Wireless Wearable Electrocardiogram

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***Abstract* — The Internet of Things has become increasingly popular as the average person has introduced more smart devices into their lives. The need for users to monitor their daily activity has been addressed by implementing biosensors into wearables to inform users of their daily activity. Our project addresses these needs. The Wireless Wearable Electrocardiogram reads the user's cardiac activity using arm mounted sensors. It sends the user data wirelessly over to a Hub that externally processes it and delivers it to an Android application for the user to monitor.**

***Index Terms* — Electrocardiography, Cardiology, Health information management, Wireless application protocol, Body sensor networks**

I. INTRODUCTION

Due to the heart's muscles depolarizing during a heartbeat we are able to measure changes in cardiac activity from the skin by using medical grade electrodes. While conventional electrocardiograms (ECGs) implement twelve leads the Wireless Wearable ECG (WWECG) implements a two lead system. This is the most practical implementation currently to ensure synchronization between sampling at both sensor locations. The WWECG is comprised of two armband sensors and an external processing Hub. Each sensor uses two electrodes to pick up the potential difference from each arm. This small signal voltage is conditioned using active analog filters and then digitized to be sent to the Hub wirelessly. Additional conditioning is implemented as an instrumentation amplifier rids the signal of common-mode noise. A digital signal processor (DSP) then takes the signal and runs it through a series of algorithms to adjust signal shape, detect the QRS complex, calculate heart rate, and detect ventricular arrhythmia. This data becomes accessible to the user through the use of an Android developed application that receives this information from the Hub using Bluetooth. The user will be able to monitor their cardiac activity in real time using the Android app. The app will store data to be accessed by the user at any time and will offer several modes of operation including the display of the ECG signal as well as calculated user heart rate.

II. SYSTEM COMPONENTS

The WWECG is comprised of three sub-systems that interact with each other. Two of the sub-systems are the arm mounted sensors, the third is the Hub.

A. Sensors

The functions of the sensors are split in between three printed circuit boards (PCBs). The PCB modules are split into the power board, the analog board, and the digital board. The power board provides power to the analog and digital boards using a detachable 3.7 volt lithium ion battery and DC-DC converters. It contains a Mini-USB port and a charge circuit that allows the battery to be recharged at 100 milliamps to prevent over-heating. A high efficiency switching converter steps down the battery voltage to 2.5 volts with a maximum output of 300 milliamps. A voltage converter takes the 2.5 volts and inverts it to -2.5 volts with a maximum output of 200 milliamps. The power board features a single-pole double-throw On/Off switch and a green indicator LED. Four header pins (2.5V, -2.5V, GND, DGND) provide power from the board to the analog and digital boards. The total current draw of the sensor at 3.7 volts is ~50 milliamps leaving power consumption at 185 milliwatts. This low power consumption keeps the sensor cool to the touch thus maximizing comfort for the user.

The analog board implements active filters to limit the bandwidth between .5 and 100 Hertz to duplicate the bandwidth used in diagnostic ECGs. 4^{th} order Sallen-Key low-pass and high-pass filters limit the upper and lower bounds of the bandwidth at an 80 decibel per decade attenuation rate. A 60 Hertz notch filter is implemented using a Tow-Thomas Biquad topology with a summing amplifier to reduce noise from utilities. Header pins allow leads to connect the electrodes to the board and for the signal output to be delivered to the digital board.

The digital board uses the ADS1114, a 16 bit delta-sigma analog to digital converter, to sample the signal and feeds it into a MSP430G2553 microcontroller via I^2C. The data is then processed by attaching a time stamp and formatted to fit within the twenty byte packet limit of Bluetooth Low-Energy.

After the processing has been completed a fourteen byte packet is created and then transmitted to an HM-11 Bluetooth module via UART where it is sent to another HM-11 master module. Also on the sensor board is an orange LED, which is located near the microcontroller, the LED will be held on whenever data is currently being transmitted and will be toggling on and off whenever the device is waiting for connection. Figure 1. Shows the overall block diagram of the sensor system.

B. Hub

The main purpose for the hub within the system is to complete processing of the Electrocardiographic signal and handle all of the inter device communication. Utilizing an Arduino Mega2560 as the central microcontroller which contains an ATMega 2560, this device will be responsible for all of the inter device communication on the hub as well as any pre or post processing that needs to be done, with the exception of the Digital Signal Processor. As the master for both sensors, the hub hosts three separate HM-11 Bluetooth modules in order to have a consistent synchronous output to the other devices within the system. With two Bluetooth modules set to master mode, these two are responsible for receiving the transmitted data from the each sensor and storing it with a buffer. While the buffer is being filled, the data is then synchronously outputted to two separate Digital-to-Analog Converters synchronously using I^2C. The MCP4725 is a 10bit DAC which will be used as inputs for the instrumentation amplifier. Once the data has been synchronously outputted to the instrumentation amplifier the last portion of analog signal conditioning will take place. The instrumentation amplifier serves to aid in the further eliminating common-mode noise obtained from the body. Texas Instruments' INA333 instrumentation amplifier was chosen to be used due to its low power consumption, approx. 50µA, and its high Common-Mode Rejection Ratio (CMRR) which is a minimum of 100dB. After the instrumentation amplifier completes its common mode noise removal the signal is then passed onto the Digital Signal Processor for final processing and algorithm implementation. Due to financial and practical constraints in PCB manufacturing we chose to use the TMS320C5535 eZdsp development kit for all of

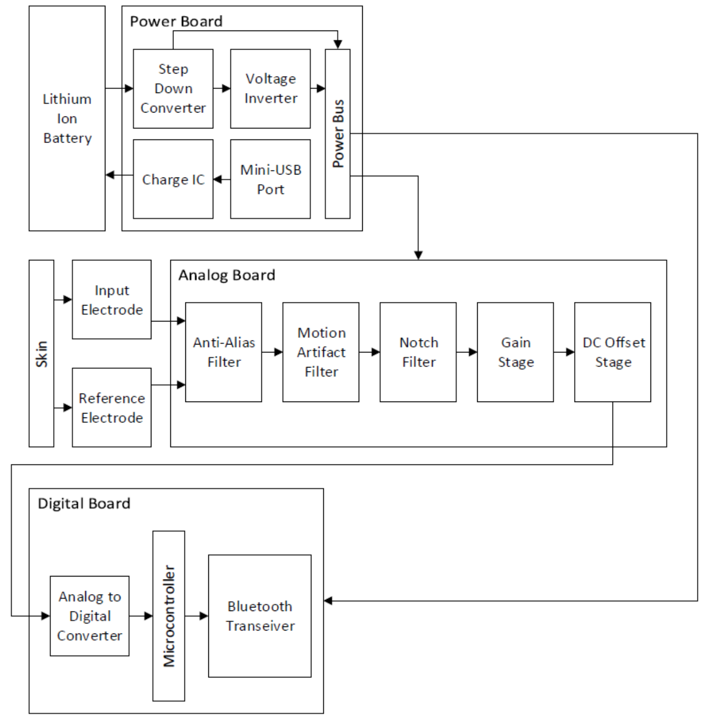


Figure 1: Sensor Block Diagram

our digital signal processing. This DSP completed the final stages of filtering of our signal to be clean enough for display on the android application once transmitted. Additionally, the DSP implements algorithms for the detection of the heart's QRS complex, calculation of the user's heart rate, and the detection of any Ventricular Arrhythmia. Using UART for the data transmission back to the main hub, the output signal, heart rate, and Arrhythmia flag is then formatted and sent via Bluetooth to the Android device.

The Hub, Microcontroller, and DSP are powered by a separate PCB. The Hub power board uses a Mini-USB port in conjunction with a Mini-USB to USB A cable to run off any 5 volt USB port for the convenience of portability. It provides USB A and USB B ports to power the microcontroller and DSP. Pin headers allow jumper access to 2.5, -2.5, and 3.3 volt outputs to power the remaining components found on the Hub.

III. HARDWARE ANALYSIS

A. Digital Signal Processor

The Wireless Wearable Electrocardiogram's hub processing unit uses the TMS320C5535 eZdsp to complete the processing and implement our various vital detection algorithms. The C5535 processor is a High-Performance, Low-Power Fixed-Point C processor that offers flexibility by implementing the signal processing via software. Specifically, the DSP is programmed using Texas Instrument's Code Composer Studio IDE. The processor operates at 100_MHz, has Dual multiply-and Accumulate Units, two arithmetic logic units, and is software compatible

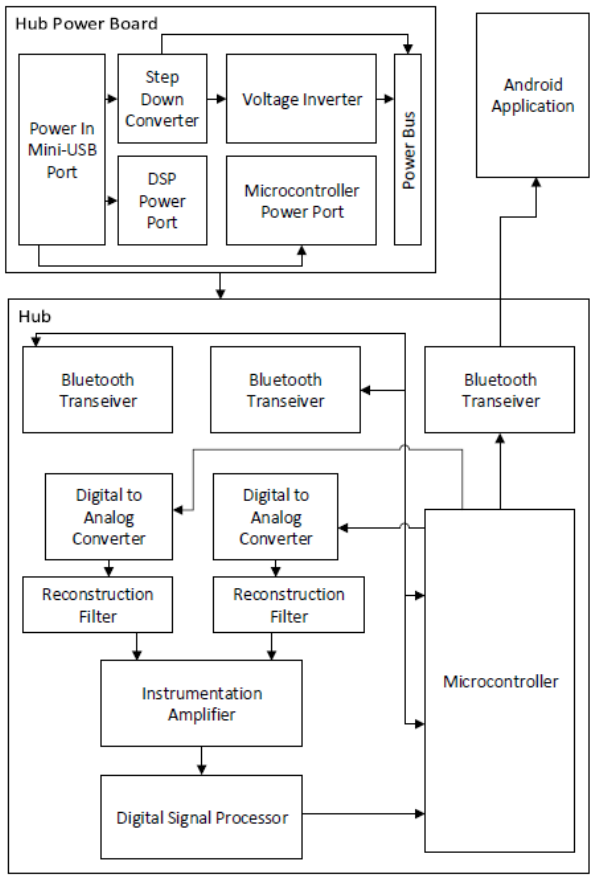


Figure 2: Hub Block Diagram

with C55x devices. Furthermore, the C5535 has a tightly coupled Fast Fourier Transform (FFT) Hardware Accelerator, this is specifically used in the arrhythmia detection algorithms. Another important peripheral is the Universal Asynchronous Receiver/Transmitter which is used to transmit and communicate between the ATMEGA2560 and the TMS320C5535. In addition, the TMS320C5535 has an on-board 10-Bit 4-Input Successive-Approximation (SAR) ADC that is used to convert the instrumentation amplifier's analog output to digital for post processing. Lastly, the TMS320C5535 eZdsp kit has a micro SD card slot where the firmware for the DSP is loaded and read from.

B. Microcontrollers

This overall design features two separate microcontroller units since the overall processing power from between the hub and the sensors differs so greatly. The first device is the MSP430G2553 which is the model commonly found on Texas instruments MSP430 Launchpad line. This device was originally selected due to previous experience along with a large source of previous applications and user resources. Operating at 16MHz the devices contains more than enough processing power within the chipset to handle both the analog to digital converter as well as all of the data transmission for the Bluetooth device. Also featuring an I^2C and UART, the device allows a simple straightforward interaction with the ADS1114 and HM-11 modules, respectively. This device will be programmed using Texas Instruments Energia application since its libraries provide functionally easy to read code along with many active community member contributing within their forums. The second microcontroller chosen was an ATMega 2560. This device provides full functionality with its four UARTs as well as both an I^2C allowing both synchronous and asynchronous communication.

C. Digital to Analog Converters

In order to transfer the signal from the digital hub's buffer input to the analog output for the instrumentation amplifier a digital to analog converter is needed. For this project, a MCP4725 was selected as it offers a bit resolution with a range from 0 to 5 volts. Because the device uses I^2C, two separate addresses can be set and outputted concurrently when sent from the ATMega2560. This synchronous output is necessary due to the nature of QRS wave construction.

IV. SIGNAL PROCESSING

A critical focus of this project was proper signal processing. Because the human body is susceptible to many forms of noise and our desired biopotential signal is so small proper analog and digital signal processing techniques were crucial. First, in the sensor portion of the system proper signal conditioning had to be taken into consideration to accurately acquire the signal for transmission. Therefore the Analog-Front-End consisted of a buffer stage that served to lower the biopotential signals' impedance which aids in lowering the common mode voltage of the signal by up to 40dB, specifically 60Hz noise seen due to impedance mismatching of electrodes. Next, a Tow-Thomas Biquad Notch filter tuned for 60Hz is implemented to further remove main's noise. This design was chosen due to the flexibility and ease in design that it offers. Once more 60Hz noise was removed the signal is passed into a 100Hz fourth order Anti-Alias filter that was designed using a Sallen-Key topology. This was to further ensure that the signal would not experience any distortion once sampled by the ADC due to aliasing. The last filter implemented was another Sallen Key topology design but this time for a high pass filter with a cutoff of .5Hz to assist with the removal of motion artifact noise. The last portion of signal conditioning consisted of putting the signal through a gain stage of approximately 1000 to take advantage of the ADC's dynamic range, since the biopotential signal is so small, and an offset addition stage to raise the minimum voltage to be above 0V since the ADS1114 cannot sample and quantize any negative voltage inputs. After all this signal conditioning the main sources of noise are moderately handled and the signal is sufficient to be transferred to hub.

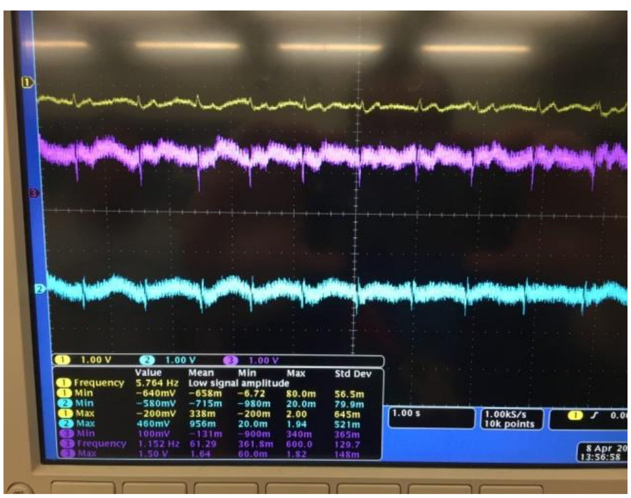


Figure 3: Output of Analog-Front-End Signals

After transmission to the hub the signal completes its processing by first going through the INA333 instrumentation amplifier to further assist in the removing of the signals common-mode noise. The signal is then quantized and sampled by the ADC on the TMS320C5535 Digital Signal Processor and the last stages of filtering are applied. These filters consist of digital FIR filter counterparts to the Analog-Front-End filters (Notch, Lowpass, and Highpass) but now on the order of 350. After implementing these filters via software on the DSP the signal is now sufficient for processing by the algorithms and for transmission/display to the android application.

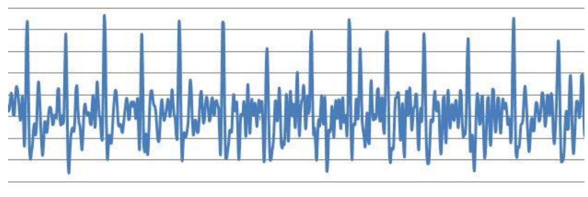


Figure 4: ECG Signal Output from DSP

V. DSP Algorithms

The Wireless Wearable Electrocardiogram implements a series of algorithms used for the detection and calculation of the user's different vital characteristics.

A. QRS Complex Detection

A person's QRS complex is the combination of three of the most distinctive changes in the user’s graphical representation of their cardiac activity on an electrocardiogram. Typically this is the central and most visually observable part of the signal's tracing. Physiologically this corresponds to the depolarization of the right and left ventricles of the heart. The algorithm for detection is based on the classical Pan Tompkins QRS detection algorithm. However, prior to implementing the algorithm it is important to do further signal conditioning/pre-processing in order to accurately estimate the energy in the QRS frequency band.

Pre-Processing/Filtering Steps:

1. Filter with a FIR Bandpass filter with pass band of 5-26 Hz.
2. Take the derivative of this bandpass filtered signal.
3. Take the absolute value of the resultant signal.
4. Average the absolute value of the signal over an 80 ms window.

Pan Tompkins Algorithm:

The Pan and Tompkins algorithm consists mainly of a peak detection algorithm. First a threshold is determined based on the mean estimate of the average QRS peak and the average noise peak. Additionally, this threshold is automatically adjusted throughout the algorithm's implementation. The peak will then be classified as either noise or a QRS complex based on whether it surpasses the detection threshold or not. There are 4 main rules that the algorithm follows to accurately differentiate between a QRS detection and a noise peak.

Rule	Description
1	Ignore any peak that either precedes or follows a larger peak by less than 196ms.
2	If a peak is detected, then now refer back to the original raw signal and determine if that segment contains both negative and positive slopes. This correlates to a peak being found in the raw signal. Otherwise, the detected peak simply represents a baseline shift.
3	Compare the reported peak to the detection threshold. If larger than the detection threshold the signal is classified as a QRS complex. Otherwise, the signal is classified as noise.
4	Lastly, classify a peak as a QRS complex if no other QRS has been classified within 1.5 R-to-R intervals, but a peak greater than half the detection threshold followed the previous QRS detection by at least 360ms.

B. Heart Rate Calculation:

The algorithm for calculating the heart rate builds off the QRS complex detection algorithm. It simply consists of counting the amount of QRS complexes occurring over a six second interval and then multiply that count by ten. The heart calculation algorithm is able to maintain its accuracy by working in conjunction with the QRS detection algorithm.

C. Arrhythmia Detection Algorithm:

Similar to the QRS complex detection algorithm, the Ventricular Arrhythmia Detection Algorithm is based on a combination of peak detection and power spectral density analysis of the ECG signal. However, prior to algorithm implementation it is important to first extract the specific features we are interested in. This involved first analyzing normal sinus rhythm signals vs. arrhythmias based on data provided by MIT's BIH Arrhythmia database. After spectral analysis it was apparent that between approximately .5Hz-20Hz arrhythmic behavior differentiated itself from normal sinus rhythm spectral characteristics but there were not any specific peaks at any specific frequency.

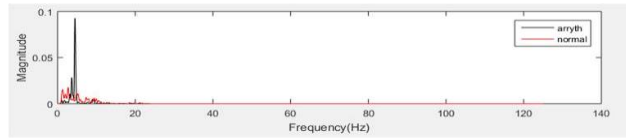


Figure 5: Arrhythmia PSD vs. Calm PSD

However, there tended to be concentrations of energy in small frequency bands for arrhythmias compared to normal sinus rhythm signals. This resulted in a sub-band total power approach to detecting the arrhythmias. For each tested signal segment of 5 seconds the power spectral density of each segment is calculated. Then in sub-bands of 2Hz the total power content in every 2Hz is determined. Afterwards a ratio comparing the power of the tested sub-band versus the characteristic normal total power is calculated. A peak detection algorithm is then implemented to determine if any peak ratios in the tested segment occur above the threshold frequency for arrhythmic activity.

Another component of the detector is a waveform morphology analysis of the time domain signal. This again employs a peak detection algorithm but this time determines the length of time between successive peaks. Arrhythmias tend to be characterized by wider intervals between peaks in their time domain waveforms whereas normal sinus rhythms tend to be more periodic.

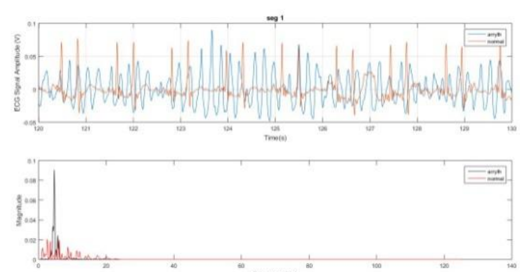


Figure 6: Time Domain Representation of Arrhythmia vs. Normal Sinus Rhythm(Top), PSD comparison(Bottom)

Lastly, a method based on clinically derived QRS complex characteristics of arrhythmias can be employed in conjunction with the QRS complex detection algorithm results to accurately detect arrhythmia. Accuracy is key because in a clinical setting it is important to have as few false alarms and missed alarms as possible, while still maintaining high accuracy. Therefore, a combination of all three approaches will most likely result in the most accurate detections. The detector will be analyzing a new segment of the ECG signal every 5 seconds. The main purpose of the algorithm is to detect any arrhythmia occurring in the patient versus differentiating between the different types of arrhythmias. Therefore the algorithm is more general in this approach.

D. Arrhythmia Detector Implementation

Since after spectral analysis it was apparent that between approximately .5Hz-20Hz arrhythmic behavior differentiated itself from normal sinus rhythm spectral characteristics the signal segment was filtered by a digital FIR lowpass and highpass filter with cutoff frequencies of 20Hz and .5Hz respectively. This is only for arrhythmia detection to preserve the desired signal characteristics in this specific bandwidth. After conditioning the signal the autocorrelation function of the 5 second segment was determined and then an FFT was performed on the segment of the signal to determine the Power Spectral Density of the segments.

$$R_{xx}(m) = \sum_{n=0}^{N-1} x(n)x(n-m)$$

$$S_{xx}(\omega) = \mathcal{F}\{R_{xx}(m)\} = |X(\omega)|^2$$

The FFT was performed using the TMS320C5535's FFT hardware accelerator. After the PSD of the segment is computed the DSP then follows the algorithm explained earlier which is additionally outlined in Fig 7.

VI. Firmware

One of the largest contributing factors to using both Energia and Arduino was the communities that both software distributions. Another great feature of using both of these was the ability bounce back and forth between devices to assist in the debugging process since both used the same or similar overall libraries. The most effective library that were consistently used for debugging would be the Serial library since allowed for simple easy data viewing.

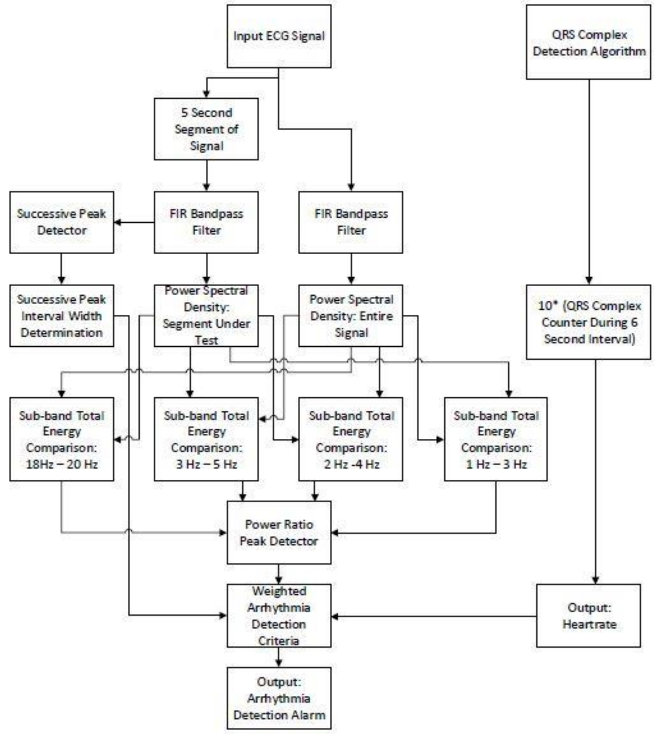


Figure 7: Arrhythmia Detection Algorithm Block Diagram

A. Sensor Firmware:

After the Bluetooth modules overall firmware was already prepared during setup. The main items that the microcontroller was used for was the LED notification, Bluetooth communication and the I^2C for the analog to digital converter. Upon power received the Bluetooth module automatically connects with the master, however no data is transferred until the master on the hub receives a start. This start the starts the software timer as well as the read from the analog to digital converter. When both of these readings are ready, they are combined into a string that represents the voltage value combined in a timestamp. This data is then outputted via the HM-11 module. Because of Bluetooth’s password settings and AES-256 encryption that comes within the Bluetooth standard there is also need to concern data being intercepted. This loop will continue until the power has been disconnected from the module.

B. Hub Firmware

Starting with hardware initialization, the hub starts and sets all of the necessary addresses and power up commands needed before entering its cyclic loop. Beginning with data that’s received from the sensors

Bluetooth modules, the values are store in a buffer. The buffer is then sorted according to timestamp within the buffer in order to consistently match signals. When the buffer has a set amount of data, it is then outputted to the next set of devices. Once the overall system reaches a set threshold of data inputs it begins transmitting to the instrumentation amplifier. After the hardware processes on the hub the data is moved to the DSP for final processing, the data is then re-inputted into another separate output buffer than transmits it back to the app. Because the Bluetooth modules throughout the device are all the same not only is the method of communication between the modules and the HM-11 the same, but security is also constantly maintained by using the encryption and passcode system stated above.

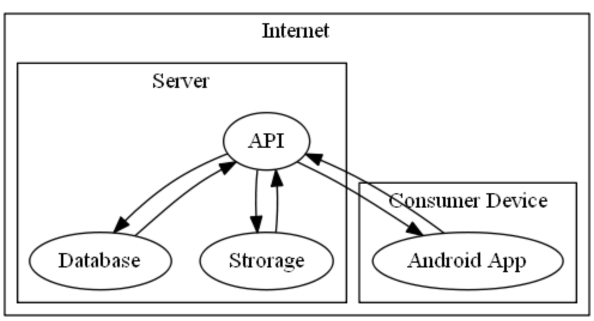
C. *Bluetooth Firmware*

Setting up the Bluetooth was a simple task of enabling certain firmware commands to auto connect with the past last slave. These AT commands are built into HM-11 module. In order to start data transmission upon power up the device set to Master must first receive a AT+START which completes the connection and disables the firmware access.

VII. Android Application

Fig 8. displays basic flow chart of how data is being transferred throughout the application, server and database storage.

Fig 8. Android Flow Chart



A. *User Interface*

After receiving the input via the Hub's Bluetooth slave, the client application handles this data by checking for an arrhythmia, storing the user's heart rate and full QRS rhythm strip representation. Once these are received and displayed via the screen, and the data is processed within the local API and sent to the server.

B. *Database Storage*

The data is stored within a MySQL database and data is moved with a key to maintain client authentication. Maintaining security within the database, the ID of each user will never be displayed by the API. With only the admin having access, the only other way to access the database would be through the Service API. The flow chart below in Fig 9. displays a simplified version of the overall authentication process.

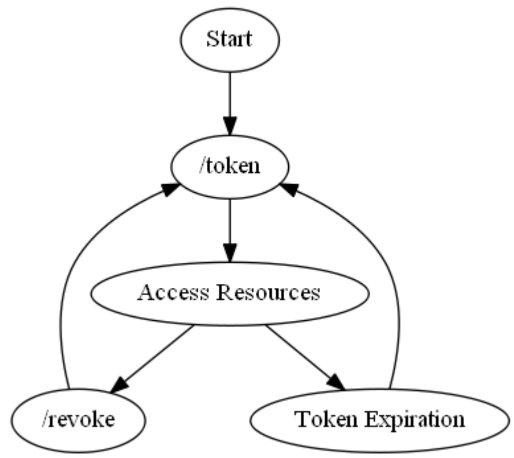


Figure 9. Simple overall authentication

VIII. Conclusion

This two semester project was a culmination of everything that we have learned throughout our undergraduate careers. We were able to successfully put into practice all of the different topics we have learned. From signal processing to electronics to embedded systems programming. Furthermore, this project was a valuable experience in working in a team and getting a full system level design and implementation underway. All of this experience will be key in any post-undergraduate career path that we choose to pursue. Lastly, at the conclusion of our project we have proven that it is possible to successfully acquire, process, and display a person's cardiac activity wirelessly without the need of a right leg driven circuit.

IX. References

Narayanaswamy, S. "High Resolution Electrocardiography." *Indian Pacing and Electrophysiology Journal*. Indian Pacing and Electrophysiology Group, 1 Apr. 2002. Web. 06 Nov. 2015.

Pan, Jiapu, and Willis J. Tompkins. "A Real-Time QRS Detection Algorithm." *IEEE Transactions on Biomedical Engineering IEEE Trans. Biomed. Eng.* BME-32.3 (1985): 230-36. Web.

Gari Clifford, John Fisher, Julie Greenberg, and William Wells. *HST.582J Biomedical Signal and Image Processing, Spring 2007*. (Massachusetts Institute of Technology: MIT OpenCourseWare), http://ocw.mit.edu (Accessed 9 Dec, 2015). License: Creative Commons BY-NC-SA

Yanowitz, Frank G., MD. "3. Characteristics of the Normal ECG." *ECG Learning Center*. University of Utah School of Medicine, n.d. Web. 25 Nov. 2015.

Oweis, Rami J., and Basim O. Al-Tabbaa. "QRS Detection and Heart Rate Variability Analysis: A Survey." *Biomedical Science and Engineering* 2.1 (2014): 13-34.

Smart Wearable ECG EKG Monitor - QardioCore. (n.d.). Retrieved October 9, 2015, from https://www.getqardio.com/qardiocore-wearable-ecg-ekg-monitor-iphone/

Delano, Maggie K.; Sodini, Charles G., "A long-term wearable electrocardiogram measurement system," in *Body Sensor* Networks (BSN), 2013 IEEEInternational Conference on , vol., no., pp.1-6, 6-9 May 2013 URL: http://ieeexplore.ieee.org/stamp/stamp.jsp?tp=&arnumber=6575459&isnumber=6575455

X. Acknowledgements

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XI. Biography

Ryan Shifman is a senior at the University of Central Florida. He plans to graduate with his Bachelor's of Science in Electrical Engineering in May of 2016. He is presently working at Voxx International and hopes to pursue a Master's degree specializing in power electronics.

Alex Consunji is a senior at the University of Central Florida. He plans to graduate with his Bachelor's of Science in Computer Engineering in May of 2016. He is presently working at Lockheed Martin and plans to attend graduate school.

Karson Kimbrel is a senior at the University of Central Florida. He is graduating in May of 2017 with a Bachelor's of Science in Computer Science. He is currently employed by Leopard Labs. Karson's average app rating on the Play Store is 4.5.

Ulbert Botero is a senior at the University of Central Florida that is graduating May of 2016 with a Bachelor's of Science in Electrical Engineering and a minor in Computer science. He plans to spend the summer as an Applications Engineering Intern with Texas Instruments before starting graduate school at the University of Florida pursuing a Ph.D. in Electrical Engineering with a research focus on Biometrics and Hardware Security.