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A Wearable Healthcare System for Cardiac Signal Monitoring Using Conductive Textile Electrodes*

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***Abstract*— Accurate cardiac signal monitoring feasible for long-term monitoring is important for a practical, cost-effective health monitoring system. In this study, we propose a wearable healthcare system based on conductive fabric-based electrodes allowing monitoring of electrocardiogram (ECG) waveforms and demonstrated the potential for arrhythmia detection using the system. The measurement system uses conductive fabric-based electrodes arranged in a modified bipolar electrode configuration on the chest area of the patient. An adaptive impulse correlation filter (AICF) algorithm and a band pass filter to enable accurate R-peak detection in noisy environments.**

I. Introduction

As of recent years, a shift of emphasis from hospital-centered to more individual-centered, e-health technology development has been observed. Following this trend, unconscious, unconstrained, continuous monitoring of vital signals for patient health has become an important area of research. One of the most basic parameters used to diagnose and manage the health of an individual is the electrocardiogram (ECG). The ECG has the advantage of being relatively convenient to obtain while still providing valuable information about the current state of the patient such that it may also be used as an efficient method for continuous remote monitoring of patients with chronic diseases [1][2].

Previous work introduced remote monitoring systems of ECG which required the user to attach devices which can potentially be uncomfortable and inconvenient when used for prolonged periods. Wearable systems using conductive textile materials have gained interest as a solution for vital signal monitoring as user discomfort is reduced by integrating monitoring devices into a single piece of clothing [3].

Traditional methods of using gel-type Ag/AgCl electrodes have the inherent need for constant replacement due to drying and the possibility of irritation to the gel solution. Fabric electrodes overcome this disadvantage making them suitable for wearable monitoring systems.

However, due to characteristics of the materials, fabric-based electrodes are more susceptible to baseline wander noise from respiratory movements and other motion artifacts. Compared to traditional methods, motion artifacts are incorporated into the measurement at a much larger scale and can cause large distortions in the signal making diagnosis inaccurate or near impossible [4].

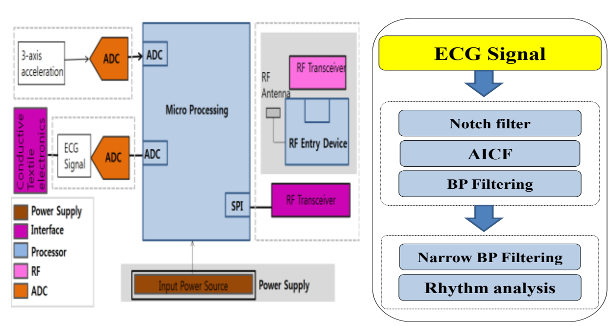
Therefore, in this paper we propose a conductive fabric-based wearable cardiac signal monitoring system which combines output signals from a 3-axis accelerometer to reduce baseline wander and other noise artifacts. The signal from the accelerometer is used as a reference signal to an adaptive impulse correlation filter (AICF) algorithm in combination with a band pass (BP) filter for more accurate R peak detection.

The rest of the paper is organized as follows. First we discuss the organization of the measurement system, followed by a discussion of the procedure for signal acquisition and processing. The performance of the system is evaluated and then the paper is concluded with a summary of the results.

II. Materials and Methods

A. ECG Measurement System Organization

The ECG measurement system was designed to meet constraints generally needed for wearable mobile monitoring systems. The system is battery-operated and uses the PIC16F microprocessor as the core component. The ultra-low power nRF24AP2 ANT IC from Nordic Semiconductors is used for communication and data transmission. The ADXL345 3-axis accelerometer from Analog Devices is used to obtain a reference signal for the subsequent signal processing. Fig. 1 shows the block diagram of the implemented system including the ECG signal acquisition bock and data transmission block (left) and the digital processing stage (right).



(a) (b)

Figure 1. Block diagram of the overall system. The figure shows the organization of ECG signal acquisition block and data transmission block (a) and the digital processing stage (b).

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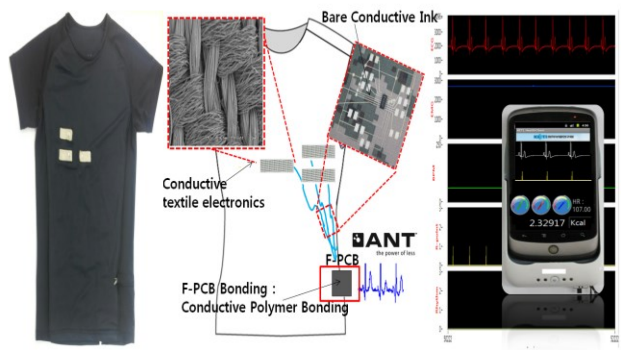
Biological signals have relatively small amplitudes compared to the offset of measurement electrodes, especially when using fabric electrodes, limiting gain in the measured signal. Therefore, an AC coupling circuit was used in the input stage to achieve higher gain and a high CMRR to improve overall output signal quality. The AC coupling circuit and data lines are realized using the conductive fabric.

The analog ECG circuit was designed with an amplification stage with gain of 37dB, and a BP filter with cutoff frequencies at 0.5Hz and 500Hz. The signal was sampled with a sampling frequency of 200Hz. The acquired analog signal is quantified using the A/D conversion unit in the microprocessor. The digital filter and an algorithm for R-peak detection and arrhythmia detection are implemented with the PIC16F microcontroller. To enable continuous long-term monitoring as with Holter monitoring, the nRF24AP2 wireless chipset (Nordic semiconductor), which implements the ANT communications protocol for low-power, short-range, multi-point wireless communication, was used.



(a) (b) (c)

Figure 2. Organization of proposed wearable healthcare system: (a) Exterior of wearable sensor shirt (b) Organization within shirt (c) Montoring system based on x86 and Android OS platform

B. Placement of Fabric Electrodes and acquisition of ECG Signals from Wearable System

Fig. 2 shows the diagram of the wearable system and the overall organization of the system. Initial placement of the conductive fabric-electrode is placed on the chest in a bi-polar electrode configuration based on the measured potential difference. Placement in a modified bipolar configuration was determined by comparing the signal obtained from the placing electrodes at standard locations and the signal obtained from Ag/AgCl electrodes placed in the standard limb lead configuration. From the results, the chest lead V4 showed the highest correlation in acquired ECG signal. Table 1 shows the calculated Pearson Correlation coefficients for each of the signals acquired at various lead locations.

TABLE I. PEARSON CORRELATION COEFFICIENTS FOR SIGNAL FROM STANDARD LIMB LEAD CONFIGURATION AND PROPOSED SYSTEM

Lead	Pearson Correlation(R)							Mean±SD
V1	0.51	0.60	0.57	0.48	0.50	0.54	0.53	0.525 ±0.041
V2	0.61	0.73	0.55	0.65	0.73	0.78	0.64	0.659 ±0.074
V3	0.95	0.93	0.92	0.80	0.76	0.85	0.88	0.873 ±0.072
V4	0.92	0.82	0.94	0.92	0.95	0.88	0.89	0.903 ±0.036
V5	0.68	0.69	0.65	0.83	0.78	0.75	0.70	0.709 ±0.058
V6	0.43	0.63	0.45	0.64	0.50	0.52	0.52	0.516 ±0.078

It well known that compared to the signal obtained from a 12-lead ECG used in practice, reducing the number of electrodes has drawbacks such as a reduced dynamic range in the obtained signal and the inability to measure ventricular and arterial signals simultaneously. In the case of the fabric electrodes, the signal obtained achieves similar performance to the lead 2 configuration of the standard limb lead. In order to improve the quality of the signal an AICF algorithm and subsequent BP filter is employed.

C. Adaptive Impulse Correlation Filter

To remove the noise signal in the ECG signal an AICF is used to estimate the deterministic component of the signal. The AICF has been shown to have improved performance in removing the noise component of the signal compared to the traditionally used ensemble averaging technique [5].

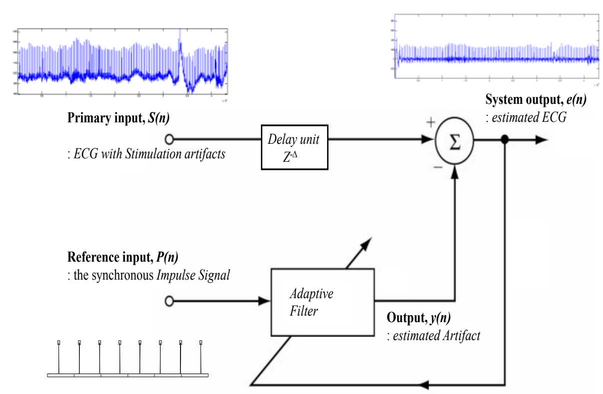


Figure 3. Block diagram of adaptive impulse correlation filter. The impulse signal is derived from the samples of the 3-axis accelerometer

AICF is similar to other adaptive noise filter methods with the difference in the reference signal used as input. General adaptive noise filters use a primary signal which includes the deterministic signal component and use a reference signal with a strong correlation with the deterministic signal. However, with the AICF, the reference signal is an impulse signal in synchronization with the activation of the deterministic signal. As shown in Fig. 3, in order to reduce the noise contained in the ECG signal, a synchronized impulse signal sampled from the signal magnitude area (SMA) of the accelerometer output is used to estimate noise and reduce artifacts due to respiration and other body movements. A normalized least mean squares (NLMS) algorithm was used to update the weight vector of the adaptive filter. Compared to least mean squares (LMS) methods, the NLMS algorithm has less computational requirements and had the advantage of being less affected by the change in power of the signal. Fig. 4(a) shows the raw ECG signal, (b) the synchronized impulse signal, and (c) the filtered signal with baseline removed.

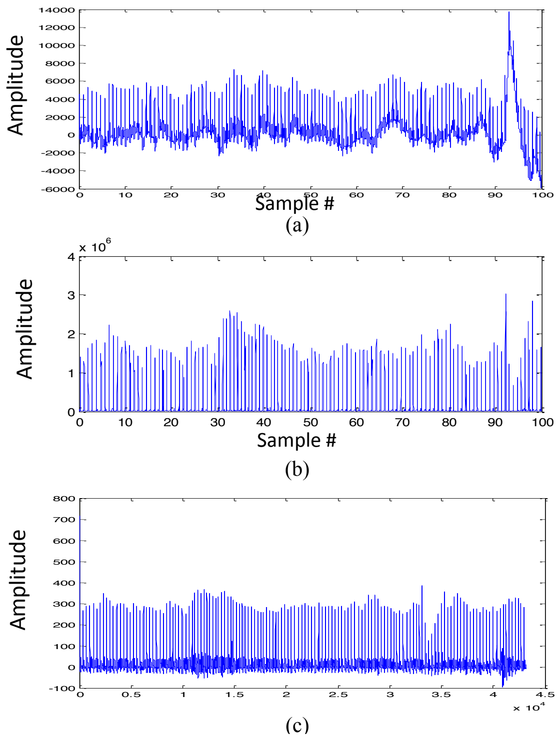


Figure 4. Baseline removal with AICF. (a) Raw ECG signal (b) the synchronized impulse signal derived from the accelerometer (c) the entire filtered output of the AICF

D. *Notch filter for 60Hz Noise and Band Pass Filter*

The obtained signal is passed through an additional notch filter to reduce 60Hz power-line noises. Fig. 5 shows the block diagram of the notch filter. Fig. 6 shows the original unfiltered signal and the signal after passing through the notch filter. The figure depicts the signal with baseline wander minimized after applying AICF. The AICF is known to cause distortion in the ST segment of the ECG waveform and therefore is not often used in practice. To complement the algorithm, a BP filter is applied additionally before R-peak detection.

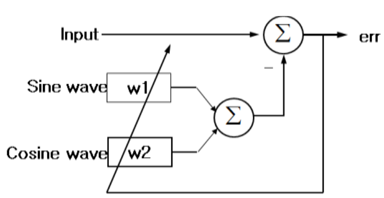


Figure 5. Block diagram of 60Hz Notch filter

E. *R-peak Detection Algorithm*

Accurate R-peak detection is needed for to diagnose the state of the patient. After removal of baseline wander, the signal is passed through an additional BP filter with a pass band of 0.5Hz to 50Hz. The signal is then passed through a narrow-band pass filter of degree 20 and a 50 point average smoothing filter. The R-peaks are detected using a threshold method. Fig. 7 depicts R peaks detected using the algorithm

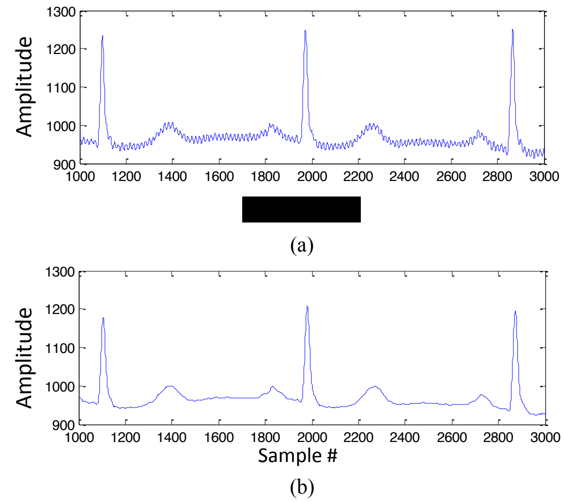


Figure 6. Results of the ECG signal (a) before filtering with the notch filter and (b) after the notch filter.

Fig. 7 shows that after filtering with the above procedures R-peaks can be accurately detected even in the event of noise from sudden movement.

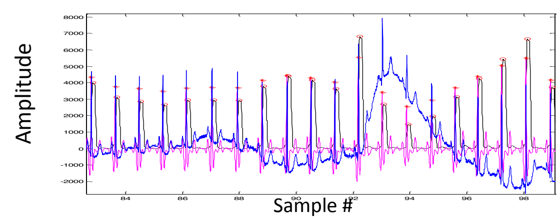


Figure 7. R-peak detection after signal processing with the proposed algorithm. The original signal is depicted in blue and processed signal in depicted in pink. The detected R-peaks are indicated with red crosshairs. Waveform is from record 105 of MIT-BIH Arrhythmia database.

F. *Arrhythmia Detection with RR Intervals*

By calculating RR intervals from the detected R-peaks, it was demonstrated that normal SR rhythms and VT rhythms could be classified by using a threshold of 500 milliseconds. Regions with RR intervals shorter than 500 milliseconds were classified are arrhythmia regions

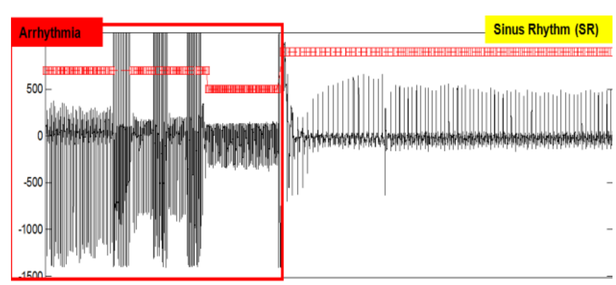


Figure 8. Detection of intervals containing arrhythmia based on RR intervals caculated from R-peaks detected with proposed algorithm. The detected arrhythmis region is indicated with a red box.

III. RESULTS AND DISCUSSION

The MIT-BIH Arrhythmia Database was used to evaluate R-peak detection performance of the system. For each of the records, the R peaks were detected using the proposed algorithm and sensitivity (Se), detection error rate (De), and overall accuracy (OA) was calculated. A false positive(FP) occurred when no R peak existed but a peak was detected due to noise or other reasons and a false negative(FN) occurred when an R peak existed, but was not detected. A total of 61088 ECG beats were extracted from 28 records existing in the database. Of those beats, 61028 beats were accurately detected with 68 FP and 58 FN. The results are summarized in Table 2.

TABLE II. RESULTS FOR R PEAK DETECTION WITH MIT-BIH ARRHYTHMIA DATABASE

Tape #	ECG peak detection						
	Total beats	***FP***	***FN***	***TP***	***De (%)***	***Se (%)***	***OA (%)***
100	2273	0	0	2273	0.00	100.00	100.00
101	1865	1	2	1865	0.00	99.89	99.95
102	2187	0	0	2187	0.00	100.00	100.00
103	2084	0	0	2084	0.00	100.00	100.00
104	2229	1	5	2228	0.00	99.78	99.96
105	2572	8	24	2564	0.01	99.07	99.69
106	2027	0	9	2027	0.00	99.56	100.00
107	2136	0	1	2136	0.00	99.95	100.00
108	1763	1	5	1762	0.00	99.72	99.94
109	2532	0	0	2532	0.00	100.00	100.00
111	2124	0	1	2124	0.00	99.95	100.00
112	2539	0	0	2539	0.00	100.00	100.00
113	1795	0	0	1795	0.00	100.00	100.00
114	1879	11	4	1868	0.01	99.79	99.41
115	1953	0	0	1953	0.00	100.00	100.00
116	2412	1	0	2411	0.00	100.00	99.96
117	1535	0	1	1535	0.00	99.93	100.00
118	2278	0	0	2278	0.00	100.00	100.00
119	1987	3	0	1984	0.00	100.00	99.85
121	1863	0	0	1863	0.00	100.00	100.00
122	2476	2	0	2474	0.00	100.00	99.92
123	1518	1	0	1517	0.00	100.00	99.93
124	1619	2	0	1619	0.00	100.00	99.88
202	2136	2	0	2134	0.00	100.00	99.91
205	2656	9	0	2647	0.00	100.00	99.66
208	2995	11	4	2984	0.01	99.87	99.63
209	3005	1	0	3004	0.00	100.00	99.97
210	2273	14	2	2641	0.01	99.92	99.47
Total	61088	68	58	61028	0.00	99.91	99.89

De(%)= FP+FN/TOTAL BEATS*100
Se(%) = TP/TP+FN*100
OA(%)= TP/TP+FP*100

In the case of record 105, Fig. 7 shows that even with baseline wander and other noise artifacts the R peaks are accurately detected. Even when the R peaks were inverted as in the case of record 210, the proposed system was able to detect R peaks as shown in Fig. 9.

In general, the proposed system showed very good performance in R peak detection. However, in the case of record 205 the number of FP and FN is greater compared to other records. This record is considered an exception as the signal integrity has been compromised to an extent that peak detection accuracy is difficult with previously verified algorithms as well as the proposed algorithm of this study. In addition, the system showed reasonable performance in R-peak detection even in the presence of arrhythmia and other noise artifacts.

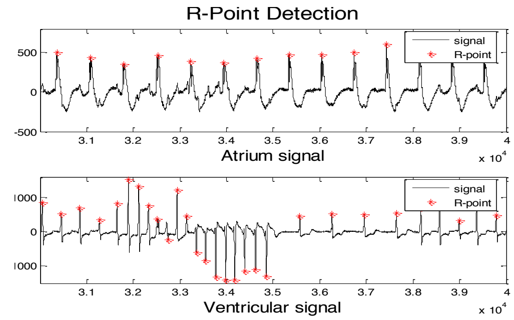


Figure 9. R-peak detection from record 210. The results show accurate R-peak detection even in the presence of inverted signals and arrhythmia.

The possibility of accurate RR interval-based arrhythmia detection using R-peaks detected by the system can be suggested by the results. Previous work have detected arrhythmia using similar methods, however these methods have had limitations that made them suboptimal for the current application. The performance of the system is promising and is expected to show better results in arrhythmia detection than previous works.

IV. CONCLUSION

In this paper a wearable system ECG measurement system was implemented using conductive fabric electrodes and an ECG signal with high correlation (R=0.903) to a lead 2 configuration of the standard limb lead using Ag/AgCl electrodes was successfully obtained. From the measured signal, our proposed signal processing algorithm using an AICF algorithm, notch filter and BP filter was used to reduce noise artifacts and detect characteristic R-peaks. The performance of the system was evaluated with recordings from the MIT-BIH database and showed very good overall accuracy of 99.89%. The relatively low computational requirements and low power requirements make it suitable for implementation in a real-time monitoring system. Useful investigations in the future will be for applications in arrhythmia detection and other health management based on other types of signal monitoring.

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