The Deployment of Novel Techniques for Mobile ECG Monitoring

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Abstract

In this paper, we address the critical issues raised inmobile ECG monitoring and propose new solutions to resolve respective problem. A chest-belt type two-electrode wireless ECG measurement system and the specific android ECG monitoring application are presented. Furthermore, we also propose optimum filter-set for motion artifact removal in ECG recording. Average R-peak detection success rate is seen to be improved from 71.8% up to 97.4% after deploying the optimum filter-set coefficient. Besides that, we also suggest a simple and reliable arrhythmia ECG detection algorithm. The algorithm is proved with a promising accuracy of 95.4% in detecting arrhythmia ECG pattern. Lastly, a real time experiment is presented to demonstrate mobile ECG monitoring using a 10.1 inch Galaxy Tab.

Keywords: *Mobile ECG Monitoring, Two-Electrode ECG, Optimum filter, Motion artifact removal, Arrhythmia ECG detection*

1. Introduction

According to World Health Organization, WHO as estimated 17 million of people die of cardiovascular disease [1]. Every 34 seconds a person in United State dies from heart disease [2]. From the statistical study, heart disease is still the top killer worldwide. Real time and long term ECG monitoring could potential screen any abnormalities in the early stage. Therefore, various kinds of studies on wearable and mobile ECG monitoring have been initiated in the early of 21st century. In Euro, user trial experience on mobile healthcare has concluded that the demand of telemedicine in future healthcare application [3].

Electrocardiography (ECG or EKG) is a transthoracic interpretation of electrical activity of the heart over a period of time, as detected by attachable electrodes on the skin and recorded by an external device [4]. In this paper, we present a convenience and real time reliable ECG monitoring solution for mobile ECG monitoring application. Power line noise and skin impedance are the most critical issue for a ground-free (two-electrode) ECG measurement system [5-8]. Therefore, a new front-end low-pass buffer that reserve high input impedance and high CMRR is designed and implemented in this paper. Motion artifact and respiratory artifact in daily life induces motion noise into the ECG signal, causing measurement difficulty and low R-peak detection accuracy [9-11]. Thus, a new adaptive filter with optimum filter-set selection is suggested to reduce motion artifact based on different activity level. Finally, a simple and reliable arrhythmia ECG detection algorithm is proposed for arrhythmia ECG detection.

2. Hardware Architecture

ECG signal is a bio signal from human body. Signal acquit directly from body is apparently to be low in amplitude and often noise contaminated. Therefore an analog signal conditioning circuitry is required for signal amplification and noise removal is necessary. The amplified and noise filtered signal is 12-bits sampled at a sampling rate of 360Hz by using the ADC pin available in AT8mega8L microcontroller. Wireless transceiver is integrated with the microcontroller to provide wireless data transmission interface. The wireless transceiver can either be a Zigbee transceiver or a Bluetooth transceiver depending on the wireless technology used by the receiver.

2.1. A New Analog Front-end Low Pass Input Buffer

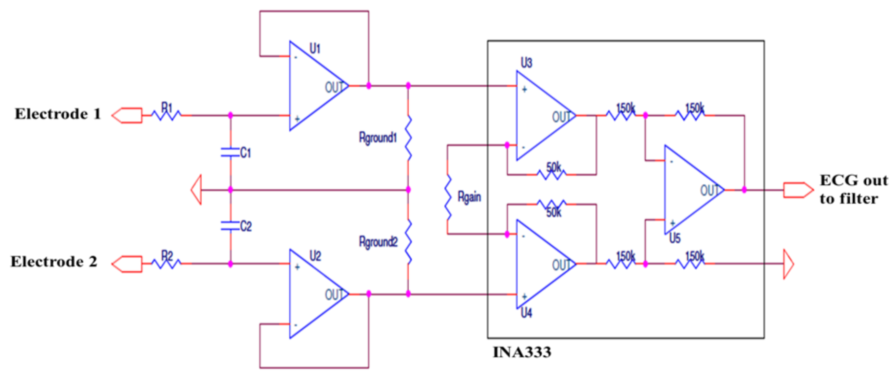

Figure 1. Circuit Diagram for Front-end Low Pass Input Buffer

Unlike conventional three-electrode ECG measurement system with right-leg driver (RLD), two-electrode system does not contain a ground electrode to feedback power line interference to body [5-8]. Therefore a very high input isolation impedance and common-mode rejection analog front-end circuit is designed for input noise isolation. Figure 1 shows the simplified circuit diagram for the proposed front-end low pass input buffer. The topology of low pass filter is to provide first level filtering to filter high frequency noise. The high cut-off frequency roll-off at 150Hz. U1 and U2 are the ¼ of quad general MCP6004 OPAMP. By making the ¼ MCP6004 a buffer, it will give a very high input isolation impedance of 1013 Ω [12]. A grounding resistor of RG is connected before the input of the instrument amplifier to present the low common-mode input impedance ZC solution as discussed in [8]. The grounding resistor provides DC path for instrument amplifier input bias current. If the input buffer is not grounded, there will be no current flow through the instrument amplifier, and all nodes in the front-end circuit will achieve same potential. High input impedance at front-end could allow a very high gain pre-amplifier possible without saturating the instrument amplifier. Unlike most of the conventional pre-amplifier technique, we suggest a gain of 100 at pre-amplifier, instead of conventional gain of 10 [13-17]. Figure 1 shows the circuit diagram for the proposed front-end low pass input buffer.

2.2. ECG Signal Conditioning Circuit

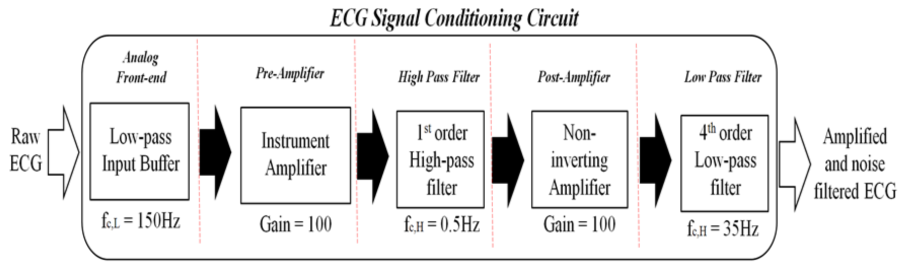

Figure 2. Block Diagram for ECG Signal Conditioning

Figure 2 shows the block diagram of ECG signal conditioning for the implemented ECG measurement system. Raw ECG is noise contaminated and very low in amplitude. Low-pass input buffer is implemented to isolate skin impedance and roll-off any high frequency component at the front-end. Then, the ECG signal is feed into an instrument amplifier, INA333 for pre-amplification. Considering the extremely high input impedance of the INA333 (~100GΩ) [18] and the very high input impedance of 10^{13} Ω from the input buffer. High input impedance will give a favorable common mode voltage rejection capability. This is the general understanding in analog circuitry design. Since we are expecting a very good CMRR at the front-end, we do not expect a very high gain at the pre-amplifier stage would cause output saturation due to common mode voltage. Thus, the gain of the pre-amplifier INA333 is set to be 100.

The very first requirement in mobile ECG recording is that the ECG measurement system must be worn around the user chest, allowing good user flexibility and free to move. Upper body movement would incur motion artifact noise and thus corrupt the ECG signal. Motion artifact is known to be a low frequency component in literature. Thus, a 1st order high pass filter is implemented to filter low frequency noise. Any low frequency component will be rejected at the cut-off frequency of 0.5Hz.

As discussed in previous chapter, ECG signal is apparently very low in amplitude (<1mV) in practical measurement. Therefore post stage amplification is required to further pull-up the signal level so that it can be measurement sensible. We applied a non-inverting amplifier with gain =100 for post-amplification.

Finally, the ECG signal is band limited at 35Hz by using a 4th order low pass filter. A typical ECG pattern appears in the bandwidth of 0.5~35Hz. The signal measurement between this typical bandwidth is sufficient to extract standard ECG pattern for mobile monitoring purpose. Another key to note, the overall system gain = pre-amplifier gain*post-amplifier gain, and yield an overall system gain of 100*100 = 10000V/V. A high-gain system promotes better signal quality and amplification. In comparison with other authors, our system gain is 10 times higher than those mentioned in [13-17]. No OPAMP saturation and no analog signal distortion are observed at the output of the signal conditioning circuit. Figure 3 Oscilloscope measurement of author's ECG signal at the output of ECG signals conditioning circuit.

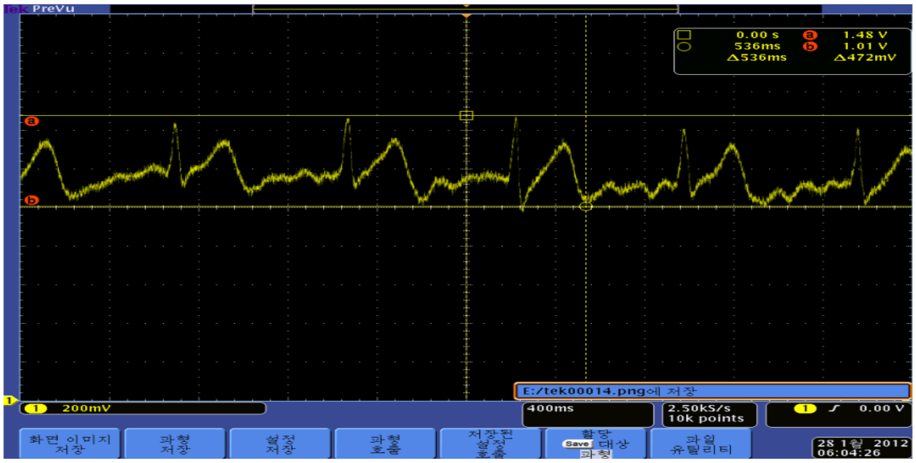

Figure 3. Oscilloscope Measurement of Author's ECG Signal at the Output of ECG Signals Conditioning Circuit

2.3.Mobile ECG Monitoring using 10.1 inch Galaxy Tab

The wearable chest-belt type ECG measurement system consists of a conductive fiber chest-belt, a signal conditioning circuit, a 3-axis accelerometer, a microcontroller and a wireless transceiver. Chest-belt type electrode is worn around user body to acquit the potential difference between two chest points (shown in Figure 5). The potential difference between two chest-points is amplified through ECG signal conditioning circuit as described in section 2.2. The output of ECG conditioning circuit is connected to the ADC0 pin at the microcontroller (MCU) for analog-to-digital conversion. The x,y,z output of 3-axis accelerometer are connected to ADC1, ADC2, and ADC3 pins accordingly. All the ECG and 3-axis data are sent to a mobile device (a 10.1 inch Galaxy Tab) for ECG and 3-axis data recording. Only ECG data are displayed on the Galaxy Tab and 3-axis data are recorded back end. Figure 4 shows the overall view of block diagram of streamline ECG measurement using mobile device, Figure 5 shows the in-house designed and implemented wireless wearable ECG measurement system, and Figure 6 shows the appearance of wearing the ECG measurement system and the real- time display of ECG signal on the Galaxy Tab for ECG monitoring purpose.

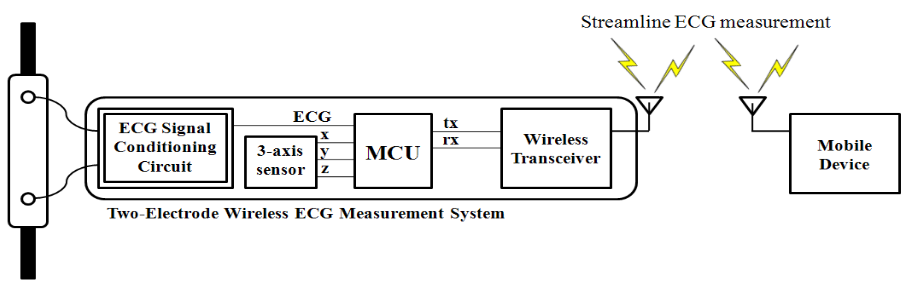

Figure 4. Block Diagram of Streamline ECG Measurement using Mobile Device

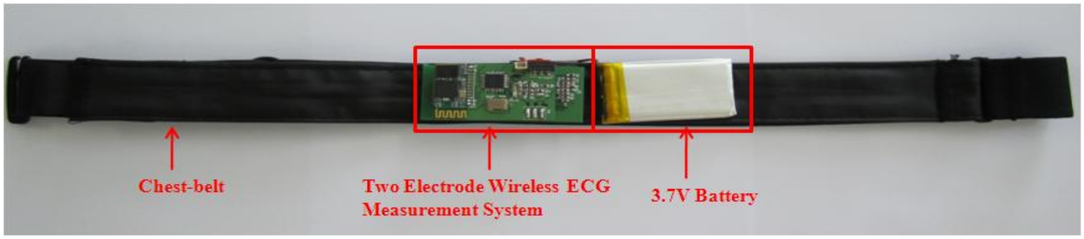

Figure 5. Implemented Wireless and Wearable ECG Measurement System

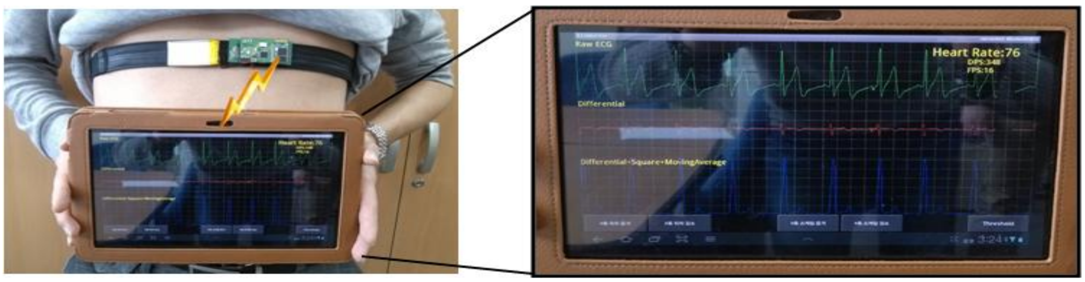

Figure 6. The Appearance of Wearing the ECG Measurement System and the Real- time Display of ECG Signal on the Galaxy Tab

2.5. A New ECG Monitoring Experience for Android Tablet

A new android application for ECG monitoring is developed and presented. The android application provides new experience for mobile ECG monitoring. First, we display raw ECG data received from the ECG measurement system through Bluetooth transmission. At the top right of the screen, we display heart rate, data per second (DPS), and frame per second (FPS). The heart rate is calculated based on Pan-Tompkins algorithm [19]. Pan-Tompkins presented a very reliable heart beat detection algorithm. He suggested that, ECG data should first to be differentiated, and then being square and moving averaged. The final outcome of the processing technique would give an enhanced QRS feature (R-peak) for heart beat detection and calculation. Raw ECG, differential ECG data and differential + squaring + moving averaged ECG are displayed on the Galaxy Tab as shown in Figure 7.

Figure 7. Snapshoot of the Android ECG Monitoring Program

3. Experiment and Results

3.1. Optimum Filter-set Coefficient

ECG recording in the presence of motion artifact is a critical challenge in mobile ECG monitoring application. Motion artifact will distort the shape of the ECG signal, resulting very low accuracy in detecting actual heart beat. Motion artifact being a low frequency component, it can be reduced by applying a high pass filter (HPF). However, a conventional high pass filter coefficient is a pre-defined value. A too low fixed HPF coefficient has less efficiency in detecting heart beat; a too high fixed HPF coefficient will result high distortion in ECG recording. Therefore, an optimum filter-set coefficient is proposed to apply variable HPF base on different activity state. The activity state (state 0, 1, 2, 3, 4, and 5) can be represented by walking speed (0, 2, 4, 6, and 8 km/h) accordingly. The calculated ISVM and the activity state classification is presented in Figure 8.

ISVM data are segregated into 14 sections for optimum HPF filter determination. The selection of HPF coefficient is auto-mated base on different state of activity. At low activity state, a low HPF coefficient is selected whereas at high activity state, a high HPF coefficient is selected. In comparison with general filter (fixed HPF) the proposed optimum HPF has a much higher average success rate in detecting the heart beat [20].Table 1 shows the result table of average heart beat detection rate (%) by comparing the use of general filter and the use of proposed optimum filter-set. At low activity state 1 and 2, HPF of 0.1Hz is selected. At medium activity state 3, HPF of 0.3Hz is selected. At high activity state 4 and 5, HPF 0.8 and 0.9Hz is selected for motion artifact removal in respective state.

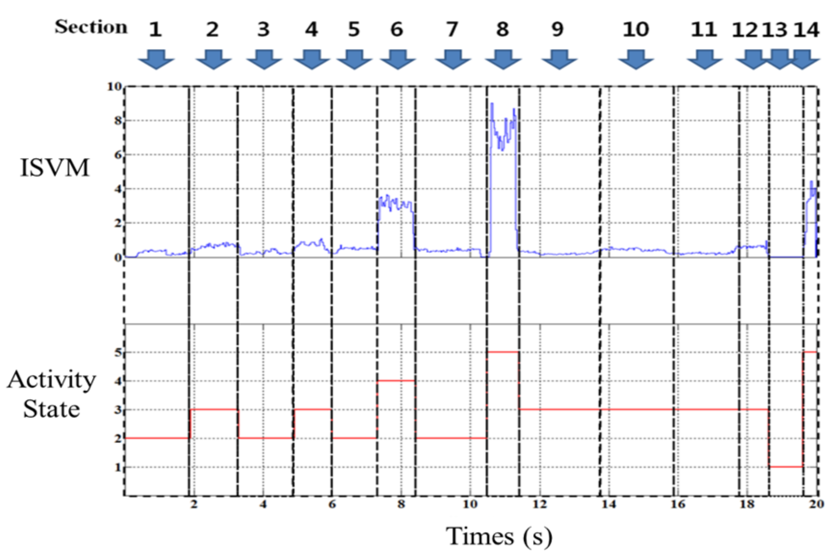

Figure 8. Activity Classification is based on ISVM calculation. Real time data are segregated into 14 sections to cross reference with ECG data in the respective activities states, so that, optimum HPF filter coefficient is selected.

Table 1. Comparison Table of Average Heart Beat Detection Rate between General Filter and the Proposed Optimum Filter-set Selection

State#	Section#	HPF (Hz)	Average Heartbeat Detection Rate (%)	
			General HPF	Optimum HPF
1	1,3,5,7,13	0.1	100	100
3	2,4,9,10,11,12	0.3	100	100
4	6	0.8	95.5	99
5	8, 14	0.9	81.8	90.5
	Average		**71.8**	**97.4**

3.2. Arrhythmia ECG Detection

A simple real-time arrhythmia ECG detection algorithm to classify morphologically arrhythmia ECG such as PVC, PAC, RBBB and LBBB is deployed and implemented. In this arrhythmia ECG detection algorithm, R-peak is detected using variable threshold method which has been widely discussed in [21, 22]. Thus, we would not further illustrate the R-peak detection method in this paper. From the point of R-peak, we define QRS window based on the estimation below:

```
QRS_start = 64ms before R peak (16 sample points)
QRS_end  =  188ms after R peak (47 sample points)
```

The first 30 seconds of normal ECG data is collected. All the QRS patterns within the 30 seconds period of time will be extracted for matrix generation. The extracted QRS pattern is in form of 64 x 32 samples in resolution. Each of the extracted QRS pattern is overlapped with each other, generating a normal ECG matrix template. In other words, the matrix template is formed by a 64 x 32 matrix cell. With every +1 overlapping on the same matrix cell, the cell value will increase by 1. The maximum cell value is set to be 10. If cell value is 10, this mean there is 10 successive overlapped matrixes on the same cell is in good match. The simplified Matlab pseudo code for matrix generation is shown as per below:

```
If PM(i, PW(i,j))<10->PM(i, PW(i,j)) = PM(i, PW(i,j))+1
If PM(i, PW(i,j)-1)<10 -> PM(i, PW(i,j))-1) = PM(i, PW(i,j)-1)+1
If PM(i, PW(i,j)+1)<10 -> PM(i, PW(i,j)+1) = PM(i, PW(i,j)+1)+1
```

After the generation of matrix template, the real time ECG signal is input to compare with the matrix template. If there is any out of match with the normal ECG matrix template, the particular ECG pattern is classified as an arrhythmia ECG. The ratePM is the index to classify the "goodness of match". If ratePM is high, it means the real time ECG pattern is matched with the matrix template. If the ratePM is low, it means the real time ECG pattern is not matched with the matrix template. Thus, arrhythmia ECG pattern can be classified in real time. The detail processing steps of creating a matrix template have been discussed in [20].

We evaluated the performance of arrhythmia ECG detection by using MIT/BIH arrhythmia database. The proposed algorithm has superior success detection rate in detecting premature ventricular contraction (PVC) arrhythmia ECG. The success rate of R-peak detection and PVC detection is tabulated in Table 2. Figure 9 and Figure 10 shows the result of classifying normal ECG and arrhythmia ECG. Black bolded line represents normal ECG whereas grey bolded line represents arrhythmia ECG.From the calculation result of ratePM, normal ECG and arrhythmia ECG are classified. Figure 11 shows the example of detecting arrhythmia ECG using record 105.

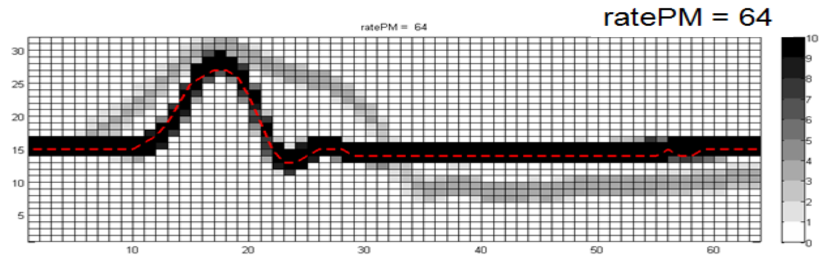

Figure 9. Result of Template Matching: The Black Bolded Line Represents Normal ECG Pattern and Shows a Maximum ratePM of 64

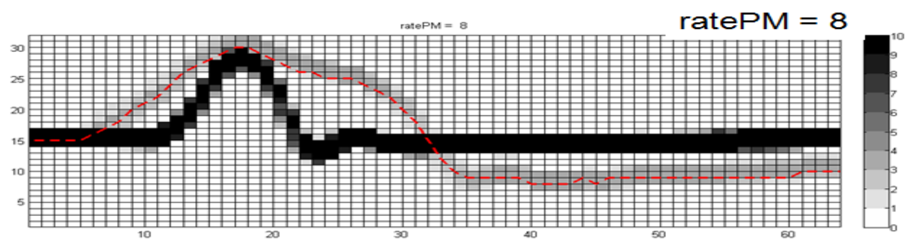

Figure 10. Result of Template Matching: The Grey Bolded Line Represents Arrhythmia ECG Pattern and Shows a Low ratePM of 8

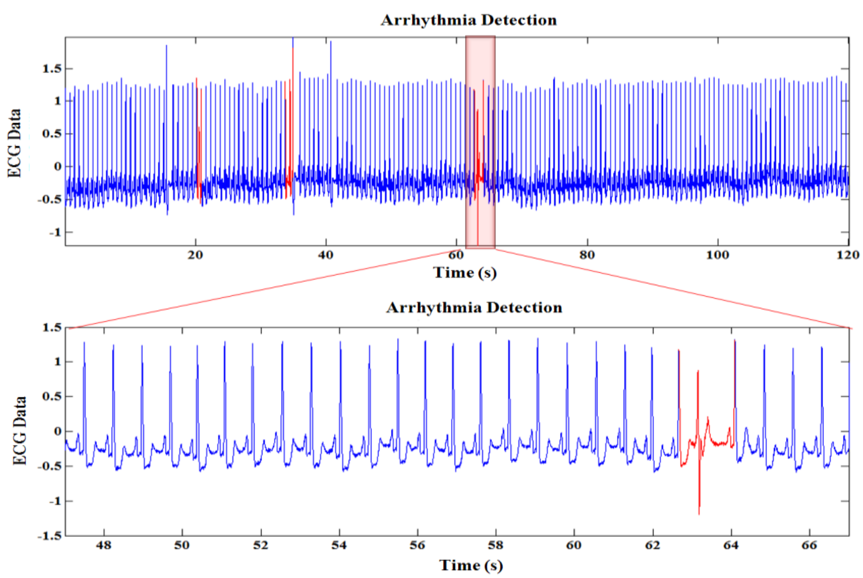

Figure 11. Result of Template Matching: The Grey Bolded Line Represents Arrhythmia ECG Pattern and Shows a Low ratePM of 8

Table 2. Result Table for R-peak and Arrhythmia ECG Detection by using the Proposed Template Matching Algorithm

MIT/BIH Record	MIT/BIH		Arrhythmia ECG detection algorithm			
	No. of Peak in record	No. of PVC in record	No. of Peak detected	Peak Detection Accuracy (%)	No. of PVC Detected	Arrhythmia ECG Detection Accuracy (%)
100	2273	1	2272	99.9	1	100
102	2187	4	2187	100	5	75
119	1987	444	1987	100	439	98.9
121	1863	1	1861	99.9	1	100
123	1518	3	1518	100	3	100
124	1619	47	1619	100	52	89.4
230	2256	1	2256	100	1	100
231	1573	2	1571	99.9	2	100
Average				**99.9**		**95.4**

3.3 Real Time ECG Monitoring

A real time experiment is conducted to demonstrate mobile ECG monitoring using the 10.1 inch Galaxy Tab. Author is wearing the chest-belt type ECG measurement system and walk around in Dongseo University compound following a pre-defined path as shown in Figure 12 section 1 and 3 are examining walking on slope, section 2 is examining a 4m circular walk on flat ground, section 4 is examining jumping at a stationary point for 3 minutes, and section 5 is examining running at constant speed for about 40 seconds. Author is monitoring his ECG pattern and heart rate (heart beat per minute) throughout the experiment. An estimated heart rate is recorded in Table 3.

In this experiment, the chest-belt type two-electrode ECG measurement system provide streamline ECG measurement to the Galaxy Tab. Real-time ECG is displayed and monitored on the Galaxy Tab as described in previous chapter. When the experiment is completed, all of the ECG data are saved in a .txt file format. The respective .txt file will be email to author's email via 4G Ethernet connection. Author downloads the .txt file from mail box and offline analysis is performed using Matlab. Figure 14 shows the snapshoot of the android ECG monitoring program while author is performing a 4 m circular walk on flat ground (section (2)) Figure 13 shows the 3D plotting of the walking path to provide better understanding on the landscape view

Table 2. Result Table for Real-time ECG Monitoring

State#	Activity under test	Heart Rate (bpm)
1,3	Walking on slope	120-185
2	Walking on flat ground	100-125
4	Running at constant speed	120-180
5	Jumping at stationary point	120-180

Figure 12. Walking Path: Section (1) and (3) is Walking on Slope, Section (2) is Performing a 4 m Circular Walk for 4 Minute on Flat Ground, Section (5) is Running at Constant Speed for 40 Seconds

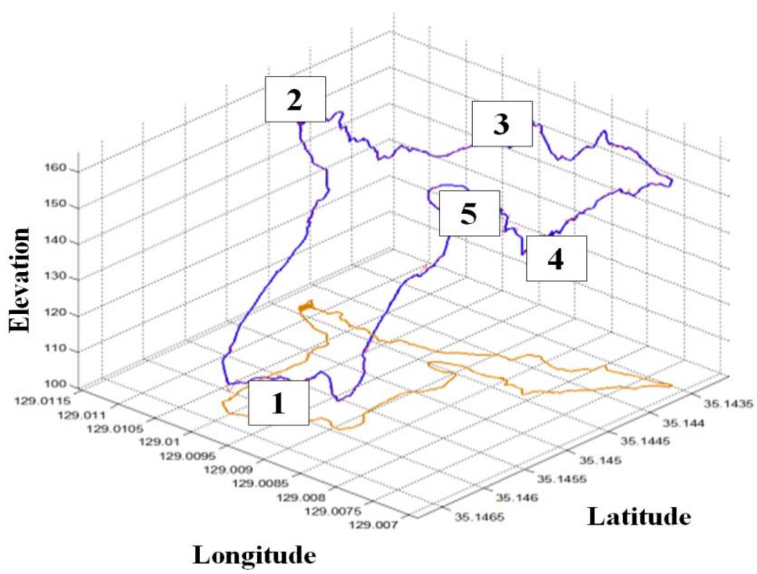

Figure 13. The 3D Plotting of the Walking Path to Provide Better Understanding on the Landscape View

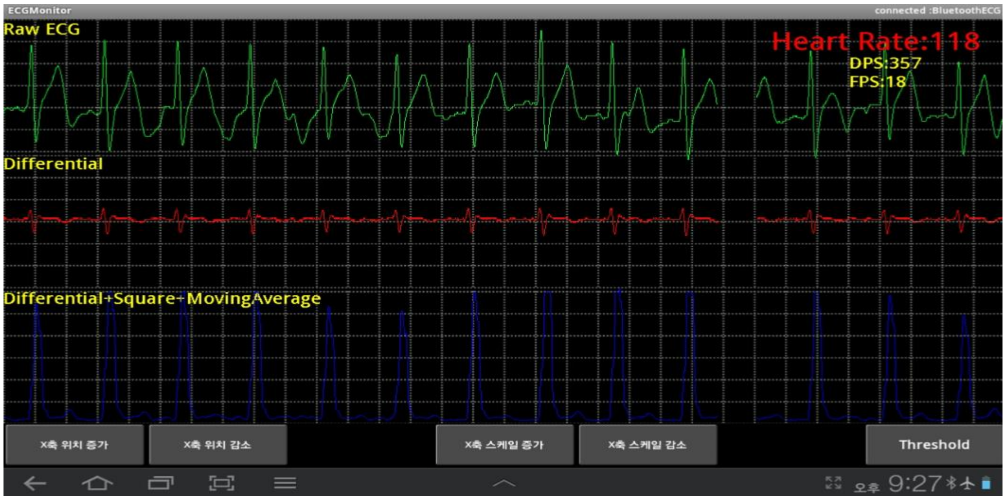

Figure 14. (a) Snapshoot of the Android ECG Monitoring Program while Author is Performing a 4 m Circular Walk on Flat Ground (section (2))

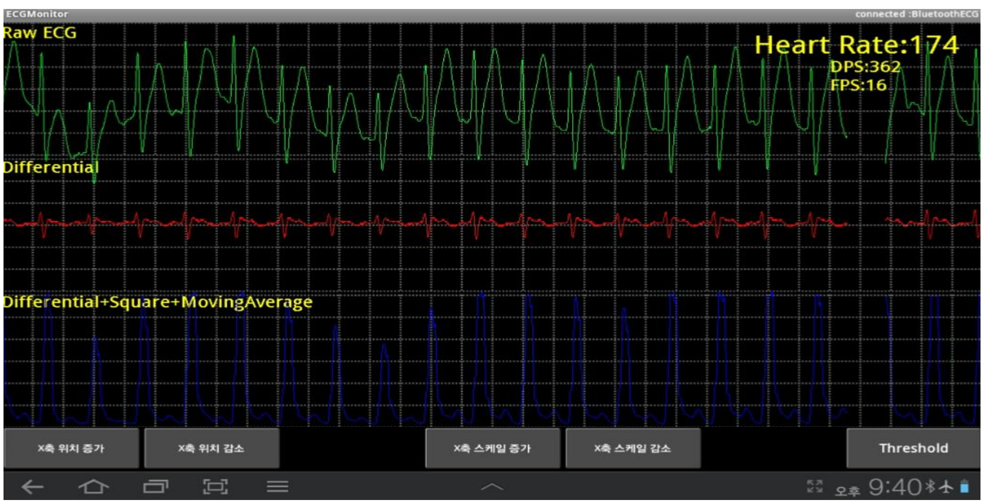

Figure 15. (a) Snapshoot of the Android ECG Monitoring Program while Author is Walking on Slope (section (1))

4 Conclusions

In this paper, several newtechniques for mobile ECG monitoring are presented and evaluated. A two-electrode wearable ECG measurement system is designed and implemented for mobile ECG monitoring.A very high isolation impedance at front-end is implemented in the two-electrode ECG system to isolate power line noise from the body. An optimum filter-set coefficient for motion artifact removal is proposed and implemented. At activity state 4 (6km/h) and state 5 (8km/h), average heartbeat detection has improvement of 71.8% to 97.4% after the deployment of optimum filter-set coefficient. A simple arrhythmia ECG detection algorithm using template matching method is suggested. From MIT/BIH database evaluation, the proposed algorithm achieves promising result of 97.4% detection rate for VPC arrhythmia detection. In the experiment of real time ECG monitoring, we demonstrate the capability of real time ECG display on a 10.1 inch Galaxy Tab for monitoring purpose. An estimated heart rate is recorded in Table 3 based on different activities. The rough result presented in Table 3 does not mean to prove any physics or medical hypothesis. The result presented in Table 3 is to show the capability of real time ECG monitoring using our mobile ECG monitoring system.

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