Chapter 2

Design and development of blood volume pulse sensor and heart rate meter

Abstract

A low power, low cost sensor has been developed for sensing the blood volume pulse using transmission mode photoplethysmography (PPG) from the finger tip. A PIC microcontroller is used for the implementation of the measurement protocol. The wavelet denoising algorithm is used to suppress the noise component in the PPG signal. The main advantages of the proposed approach are the reduction in cost, dimensions and power consumption. The probe is well suited for the proposed measurements and is self-contained and portable.
2.1 INTRODUCTION
Photoplethysmography (PPG) is a non-invasive method for the detection of cardiovascular pulse waves propagating around the human body (1). It is based on the determination of optical properties of vascular tissue using a light source and a photodetector. The emitted light is reflected, absorbed or scattered by the blood and tissues, and the resultant transmitted light is measured using a suitable photodetector. There are two different modes of detection: transmission mode and reflection mode. In the transmission mode, the light source is placed on one side of the tissue and the detector on the other, and is limited to areas such as the finger, earlobe, or toe. In the reflection mode, where the light source and the photodetector are placed in parallel, the measurement of backscattered light from any skin surface is allowed. The intensity of light reaching the photodetector is measured and the variations are amplified, filtered and recorded as a voltage signal – PPG. Variations in the photodetector signal are related to changes in blood flow and blood volume in the underlying tissue (2-5).

The PPG signal has an ac component and a dc component. The ac component is synchronous with the heart rate and depends both on the pulsatile pressure and pulsatile blood volume changes. The dc component of the signal varies slowly and reflects variations in the total blood volume of the examined tissue. The ac pulse shapes are indicative of vessel compliance and cardiac performance and the amplitude is usually 1 to 2% of the dc value. The pulsatile component of the PPG signal (blood volume pulse) has a
contour which includes a content descriptive of vascular health. Two such parameters are the minimum rise time (MRT) and the stiffness index (SI) which can be derived from the contour analysis of the peripheral pulse wave (6-9). Also, PPG has considerable potential for tele-medicine including the remote home monitoring of patients. Miniaturization, ease-of use and robustness are key design requirements for such systems (9). So, the design of a high fidelity photoplethysmograph transducer with signal conditioning circuit is indispensable to obtain an extremely accurate and noise-free signal of the blood volume pulse (BVP). There are quite a few developments in this field and researchers around the world have been attempting to design PPG sensors for long term, short term and ambulatory recording. An artifact resistant power efficient finger-ring PPG sensor which could be used for long term recording with much emphasis on mechanical design is available in the literature (10). A pulse oximeter that could be fully powered by human heat was designed with considerable power saving (11). This design demands for fairly expensive thermocouples. A reflective photoplethysmograph earpiece sensor for long-term pervasive monitoring was developed for long term pervasive monitoring (12). During recent times a LED-LED based photoplethysmographic sensor which is low in cost and power consumption was proposed (13). The proposed sensor in the present studies has good spectral sensitivity, increased and adjustable resolution, and avoids the need for expensive and precise operational amplifiers although its performance was not compared with that of a commercial equipment. In this chapter the design and development of a simple blood volume pulse sensor and a heart rate meter using a PIC
Design and development of …

microcontroller is described using the acquired blood volume pulse signals. The use of a microcontroller makes the system compact, cost effective and easily programmable. The use of wavelet transform helps in removing the high frequency noise from the acquired blood volume pulse.

2.2 PRINCIPLE OF MEASUREMENT

Photoelectric plethysmography, also known as photoplethysmography (PPG) or optical plethysmography is a non-invasive method to detect cardio-vascular pulse wave that propagates through the body using a light source and a detector. Hertzman and Spielman (2) were the first to use the term “photoplethysmography” and suggested that the resultant “plethysmogram” represented volumetric changes in the blood vessels. Light traveling through biological tissue (e.g., the fingertip or earlobe) undergoes scattering and absorption by different absorbing substances such as skin pigmentation, bones and arterial and venous blood. The arteries contain more blood during systole than during diastole and their diameter increases due to the increased pressure. This blood volume change occurs mainly in the arteries and arterioles but not in the veins. The pulse shape and amplitude can vary with the relative position between the detector and the vessel under study. The amplitude of the volume pulsations with each heartbeat is correlated with the flow. In PPG, the volume under study, depending
on the probe design, can be of the order of $1 \text{ cm}^3$ for transmission mode systems.

The principle of operation of PPG is based on the fact that light is attenuated when it is shown onto the skin and the attenuation shows variation depending on the volume of blood entering the tissue under observation. The optical system arrangement for acquiring the PPG signals in the transmission mode is shown in figure 2.1.

![Fig 2.1: Optical system arrangement for transmission mode photoplethysmography](image)

The Beer-Lambert law for light transmission in an absorbing medium, as shown in equation 2.1, is the primary basis for the functioning of the photoplethysmograph.
\[ I = I_0 e^{-\mu_a d} \] (2.1)

where \( I \) is the transmitted intensity, \( I_0 \) is the input light intensity, \( \mu_a \) is the absorption coefficient, and \( d \) is the distance between source and detector.

Since blood is a highly scattering medium, the Beer Lambert's law must be modified to include an additive term ‘G’ due to scattering losses and a multiplier ‘B’ to account for the increased optical path length due to scattering and absorption. Now, the modified Beer Lambert’s law which incorporates these two additions is shown in equation 2.2 and this equation has been used as the basic theory for our measurement (14,15). This approach helps to develop an understanding of the absorbance of light as it passes through living tissues and explains why and how PPG works.

\[ I = I_0 e^{-\left(\mu_a dB + G\right)} \] (2.2)

Here \( G \) is a factor dependent on the measurement geometry and the scattering coefficient of the tissue interrogated. The wavelength of the source used is of significant importance in PPG. Light sources that operate in the near red (600–700 nm) and near infrared (880–940 nm) of the spectrum are most effective because whole blood has a relatively small absorption at wavelengths greater than 620nm (9).
2.3 DESIGN OF THE HARDWARE SETUP AND THE HEART RATE METER

PPG has been used in oxygen saturation measurement, heart rate monitoring, and the assessment of peripheral circulation and large artery compliance. However, the waveform of the PPG signal may be affected by the contacting force between the sensor and the measurement site (16). The sensor head was designed by taking this factor into consideration. The PPG signals were recorded by using this self-designed transmission mode PPG sensor. The GaAs infrared light emitting diode (LED55c) with a peak wavelength of 940nm and a matching silicon phototransistor LF141 were used for the present studies. The intensity of the LED is sufficiently low to minimize excessive local tissue heating. The LED is driven by a continuous current between 10mA and 15 mA, the value of which can be varied by signals from the microcontroller depending on the tissue perfusion characteristics (17). We have not used any modulation and the power level used is around 5mW. The sensor probe is fabricated using a commercial material called delrin and is enclosed within a compartment so that the effect of ambient light is minimized. Delrin is a crystalline plastic which offers an excellent balance of properties that bridge the gap between metals and plastics. It has good machinability, high fatigue endurance, and low moisture absorption. The acquired blood volume pulse waveform is fed to a microcontroller (16F873) which is programmed to calculate the heart rate in beats per minute (bpm), as each heart beat gives rise to a blood volume pulse. The details involved in the mechanical design of the probe and the block diagram of the hardware setup are as shown in fig 2.2.
2.3.1. Mechanical design

The material that we had used for making the springs in the probe head is SS302 grade steel. Using the standard data available for the material we were able to calculate the shear stress (18). We have used a spring of diameter 4mm and the wire diameter was 0.5 mm. Using these values and by assuming the deflection of the spring to be 10 mm, we arrived at the value of the applied force. In the probe head we have used six springs which were arranged in parallel. Assuming the value of the modulus of rigidity as $0.79 \times 10^5$ Pa for the material used and correlating the same to the applied force, one can calculate the number of turns to be used for the spring. Using standard relations (18), the length of the spring was evaluated as 39 mm. As shown in figure 2.2, the portion of the sensor head above the dotted line can be lifted so as to accommodate the finger of the subject. The contacting force was determined using a flexiforce sensor which was attached to the inner side of the probe. The contacting forces were determined by using a force to voltage circuit and the PPG signals were recorded for different contacting forces. The contacting forces could be varied from 0.2 to 1.8 N, but however it has been observed that the contacting force is within the range of 0.2 to 0.5N for satisfactory acquisition of PPG signals from various subjects.
The sensor head was designed such that it was small in size, light in weight and rigid in construction. It was designed in such a way that it does not either occlude the flow of blood, or allow the finger to move freely so that motion artifact can be greatly reduced and a nominal contacting pressure can be applied to the finger to get a satisfactory PPG signal. Ambient light is prevented from reaching the detector so that the effect of artifact on signal acquisition is minimized. The LED and the detector inside the probe were mounted on a reflection free
surrounding. To prevent the signal from being affected by electromagnetic interference a shielded cable was used to pick up the signal from the probe end.

2.3.2. Hardware design

The block diagram of the designed hardware setup is shown in figure 2.3. The sensor probe houses the LED and the photodetector. The slot for introducing the finger is elliptical in shape with major axis and minor axis of 2.1 cm and 1 cm, respectively. The design of our hardware is for short term recording of PPG signals. The output from the photodetector is fed to a voltage follower so that there is no loading of the original signal. The circuit is implemented using operational amplifiers (OP 07 and LF 356) which are ultra low offset and FET input amplifiers, respectively. The baseline restoration circuit helps in eliminating the low frequency baseline fluctuations, which if present will corrupt the acquired signal. A switched capacitor filter (low pass, 8th order) helps in eliminating signals above 10 Hz, within which limit the BVP signals are expected to occur. The special feature of switched capacitor filter is that it has a cut off frequency which has a typical accuracy of ± 0.3% and is less sensitive to temperature changes. The band-limited signal is then amplified, clamped so that only the pulsatile component of the PPG waveform is recorded. The final output is fed to a microcontroller for calculating the heart rate and displayed using an LCD module. The program has been developed using assembly language.
The heart rate acquisition circuit is shown in figure 2.4. Depending on the amplitude of the signal that is acquired from the subject, a proper thresholding is also incorporated so that erroneous beats are eliminated. The PPG signal that is acquired from different subjects depends on their tissue perfusion characteristics. So the peak to peak value (Vp-p) of the acquired signal is calculated from which the value of 0.5Vp-p is evaluated. The middle values of 0.5Vp-p to Vmax and 0.5Vp-p to Vmin are then calculated. Only if the incoming signal has amplitudes above and below these two values respectively it will be considered for incrementing the counter which counts the number of beats. The above procedure is adopted to eliminate erroneous beats from being counted. The resolution of the ADC used is 10 bits. The PPG signal was sampled at a rate of 500 Hz. There is a beep circuit which is included along with the heart rate meter so that once the heart
rate goes below 60 bpm or above 100 bpm an alarm will be produced. A dual EIA driver/receiver (MAX232) chip can be used for further interfacing with a computer. The use of a PIC microcontroller makes the system very cost effective, with a very low requirement for additional external components. Programs can be modified with much ease and the whole system can be made compact by using microcontrollers.

Fig 2.4: Heart rate acquisition circuit
2.3.3. Denoising of PPG using wavelet transform

In order to avoid the high frequency noise superimposed on the PPG signal denoising technique was employed. Wavelet analysis is a time scale analysis rather than a time frequency analysis. This also helps in extracting the relevant time-amplitude information from a signal (19). It can also be used to improve the signal to noise ratio based on our prior knowledge of the signal characteristics. The wavelet transform decomposes the signal into different scales with different levels of resolution by dilating a single prototype function, the mother wavelet. The differences between different mother wavelet functions (e.g., Haar, Daubechies, Coiflets, Symlet etc) consist in how the scaling of signals and the wavelets are defined. To have a unique reconstructed signal from wavelet transform, we need to select the orthogonal wavelets to perform the transforms. The wavelet decomposition results in levels of approximated and detailed coefficients and the high frequency noise is normally seen in the detailed coefficients. By proper low pass filtering we were able to remove noise to a satisfactory level. The acquired signal was subjected to denoising using wavelet transform (19,20). The Daubechies2 (db2) wavelet was used and it was found that with five levels of decomposition, the high frequency noise could be satisfactorily eliminated. The denoised version of the acquired PPG signal is shown in figure 2.5. The denoised PPG signals were employed for analysis.
2.4. RESULTS AND DISCUSSION

The blood volume pulse sensor developed here has been tested in its heart rate configuration. The BVP was recorded from 5 individuals using the proposed setup. The amplitude of the signals obtained from the designed setup, which were in the range of 100 to 200 mV, were boosted in the last stage (as shown in figure 2.2) to 5V before being fed to the analog input channel of the PIC microcontroller. The heart rate displayed using the designed HR meter seemed to correlate well with

Fig 2.5: Recorded PPG and its denoised version
the beat-beat estimates calculated using the R-R interval of an electrocardiogram. The ECG and PPG signals recorded using a clinical instrument (L&T Micromon 7142 L) and the PPG signal recorded using the developed setup and their respective FFT’s in respect of a subject are shown in figures 2.6–2.8, respectively.

Fig 2.6: ECG recording and its FFT (using L&T Micromon)
Fig 2.7: PPG recording and its FFT (using L&T Micromon)

Fig 2.8: PPG recording and its FFT (using developed sensor)
There seems to be a close correlation among all of them with respect to the heart rate parameter. From the plots of the heart rate variation over a period of one minute derived from both the developed sensor and the clinical instrument, as shown in figure 2.9, we were able to see that the sensor developed by us is quite comparable with that of a clinical instrument. The accuracy of our present sensor was found to be within ±3 bpm for the subjects we had studied. The use of wavelet transforms for denoising the acquired signals makes the developed method more versatile. The heart rate determined using the present method and those calculated using the R-R interval of the electrocardiogram for five subjects are shown in Table 2.1 and they show good agreement. The calculation of the parameters can be made simple by making use of a microcontroller. The signal when recorded with the forearm of the subject placed at a comfortable position, and approximately at the heart level is found to be of good quality. The acquired signals may be analysed offline using Matlab.

Fig 2.9: Heart rate variation with respect to time for recordings from the developed sensor and ECG.
(a) heart rate (bpm) from proposed sensor and (b) heart rate (bpm) from ECG.
2.5. CONCLUSIONS

The optoelectronic setup designed and developed here is simple, easy to use, and can be used in home health care monitoring because of its small size and portability. The use of microcontroller for heart rate monitoring makes the system miniaturized and also modular in nature. The complex interfacing of external peripherals like ADC/DACs can be eliminated. The material that has been used for making the sensor head has excellent machinability and hence can be used for mass production of cost effective, self contained portable probe heads. The probe head can be operated in wet environments or at elevated temperatures with little effect on the performance or dimensions.

Simple and cost effective stand-alone systems could be developed for

Table 2.1: Heart rate determined using the proposed method and ECG

<table>
<thead>
<tr>
<th>Sl. no.</th>
<th>Heart rate (bpm) using the proposed method</th>
<th>Heart rate (bpm) using ECG</th>
</tr>
</thead>
<tbody>
<tr>
<td>1.</td>
<td>72 ± 2</td>
<td>72 ± 1</td>
</tr>
<tr>
<td>2.</td>
<td>65 ± 3</td>
<td>67 ± 4</td>
</tr>
<tr>
<td>3.</td>
<td>82 ± 1</td>
<td>83 ± 3</td>
</tr>
<tr>
<td>4.</td>
<td>78 ± 3</td>
<td>80 ± 2</td>
</tr>
<tr>
<td>5.</td>
<td>69 ± 1</td>
<td>68 ± 2</td>
</tr>
</tbody>
</table>

41
deducing indices related to cardiovascular system non-invasively using the present setup.

References