ELEC 243: Electronic Measurement Systems

Lab Report #10: Pulse Oximeter Design Project
Group 2: A. Alem and K. Francis
Group 7: M. Gagneja and Y. Zhang

I. Introduction

Electronic measurement systems play a key role in everyday life and are fundamental to the technologies in use today. The importance of electronic measurement systems in the field of healthcare is immense, as they are necessary for monitoring patient vitals. In this lab, we seek to design and build a pulse oximeter that would function in monitoring an individual's oxygen saturation and heart rate. This technology is implemented around the world on most patients in hospitals or those undergoing emergency care.

II. Design Goals and Requirements

In designing and building our pulse oximeter, we seek to meet a series of requirements and design goals. First and foremost, we must meet the definitions for a pulse oximeter. Pulse oximetry measures pulse rate and blood oxygen concentration by sending infrared and red light through a peripheral body part, such as a fingertip or earlobe, and interpreting the output.

The absorption of the light, which is in two different wavelengths, can be interpreted as a ratio to determine pulse rate and blood oxygen concentration. This occurs because oxygenated hemoglobin absorbs more infrared light and transmits more red light, while deoxygenated hemoglobin transmits more infrared light and absorbs more red light. The R/IR ratio can be calculated from the photodiode signal with each of the wave's respective root-mean-square values of intensity. A ratio of approximately 0.43 is expected for a perfect SpO_2 level, which would be 100%. A healthy SpO_2 value ranges from 95-100%.

The input to our system consists of the red and infrared light that travels through a fingertip. Our system then takes in this information and performs a series of signal amplifications, filtrations, and calculations to interpret the waveforms into the desired measurement values.

The output of our system is ideally the heart rate in beats per minute and the SpO_2 in percent. Typically, we expect an output of a stable indication of pulse at approximately 50-80 BPM if the subject is resting, and an SpO_2 at approximately 90-100 percent. The system must function accurately under ambient lighting conditions. We utilize a Nellcor compatible finger probe containing an emitter/detector combination to send light through the finger.

III. The Design Process

In order to organize the signal conditioning processes necessary, we first created a block diagram laying out the steps to reaching the desired output provided the input waveform. Our design was inspired by the process used by Texas Instruments in their "Single-Chip Pulsoximeter Design Using the MSP430" (the MSP430 is a microcontroller)[1]. Our initial block diagram can be seen in Appendix A. The steps within our block diagram outline the process, which is separated into analog and digital components. First, the LED driver sends red and infrared light through the LEDs into the finger, and the photodiode detects the light on the other side of the finger. This signal then is amplified through a transresistance amplifier to be converted to a voltage, followed by an inverting amplifier because our understanding of TI's model led us to believe that our signal would be inverted at this point. The signal enters the DAQ after which point it is in a digital form. We then apply an Infinite Impulse Response (IIR) filter, or bandpass filter, in order to take only the certain range of frequencies that would correspond to the desired signal and also to reduce noise.

However, once we actually began to create the device, we modified and specified our initial design. The first step of the process was to verify the working condition of the LED; we generated an input square wave for the LED and found that there was indeed a large DC component of our signal, approximately one order of magnitude larger than the AC peak to peak voltage.

We then needed to convert the current, which was outputted by the photodiode, to a voltage with a reasonable magnitude. This was necessary because the signal needed to be in voltage form to enter the DAQ. Since the signal's final destination is the DAQ, the exact voltage inputted is trivial since the computer can essentially analyze any voltage. However, we wanted to output a large enough voltage that can be easily discernable and checked for the sake of debugging with an oscilloscope. Thus, we decided to design a transresistance amplifier than can output a voltage magnitude between .1 V and 1V. This is a reasonable voltage range that can easily be generated with the resistors in our kit as well as large enough to be detected and observed. After a series of calculations, it was determined that a 680 kΩ resistor in parallel with a 10 pF capacitor generated the impedance necessary to convert the input current into a voltage within our desired range. The calculations are located in Appendix C-2.

We considered several different approaches to removing the DC component of the signal. We initially ran into trouble with trying to take the average RMS and input two signals into the DAQ. However, once we were able to troubleshoot the latter, we realized we could use a high-pass filter on the DC signal since it is at a signal frequency of 0 Hertz. We tried multiple values on the high-pass filter ranging from 0.25 to 500 Hertz, and set to 70 Hertz which both reduced the 60 Hz noise and removed the DC component

Initially we attempted to demultiplex our signal by using a combination of analog and digital techniques, which would have required each DAQ assistant to have multiple inputs and outputs. Although we later learned that this is possible to do using different channels, at the time we decided to try another, simpler approach because we did not know how to set up the multiple channels. Instead, we were able to complete the whole process of demultiplexing on LabView by devising a method in which we manipulated the vertical shift of the square wave to separate the R and IR components of the signal. The details of this method are further explained in our Final Design section.

IV. Final Design

The final design of our pulse oximeter includes a virtual instrument on LabView and the analog components on our breadboard. The process starts with the generation of a square wave in our VI, which is sent to our breadboard to power the LEDs within the finger probe. The light from the red and infrared LEDs passes through the finger and is sensed by the probe's photodiode, and the resulting signal is first processed by analog circuits on the breadboard, and then sent through the analog to digital converter and further conditioned by the LabView VI, which also displays the resulting measurements of SpO_2 and pulse rate values. A general block diagram is located in Appendix B.

Analog Components

The first component of our device is the LED driver circuit, which powers the LED in the Nellcor compatible finger probe. A schematic for the circuit is shown below.

Figure 1: LED Driver Circuit
(DRAWING)

The initial signal is generated by Labview, which sends out a simulated 500 Hz, 10 V peak to peak square wave into the breadboard through the DAQ. The resulting current is then sent through a

470 Ω resistor in order to decrease the current before it reaches the LED circuit. By representing our diode circuit with ideal diodes, we were able to calculate that a resistor value of 470 Ω was a sufficiently high value to not blow the LEDs with too large of an input current. This resistor also does not severely reduce the current and can still produce enough to stimulate the LEDs with a 10 V input. Calculations are included with further detail in Appendix C-1. Due to the diode arrangement in the circuit, the red (R) LED (660 nm) is only illuminated when the voltage across the diode is positive, and the infrared (IR) LED (910 nm) is only illuminated when the voltage across the diode is negative. Therefore, we alternate the ON state of both LEDs at a 500 Hz frequency and 5 V voltage for each.

The alternating light wavelengths then travel through the finger being measured. Some light is absorbed by the skin, tissue, and blood, and the rest is received by the photodiode. In response, the photodiode produces a current which alternates between the value representing the transmitted R wave and the value representing the transmitted IR wave. The resulting multiplexed waveform includes a DC voltage component, caused by constant light absorption by non-pulsatile blood, skin, and tissue, and an AC component caused by the pulsatile volume of arterial blood.

Our next analog component amplifies the photodiode signal in order to get values that we can work with more easily in mathematical manipulation. The photodiode amplifier circuit is shown below.

Figure 2: Photodiode Amplifier Circuit
(DRAWING)

In this circuit, we utilize a transresistance amplifier. After a series of tests, we decided on a resistance value of 680 kΩ for amplification in parallel with a small capacitance value of 1 pF to reduce clipping of the waveform. This waveform signal is then sent into LabView through the DAQ converter, and further analyzed with our pulse oximeter virtual instrument (VI).

Digital Components

The block diagram for our VI is located in Appendix D. Our LabView VI runs within a while loop, and contains three main components: the square wave generator, the SpO_2 signal processing and calculations, and the pulse rate signal processing and calculations. The square wave generator simply consists of a Simulate Signal subVI, which sends a square wave with 500 Hz frequency and 5 V amplitude to a DAQ Assistant, which sends that square wave signal to the breadboard. On the breadboard, this signal is sent to the probe, and is also sent back to LabView.

The second DAQ Assistant in our VI obtains two signals: the square wave and the signal from the photodiode. These two signals are first multiplied by ten, and then split to separate the square wave from the photodiode signal. The photodiode signal is sent to both of the remaining components, to be processed and return the SpO_2 and pulse rate measurements.

In order to determine a measurement for SpO_2, we first had to demultiplex the signal, to obtain V_{RMS} values for both the red light and infrared light components of the signal. To do this, we multiplied the signal by shifted and scaled versions of the square wave. We divided the square wave by 100, so that it would have an amplitude of 0.5 V, and then sent the wave along two branches of wire; we shifted one of the waves up by 0.5 and the other we shifted down by 0.5 and multiplied by -1. This resulted in two square waves, both alternating between 0 and 1, which are exactly out of phase, so while one wave is at 1, the other is always at 0, and vice versa. We then split the photodiode signal into two wires, and multiplied one of the square waves by each of the copies of the photodiode signal. This gives us two signals, one of which contains the transmitted signal from the red LED and one of which contains the transmitted signal from the infrared LED. From this point, we send each signal through a highpass filter with a cutoff frequency of 70 Hz, which removes the DC component of

the signals, as well as blocking any 60 Hz noise that might be in our signal. Each of these filtered signals is sent through a Statistics subVI, which returns the RMS values of the signal. We divide the V_{RMS} from the red light signal by the V_{RMS} from the infrared light signal to obtain a ratio of voltages. This ratio is squared to obtain a power ratio ($P=V^2/R$), since power is proportional to light intensity. This gives us the R/IR ratio that we need to calculate SpO_2. Based on our research, we determined that the relationship between R/IR and SpO_2 can be described by the following equation:

SpO_2 = -25 x (R/IR ratio) + 110 (Oak et al.)

This equation is implemented on LabView, and then the calculated SpO_2 value is displayed on the front panel.

In order to determine the pulse rate, we were able to use a much simpler signal processing procedure. First, the amplified photodiode signal was sent through a bandpass filter with cutoff frequencies of 0.5 Hz and 2.5 Hz. These frequencies were chosen because they correspond to pulse rates of 30 bpm and 150 bpm, which are on the outer ranges of the rates we would expect to measure. Additionally, using this relatively small range, we were able to filter out noise that could negatively affect the accuracy of our measurements. The filtered signal is analyzed using the Tone Measurements subVI, which returns the frequency which corresponds to the largest magnitude in the signal's spectrum. This frequency should correspond to the pulse rate in Hertz, so we just multiply by 60 seconds/1 minute to obtain a value in bpm. In order to get a stable reading, we set up a running average using a shift register. So, the calculated bpm value is send to the shift register, which sends the value to the next iteration. The bpm value from the current and previous iteration are averaged, and the resulting value is displayed on the front panel as the BPM measurement.

V. Testing

Our device is simple to use. The patient inserts their finger into the probe, and holds still. The resulting R/IR, SpO_2, and BPM are numerically displayed on the Front Panel. Additionally, there is a graph of the pulse displayed along with the values.

We tested our device on 4 different individuals, though they were all from a similar age, gender, and lifestyle demographic. The test subjects were four female Rice University students ranging from 19-20 years old. Each subject placed their finger in the probe and held it as still as possible. The values resulting from their tests are displayed in the table below. The front panel for Mehek's test is attached in Appendix E.

Table 1: Pulse Oximeter Testing Values for Four Trials

Subject	R/IR	SpO_2	BPM
Mehek	0.4309	99.23	74.34
Anita	0.5757	95.61	79.90
Kathleen	0.4376	99.06	63.99
Julia	0.5896	95.26	77.89

We had some issues with our testing as far as stability of values at the beginning of each trial. Due to the mechanical issues with our finger probe (as it is not the highest quality), the values for BPM would jump very suddenly whenever the finger was moved even slightly. Additionally, the SpO_2 values do change slightly based on the tightness of the probe around the fingertip. Most of the

fluctuations or deviations in our values from what is expected are likely due to this problem. However, the trends remained the same for each subject over a number of trials. That is, Julia's and Anita's SpO_2 values were always less than Kathleen's and Mehek's, so the pulse oximeter was consistent.

VI. Conclusion

One advantage of our design is that the small bandwidth of our filter in the signal conditioning for our pulse rate calculations gives us a very clean signal, which displays as a smooth pulse waveform and returns a stable pulse rate value. However, this can also be a disadvantage, because it limits the range of pulse rates that our device is able to accurately measure. Additionally, the running average incorporated into our VI allows for more accurate readings of pulse rate. By averaging two iterations to calculate our pulse rate, we obtain more stable values, which would not fluctuate as widely with small irregularities in pulse detection.

In this lab, we learned a lot about signal conditioning and the capabilities of LabView programs, and we were able to apply what we have learned this semester about electronic measurement systems to a real-world bioengineering application. We succeeded in meeting our design goals and requirements, and in building a functional pulse oximeter using a Nellcor finger probe, our breadboard, basic circuit components, and a LabView virtual instrument.

References:

1. Chan, Vincent, and Steve Underwood. *A Single-Chip Pulsoximeter Design Using the MSP430*. Tech. Dallas: Texas Instruments Incorporated, 2012. <http://www.ti.com/lit/an/slaa274b/slaa274b.pdf>.
2. Oak, Sang-Soo, and Praveen Aroul. Rep. no. SLAA655. Texas Instruments, Mar. 2015. Web. 22 Apr. 2015.