CSE 474 Winter 2018 Final Project Report

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Abstract

Electrocardiography (ECG) is an essential technique for monitoring heart health and detecting common cardiac disorders. However, ECG is usually performed using expensive equipment in hospitals and clinics. To enable more ubiquitous individual heart monitoring, we present a prototype of an inexpensive and portable ECG monitor. By empowering users to record, visualize, and interpret their own cardiac signatures, this technology can provide users with potentially critical heart health diagnostics [Saxon 2013]. Our design records the user's ECG with the user's fingers placed on a pair of electrodes. The ECG is previewed in real time on a touchscreen, with the P wave and QRS complex annotated. The device can also connect over bluetooth to a smart phone to display and record the heart rate. After the user's ECG is completed, our device provides the user a cardio summary with information about various arrhythmias that may have been detected.

1 Introduction

Electrocardiography (ECG) is a technique used to measure the electrical impulses in the heart by recording a voltage potential difference between two electrodes placed on the surface of the body. In a typical ECG trace (see Fig. 1), each heart beat exhibits the following common features:

1. **P Wave** caused by atrial depolarization
2. **QRS Complex** caused by ventricle depolarization
3. **T Wave** caused by ventricle repolarization

Timing information extracted from an ECG can provide insight into the behavior and health of the heart. The time between subsequent R waves is called the R-R interval (RRI). The heart rate (HR) is the inverse of the RRI, typically reported in units of beats per minute (bpm). An abnormal or uneven heart rate is called arrhythmia. In particular, bradycardia and tachycardia are two common arrhythmias characterized by heart rates that are less than 60 bpm or higher than 100 bpm respectively [Kusumoto and Bernath 2011]. The time between the start of the P wave and the start of the QRS complex is called the P-R interval (PRI). An PRI greater than 0.2 seconds can indicate heart block [Doraiswamy 2009]. Finally, a total QRS complex duration (QRS) greater than 0.12 seconds is an indicator of premature ventricular contraction (PVC) [Patel and Cheeran 2012].

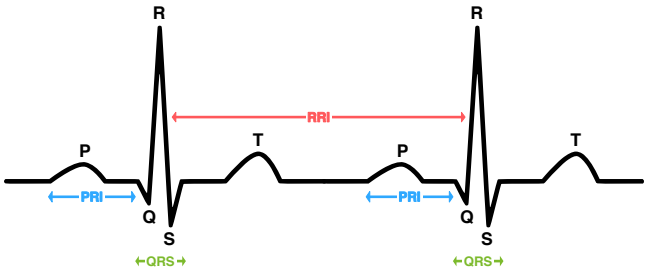

Figure 1: *A typical ECG trace annotated with P-R interval (PRI), QSR complex (QSR), and R-R interval (RRI).*

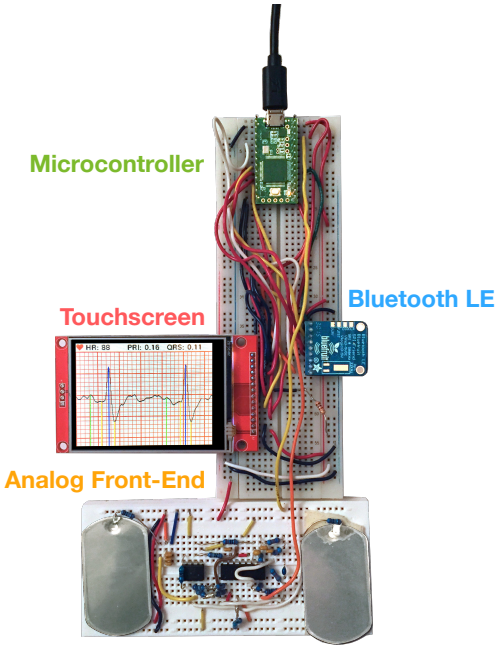

Figure 2: *Photo of our ECG device*

2 System Overview

2.1 Hardware Modules

Our system is composed of four separate hardware modules (see Figures 2 and 3).

2.1.1 Microcontroller Module

The microcontroller module is a PJRC Teensy 3.2 development board, which has a MK20DX256VLH7 Cortex-M4 microprocessor with 256 kB of flash memory and a built-in analog-to-digital converter (ADC). This module provides power to the other modules, performs all of the system's digital signal processing, and runs the main application.

2.1.2 Analog Front-End Module

The analog front-end module is an all-analog circuit that filters and amplifies the differential input from our two electrodes using an instrumental amplifier and multiple filters. This module performs analog scaling and pre-filtering of the signal, the output of which is sampled by the microcontroller via the ADC.

2.1.3 Touchscreen Module

The touchscreen module is a PJRC ILI9341 320px×240px full color thin-film-transistor liquid-crystal display (TFT LCD) featuring an XPT2046 touchscreen controller. It provides the user display of the main application, showing an annotated ECG trace during standard use. This module communicates with the microcontroller using the Serial Peripheral Interface (SPI) protocol to update the LCD and detect touch events.

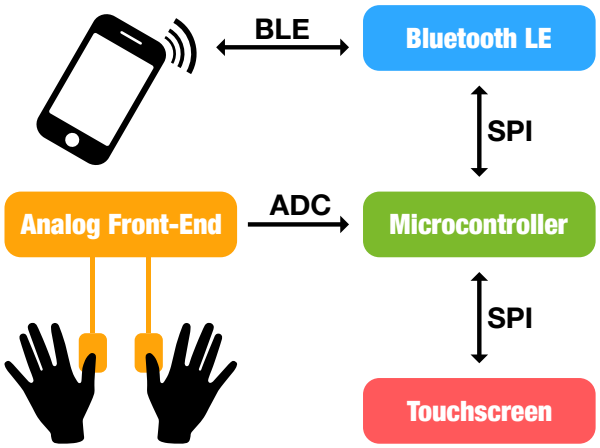

Figure 3: *Block diagram of module interactions.*

2.1.4 Bluetooth LE Module

The Bluetooth LE module is an Adafruit Bluefruit LE peripheral featuring Nordic's nRF51822 Bluetooth Low Energy (BLE) transceiver. It uses the GATT BLE standard to advertise to phones and other devices measured heart rate, making our ECG compatible with currently available heart rate monitor smartphone applications. It communicates with the microcontroller with the SPI interface.

2.2 Hardware Theory of Operation

The analog front-end module measures and filters the subject's ECG from the electrodes. It is comprised of four major components (see Figure 4).

1. Instrumentation Amplifier
2. Virtual Ground Circuit
3. Band Pass filter
4. 60 Hz Notch filter

2.2.1 Instrumentation Amplifier

The instrumentation amplifier (INA) circuit uses a Texas Instruments INA128U chip to provide an amplified differential signal of the voltage from each of our electrodes. In addition, it has a high common mode rejection, meaning that it eliminates a significant amount of interference produced by AC power line noise from output. Due to the fact that the heart is offset in the body, electrical pulses from the heart reach one finger before the other, meaning the differential signal provides a good measurement of the electrical activity of the heart.

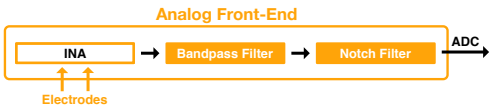

Figure 4: *Block diagram of analog signal processing*

2.2.2 Virtual Ground Circuit

To compensate for not using a third electrode to ground the subject and ensure that all measurements are made to a common baseline, we use a Virtual Ground circuit to produce a voltage of 1.65 V with respect to our circuit's actual ground. We use this voltage as a reference for our INA and various filters so that the differential signal ranges between 0 V and 3.3 V when compared to our original ground reference, instead of -1.65 V and 1.65 V.

2.2.3 Band Pass filter

To further eliminate noise from our signal, we use a band pass filter to eliminate any frequencies less than 1 Hz or greater than 40 Hz. For the purposes of our ECG, we are studying heart signals of hearts that normally have beats per minute (BPM) of around 60 to 100 BPM, which corresponds to 1 Hz to 1.7 Hz frequencies. Eliminating frequencies outside of this range helps make clean measurements of our subject's ECG.

2.2.4 60 Hz Notch filter

In a further attempt to reduce common sources of noise, we have a final filter that we pass our measured signal through that eliminates 60 Hz frequencies, which reduces noise received from AC power lines in the environment.

2.3 Digital Signal Processing Theory of Operation

Input to the microcontroller via the ADC is stored in an ADC buffer $x[t]$ at a sampling frequency of 250Hz. These values are then fed through a second order Butterworth IIR high-pass filter $h[t]$ with a -3dB cutoff at 1 Hz, a second order Butterworth low-pass IIR filter $l_1[t]$ with a -3dB cutoff at 20Hz, and then again through an identical low-pass filter $l_2[t]$:

$$\begin{aligned} h[t] &= (0.982)x[t] - 2x[t-1] + x[t-2] \\ &\quad + (1.964)h[t-1] - (0.965)h[t-2] \end{aligned}$$

$$\begin{aligned} l_1[t] &= (0.046)h[t] + 2h[t-1] + h[0] \\ &\quad + (1.307)l_1[t-1] - (0.492)l_1[t-2] \end{aligned}$$

$$\begin{aligned} l_2[t] &= (0.046)l_1[t] + 2l_1[t-1] + l_1[0] \\ &\quad + (1.307)l_2[t-1] - (0.492)l_2[t-2] \end{aligned}$$

The filtered values are stored in a filter buffer. A pulse filter reads from the filter buffer and outputs a signal that accentuates QRS pulses. The pulse filter output $p[t]$ is generated from filter buffer data $l_2[t]$ by the following relation:

$$p[t] = \left(\sum_{i=t}^{t-4} l_2[i]^2 - \min_{i=t}^{t-4} (l_2[i]^2) \right)^2$$

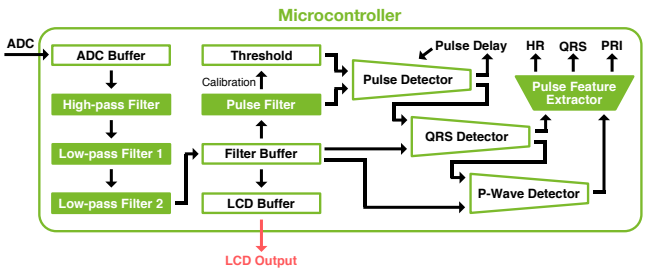

Figure 5: *Block diagram of digital signal processing.*

Pulses are detected by the pulse detector with input from the pulse filter and a threshold determined at calibration time; the threshold is set at 0.5 standard deviations above the mean of the filtered data during the 3 second calibration period.

2.4 Feature Extraction Theory of Operation

After the pulse detector recognizes that a pulse has occurred, the QRS detector determines the exact times of the Q, R, and S points ($T_Q[n]$, $T_R[n]$, and $T_S[n]$). To prevent repeated detection of a single pulse or false detection of a following T wave, the pulse detector is inhibited by a pulse delay of 0.4s (see Figure 5). The output of the P wave detector determines the P wave time $T_P[n]$ as the local maximum of the filtered signal between $T_R[n] - 0.18$s and $T_R[n] - 0.06$s. The instantaneous HR, PRI, and QRS are then calculated by the pulse feature extractor as follows:

$$\text{HR}[n] = \frac{60.0\text{s}}{T_R[n] - T_R[n-1]}$$

$$\text{PRI}[n] = T_R[n] - T_P[n]$$

$$\text{QRS}[n] = T_S[n] - T_Q[n]$$

These values are stored in separate feature buffers, and are averaged over the most recent 5 measurements values before displaying to the user (see Figure 5).

3 Future Work

To bring this ECG monitor design to a consumer market, the electronic components should be produced on a printed circuit board (PCB) and enclosed in a protective encasement. To improve portability, it should be powered by an internal rechargeable battery. In order to allow users to track their heart's health over time, the ECG monitor could store a record of cardiac traces on internal flash memory. Alternatively, a dedicated smartphone application may stream ECG metrics or complete traces to clinicians to improve diagnosis.

In addition to screening for tachycardia, bradycardia, heart block, and PVC, future work could incorporate prediction of premature atrial contraction (PAC), atrial fibrilation (AFib), and other heart maladies diagnosable from an ECG trace.

As a motivating example of related work, the AliveCor KardiaMobile is an FDA-approved mobile ECG monitor (Figure 6), which can detect arrhythmias [Richley and Graham 2015] and heart palpitations [Newham and Tayebjee 2017]. It consists of a compact electrode pair with embedded analog filtering and signal amplification, which communicates with a smartphone through frequency modulated ultrasound audio.

4 Conclusion

Our design demonstrates the feasibility of a portable, interactive ECG monitor that is compatible with preexisting heart rate monitor software. This technology is reasonably affordable (see Appendix A) and has high potential for mobile health applications.

The system records a high-resolution digital ECG trace and displays it in real time with each P wave and QRS complex annotated. Using timing information from these extracted features our design displays the user's average heart rate, P-R interval, and QRS complex duration. Cardiac abnormalities are displayed to the user in a medical summary of possible heart maladies. Heart rate information can also be transmitted via Bluetooth to existing health and fitness smartphone applications.

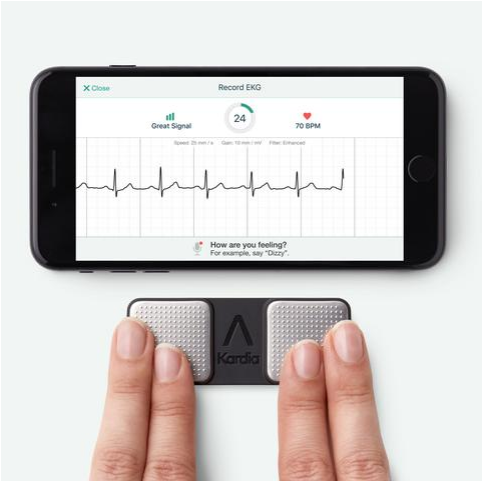

Figure 6: *AliveCor KardiaMobile portable ECG monitor and smartphone application (image courtesy of* `https://store.alivecor.com/products/kardiamobile`*).*

The most valuable aspects of our design are the user-friendly touchscreen interface and the consistent feature extraction. The analog and digital signal filtering result in a signal with minimal external noise. In addition to observing their ECG in real time, users can review a 30 second trace with intuitive touch scrolling after measurement. In addition, due to the design of the pulse filter and calibration algorithm, we observed that our feature extraction algorithm correctly identifies P waves and QRS complexes on a variety of individuals with different ECG morphologies, including those with unusually weak R waves and strong T waves. Consistently identifying heart trace features is critical for a successful ECG monitor.

During the design process, we were faced with a trade-off between ensuring the detection of every R wave and preventing false detections triggered by accentuated T waves, which we found in some individuals to have amplitudes comparable to or greater than R waves. Due to the importance of reliably detecting R waves in order to provide accurate timing metrics, and the extreme diversity of individual T wave morphologies, we chose not to detect T waves in our design. T wave timing information is unnecessary in our design, since it is unnecessary for detecting arrythmias, heart block, and PVC. However, it is plausible to extend this technology to identify T waves in addition to the P waves and QRS complexes, enabling detection of heart conditions associated with T wave abnormalities.

Overall, our design demonstrates potential to help individuals and their health providers diagnose possible heart abnormalities before they manifest as serious medical problems.

Appendix A: Component Costs

Below is an estimate of design's hardware component costs.

Table 1: *Approximate hardware component cost breakdown.*

Component	Cost
Teensy 3.2 Development Board	$19.80
ILI9341 LCD Touchscreen	$15.00
Adafruit Bluefruit LE SPI	$17.50
Analog Circuit Components	$7.00
TOTAL	$59.30

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