45 dB</td>
</tr>
<tr>
<td>Without DF</td>
<td>45.89 dB</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>TABLE II</th>
<th>MAE AND MAPE BELONG TO HR RECORDED IN THREE CONDITIONS</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>1. Person</td>
</tr>
<tr>
<td>Standing</td>
<td>1.10</td>
</tr>
<tr>
<td>MAPE</td>
<td>1.83%</td>
</tr>
<tr>
<td>Walking</td>
<td>2.10</td>
</tr>
<tr>
<td>MAPE</td>
<td>2.80%</td>
</tr>
<tr>
<td>Going up stairs</td>
<td>MAPE</td>
</tr>
</tbody>
</table>
interrupt only when necessary. Due to these precautions and micropower IA, OA used in the front-end circuit, low current and power consumption are achieved. The preferred battery has been tested first using just ECG front end only, then using ECG front end with MCU; 0.25- and 1.85-mA battery current have been measured for both setups, respectively. The used battery has an energy capacity of 620 mAh. With this battery installed, the system can run 335 h (approximately 14 days) and 250 sample/s data have been packed and transmitted to the cell phone free of problems.

For the software, smartphone applications (for both user and doctor), server, and web interface have been tested and evaluated. During data plotting, all data are transmitted to the server without any problems and stored in the background. “HELP” request from user smartphone has been sent and received from doctor’s smartphone with the location marker on the map. The web interface is easy to use and adaptable for any request. New patients can be added and deleted promptly. ECG and HR are checked with different scenarios and with different people. All the functions have worked through the testing without failure. The order of the designed web and smartphone interfaces can be easily modified on request.

VI. DISCUSSION

The proposed Tele-ECG system has been compared to the previous work for different features given in Table III. Considering the wireless ECG health monitoring systems, use of the conventional Ag/AgCl electrodes is still used despite their adversely effect on the human body when used for extended periods [34], [35]. As an alternative to the conventional Ag/AgCl electrodes, some researchers have reported the use of dry electrodes [6], [36], [38], capacitive electrodes [3], and TE [8]–[13], [21], [37], [39]. In a recent study, Mahmud et al. [6] have proposed to measure instant ECG with a smartphone accessory but the displayed ECG signal has discernible BD and noise and their data are saved locally on their cell phone. With the limited cell phone memory, only one patient can be monitored in real time. Competitively, the BD and noise performance of the measured ECG signal of the proposed system in this paper is superior as demonstrated in Fig. 9 and this system saves the data either locally or in a cloud and is not limited to one patient. The remote server provides enough memory for data storage and the physicians can add or delete patients on the server. Furthermore, the smartphone app in the previous report has an emergency help button that sends location information to a physician through an short message service (SMS). In this paper, in case of an emergency, the proposed system sends the location information through an SMS as well but in addition to that, the physician has access to the webpage to monitor patient’s location and condition in real time. The SMS can be sent by two methods: the first method is manual, where the “HELP” button on the smartphone screen, can be used to send an alert message to the doctor in case of an emergency and the second method is automatic, where the system notifies healthcare specialist if the calculated HR reaches beyond a set range. In other studies, Spano et al. [36] and Wang et al. [38] have demonstrated the remote healthcare system by using dry electrode placing a belt and a medical vest with battery life approximately 7 and 2 days, respectively. Their studies have different advantages among them such as [36] tracks multiple patients while [38] provides emergency alert and geographical location information. In this paper, the proposed system has more advantage than them having all features in the same system with approximately 14 days of battery life. In one study, a capacitive electrode has been used to monitor ECG with a battery life of less than 1 day [3].

In the literature, there are several TE usage for ECG measurement since the TE does not contain adhesives or gel, which is a big advantage. Chamadiya et al. [8] have proposed redesigning the patient’s bed, wheelchair, and stretcher to measure ECG by TE. This system is appropriate for use in a hospital but not appropriate for home care. In their system, the patient position has to be adjusted and stayed stable to achieve a good quality ECG signal, which can be difficult and inconvenient for the patient’s lifestyle. In this paper, a modified singlet with TE is worn and the TE maintains contact with the body and users do not have to repose against
any electrodes. Some researchers have compared different types of TE performance [11, 13] and some of them have fabricated new TE [12, 21] in their studies.

Researchers in two studies have used TE to design a remote healthcare system [9], [37], where they focused on a good quality ECG measurement with long time use. However, it is quite important to provide the users with ease-of-use with minimal impact on daily life and easily washable wearable parts for hygiene along with ECG telemedicine technology. Also, adjusting the TE pressure for maximum contact with the skin surface is very important. Since they have sewn the TE to a T-shirt directly both aspects of TE pressure on the body and TE cleaning procedures are not addressed.

To mitigate the problem with washing, a recent study has fabricated different types of TE and evaluated the performance after washing. The only focus of the study is to create washable TE and show the ECG measurement performance after washing the TE 50 times. They demonstrate that the washable properties of TE for hygiene of wearable sensors are quite important. The TE is sewn on a bra and as TE and bra together required dry-cleaning after long-term use [39], which increases the cost of cleaning since bra does not need dry-clean.

In this paper, a novel TE-based wearable Tele-ECG monitoring system architecture has been proposed. The TE can be attached to, removed from, and moved on the singlet easily via the Velcro. The correct position of TEs on the singlet can be adjusted by the Velcro, if the users’ singlet size is different, allowing the system to be easily adapted for everyone who has different weights. The system architecture does not have a traditional electrical cable and the signal is carried by the textile thread. There snap fasteners receive the signals from the textile threads and transmit them to the electronic circuit. The singlet can be washed and the TE can be dry-cleaned, which is an advantage of the compared to previously reported methods. An elastic and soft sponge is provided between stretchable singlet and TE. The rubber sponge applies a gentle pressure to the body allowing the entire TE surface to touch the body. The proposed Tele-ECG system has been designed for comfortable use, ease of cleaning, good quality ECG, long battery life, and several telemedicine benefits such as emergency request, both manually and automatically, and the ability to follow multiple patients. In addition, medical histories can be added to the web page to track the progression of patient’s health condition, a feature not reported in the literature. Although the BLE technology with the chosen MCU consumes low energy and has long battery life, there is an automatic battery replacement alert for users to continue uninterrupted usage. As given in Table III that the previously reported remote healthcare systems have some of features included this paper; however, the proposed Tele-ECG system in this paper has all mentioned features seamlessly integrated for superior user experience. As a result, the designed Tele-ECG system would provide comfortable ECG monitoring system to users.

VII. Conclusions

In this paper, a wearable wireless Tele-ECG monitoring system is demonstrated. The system has a novel architecture which includes a singlet redesigned by attaching TE, textile thread, snap fasteners, Velcro, soft sponge, ECG front end, and MCU. The instruments on the singlet acquire the ECG signal and transmit it to a cell phone. The ECG signal along with the measured HR is depicted on the cell phone and then transmitted to a server, which allows the physician to observe the signals through the webpage and the smartphone. In addition, a holter-based ECG measuring system has been designed to compare TE-based system. Both systems have been evaluated with 30 volunteers in standing, walking, and going upstairs conditions. The highest SNR value of TE- and holter-based systems is 45.62 and 45.89 dB, respectively. The average correlation of ECG measurements of two systems in each condition is 99.23%. In addition, the HR is measured with both systems for each group and with fingertip pulse oximeter for all patients. The smallest MAE and MAPE are obtained as 1.1% and 1.83% between the TE-based system and the pulse oximeter. The wearable system merges the latest technologies such as TE, BLE, smartphone, server, and webpage to utilize and combine all the advantages into one telemonitor. The significant advantage and novelty of the proposed telemedicine technology is the combination of the best available technologies and the addition of few other features. No single reported device has all the features and benefits of the proposed device: comfort in daily life, easy to clean, high-quality ECG by TE, fast data transmission, long battery life with approximately 14 days, remote multiple-patient following by physicians, manual and automatic emergency requests, medical history, and geographical location tracking. The proposed system can potentially reduce congestion of hospitals and the cost of the medical examination since the patients can be monitored remotely for heart problems. As the future work, additional sensors can be added to the proposed system to measure the temperature, SpO₂, and movement of users and the TE can be employed to detect EEG or EMG.

Acknowledgments

The authors would like to thank Dr. B. Karagözoglu for his suggestions and comments, B. Gun for his official support, and Y. Arik and S. Kocakaplan for their SolidWorks program support and Aluminum test, training, and research center for 3-D printing.

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