PROJECT TRANSCRIPTION

EEE-426

Biomedical Instrumentation Laboratory

PORTABLE CARDIAC ACTIVITY MONITROING DEVICE OFFERING ANDROID APPLICATION

Presented By- Group-**10**

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Further, we would like to thank Joel Murphy and Yury Gitman, the designers and developers of Pulse Sensor Amped. Without their open source technical affinity, this work would be very much tougher and more complicated. Accordingly, we acknowledge the online tutorials and technical articles about Pulse Sensor Amped available at **http://pulsesensor.com.**

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ABSTRACT

A substantive framework of a wearable cardiac activity monitoring device has been presented which is integrated with a Human Machine Interface (**HMI**) via an **Android** application. The concept is to analyze fingertip Photoplethysmographic (**PPG**) signal and estimate certain physiological parameters like Heart Rate (**HR**) and Heart Rate Variability (**HRV**). **Pulse sensor Amped** has been utilized to extract PPG signal which evaluates Photoplethysmograph of a non-invasive signal that is a quantitative marker of autonomic nervous system of a human body. After being extracted, the raw signal has been processed by a microcontroller programming tool and then the evaluated data have been transformed to Android cell-phone via a wireless communication protocol. In this respect, a customized Android application named **Impulse** has been developed which compiles the obtained signal and displays sinus activity and rhythmic variation of several physiological elements like Beat per Minute (**BPM**), HRV, mean and covariance of BPM on the basis of time-to-time frame analysis. The application provides consistent monitoring of the PPG waveform and proportionately it simulates the stimulus in a continual manner. Thus a complete HMI system has been fabricated which enhances the scope of self-diagnosis of a human heart in a perpetual duration of evaluation. Moreover, the hardware segment of this system is quite simple and compact and this practice ratifies the portability and reliability of the demonstrated project. Additionally, relevant data processing and filtering algorithms have been implemented on **Matlab** to exterminate dc errors and other baseline problems of a sensed PPG signal. The simulation results obtained from Matlab are consistent with the Android based supplementation. The project has been tested for different users under several physical consequences and the performance analysis affirms its accuracy.

0.1 Introduction

A conceptualized prototype of an Android-based HMI system to monitor heart issues and to analyze physiological parameters is presented here. This work is oriented to extract and process Photoplethysmography (**PPG**) signal from a pulse sensor and estimate about Heart Rate (HR) and Heart Rate Variability (HRV) of the autonomic sinus activity of a human body. The ace promising feature is to introduce a customized Android App named **Impulse** in order to evaluate BPM, HRV and deviations in a consistent manner. The data transformation takes place via wireless communication protocol and users can frequently utilize this diagnostic analysis. The ultimate formation of the project is a portable Cardiac activity monitoring gadget which is interfaced with a user-friendly smart phone application. However, the basics of PPG, physiological indexes, configuration of Pulse sensing device, features of the self-acclaimed Android app and MATLAB filtering techniques are articulated here in a segment-by-segment manner.

0.2 PPG Signal Analysis

The following excerpt documents the basic salient features of Photoplethysmographic signal.

0.2.1 Definition

Photoplethysmography is a non-invasive technique that measures relative blood volume changes in the blood vessels close to the skin. By definition, Photoplethysmogram (PPG) is an optically obtained plethysmogram, a volumetric measurement of an organ of human body. PPG signal is used to estimate the skin blood flow using infrared light.

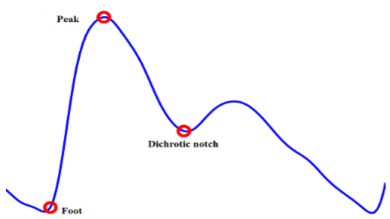


Figure 1: PPG wave segment

0.2.2 Interpretation

With each cardiac cycle the heart pumps blood to the periphery. Blood pressure is the pressure that blood exerts on the walls of a vessel in the arteries. During the contraction of ventricles blood is expelled from the heart; pressure is generated and it is at its maximum in the arterial system. This is called the **Systolic** blood pressure when the muscle compression happens. When the heart is relaxing and the ventricles are refilling with the blood returning from the rest of the body, the pressure is very low which is referred to as **Diastolic** blood pressure.

Even though this pressure pulse is somewhat damped by the time it reaches the skin, it is enough to distend the arteries and arterioles in the subcutaneous tissue. If a competent pulse sensing device is attached without compressing the skin, a pressure pulse can be observed from the venous plexus as a small secondary peak. The change in volume caused by the pressure pulse is detected by illuminating the skin with the light from a light-emitting diode (LED) and then measuring the amount of light either transmitted or reflected to a photodiode. Each cardiac cycle appears as a peak and sinus activities and relevant parameters depicting the health issues can be specified from the obtained PPG signal.

Therefore it can be stated that Photoplethysmography is an optical technique for measurement of pulsating blood flow. It consists of LED modules and detector for photocell. The amount of light absorbed depends upon the blood density in the finger tip and remaining amount of light is transmitted or reflected which is captured by the sensor. The Plethysmographic pulse obtained by the optical sensor is used for various cardiovascular parametric calculations.

0.2.3 Elementary Components

The PPG signal reflects the blood movement in the vessel which goes from the heart to the fingertips and toes through the blood vessels in a wave-like motion. The period of the PPG waveform consist of two phases. The rising edge is called the Anacrotic phase and is the systolic upstroke time [**1**]. The Catacrotic phase is characterized by the diastole on the falling edge. A Dicrotic notch is on the Catacrotic phase. The Dicrotic notch is due to the closing of the aortic valve and thus increases blood volume in the arteries. The peak height is the difference between the maximum of a cardiac cycle and the previous minimum. This is the height of the pulsating (AC) component of the PPG. It is proportional to the difference between the arterial systolic and diastolic pressures [**1**].

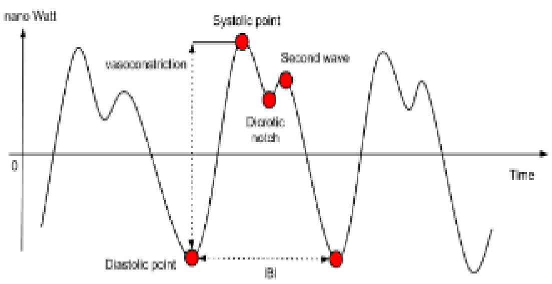


Figure 2: Distinctive components of a PPG signal

0.2.4 Detection Process

Photoplethysmography is an optical measurement phenomenon that uses an invisible infrared light sent into the muscular tissue and the amount of the backscattered light corresponds with the variation of the blood volume. In accordance with the consistent variation of blood flow, the concentration of blood vessels and other microscopic particles changes randomly and this practice causes subsequent changes in the mechanism of blood fluid. When a pulse sensor is attached to the skin tissues, the variation of incident and reflected light beams is evaluated. In this follow-up, the photo detector substantiates the particular amount of reflection through a reflected light screening method. In this manner, the blood flow variation in terms of plethysmogram is detected.

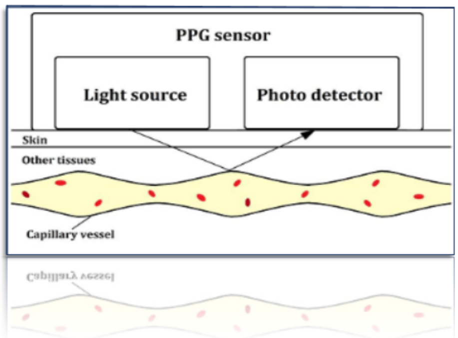


Figure 3: PPG detection from volumetric blood cell variation

0.2.5 Applications

Researchers from different domains of science have become increasingly interested in PPG because of its advantages as non-invasive, inexpensive, and convenient diagnostic tool. Traditionally, it measures the oxygen saturation, blood pressure, cardiac output, and autonomic functions. In addition, PPG is a promising technique for early screening of various pathological substances and could be helpful for regular assessment. The paramount factors for utilizing PPG signal in diagnosis process are stated here.

Non-invasive Character

This feature affirms the fact of not using any harmful inherent chemical or medical instrument in a human body to extract and process sinus signal. PPG is detected via a photo sensitive device which is externally connected to the wrist or fingertips or ear lobes of a human body.

Cost-effectiveness

The device fabrication and interfacing modules are cheap and the detection process of PPG is proven quite reasonable in the sense of economic feasibility in case of frequent applications.

Sustainability

The PPG extraction and operative devices are compact and easy to carry objects. The lenient package units of PPG monitor and processor are the underlying reasons of being the device sustainable and reliable.

Simplicity

The overall detection procedure and active observation of PPG-based sinus activity are estimated quite simple and user friendly.

0.3 ECG versus PPG

Electrocardiography (**ECG** or **EKG**) a representation of the electrical activity of the heart muscle as it changes with time, usually printed on paper for easier analysis. Like other muscles, cardiac muscle contracts in response to electrical depolarization of the muscle cells. In this respect, ECG is the process of recording the electrical depolarizing of the heart over a period of time using electrodes placed on a patient's body. These electrodes detect the tiny electrical changes on the skin that arise during each heart beat.

On the other hand, PPG is a medical topology of estimating cardiac activity on the basis of speculative evaluation of volumetric changes in blood cells and vessels of a human body. The PPG extraction involves retrospective use of light sensors and Infrared modules for accurate signal acquisition and processing.

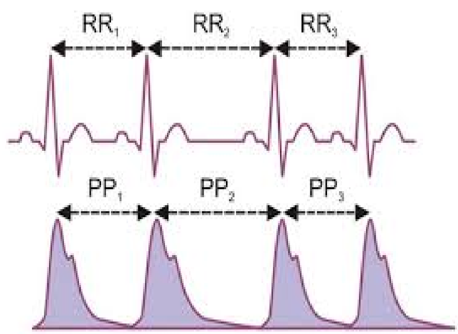


Figure 4: Comparative analysis of ECG and PPG signal

The basic differences between ECG and PPG analysis are articulated here.

0.3.1 Sensing Operation

ECG (electrocardiography) sensors measure the bio-potential generated by electrical signals that control the expansion and contraction of heart chambers. Whereas, PPG (photoplethysmography) sensors use a light-based technology to sense the rate of blood flow as controlled by the hearts pumping action.

0.3.2 Heart activity Measurement

ECG sensors directly use electrical signals produced by heart activity.
On the contrary, PPG uses electrical signals derived from light reflected due to changes in blood flow during heart activity.

0.3.3 Accuracy

ECG is a reference standard signal that is used for monitoring cardio health and wellness by healthcare providers.
PPG sensors, on the other hand, typically use ECG signals as a reference for static HR (Heart Rate) comparison.

0.3.4 Heart Rate measurement

With ECG, instantaneous as well as mean HR can be measured accurately.
HR can be measured with PPG, however, it is only suitable for average or moving average measurements.

0.3.5 HRV estimation

HRV can be reliably derived from ECG data as R-Peak Intervals can be extracted with millisecond accuracy.
And with PPG sensors, Peak Interval accuracy is limited by usable sampling rate due to the high power consumption of LEDs.

0.4 Physiology of PPG signal

The paramount physiological parameters of PPG signal considered in this project walk-through are Heart Rate (HR) and Heart Rate Variability (HRV).The fundamental prospects of HR and HRV analysis are briefly described here.

0.4.1 Heart Rate

Heart Rate can be classified into two types in respect to the time frame evaluation of the indexes.

Mean HR (MHR)

From PPG signal, counting number of Dichrotic notch (prominent peaks of the downward spike) provides MHR. It is normally sampled for a 1 minute duration of activity. In general MHR remains within the range of 60 to 120 BPM for a healthy middle-aged person. However, MHR does not provide too much informative assistance in case of assessing human health issues. There are two forms of MHR evaluation- a) Resting Heart Rate and b) Recovery Heart Rate.
a) Resting Heart Rate
it is depicted as the number of beats in one minute when the person is at complete rest.
b) Recovery Heart Rate
It is depicted as the HR that a human body would decrease to after an exercise session. For 30-35 minutes walking session, Recovery HR has been found for a 25 year old man varies from **95-155 BPM**.

Instantaneous HR (IHR)

IHR is defined as the spontaneous beat-to-beat interval which is presented as **Tachogram**. This non-linear time series is prevalent in the analysis of autonomic nervous system (**ANS**).

0.4.2 Heart Rate Variability

Theoretically HRV is referred to as the systematic regulation of SA node by two branches of Autonomic Nervous system (ANS). It is an indirect measure of heart health, as defined by the degree of balance in sympathetic and vagus nerve activity.
HRV analysis is performed on three domains.
Time domain analysis
This is the conventional and most widely accepted method for analyzing HRV which is undergone for a period of 24 hours.
Frequency domain analysis
Fourier domain analysis consists of two major sections named High frequency (**0.18-0.4 Hz**) and Low frequency (**0.04-0.15 Hz**). Ratio of the low-to-high frequency component is an index of **parasympathetic to sympathetic balance**.

Geometric method
This is shown as a phase space portrait which is called **Poincare plot**. It is estimated on the basis of a predefined delay like x$[n + D]$ vs x[n] where D is the delay and n means sample numbers

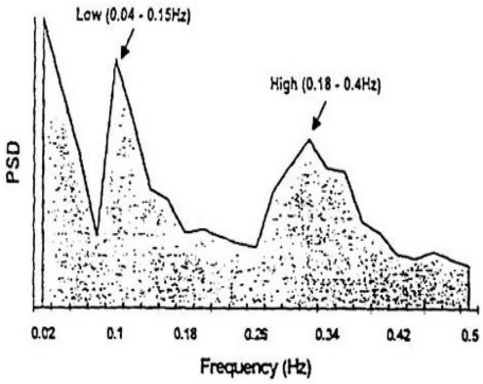


Figure 5: Frequency spectral depiction of HRV

Significance of HRV:
i. HRV predicts about the survival ability after heart attack.
ii. Reduced HRV is an indication of MI disease.
iii. Reduced HRV is an indication of fatal ventricular arrhythmia.
iv. Reduced HRV is related to negative emotions.
v. HRV decreases with age.

0.5 Implemented PPG Extraction Device

This segment articulates the description of the demonstrated sensory modules, wireless interfacing devices and relevant programming kits and their operational schemes.

0.5.1 Pulse Sensor

In this demonstrated work Pulse Sensor Amped has been utilized to detect PPG signal from the variations of volumetric fluid excursion.

Background

The Pulse Sensor, that evaluates a photoplethysmograph, is a renowned medical device used for non-invasive heart rate monitoring. Generally Photoplethysmographs measure blood-oxygen levels (SpO2). The heart pulse signal that comes out of a photoplethysmograph is an analog fluctuation in voltage, and it has a predictable wave shape. The depiction of the pulse wave is called a photoplethysmogram or PPG. The implemented version of PPG extracting module called **Pulse Sensor Amped** amplifies the raw signal of the obtained waveform and normalizes the pulse wave around **V/2** (midpoint in voltage). Pulse Sensor Amped responds to relative changes in light intensity. If the amount of light incident on the sensor remains constant, the signal value will remain close to 512 (midpoint of ADC range). If more light is incident, the signal will go up as

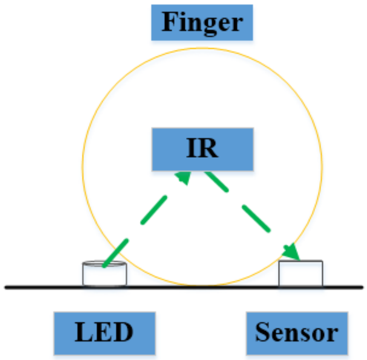


Figure 6: Light detection mechanism of Pulse sensor

well and with less light, the opposite happens. Light from a LED panel, that is reflected back to the sensor, changes during each pulse.

In order to find successive moments of instantaneous heart beat and to measure the time between, called the Inter Beat Interval (IBI), the predictable wave pattern of the extracted PPG wave has to be considerably implicated.

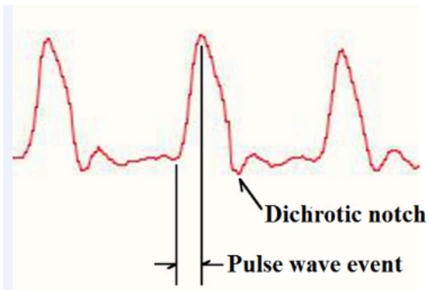


Figure 7: Detected PPG waveform

When the heart pumps blood through the body, with every beat there is a pulse wave (kind of like a shock wave) that travels along all arteries to the very extremities of capillary tissue where the Pulse Sensor is attached. Actual blood circulates in the body much slower than the pulse wave travels. The events are followed as they progress from point **T** on the PPG. A rapid upward rise in signal value occurs as the pulse wave passes under the sensor. Then the signal falls back down toward the normal point. Sometimes, the **Dicroic notch** (downward spike) is more pronounced than others but generally the signal settles down to background noise before the next pulse wave washes through. Since the wave is repeating and predictable, any recognizable feature can be selected as a reference point, say the peak value and the heart rate can be measured by applying arithmetic formula and mathematical computations on the time between each peak. This, however, can run into false readings from the Dicroic notch, if present, and may be susceptible to inaccuracy from baseline noise as well. There are other good reasons not to base the beat-finding algorithm on arbitrary wave phenomena. Ideally, the instantaneous moment of the heart beat are supposed to be found. This is important for accurate **BPM** calculation, Heart Rate Variability (**HRV**) studies, and Pulse Transit Time (**PTT**) measurement.

Circuit Configuration

Pulse sensor is basically an open-source hardware project presented by Joel Murphy and Yury Gitman. The latest version which has been used in this context is called Pulse sensor Amped.

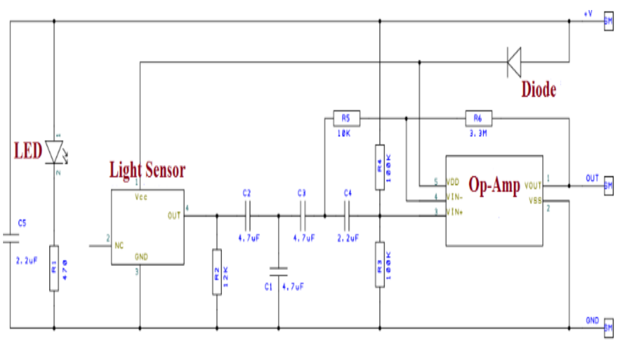


Figure 8: Design of the Pulse sensing circuitry

Table 1: Specifications of the Pulse Sensor Amped

Parameters	Measurement range
Diameter	0.625 in
Thickness	0.125 in
Cable length	24.00 in
Operating voltage	3-5V
Current consumption	4mA at 5V

Operational Topology

The Current to Voltage Converter is an Op Amp circuit that invariably uses a photodiode as a current source and is often used as a starting point for developing optical heart-rate monitors. A fairly universal Low Pass Filter is designed for the stable output waveform. The filtering circuitry is a passive RC module which contains **R: 100 ohm** and **C: 4.7uF**. Ambient light sensor device named **APDS-9008** has been imbibed onto the circuit board. This is an integrated photodiode, Op Amp and resistive feedback network. The peak sensitivity for this sensor is **565 nm**, hence the green LED unit has been utilized for illuminating the Infrared operation. Additionally, several series-parallel combinations of 22k and 470 ohm resistors have been included in the design to emulate the voltage and current division process for the optimum operation of the sensor.

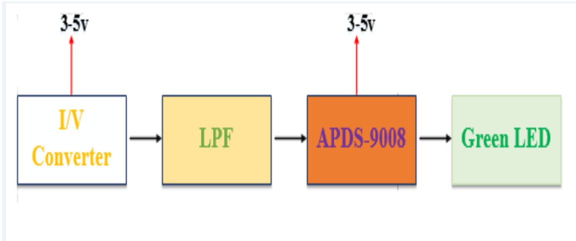


Figure 9: Manipulative stages of the developed sensor

0.5.2 Wireless Module

Bluetooth communication protocol has been applied in this project to substantiate wireless data transformation between the inter-connected devices. In this respect, **HC-05** module has been used. This is an easy to use Bluetooth **SPP** (Serial Port Protocol) module, designed for transparent wireless serial connection setup. Serial port Bluetooth module is fully qualified Bluetooth **V2.0+EDR** (Enhanced Data Rate) 3Mbps Modulation with complete 2.4GHz radio transceiver and baseband. It uses CSR Bluecore 04-External single chip Bluetooth system with CMOS technology and with **AFH** (Adaptive Frequency Hopping Feature). It has the footprint as small as $12.7mm \times 27mm$ [**2**].

Specifications

Hardware features:
a. Typical **-80dBm** sensitivity
b. Up to **+4dBm** RF transmit power
c. Low Power 1.8V Operation, **1.8 to 3.6V** I/O
d. PIO control
e. UART interface with programmable baud rate
f. With integrated antenna and edge connector

Software features:
a. Default Baud rate: **38400**, Data bits: **8**, Stop bit: **1**, Parity : No parity Given a rising pulse in PIO0, device will be disconnected
b. Status instruction port PIO1: low-disconnected, high-connected
c. PIO10 and PIO11 can be connected to red and blue led separately
d. Auto-reconnect in 30 min when disconnected as a result of beyond the range of connection

0.5.3 Micro-controller

ATmega328P microcontroller chip has been utilized for developing the firmware of extraction of the signal from the pulse sensor and calculation of heart rate.

Table 2: PPG variables and their updating intervals

Variables	Refresh rate	Significance
Signal	2ms	Raw signal
IBI	Every beat	Time between beats
BPM	Every beat	Beats per minute
QS	Set true every beat	Cleared by user
Pulse	Set true every beat	Cleared by ISR

This is a 8-bit microcontroller chipset which operates at a nominal voltage of 5V.
The intermittent stages of the developed program are presented here whereas the complete code blocks are attached in **Appendix A**.

1. First of all, Timer2, an 8 bit hardware timer is set for maintaining a regular sampling rate so that it throws an interrupt every other millisecond. That gives a sample rate of **500Hz**, and beat-to-beat timing resolution of **2ms**. This will disable PWM output on pin 3 and 11.

2. The register settings above tell Timer2 to go into **CTC** mode, and to count up to 124 (0x7C) over and over and over again. A pre-scalar of 256 is used to get the timing right so that it takes 2 ms to count to 124. An interrupt flag is set every time **Timer2** reaches 124, and a special function called an Interrupt Service Routine (**ISR**) is run at the very next possible moment, no matter what the rest of the program is doing. **sei()** ensures that global interrupts are enabled.

3. When the board is powered up and running with Pulse Sensor Amped plugged into analog pin 0, it constantly (every 2 ms) reads the sensor value and looks for the heart beat. This function is called every 2 ms. First thing to do is to take an analog reading of the Pulse Sensor. Next the variable **sampleCounter** is incremented. A variable named N will help avoid noise.

4. To keep track of the highest and lowest values of the PPG wave, accurate measure of amplitude is necessary. Therefore, variable P and T hold peak and trough values, respectively. The thresh variable is initialized at 512 (middle of analog range) and changes during run time to track a point at 50

5. In order to check the pulse, a minimum amount of time has to pass. This helps avoid high frequency noise.

6. To calculate realistic BPM value, a boolean **firstBeat** is initialized as true and **secondBeat** is initialized as false on start up, so the very first time a beat is found.

7. Pulse has been declared true during the upward rise in Pulse Sensor signal when we found the beat, above, so when the signal crosses thresh going down, the pulse is over. A little housekeeping in clearing **pulsePin** and the Pulse Boolean is considered. Then the amplitude of the wave that just passed is measured and thresh is updated.

0.6 Android Application

To establish a complete HMI system for consistent monitoring of cardiac activity and its variations, a self-contained Android application named **Impulse** has been developed. This smart phone app is oriented to display physiological parameters like HR, HRV, Standard Deviation (**SD**) and Coefficient of Variance (**CoV**). Through a wireless data transformation unit, Impulse continually takes input from the pulse sensor and processes the data and ultimately provides an estimation of human diagnostic aspects.

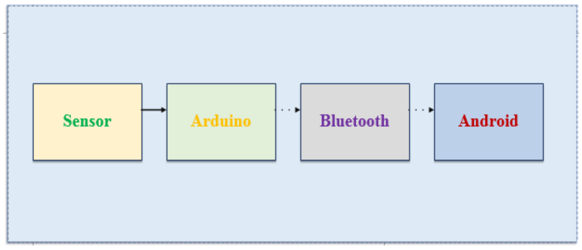


Figure 10: Overall functional diagram of the system

The complete software has been developed on Java platform and several self-defined library packages and functional code blocks have been compiled on the basis of Android compatible programmer. The entire algorithm of the software and online location of the customized code are attached in **Appendix B**.

0.6.1 Software Manual

Impulse, the fragmented Android App, is user-friendly and very much compatible with the available versions of Android operating system. In order to maximize the applicability of the demonstrated software, a complete operative manual is articulated here.

Step-1: Start
When a user clicks on the icon of the app, a preface window opens. This window is segmented into four empty display panels and some indicative options. This is the initiation of opening the console of the app.

Step-2: Configuration
When a user clicks on the notation set at the rightmost corner of the app window, there appears a box texted **Configure Connection** to implant Bluetooth connectivity of the smart phone with the relevant wireless module.

Step-3: Confirmation
After configuring the user has to establish a certain wireless communication with sensor. Therefore, the Bluetooth option of the cell phone has to be activated. A message box titled **Bluetooth Permission Request** appears which ensures the Bluetooth connectivity of the host device. This box urges the switching on

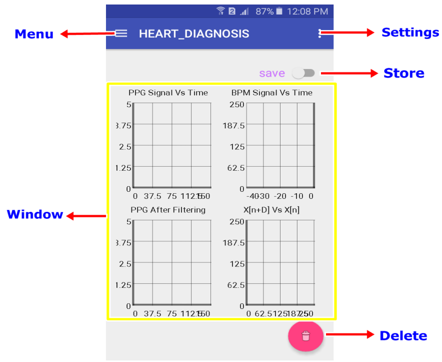


Figure 11: Initiation of the Android app

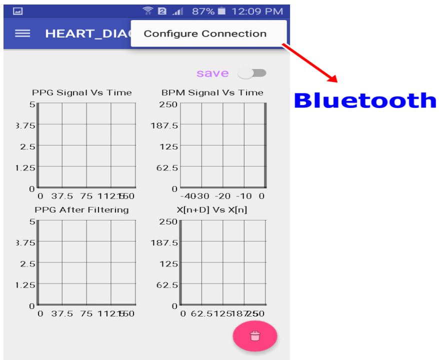


Figure 12: Configuration stage of the app

of the Bluetooth of the phone.

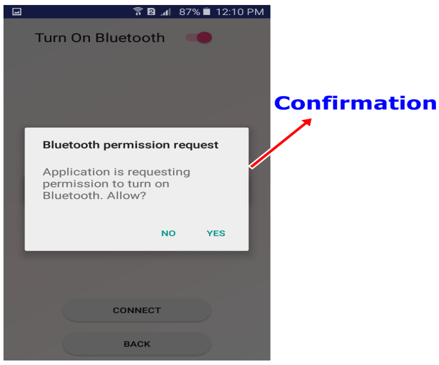


Figure 13: Confirmation of the wireless protocol

Step-4: Module

After turning on the Bluetooth option, there appears a display panel which shows the Mac address of the wireless module and also the device IP address is shown. There is also a provision of checking the status of the connection protocol via an authentication toolbox referred to as **Turn on Bluetooth**.

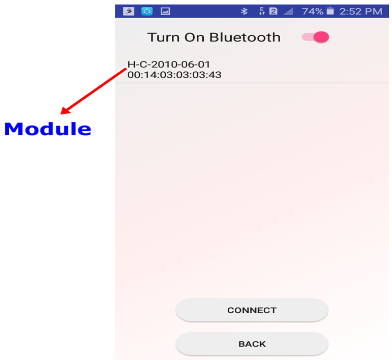


Figure 14: Device notations of the Bluetooth module

Step-5: Connection

When the authentication is done, another toolbox segmented into **Connect** and **Back** appears. **Connect** option takes forward the data acquisition process and **Back** option is to recheck the connection with the wireless device. If everything is okay, then a user is supposed to press the **Connect** button and enjoy the monitoring cardiac phenomena.

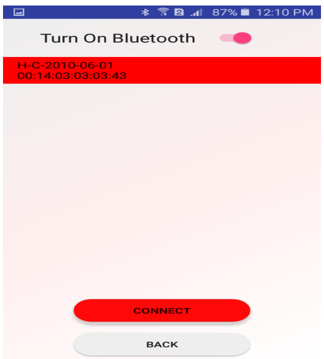


Figure 15: Wireless interface between the app and device

0.6.2 Operational Principle

As stated earlier, the overall fabrication of the presented Android app is executed on the Java and Android platform. The displays of the app console show time domain analysis of the raw PPG signal, time domain analysis of the filtered PPG signal, HRV in time frame in terms of BPM versus time plot and Poincare plot (Geometric analysis) of the anticipated HRV signal. Moreover, some mathematical parameters like Mean, SD and CoV are integrated with the console operation. When a user successfully implicates the software, his/her current BPM is supposed to be shown in the window. As it is a dynamic approach of

BPM calculation, the user can observe frequent changes in the estimated BPM values and also in the signal waveforms.

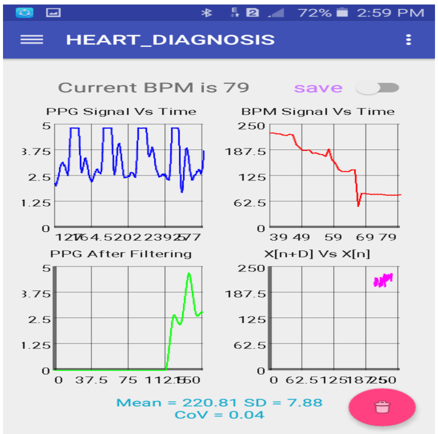


Figure 16: Cardiac analysis during a resting session

The time domain analysis is executed considering the time frame segmentation into 40ms slots. Thus a bunch of data is processed for every **40ms** time interval. The Poincare plot has been designed for a delayed version of sampled PPG values and this delay amount has been set for a **5ms** duration. Consequently, there is an option named Save which ascertains the accumulation of different time framed analyzed BPM values. By dint of this provision, there is always a scope of saving the diagnostic parameters for further application.

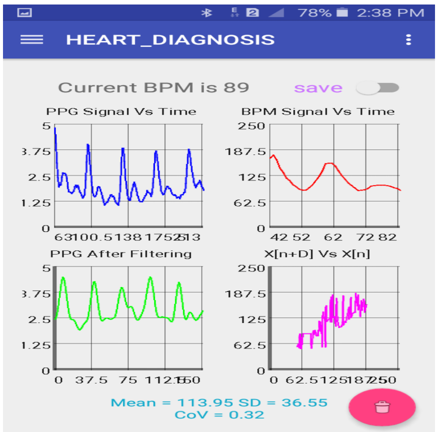


Figure 17: Cardiac analysis during steady walking

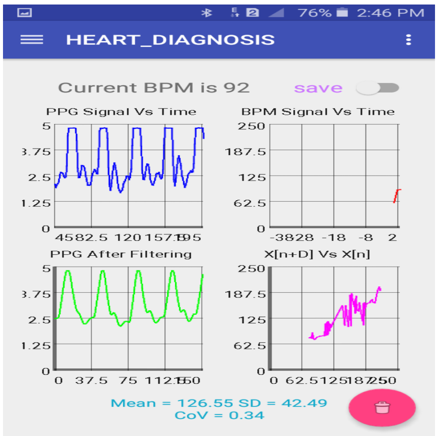


Figure 18: Cardiac analysis during swift walking

0.7 MATLAB Simulation

The data acquisition and processing methods have been simulated on MATLAB to develop a substantive framework for modifying the demonstrated work. Since the proposed Human Machine Interface (**HMI**) system is dependent on the accurate data extraction from the pulse sensor, there is a certain prospect regarding the discrepancies of the designed hardware block. The raw signal is quite noisy and contaminated with motion artifacts and random interference. Hence, the BPM calculation from the acquired PPG signal is prone to inaccuracy. To modify the erroneous PPG signal, various sophisticated signal processing algorithms and functional postulates have been presented. Here the attempt of filtering the sensed PPG signal is briefly explained.

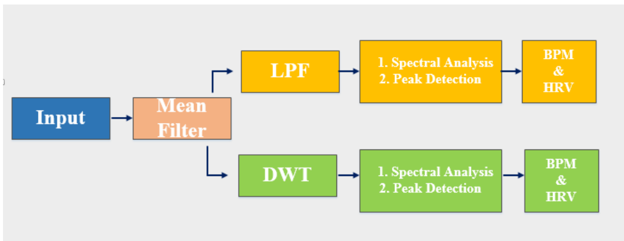


Figure 19: Functional diagram of MATLAB simulation

0.7.1 Functional Topology

Two distinctive methods are proposed here- a) Butterworth Filter b) Discrete Wavelet Transformation (DWT).

a) Butterworth Filter:

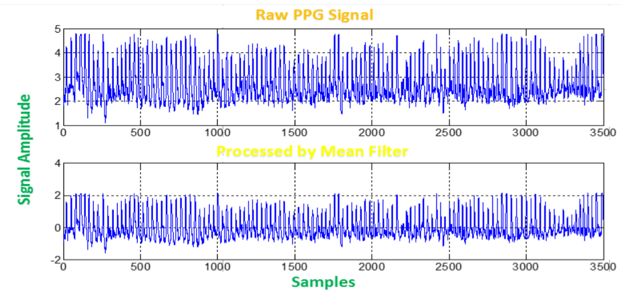


Figure 20: Processing of raw PPG by mean filter

At first the raw PPG signal is processed through a **Mean** filter which reduces the baseline error of the unprocessed data. Then a **fourth order Low Pass** Butterworth filer is designed [**3**]. The cut-off frequency is set to be approximately the sampling value (**25Hz**). After that Discrete Fourier Transform (**DFT**) algorithm is used to analyze the signal in Frequency domain. In this case, Fast Fourier Transform (**FFT**) method is applied with a bin size of **4096**. For clear estimation of the filtered PPG, the resolution size of FFT is kept large. Consequently, for power spectral analysis of the processed data, **Welch PSD** (Power Spectrum Density) methodology is executed. In addition, for evaluating the parasympathetic and sympathetic autonomic nervous activities, HRV spectrum is estimated in terms of Two sided Pwelch waveform [**4**].

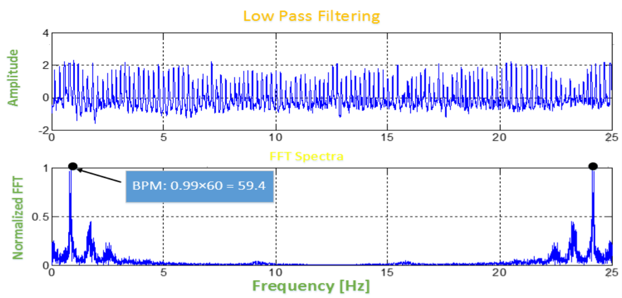


Figure 21: Implemented Low Pass filtering mechanism

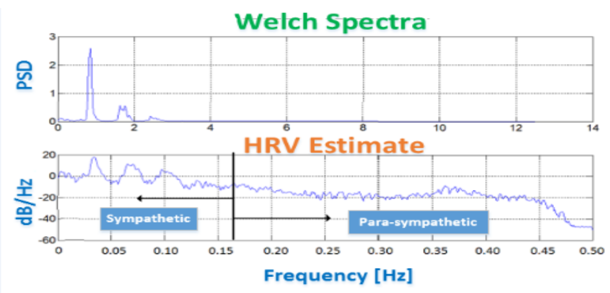


Figure 22: Welch spectral estimation

b) Discrete Wavelet Transformation (DWT):

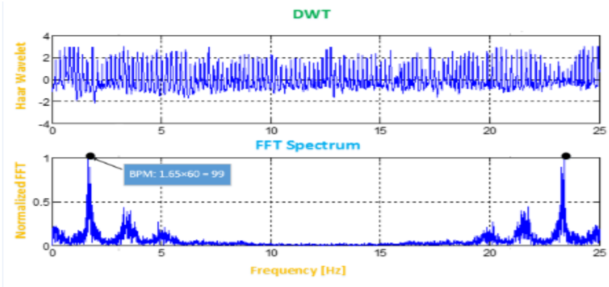


Figure 23: Processing of PPG signal via haar wavelets

In the beginning, similar to that of Butterworth filter mechanism, the raw PPG signal is processed through a Mean filter to reduce the baseline error. Then **haar** type wavelet processing block is designed [**5**], [**6**]. After that Discrete Fourier Transform (**DFT**) algorithm is used to analyze the signal in Frequency domain. Again, Fast Fourier Transform (**FFT**) method is applied with a bin size of **4096**. Consequently, for power spectral analysis of the processed data, **Welch PSD** (Power Spectrum Density) methodology is executed. And again for evaluating the parasympathetic and sympathetic autonomic nervous activities, HRV spectrum is estimated in terms of Two sided Pwelch waveform [**4**].

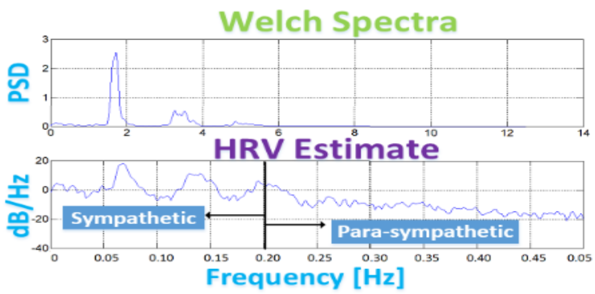


Figure 24: PSD and HRV approximation

In both of the above stated mechanisms, the BPM calculation from the FFT spectra is executed by the following formula-
$BPM = Frequencyvalueatwhichthemostprominentpeakoccurs \times 60$

However, for both case an independent code snippet for automatic peak detection is developed as well.

The overall MATLAB code for filtering algorithms is attached in **Appendix C**.

Table 3: Evaluated comparison among different modes of DWT

Modes	Correlated factor
haar	0.956
db4	0.914
sym2	0.905
coif1	0.883
bior3.3	0.862

0.7.2 Overview of MATLAB Implementation

The first method applied for processing the erroneous PPG signal is a fourth order Butterworth Low Pass filter which is suitable for average BPM estimation. Moreover, it can predict BPM accurately under locomotive or unstable conditions. This method is more applicable during resting states. However, the Low Pass filtering methodology has also been tested with **Chebysheb** and **Elliptic** structures but these are not proven effective in BPM estimation.
The second method demonstrated here is Discrete Wavelet Transformation (DWT). There are different modes of DWT such as **haar**, **db4**, **sym2**, **coif1**, **bior3.3**. Among these the most applicable in case of PPG analysis is **haar** wavelet scheme. This assessment of selecting the most appropriate mode is based on the correlated factor between the inverse DWT signal and the original one.

Moreover, there are some other methods of wavelet decomposition like Continuous Wavelet Transformation (**CWT**) and Stationary Wavelet Transformation (**SWT**). But these topologies are suitable for high frequency signal analysis. In case of PPG signal, using CWT and SWT methods filters out the low frequency components of the expected signal. Therefore, these methods are not used in PPG estimation.

In comparison to Low Pass filtering phenomenon, **DWT** is proven to be more accurate in analyzing the noisy PPG signals corrupted by motion artifacts.

0.8 Conclusive Remarks

In what follows, the presented prototype of a reliable as well as cost-effective portable human heart diagnostic system is an interpretation of a user-friendly HMI device. The consistent monitoring of cardiac activity and analysis of the variables like HRV and BPM certainly enhance the applicability of this initiative. There is a prevalent scope for further development in case of noisy and corrupted PPG signal processing.Specially, the sensing module and wireless configuration can be modified for more optimized signal acquisition. However, the MATLAB processing algorithms can be effectively utilized as well for future modification. In short, this tentative venture augments the dimensionality of analyzing cardiac signals and their inherent physiological significance.

Appendix A

Micro-controller Code:

```
//
void interruptSetup(){
  TCCR2A = 0x02;
  TCCR2B = 0x06;
  OCR2A = 0x7C;
  TIMSK2 = 0x02;
  sei();
}
//
void interruptSetup(){
  TCCR1A = 0x00;
  TCCR1B = 0x0C;
  OCR1A = 0x7C;
  TIMSK1 = 0x02;
  sei();
}
//
ISR(TIMER2_COMPA_vect) {
  Signal = analogRead(pulsePin);
  sampleCounter += 2;
```

```
int N = sampleCounter - lastBeatTime;
 if(Signal < thresh && N > (IBI/5)*3){
   if (Signal < T){
     T = Signal;
    }
  }
  if(Signal > thresh && Signal > P){
     P = Signal;
   }
  if (N > 250){
   if ( (Signal > thresh) && (Pulse == false) && (N > ((IBI/5)*3) ){
      Pulse = true;
      digitalWrite(pulsePin,HIGH);
      IBI = sampleCounter - lastBeatTime;
      lastBeatTime = sampleCounter;
       if(secondBeat){
   secondBeat = false;
   for(int i=0; i<=9; i++){
     rate[i] = IBI;
   }
  }
        if(firstBeat){
            firstBeat = false;
            secondBeat = true;
```

```
            sei():
            return;
           }
    word  runningTotal = 0;
    for(int i=0; i<=8; i++){
      rate[i] = rate[i+1];
      runningTotal += rate[i];
    }
    rate[9] = IBI;
    runningTotal += rate[9];
    runningTotal /= 10;
    BPM = 60000/runningTotal;
    QS = true;
    }
}
if (Signal < thresh && Pulse == true){
    digitalWrite(13,LOW);
    Pulse = false;
    amp = P - T;
    thresh = amp/2 + T;
    P = thresh;
    T = thresh;
  }
if (N > 2500){
```

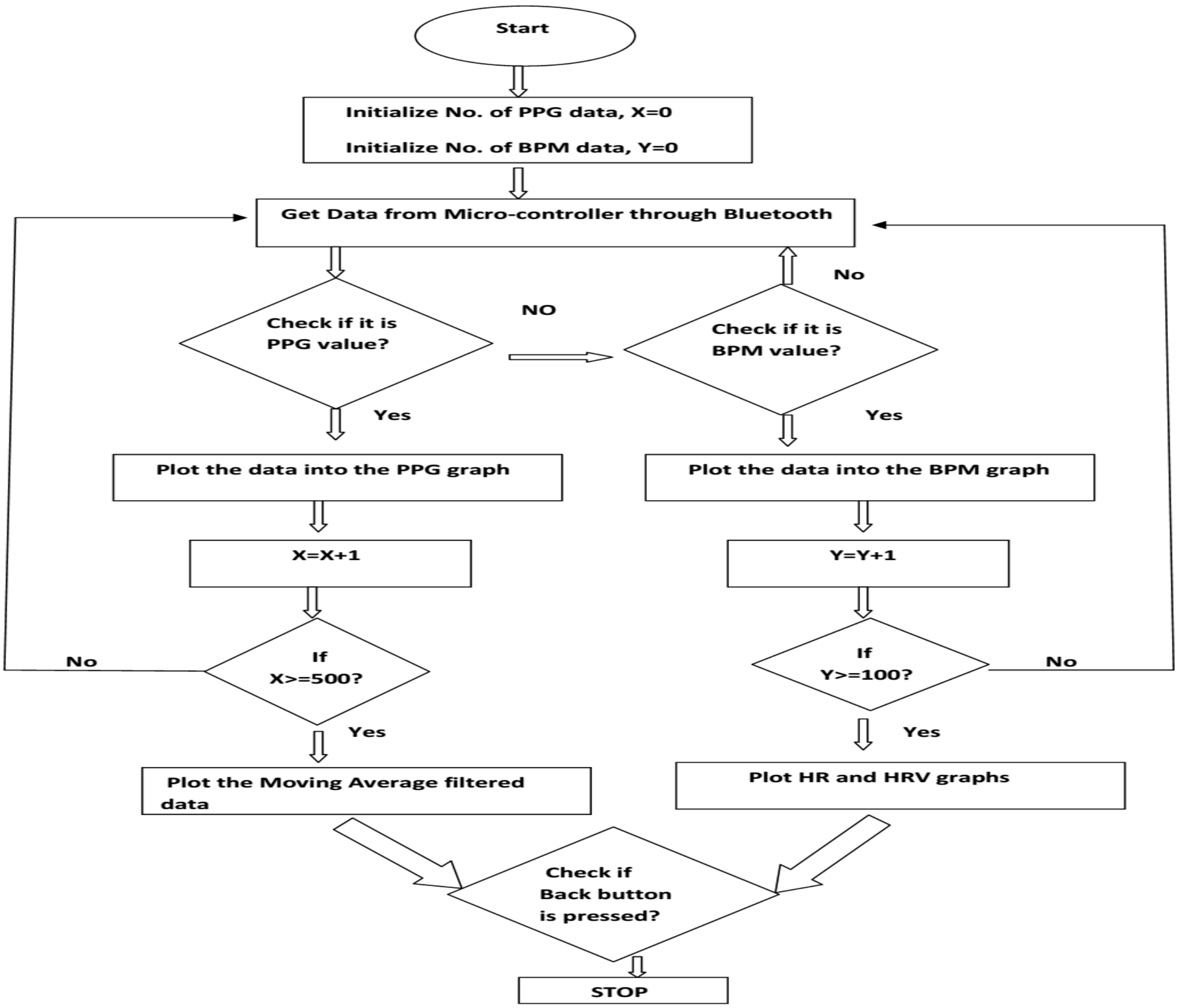
```
    thresh = 512;
    P = 512;
    T = 512;
    firstBeat = true;
    secondBeat = false;
    lastBeatTime = sampleCounter;
  }
//
int pulsePin = 0;
int blinkPin = 13;
int fadePin = 5;
int fadeRate = 0;
volatile int BPM;
volatile int Signal;
volatile int IBI = 600;
volatile boolean Pulse = false;
volatile boolean QS = false;
volatile int rate[10];
volatile unsigned long sampleCounter = 0;
volatile unsigned long lastBeatTime = 0;
volatile int P =512;
volatile int T = 512;
volatile int thresh = 512;
volatile int amp = 100;
```

```
volatile boolean firstBeat = true;
volatile boolean secondBeat = false;


void setup() {
  pinMode(13,OUTPUT);
  pinMode(10,OUTPUT);
  Serial.begin(115200);
  interruptSetup();
//  analogReference(EXTERNAL);
}
void loop() {
  sendDataToProcessing('S', Signal);
  if (QS == true){
    sendDataToProcessing('B',BPM);
    sendDataToProcessing('Q',IBI);
    fadeVal = 255;
    QS = false;
   }
   ledFadeToBeat();
   delay(20);
}
```

Appendix B

Algorithm of Android App:



The entire developed Android code is available at the following link-

https://drive.google.com/folderview?id=0ByQbqzH7RHTaQ3hOdXNneWRMcDA&usp=drive_web [Online]

Appendix C

MATLAB Code:

```
clear all
clc
fs=25;
sig=load('a_ppg_raw.txt');                    % input PPG signal
plot(sig)
xlabel('Samples','FontSize',18)
ylabel('PPG signal amplitudes','FontSize',18)
title(' Raw PPG signal','FontSize',18)
set(gca,'FontSize',18)
grid on
figure
y=detrend(sig,'linear');                      %y=sig-mean(sig);
plot(y)
xlabel('Samples','FontSize',18)
ylabel('PPG signal amplitudes','FontSize',18)
title('PPG signal through mean filter','FontSize',18)
grid on
set(gca,'FontSize',18)
figure

% Butterworth Filter

[b,a]=butter(4,.9,'low');                     % 4th order Low Pass filter
y1=filter(b,a,y);
f=fs*linspace(0,1,length(y1));
plot(f,y1)
xlabel('Frequency in Hz','FontSize',18);
ylabel('Filtered PPG signal amplitudes','FontSize',18)
title('PPG signal through low-pass Butterworth filter','FontSize',18)
set(gca,'FontSize',18)
grid on
figure
T=.04;
ts=T/100;
ycorr=xcorr(y1);                              % correlation in time domain
plot(ts*(1:length(ycorr)),ycorr/max(ycorr))
```

```
ycorr=xcorr(y1);
plot(ycorr)
xlabel('lags','FontSize',18);
ylabel('Auto-correlated PPG signal','FontSize',18)
title('Time domain analysis','FontSize',18)
set(gca,'FontSize',18)
grid on
figure
yfft=abs(fft(y1,4096));                        % DFT analysis
f1=fs*linspace(0,1,length(yfft));
plot(f1,yfft)
xlabel('Frequency in Hz','FontSize',18);
ylabel('FFT of PPG signal','FontSize',18)
title('Frequency spectrum analysis','FontSize',18)
set(gca,'FontSize',18)
grid on
figure
nfft=4096;
[pxx,F]=pwelch(y1,[],[],nfft,fs);              % Power spectrum density
plot(F,pxx)
xlabel('Frequency in Hz','FontSize',18);
ylabel('Power Spectrum Density of PPG signal','FontSize',18)
title(' Welch spectrum analysis','FontSize',18)
set(gca,'FontSize',18)
grid on
figure
pwelch(y1,[],[],[],fs,'two sided')             % HRV estimation
figure

%code for calculating BPM

beat_count=0;
for k=2:length(y1)-1
  if (y1(k) > y1(k-1) & y1(k)> y1(k+1) & y1(k) > 1)
    beat_count=beat_count+1;
  end
end
N=length(y1);
t=N/fs;
T=t/60;
bpm_1=beat_count/T

% Discrete wavelet Transformation (DWT) Filter
```

```
ydwt=dwt(y,'haar');                                  % Haar wavelets
f2=fs*linspace(0,1,length(ydwt));
plot(f2,ydwt)
xlabel('Frequency in Hz','FontSize',18);
ylabel('Filtered PPG signal amplitudes','FontSize',18)
title('PPG signal through Haar wavelet','FontSize',18)
set(gca,'FontSize',18)
grid on
figure
T1=.04;
ts1=T1/100;
ycorr1=xcorr(ydwt);                                  % correlation in time domain
plot(ts1*(1:length(ycorr1)),ycorr1/max(ycorr1))
xlabel('lags','FontSize',18);
ylabel('Auto-correlated PPG signal','FontSize',18)
title('Time domain analysis','FontSize',18)
set(gca,'FontSize',18)
grid on
figure
yfft1=abs(fft(ydwt,4096));                           % DFT analysis
f3=fs*linspace(0,1,length(yfft1));
plot(f3,yfft1)
xlabel('Frequency in Hz','FontSize',18);
ylabel('FFT of PPG signal','FontSize',18)
title('Frequency spectrum analysis','FontSize',18)
set(gca,'FontSize',18)
grid on
figure
[pxx1,F1]=pwelch(ydwt,[],[],nfft,fs);                % Power spectrum density
plot(F1,pxx1)
xlabel('Frequency in Hz','FontSize',18);
ylabel('Power Spectrum Density of PPG signal','FontSize',18)
title(' Welch spectrum analysis','FontSize',18)
set(gca,'FontSize',18)
grid on
figure
pwelch(ydwt,[],[],[],fs,'two sided')                 % HRV estimation
set(gca,'FontSize',18)

%code for calculating BPM

beat_count1=0;
```

```
for i=2:length(ydwt)-1
   if (ydwt(i) > ydwt(i-1) & ydwt(i)> ydwt(i+1) & ydwt(i) > 1)
      beat_count1=beat_count1+1;
   end
end
N1=length(ydwt);
t1=N1/fs;
T1=t1/60;
bpm_2=beat_count1/T1
```

Bibliography

[1] N.Daimiwal, M. Sundhararajan and R. Shriram, "Respiratory Rate, Heart Rate and Continuous Measurement of BP Using PPG", Int. Conf. Communication and Signal Processing, India, Apr. 3-5, 2014.

[2] ITead Studio. (2010). Robotshop. [Online]. Available: http://www.robotshop.com/media/files/pdf/rb-ite-12-bluetooth_hc05.pdf

[3] I. Reyes, H. Nazeran, M. Franco, E. Haltiwanger, "Wireless Photoplethysmographic Device for Heart Rate Variability Signal Acquisition and Analysis", 34 Annu. Int. Conf. IEEE EMBS, San Diego, California, Aug. 28-Sep. 1, 2012.

[4] V.S. Murthy, S. Ramamoorthy, N. Srinivasan2, S. Rajagopal, M. M. Rao, "Analysis of Photoplethysmographic Signals of Cardiovascular Patients", in Proc.,23 IEEE Annu. Int. Conf.,Eng. In Medicine and Biology Society, vol.3, 2001.

[5] S. Sarkar, A.K. Bhoi, G. Savita,"Fingertip Pulse Wave (PPG signal) Analysis and Heart Rate Detection", Int. Journal of Emerging Technology and Advanced Engineering, vol.2, issue 9, Sep. 2012.

[6] G. Joseph, A. Joseph, G. Titus, R. M. Thomas and D. Jose, " Photoplethysmogram (PPG) Signal Analysis And Wavelet De-noising", Int. Conf. Magnetics, Machines & Drives, AICERA-iCMMD, 2014.