Design and Fabrication of a Low Cost Heart Monitor using Reflectance Photoplethysmogram

Nur Ilyani Ramli, Mansour Youseffi and Peter Widdop

Abstract—This paper presents a low cost design of heart beat monitoring device using reflectance mode PhotoPlethysmography (PPG). PPG is known for its simple construction, ease of use and cost effectiveness and can provide information about the changes in cardiac activity as well as aid in earlier non-invasive diagnostics. The proposed device is divided into three phases. First is the detection of pulses through the fingertip. The signal is then passed to the signal processing unit for the purpose of amplification, filtering and digitizing. Finally the heart rate is calculated and displayed on the computer using parallel port interface. The paper is concluded with prototyping of the device followed by verification procedure of the heartbeat signal obtained in laboratory setting.

Keywords—Reflectance mode PPG, Heart beat detection, Circuit design, PCB design

I. INTRODUCTION

Heartbeat monitoring device (HBMD) is an electronic tool that detects physiological parameters and converts to usable heart rate reading. Heart rate is the number of times the heart beats in a minute and it is produced via depolarization at the sinoatrial and atrioventricular nodes in the heart.

The use of HBMD has evolved since its first invention. Monitoring of heart rate has become a common tool in the clinical domain because it appears sensitive to both physiological and psychological disorders. Initial use of HBMD was for clinical applications as a tool in diagnosis, prognosis and management of patients with health problem [2], [3]. The development and usage of the HBMD has taken place rapidly in other areas such as sports and also as heart rate variability (HRV) training device [4]-[6]. For example, the shape of the heart beat waveform reflects arterial changes correlated with hypertension. In addition, HRV training offers promise for the treatment of migraine, reduces asthma severity followed by improvement in cases of diabetic ulcer and frostbites. Furthermore, monitoring of human sexual arousal would contribute remarkably towards the analysis and treatment of sexual dysfunctions [6].

N.I. Ramli is with the School of Engineering, Design and Technology, University of Bradford, BD7 1DP UK and on study leave from Universiti Tun Hussein Onn Malaysia (UTHM), 86400 Parit Raja, Batu Pahat Johor MALAYSIA (e-mail: N.I.B.Ramli@bradford.ac.uk).

M. Youseffi is with the School of Engineering, Design and Technology, University of Bradford, BD7 1DP UK (e-mail: m.youseffi@bradford.ac.uk).

P.D. Widdop is with the School of Engineering, Design and Technology, University of Bradford, BD7 1DP UK (e-mail: P.D.Widdop@bradford.ac.uk).

Pulse oximetry, electrocardiography (ECG) and blood pressure monitor are amongst a few medical devices that display heartbeat reading digitally. However, the only medical device nowadays that extracts heartbeat signals and displays it in graphical form is the electrocardiography (ECG). Common problems with the ECG method is too many sensors and cables connection, fluctuations in the ECG signal baseline, drift, power line noise, motion artifacts due to electrode movement and electromyographic (EMG) interference due to muscular activity. Furthermore, ECG is not suitable for burn victims and continuous monitoring may not be tolerable and the conduction gel used may cause discomfort and inflammation to the skin.

In this paper, low cost design of heart beat monitoring device (HBMD) is proposed. The design employ reflectance mode photoplethysmography (PPG) to extract the pulse signal from the finger which is equivalent to the heart beat. The prototype consists of several stages to amplify, filter and digitize the extracted heart beat. Finally, interfacing technique via parallel port to computer is used to calculate and display the heart rate. The device will be simple to use and cost effective.

II. RESEARCH OVERVIEW

A. Normal Heart Rate

Generally, heartbeat can be felt at locations where large arteries are close to the surface of the body. The rate of heartbeat is measured in beats per minute (BPM). Normal heart rate may be indicated by the state of the natural pacemaker, SA node and the state of the conduction pathways. During resting, an infant rate of heartbeat ranges around 130-150 beats per minute, toddler's around 100-130 beats per minute and an older child's about 90-110 beats per minute. Adult's heart rate is about 60-100 beats per minute with average heart rate at about 70 beats per minute for males and 75 beats per minute for females. However, heart rate varies depending on the age and health condition of a person. In a well-trained athletes, it can be significantly low ranging around 40-60 beats per minute [1].

B. Common Methods of Heart Beat Detection

Cardiac arrhythmias are often first detected by simple but nonspecific means using a stethoscope or manual feel for peripheral pulses. These cannot usually diagnose specific arrhythmias, but can give a general indication of the heart rate

and whether it is regular or irregular. The simplest specific diagnostic test for assessment of heart rhythm is the electrocardiogram (ECG) followed by a Holter monitor which basically is an ECG recorded over 24-48 hour period in order to detect arrhythmias that may happen briefly and unpredictably throughout the day. These are all non-invasive diagnostic tools.

A more advanced study of the heart's electrical activity can be performed to assess the source of the aberrant heart beats. This can be accomplished in an Electrophysiology study which is minimal invasive equipment. The procedure uses a catheter to observe the electrical activity from within the heart. Additionally if the source of the arrhythmias is found, the abnormal cells are often ablated and the arrhythmia can be permanently corrected.

TABLE I
ADVANTAGE AND DISADVANTAGE OF DIAGNOSTIC TOOLS

Diagnostic Approach	Advantage	Disadvantage
Manual Pulse Detection	- Many site of the body - Simple and easy to perform	- Low accuracy
Stethoscope	- Cheap and Ease of use - Magnification - Speed and comfort	- Not practical in noisy environment - Specialized device
Electrocardiogram (ECG)	- Painless - Easy to perform	- Tedious preparation - Quite expensive - Specialized equipment
Holter Monitor	- Painless - Long monitoring up to 48 hours - Portable	- Cumbersome and interfere with bathing or showering - Quite expensive
Electrophysiology Study (EPS)	- Determine the precise mechanism of an arrhythmia problem - Perform curative catheter ablation - Evaluate the need for implantation of a cardioverter-defibrillator in patients	- Time consuming - Expensive - Minor and major bruising - Risky - Death

C. Plethysmography Method

The heartbeat monitoring device proposed uses Plethysmography method instead of skin electrode. Plethysmography (PG) is a method used to determine and register the variations in blood volume or blood flow in the body. These transient changes occur with each heart beat. There are several different types of PG, which vary according to the type of transducers used. The common types include air [7], impedance [8]-[10], photoelectric [9],[11]-[18] and strain gauge [19].

Each type of PG measures the changes in blood volume in a different manner and is used in different types of applications [15]. Compared to other types of Plethysmography, PhotoPlethysmography is a non-invasive technique easy to set up, simple to use and low in cost [13]-[15], [17], [20]. In addition, PPG has the ability to take measurement without having direct contact with the skin surface, as is necessary with other Plethysmography methods [18].

III. HEARTBEAT MONITORING DEVICE (HBMD)

A. Block diagram

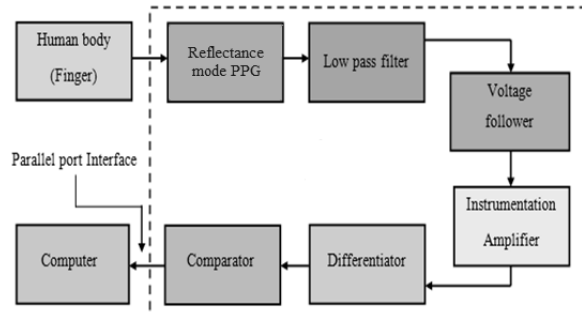Fig. 1 shows the block diagram of the proposed device composed of number of stages.

Fig.1 Block diagram of proposed HBMD

B. Sensor design

In Reflectance mode, the super bright LED and the Light Dependant resistor (LDR) are placed side by side on the surface of the finger as shown in fig.2. For this PPG Mode, red light with viewing angle of 15° (625nm, 20000mcd) is allowed to pass through the finger and the transmitted light is detected by an LDR. The LDR (photoconductive cell ORP12) is a photo resistor made up of material called cadmium sulfide (CdS) with its resistance decreases when the intensity of the light detected increases. This type of sensor can be used to measure arterial blood volume from many surfaces of the body where transmission mode cannot.
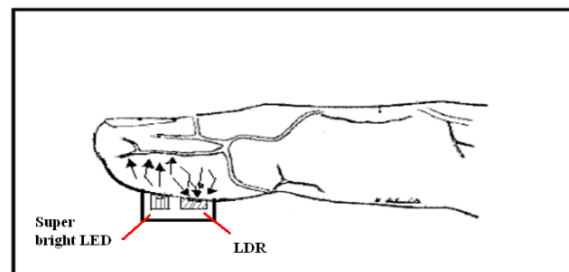

Fig.2 Reflectance mode PPG

C. Low Pass Filter Design (LPF)

Filters are often used in electronic systems to emphasize signals in certain frequency ranges and reject signals in other frequency ranges. Therefore, the low pass filter used in this circuit design aimed to filter low frequency signals from the PPG signal at a cut-off frequency of 0.16Hz thus ideally filtering out high frequency noise as well.
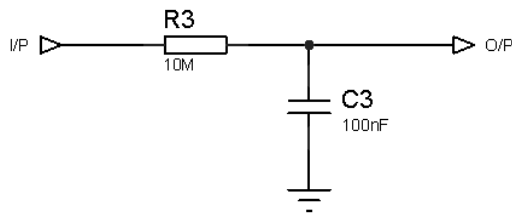
Fig.3 Passive low pass filter circuit

The values for the resistor and capacitor were calculated as follows:

$$f_{cut-off} = \frac{1}{2\pi RC} \qquad (1)$$

$$0.16 = \frac{1}{2\pi RC}$$

Choosing C = 100nF, R will be approximately to:

$$R = \frac{1}{2\pi(100 \times 10^{-9})(0.16)}$$

$$R = 10M\Omega$$

When Cut-off frequency is 0.16Hz, the phase shift is calculated as below:

$$Phase\ Shift\ \emptyset = -\arctan(2\pi fRC) \qquad (2)$$

$$= -45° \ or \ 0.785 \ rad$$

Frequency response of the filter to be nearly flat for low frequencies and the entire input signal is passed directly to the output, resulting in a gain of nearly 1, unity until it reaches the Cut-off Frequency point (fc). This is because the reactance of the capacitor is high at low frequencies and blocks any current flow through the capacitor. After this point the response of the circuit decreases giving a slope of -20dB/ Decade roll-off as signals above this frequency become greatly attenuated, until at very high frequencies the reactance of the capacitor becomes so low that it gives the effect of a short circuit condition on the output terminals resulting in zero output.

D. Voltage Follower Design

Voltage follower is made up of a non-inverting amplifier. As the input signal is connected directly to the non-inverting input of the amplifier the output signal is not inverted resulting in the output voltage being equal to the input voltage, V_{out} = V_{in}. This makes the voltage follower circuit ideal as a unity gain buffer circuit because of its isolation properties as impedance or circuit isolation is more important than amplification. The input impedance of the voltage follower circuit is very high, typically above 1MΩ as it is equal to that

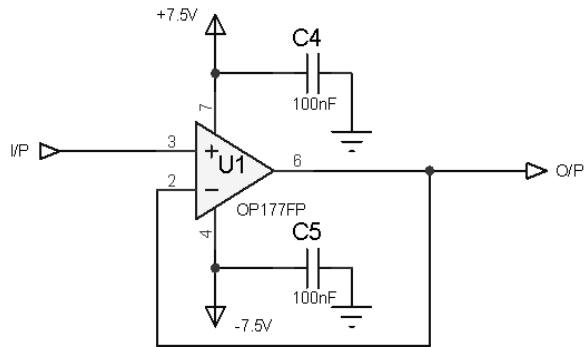of the operational amplifiers input, since an ideal op-amp condition is assumed.

Fig.4 Voltage follower circuit

For this HBMD an OP177 IC is used. The reason for choosing OP177 IC is because it is one of the highest precision operational amplifiers currently available. It is low in noise and also cost affective.

In voltage follower circuit, R_{in} has increased to infinity and R_f reduced to zero, the feedback is 100% and V_{out} is exactly equal to V_{in} (Labeled as I/P) giving it a fixed gain of 1 or unity.

As the input voltage V_{in} (I/P) is applied to the non-inverting input the gain of the amplifier is given as:

$$V_1 = \frac{R_2}{R_2 + R_F} \times Vout \qquad (3)$$

$$Voltage\ Gain,\ Av = \frac{V_{out}}{V_{in}} \qquad (4)$$

$$Then,\ Av = \frac{V_{out}}{V_{in}} = \frac{R_2 + R_F}{R_2} \qquad (5)$$

Then the closed loop voltage gain of a Non-inverting Amplifier is given as:

$$Av = 1 + \frac{R_F}{R_2} \qquad (6)$$

Using (6), the Gain of the Voltage Follower is as follows when R_2 and R_F is zero:

$$Av_{follower} = 1 + \frac{R_F}{R_2}$$

$$Av_{follower},\ G = 1$$

E. Instrumentation Amplifier Design (IA)

The type of instrumentation amplifier used is an INA118. The INA118 is a low power, general-purpose instrumentation amplifier offering excellent accuracy. Its versatile 3-op amp design and small size make it ideal for a wide range of applications especially for the medical instrumentation purpose. The INA118 features a single external resistor which can set any gain from 1 to 10,000.
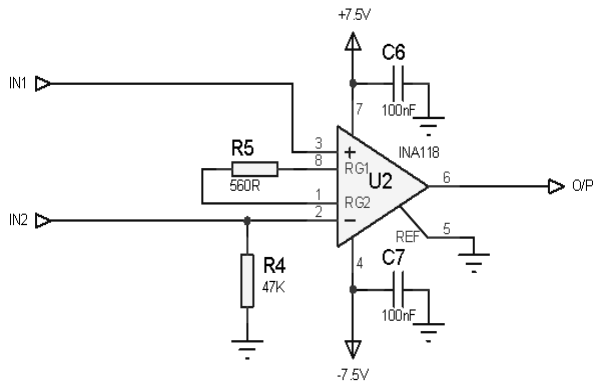

Fig.5 Instrumentation amplifier circuit

Instrumentation amplifiers are high gain differential amplifiers with high input impedance and a single ended output. They are mainly used to amplify very small differential signals, in this case the heartbeat signal. They also have very good common mode rejection (zero output when $V_1 = V_2$) in excess of 100dB at DC.
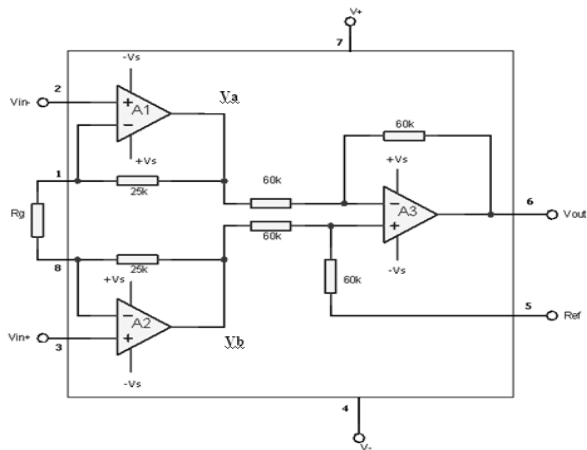

Fig.6 Internal circuit of component INA118

The negative feedback of the top op-amp causes the voltage at V_a to be equal to the input voltage V_{in-}. Likewise, the voltage at V_b is equal to the value of V_{in+}. This produces a voltage drop across R_g which is equal to the voltage difference between V_{in-} and V_{in+}. This voltage drop causes a current to flow through R_g, and as the two inputs of the buffer op-amps draw no current (virtual earth), the same amount of current flowing through R_g must also be flowing through the two resistors of the 25k. This then produces a voltage drop between points V_a and V_b equal to:

$$\frac{V_a - V_b}{vin^- - vin^+} = 1 + \frac{2(25k)}{Rg} \qquad (7)$$

The voltage drop between points V_a and V_b is connected to the inputs of the differential amplifier which amplifies it by a gain of 1 (assuming that all the "R" resistors are of equal value, in this case all resistors is equal to 60k). Therefore the general expression for overall voltage gain of the instrumentation amplifier circuit is:

$$Vout = (Vin^- - Vin^+)\left[1 + \frac{2(25k)}{Rg}\right] \qquad (8)$$

From (8), the value of R_g needed to achieve gain of 100 is calculated as below:

$$\frac{Vout}{(vin^- - vin^+)} = \left[1 + \frac{2(25k)}{Rg}\right]$$

$$G = 1 + \frac{50k}{Rg} \qquad (9)$$

$$Rg = 505.1\Omega$$

Nearest 1% Rg

$$Rg = 511\Omega$$

However, the nearest value of R_g available is 560Ω. Therefore, the Gain of the Instrumentation amplifier is not 100 but less as calculated below:

$$G = 1 + \frac{50k}{Rg}$$

$$G = 1 + \frac{50k}{560}$$

$$G = 90.3$$

F. Differentiator Design

This circuit performs the mathematical operation of differentiation that is it produces a voltage output which is proportional to the input voltage's rate-of-change and the current flowing through the capacitor. The resistor and capacitor forms an RC Network across the operational amplifier and the reactance (X_c) of the capacitor plays a major role in the performance. For the differentiator stage, an OP177 is used. This is the same type of IC used as the voltage follower.

As seen in Fig.7, the capacitor C_8 blocks any DC content only allowing AC type signals to pass through and whose frequency is dependent on the rate of change of the input signal. At low frequencies the reactance of capacitor C_8 is *high* resulting in a low gain (R_7/X_c) and low output voltage from the op-amp. At higher frequencies the reactance of the capacitor is much lower resulting in a higher gain and higher output voltage from the differentiator amplifier. Therefore the differentiator circuit becomes unstable and will start to oscillate. This is due mainly to the first-order effect, which determines the frequency response of the op-amp circuit causing a Second-order response which, at high frequencies gives an output voltage far higher than what was expected. To avoid this, the high frequency gain of the circuit needs to be

reduced by adding an additional small value capacitor across the feedback resistor R_7 which is the capacitor C_9.
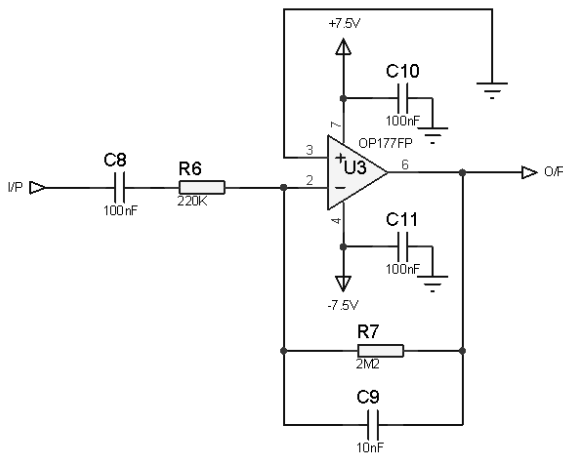

Fig.7 Differentiator circuit

Since the node voltage of the operational amplifier at its inverting input terminal is zero, the current, i flowing through the capacitor is as follows:

$$i_{in} = I_F \qquad therefore \quad I_F = -\frac{Vout}{R_F} \qquad (10)$$

$$Given \qquad Q = C \times Vin \qquad (11)$$

Where: Q is the capacitor charges
C is the capacitance
V_{in} is the voltage drop across the capacitor

Therefore the rate of change of this charge is:

$$\frac{dQ}{dt} = C\frac{dVin}{dt} \qquad (12)$$

However dQ/dt is the capacitor current i:

$$i_{in} = C\frac{dVin}{dt} = I_F \qquad (13)$$

$$Thus \qquad \frac{Vout}{R_F} = C\frac{dVin}{dt} \qquad (14)$$

From (14), the output voltage of the differentiator is as below:

$$Vout = -R_F C\frac{dVin}{dt} \qquad (15)$$

$$Vout = -R_7 C_8 \frac{dVin}{dt}$$

Therefore, the output voltage V_{out} is a constant $-R_7.C_8$ multiplies by the derivative of the input voltage V_{in} with respect to time. The minus sign indicates a 180° phase shift because the input signal is connected to the inverting input terminal of the operational amplifier. The gain of the differentiator is determined by the ratio of the feedback and input resistors:

$$A = -\frac{R_7}{R_6} \qquad (16)$$

Therefore the differentiator has gain of 10. As a result the circuit now acts like a Differentiator amplifier at low frequencies and an amplifier with resistive feedback at high frequencies giving better noise rejection.

G. Comparator Design

The final stage of the Heart beat monitoring device (HBMD) is the comparator. A comparator is a circuit that performs a comparison between two voltages (V_1 and V_2) and provides an output that indicates which of the two voltages is greater by switching its output either high (if $V_1 > V_2$) or low (if $V_1 < V_2$). For comparator design, an LM311 is used. Unlike comparator LM106 or LM710, the LM311 input currents is nearly thousand times lower. Although slower than the LM106 and LM710 (200 ns response time VS 40 ns), LM311 is less prone to spurious oscillations which is one of the desired feature for a comparator.

As with any op amp, the comparator has 3 basic behaviours: high input impedance which means very little current enters the inputs, low output impedance where the output can sink appreciable current and third very high gain where

$$V_{OUT} = \infty\ (V_{IN+} - V_{IN-}) \qquad (17)$$

The comparator generally is used simply to determine which of the inputs has a higher voltage.

- If $V_{IN-} > V_{IN+}$, then V_{OUT} goes as low as it can (ground).
- If $V_{IN-} < V_{IN+}$, then V_{OUT} goes as high as it can (the positive power supply, +V).
- If $V_{IN-} = V_{IN+}$, or more realistically if the two inputs are within some small difference of each other, the output is indeterminant.

If $V_{IN-} = V_{IN+}$, the output is uncertain. This is overcome by using positive feedback to add some *hysteresis* to the transfer characteristic.
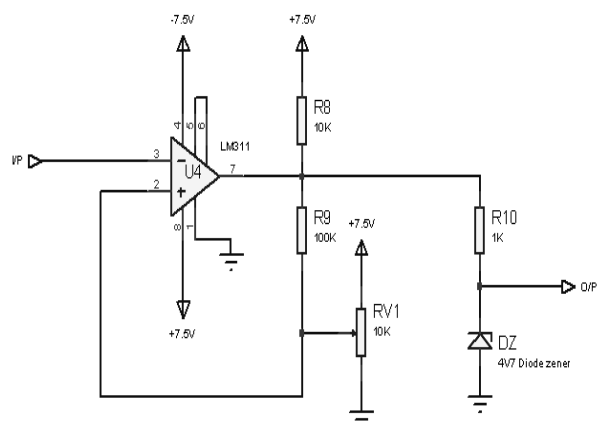

Fig.8 Comparator circuit

In the circuit, the trim pins (pins 5 and 6) act as unwanted auxiliary inputs. Since these pins are not connected to a trim-pot, they are shorted together. If they are connected to a trim-pot, a 0.01 μF capacitor C_1 between pins 5 and 6 will minimize the susceptibility to AC coupling. Furthermore, hysteresis (positive feedback) is used to prevent oscillation, and to avoid excessive noise on the output because the comparator is a good amplifier for its own noise.

This final stage is important because it involves the interface. The circuit basically produces the digital output instead of analogue which can be interfaced by using the parallel port. It is crucial that the input voltage to the parallel port is less or equal to 5 volt to prevent any damage. To achieve this, a 1KΩ resistor and a zener diode (4.7V) are connected to the output pin so that these two components will limit the output voltage to 5 volts or lower.

H. Parallel Port Interface

Parallel port is a simple and inexpensive tool for building computer controlled devices and projects. This port will allow the input of up to 9 bits or the output of 12 bits at any one given time, thus requiring minimal external circuitry to implement many simpler tasks. The port is composed of 4 control lines, 5 status lines and 8 data lines. It is commonly found on the back of the PC as a D-Type 25 Pin female connector and also known as the printer port.

There are other types of interfacing called serial port and also USB. Serial port would require a microcontroller, firmware, programmer or in-system-programmability, level translators, crystal, as well as many more passive components. Likewise, USB also requires a considerable amount of hardware, in addition to a much greater software and firmware investment. Furthermore, debugging USB hardware often requires very expensive protocol analyzers (MAXIM, 2004).

Therefore the parallel port was chosen as the interfacing technique because the goal was to build an interface which is quick, simple, and cheap. The simplicity and ease of programming makes parallel port the choice in displaying the results of the heart rate obtained. Even so parallel port also has drawback, which is a result of the operating system. Windows NT does not allow user applications to directly read/write hardware registers. Since all the computers on site are Window NT, a kernel driver known as Userport is used to overcome the problem.

For the purpose of processing and displaying of the heart rate results obtained on the computer, C programming language has been chosen. It is a powerful tool yet straight forward to use especially for beginners. Before writing the program, the programming flow chart is determined. This is to identify the important components needed to achieve the objective of processing and displaying of the heart beat results.

When the program is running, firstly, it will access and set the timer and start counting for 30 seconds. At this time, if the program detects the signal, i.e. a trigger from the hardware which means the voltage output of the hardware is high, then the program will plus the pulse by one. It will continuously record the pulse until 30 seconds. At the same time, the

program also displays HIGH in the text window. If there is no input detected, the text window will display LOW and will continue until an input is detected.

After the program has counted for 30 seconds, heart beat status will display the heartbeat result in beat per minute (Pulse in 30 sec*2) and shows the user's heart rate condition whether it is Abnormal, Normal, Weak or Dead. The user then may quit the program by pressing F2 or restart the program by pressing F1.
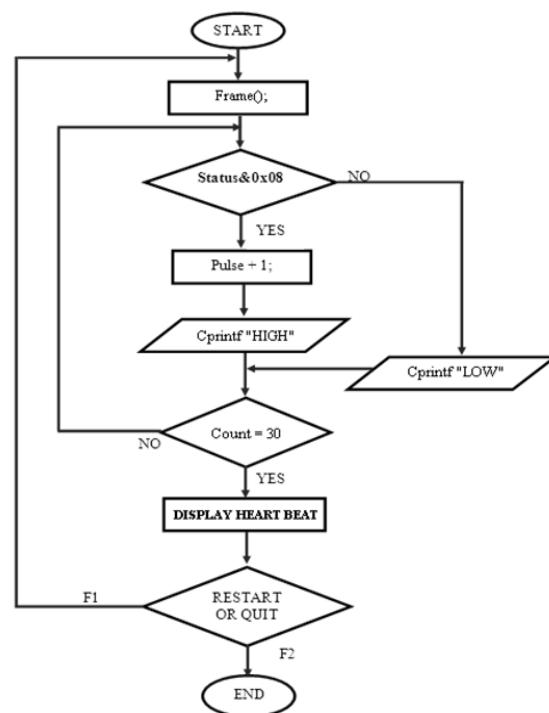
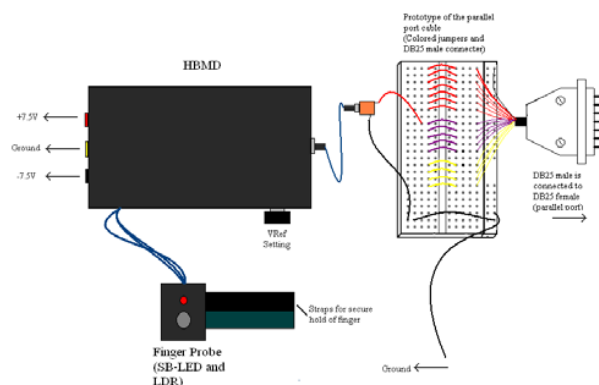Fig.9 Algorithm for heart rate analysis

Fig.10 Interconnection of the whole system of HBMD

Fig.10 presents the interconnection of the HBMD system to the computer using Parallel port interface. The procedure of using the HBMD is as follows:

1) The HBMD is set up as shown in Fig.10 ensuring the correct connection of the power supply cables and that the Ground wire from the DB25 connector is connected to the ground of the circuit (HBMD).
2) The power supply is turned on ensuring that it supplies ±7.5V.
3) Next, the male DB25 connector is connected to the female DB25 socket at the back of the CPU of the computer.
4) Index finger is then placed on the Finger probe and is secured using the strap provided.
5) Open Turbo C folder where the HBMD file was saved and double click on the Filename.EXE. The program will automatically start loading and counting the heartbeat.
6) To run the program again, F1 is pressed. Otherwise F2 is pressed to leave the HBMD program.
7) To turn off the HBMD, the DB25 connector (male) is first disconnected followed by switching off the power supply.

IV. RESULTS AND DISCUSSION

A. Prototype of HBMD

Prior to building of the actual hardware, circuit simulation using Proteus software (Isis) was performed on each stage of the circuit to ensure that the design functions according to its specification.

Experiment sessions were conducted by building and testing each functional block individually on the breadboard before proceeding to the next stage. Then the printed circuit board (PCB) was designed using Proteus software (Ares). One of the advantages of using Proteus Software (Ares) over other PCB design software is its 3D Visualization feature. Using 3D Visualization feature provided, the PCB designed can be visualized which will give an idea of what the PCB will look like when it is fabricated.

The PCB of the HBMD was placed into an enclosed rectangular box with the dimension of 5.0" X 2.7" X 1.7" (L X W X H) and the Finger Probe in a smaller rectangular box with dimension of 1.8" X 1.3" X 0.75" (L X W X H).
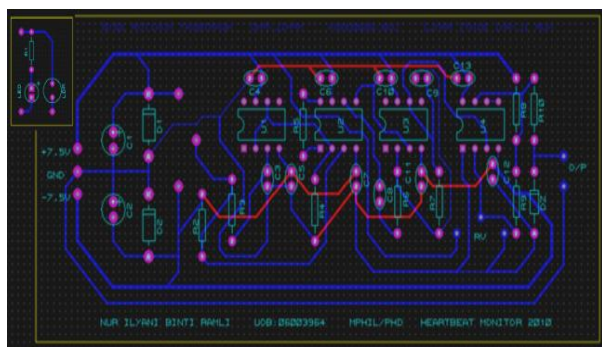

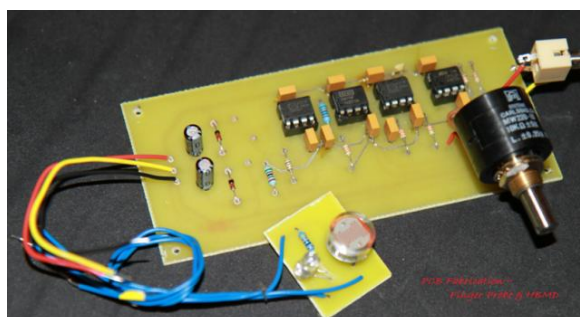Fig.11 PCB Design of HBMD (Proteus@Ares)

Fig.12 3D Visualization of PCB

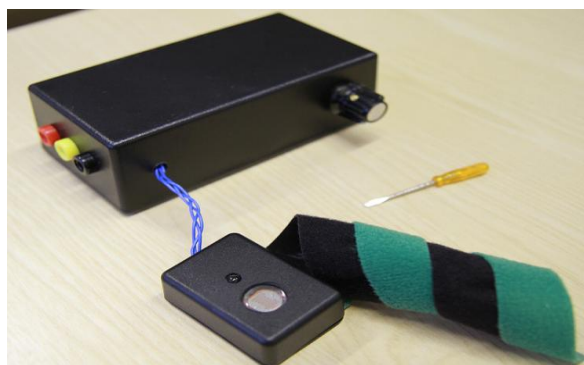Fig.13 PCB fabrication of reflectance PPG and HBMD

Fig.14 Casing for reflectance PPG and HBMD

B. Signal Verification of HBMD

The following are result obtained from experimental session conducted. As shown, the heart beat signal obtained is strong and stable but with 60Hz noise. The amplitude of the heart beat signal at this stage is approximately 45mV with sampling rate of 1s.
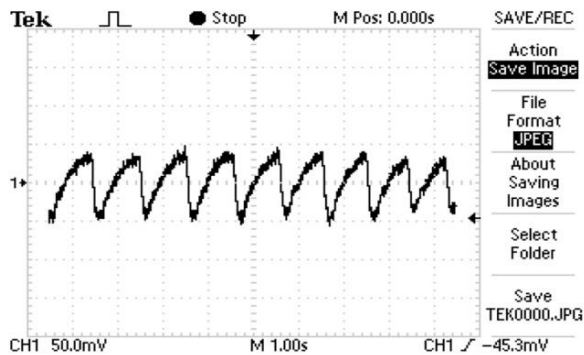
Fig.15 Heart beat signal obtained from reflectance PPG

To eliminate the high frequency component or the noise, a low pass filter is used. Comparing Fig. 15 and Fig.16, frequency noise of 60 Hz has been filtered out as expected. The filtered signal is required to have a SNR of 20 dB or greater, to ensure that the instrumentation amplifier is able to correctly convert the continuous signal to a higher amplitude signal form without producing false trigger due to noise.

In Fig.16, the dc signal obtained is very small (100mV) followed by the voltage follower which ensures the same signal is passed through to the instrumentation amplifier stage, however it verified that the filter is able to remove high frequency noise from the heartbeat signal.

The noise present in the signal has been reduced; however, the peak-to-peak amplitude has also been attenuated due to filter effects. To counteract this effect and condition the signal, the amplification is applied using the instrumentation amplifier. The signal was obtained at sampling rate of 1s.
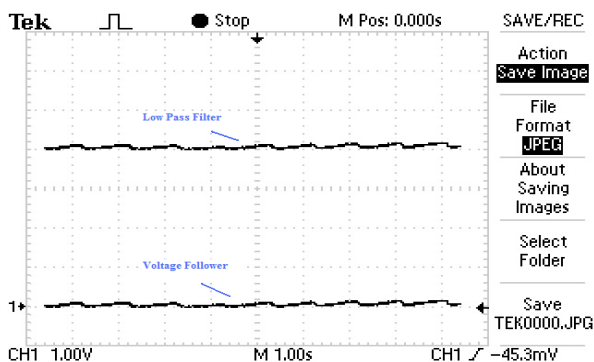

Fig.16 Heart beat signal obtained from low pass filter and voltage follower

Fig.17 shows the output signal of the instrumentation amplifier with respect to the differentiator stage. The instrumentation amplifier signal (7.5V) from the PCB is strong and the output signal of the differentiator is also clearer (≈10V). When the slope of the instrumentation amplifier signal is very steep, the signal of the differentiator also move faster in the opposite direction. Whereas when the instrumentation amplifiers signal

increase steadily, the slower the changes in signal of the differentiator.
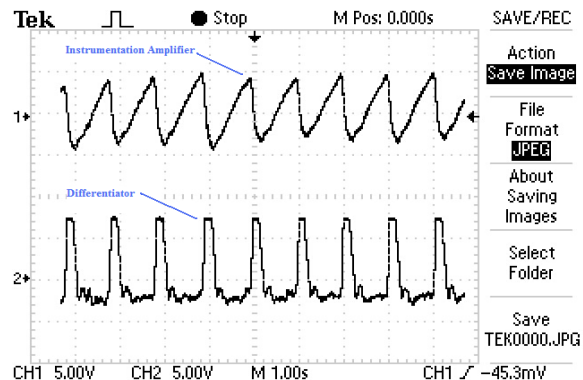

Fig.17 Heart beat signal obtained from instrumentation amplifier and differentiator

Fig.18 Heart beat signal obtained from differentiator and comparator

Finally, the output signal of the comparator is stable due to the use of positive feedback also known as hysteresis. The signal obtained at this point is around 4.7V which is below 5V. This is sufficient for the purpose of parallel port interfacing using the C programming.

C. Verification of Heart Rate Analysis Program

For heart rate analysis purpose, a reading is taken from a female age 29 in sitting position.

Heart rate of the subject is calculated using (18).

$$(BPM) = Freq\ (Hz) \times 60 \qquad (18)$$

$$Heart\ Beat\ per\ Minute = 1.04Hz \times 60$$
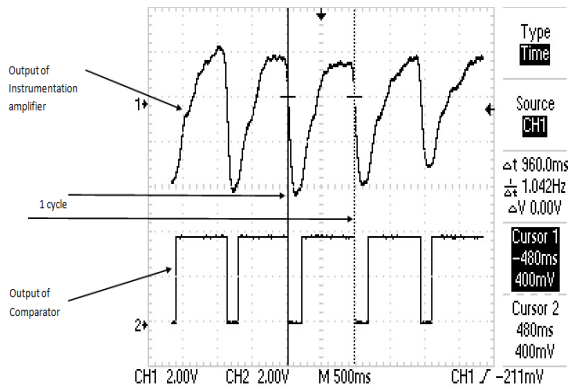
$$= 62\ Beats\ per\ Minute$$

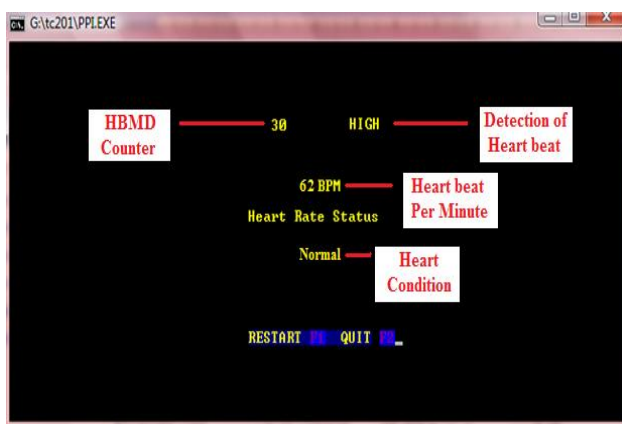Fig.19 Heart beat signal obtained from oscilloscope

Fig.20 Execution of heart rate analysis program

As depicted in Fig.19 and Fig.20, similar result is presented which indicates that the heart beat monitoring device proposed is able to perform the 3 phases stated earlier which is to detect the weak heart beat signal from the finger tip using Photoplethysmogram (PPG), filtering and amplifying the signal to be calculated and displayed on the computer using parallel port interface.

D. Cost Analysis

The electronic components were bought from RAPID and ONECALL Company based in UK. The final cost of the heart beat monitoring device prototype is less than GBP20. Therefore, objective of designing a low cost heart beat monitoring device has been achieved since most of the device on the market is more than GBP50.

V. CONCLUSION

Prototype of the heart beat monitoring device based on reflectance mode PPG was successfully designed and developed which detects the variations in blood volume from the index finger. However further validation is needed to further conclude the accuracy and the efficiency of the device in clinical setting. Therefore, future work involves testing of the device in different body position while carrying out different activities and focusing also on user finger circumferences. This is found to affect the accuracy of the device. Finally further research is required to investigate the potential of the proposed HBMD in performing Heart Rate Variability (HRV) analysis. It is crucial because HRV represents quantitative marker of autonomic activity, and a powerful tool in the recognition of the relationship between autonomic nervous system and cardiovascular mortality. HRV may be evaluated by using time domain, frequency domain and nonlinear methods of analysis such as Poincare Plot Analysis, Correlation dimension (CD) analysis and Detrended Fluctuation Analysis (DFA).

REFERENCES

[1] G. R. Faulhaber, "Design of service systems with priority reservation," in *Conf. Rec. 1995 IEEE Int. Conf. Communications,* pp. 3–8.

[2] W. D. Doyle, "Magnetization reversal in films with biaxial anisotropy," in *1987 Proc. INTERMAG Conf.*, pp. 2.2-1–2.2-6.

[3] C. Pallister, "*Blood physiology and pathophysiology,* Oxford, Butterworth-Heinemann," 1994.

[4] J. P. Boehmer, "Device therapy for heart failure," in 2003 *The American Journal of Cardiology,* 91, 53-59

[5] E.O. Robles De Medina, H. A. P. Peeters and F. H. M. Wittkampf, "Frank N. Wilson: the unipolar electrocardiogram," in 2000 *ACC Current Journal Review,* 9, 16-19.

[6] J.Achten & A. Jeukendrup, "Heart Rate Monitoring: Applications and Limitations," in 2003 *Sports Medicine* Volume 33, p. 517-538.

[7] F. O. X.Gamelin, S.Berthoin and L.Bosquet , "Validity of the Polar S810 Heart Rate Monitor to Measure R-R Intervals at Rest," in 2006 *Medicine & Science in Sports & Exercise,* 38, 887-893 10.1249/01.mss.0000218135.79476.9c.

[8] E.Peper, R.Harvey, I.M.Lin, H.Tylova and D.Moss, "Is There More to Blood Volume Pulse Than Heart RateVariability, Respiratory Sinus Arrhythmia, and Cardiorespiratory Synchrony?," in 2007 *Association for Applied Psychophysiology & Biofeedback,* 35, 54-61.

[9] A. N. Nicolaides, "Investigation of Chronic Venous Insufficiency: A Consensus Statement," in 2000 *Circulation,* 102, e126-163.

[10] F. S. Ermi & L.Marks, "Design of electrodes for pulse volume measurement/impedance Plethysmography," in 2000 *Bioengineering Conference Proceedings of the IEEE 26th Annual Northeast.*

[11] A.Santic & M.Saban, "Features of Pulse Plethysmography and a Comparison with Impedance Plethysmography," in 1992 *Engineering in Medicine and Biology Society, Vol.14. Proceedings of the Annual International Conference of the IEEE.*

[12] H. P. Schwan, "Electrical Properties of Body Tissues and Impedance Plethysmography," in 1955 *Medical Electronics, IRE Transactions on,* PGME-3, 32-46.

[13] J. N. Lygourus and P. G. Tsalides, "Optical-fiber finger photo-plethysmograph using digital techniques," in 2002 *Sensors Journal, IEEE, 2*, 20-25.

[14] V. S. Murthy, S. Ramamoorthy, N. Srinivasan, S. Rajagopal and M. M. Rao, "Analysis of photoplethysmographic signals of cardiovascular patients," in 2001 *Engineering in Medicine and Biology Society, 2001. Proceedings of the 23rd Annual International Conference of the IEEE.*

[15] J. Kraitl, H.Ewald & H.Gehring, "Analysis of time series for non-invasive characterization of blood components and circulation patterns," in 2008 *Nonlinear Analysis: Hybrid Systems, 2*, 441-455.

[16] J. Allen, "Photoplethysmography and its application in clinical physiological measurement," in 2007 *PHYSIOLOGICAL MEASUREMENT, 28*, R1–R39.

[17] A.Reisner, P. A.Shaltis, D.McCombie and H. H. Asada, "Utility of the Photoplethysmogram in Circulatory Monitoring," in 2008 *Anesthesiology, 108*, 950-958 10.1097/ALN.0b013e31816c89e1

[18] Y. S. Peck and P. R. Smith, "An Overview of Non-contact Photoplethysmography," in 2003 Loughborough, UK, Department of Electronic and Electrical Engineering, Loughborough University.

[19] H.Ugnell and P. Å. Oberg, "The time-variable photoplethysmographic signal; dependence of the heart synchronous signal on wavelength and sample volume," in 1995 *Medical Engineering & Physics, 17*, 571-578.

[20] K.Nakajima, T.Tamura and H.Miike, "Monitoring of heart and respiratory rates by photoplethysmography using a digital filtering technique," in 1996 *Medical Engineering & Physics, 18*, 365-372.

[21] B. Hyun Jae, C. Gih Sung, K. Ko Keun, K. Jung Soo and P. Kwang Suk, " Photoplethysmogram measurement without direct skin-to-sensor contact using an adaptive light source intensity control," in 2009 *Trans. Info. Tech. Biomed., 13*, 1085-1088.

[22] R. J.Whitney, "The measurement of volume changes in human limbs," in 1953 *Journal of Physiology, 121*, 1-27.

[23] R. Krishnan, B. Natarajan and S. Warren, "Two-Stage Approach for Detection and Reduction of Motion Artifacts in Photoplethysmographic Data," in 2010 *Biomedical Engineering, IEEE Transactions on, PP*,1-1.