Optical Glucometer Interface

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Developing a Data Collecting System for Near-Infrared Biosensing Applications

Prepared for EE4BI6 - McMaster University

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Developing a Data Collecting System for Near-Infrared Biosensing Applications

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ABSTRACT

Near-Infrared spectroscopy has been used reliably for glucose identification and quantification in chemistry lab environments. With an appropriate processing algorithm, it should be possible to measure glucose concentration in blood non-invasively. The goal of this project is to develop a data-collecting system that can be used with a range of IR photo detectors. The primary concern is to eliminate as much noise and external interference as possible, in as high of a capture resolution as possible to allow for more detailed analysis. By utilizing a combined analog current to digital converter packaged in a single chip, noise can be minimized and the transference of data can be better preserved due to the higher resilience of digital information towards noise. The cleaner signal combined with the high resolution ADC makes for an ideal experimentation setup for further research in this topic, as well as other related optical sensing applications. The device should be able to stream 20-bit of live sensor data in real-time through a serial port (or USB) into a computer for further analysis in MATLAB or LabView. The theory behind the device, hardware design, experimental results, efficacy and further research potentials of the device will be discussed.

Key words: Glucometer, Near-Infrared, Optical, Non-Invasive, human-computer interfacing, high-resolution, data collection, customizable sensor

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Ryan Tirtariyadi

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INTRODUCTION

As smartphone usage grows, consumers are carrying powerful personal computer that are as compact as a candy bar. The main advantage of carrying a smart phone is the convergence of one's mobile devices. People are now carrying a single electronic device for their personal organization, communication and entertainment. Some examples are Windows Mobile based devices, BlackBerrys, iPhones and Android based devices.

The goal of this project was to integrate the glucometer with the smartphones by offloading the processing and user interface requirements of glucometers to the smartphone. Modern smartphones carry CPUs with clock speeds going as fast as 614MHz. Calculations required to process near-infrared absorption data are relatively simple and do-able on widely available smartphone CPUs. Therefore, a biosensing attachment to a smartphone that utilizes its processing power should be feasible.

Diabetics have two options to monitor their blood glucose. The first is by periodically piercing their skin and collecting a sample of blood to be tested on a glucometer. Then they administer a dose of insulin to regulate their blood glucose levels as necessary. The second method is by having a continuous monitoring system, either implanted under the skin or integrated into an insulin pump. The latter option is more expensive and less accessible. [3,4,5]

The pinprick method of blood glucose testing is uncomfortable and causes reluctance to regular testing. This introduces risk of devastating complications, most of which are also costly to treat. A less invasive testing method can promote regular testing and better glucose control. In addition, by minimizing the device into a simple attachment to something as ubiquitous as a mobile phone, it will be less cumbersome to carry and more accessible to a wider range of patients.

The primary complication to this project is the difficulty in extracting the glucose information from the NIR absorption data. The challenges in designing such a system will be discussed in further detail. Due to this roadblock, the project has been refocused.

The primary goal is now to design and construct a data collection platform on which large amounts of relevant data can be gathered for further analysis. While the mobile or wearable integration is still a possibility, the scientific basis to ensure reliable and accurate glucose measurement still needs much groundwork.

This report will discuss the challenges in designing a widely versatile device while maintaining accuracy and low cost.

LITERATURE REVIEW

The data-gathering phase of the project has been very discouraging towards the prospect of implementing an accurate and reliable optical glucometer for in vivo glucose measurements. The experimental results that were successful or relevant were done with very elaborate laboratory setup under very controlled conditions. This suggests difficulty for any real-world adaptation, if at all possible. The major issue with data gathering is that there are no confirmed example of a optical glucose monitoring system that were deemed to be sufficiently accurate and/or reliable. Therefore, this project is founded purely on an experimental topic much more suited for graduate or doctorate research subject. [1,7]

A recent review of the various companies attempting to produce and manufacture a non-invasive glucose-monitoring device revealed that this is a topic of high-interest among the biomedical engineering community, and that the solution has eluded scientists and engineers for over two decades. Referred to as 'Hunting the Elusive Turkey', the endeavour has cost dozens of companies hundreds of millions of dollars. [3]

The general consensus at this point is that not one company has reliably managed non-invasive glucose monitoring, and any device that were advertised to be capable of the task has been proven to be unreliable, inaccurate, unsafe or all of the above. [3, 5, 7]

However, bits and pieces of information from dozens of journal articles and spectroscopy literature revealed that glucose does have an absorption characteristic in the NIR range. Some studies concluded that it is possible to perform such a measurement given the appropriate source-detector separation using a 1600nm IR source. [13] Another experiment successfully implemented non-invasive glucose measurement by using impedance spectroscopy.[1] However, the study also learnt that their sensor was detrimentally affected by changes in temperature when used outside a controlled lab environment (when patients go home). [3]

What are currently available in the field are various methods of sugar detection and quantification in isolated lab settings. For example, near-infra red spectroscopy is being used to measure the sugar content of fruits and other food products. These applications exploit the various scattering and transflectance properties in the 1300, 1450 and 1550 nm wavelengths. [3,7,11]

DESIGN CONSIDERATIONS

The working principle is essentially an extension to the already established pulse oximeters. Red and Infrared transmission through the skin is measured and compared at corresponding pulse peaks in order to correlate to values to Beer-Lambert predictions relating to blood oxygen concentrations. Molecules inside the human body may absorb a light source of the correct wavelength and intensity. Every molecule has different absorption spectra due to their physical shape and size differences. If there is a specific absorption signature or physical characteristic to the glucose molecule, then it is plausible to theorize that it may not only be detectable, but also quantifiable. [17,18]

Hardware

In order to make this a very versatile system, a modular platform is preferred. The source-detector pair can be mix-and-matched to accommodate different experimentation setup. The interface unit is also designed in such a manner that allows the manipulation of the source with a pulse sequence. The method of data extraction is also by nature a very flexible affair because every aspect of the ‘measurement’ can be manipulated. Thee two important details will be discussed in detail.

The Sources

It is often necessary to pulse the source in a specific duty cycle in order to multiplex the different wavelengths. In pulse oximetry, the most common configuration used is two LEDs of different wavelengths and a single photodetector. The photodetector is aligned to detect light coming from a single LED by time-division. At high frequencies, way beyond the human heartbeat, the pulsing is practically transparent to the system and contributes no error to the measurements.

The interface board has a DB-9 serial port connector built onto them to allow for different kinds of light sources to be attached. The switches built onto the board can allow for selective source activation if a single source module were built with multiple sources. They are simply mapped individually to the different pins of the serial connector. The benefit of this is the ease of setup configuration when setting up complicated experiments. The hardware needs only minor adjustments for a wide-range of tests.

The source module can be easily sterilized. A glass tube protects the contact point to subject or sample, which can be easily washed/rinsed for repeat use. If the glass should break under use, it's a simple matter to replace the protection if the sensor assembly inside was not damaged. The glass enclosure also enables various possible sensor/source placements otherwise impossible when using bare sensors and sources. (ie. inside the mouth)

To maintain a compact size, great care was taken in calculating the minimum clearance of the components (LEDs, resistors, etc) mounted on the breadboard inside the test tube. The calculation methods are displayed in detail in Appendix B.

The connection is completely shielded because it is based off of a DB-9 serial cable. In addition to the ground pin, the cable is wrapped in foil and has ground shield conduction to the board, which is then grounded to the same reference as the chip's analog ground. The details of the pin-out connections are displayed in detail in Appendix C

This system is interested in extracting glucose; therefore wavelengths of 940nm, 1300nm, 1450nm and 1550nm were selected. Studies have shown these to be the primary wavelengths of interest when examining various sugar absorbances.

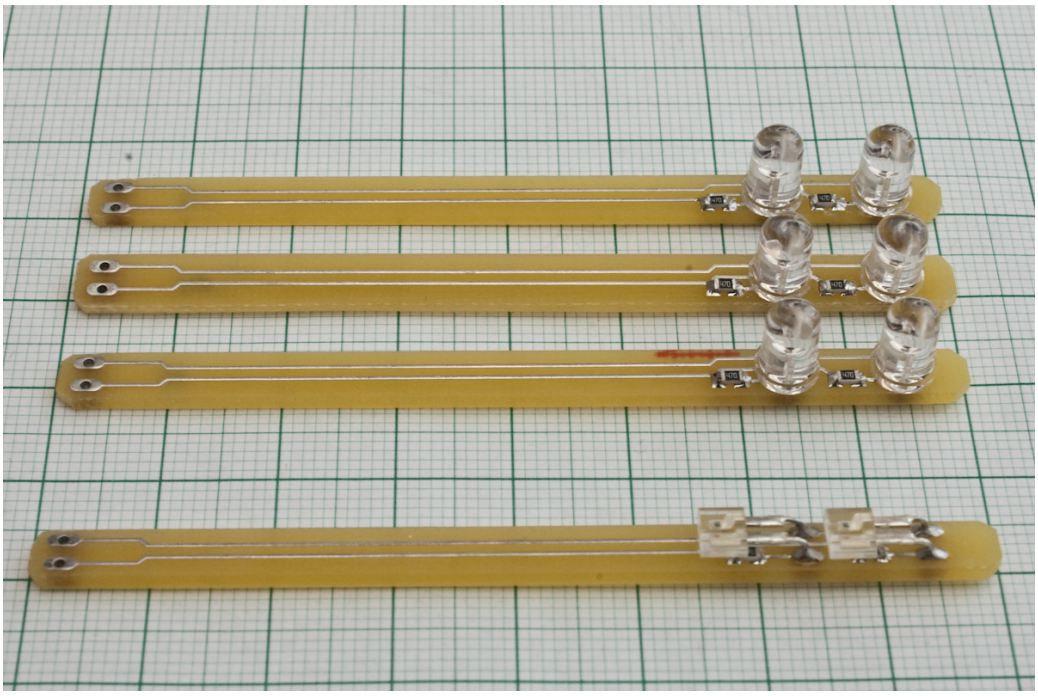


Figure 1: Source assemblies: from top to bottom: 1550nm, 1400nm, 1300nm and 940nm.

The Detectors

The photodetector assembly is also similarly designed to be flexible. The detector module is identically shielded and protected by a glass casing to allow for easy sterilization and reuse. The detector assembly can have a light source built-in, to allow for transflectance or reflectance tests. Switches on the interface board controls the activation of this source (Dual-Mode). Two photodiodes were selected; a typical 600-1100nm range serving to sense the more typical 660nm (red) and 940nm (IR), and a second one at 700-1800nm for the NIR applications.

The primary photodetector of interest is the THOR FGA10, which works in the 700-1800nm range. In spectroscopy, glucose and various other sugars absorb primarily in the 1400-1500nm range. The appropriate range, combined with a relatively flat and linear spectral response makes this InGaAs photodiode the perfect sensor for the purposes of this project. The FGA10 has an effective active area of 1mm in diameter. This allows for highly sensitive readings.

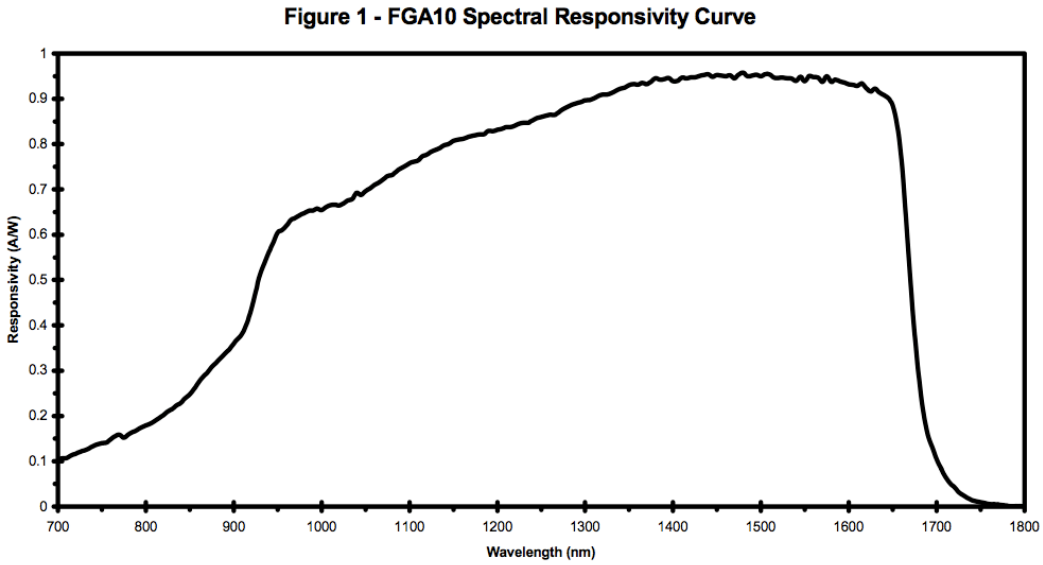


Figure 2: THOR FGA10 Spectral Response.

The same glass enclosure is applied to the detector assembly. This allows for easy cleaning and re-use. The ability for quick re-use is important when considering applications where the sensor assembly needs to be as close as possible to a non-epidermal tissue, or mucosal tissue. By using a glass tube, the sensor can be placed inside subject's mouth, against the inner cheek – where there are no epidermal layers and reflectance is minimized.

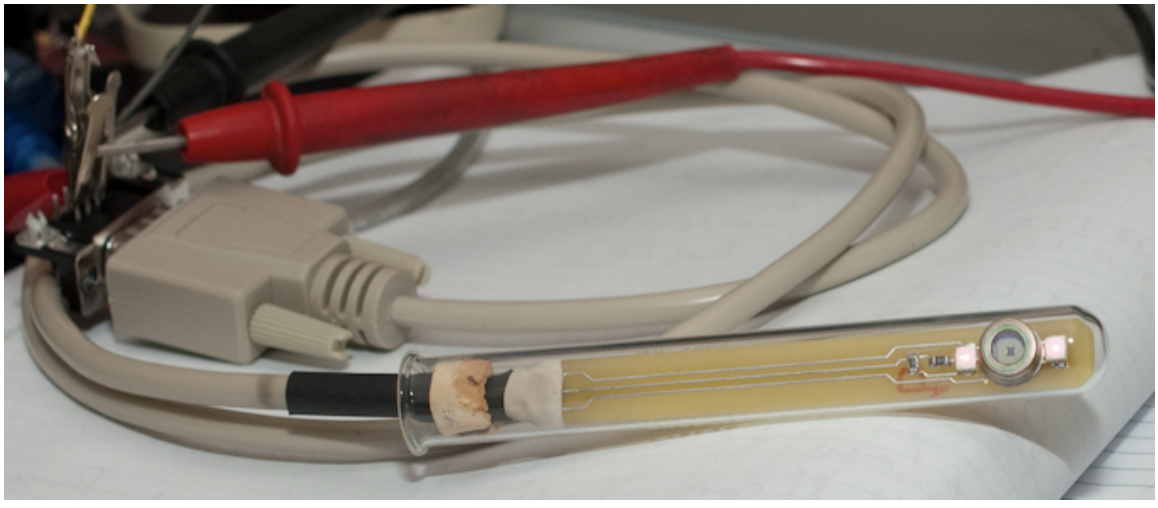


Figure 3: Detector Assembly (Dual-Mode) in test tube wired to a Serial cable.

The Chip – Texas Instruments DDC112

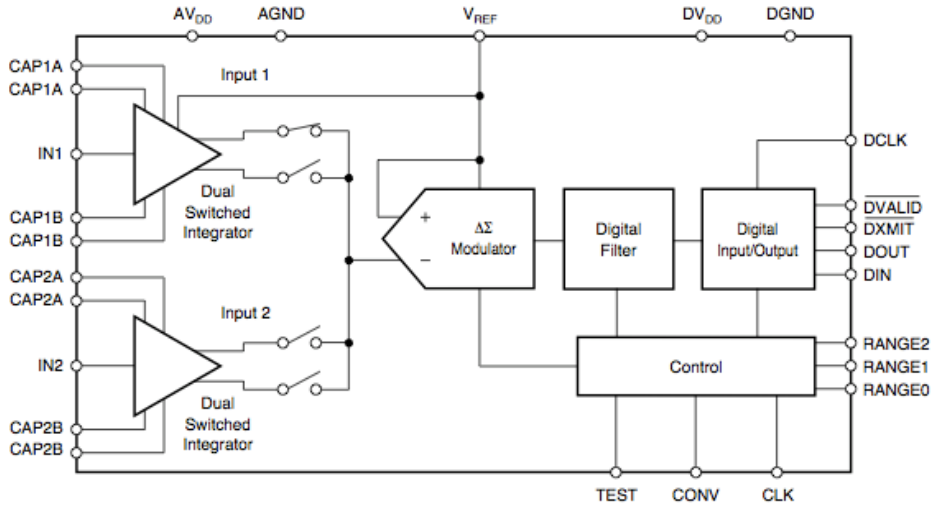


Figure 4: DDC112 Block Diagram.

A 20-bit analog charge to digital converter was chosen for this project due to its high resolution and low-noise potentials. As a single-chip solution, the DDC112 combines an analog charge to voltage integrator with digital filtering and a 20-bit A/D.

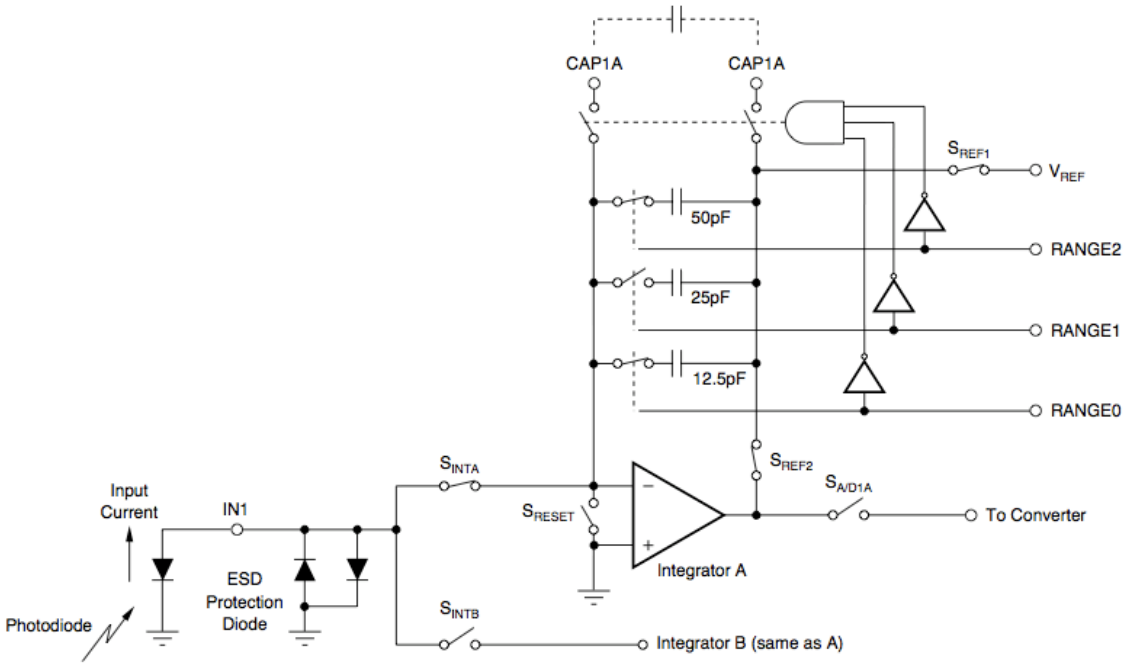


Figure 5: DDC112 Operational Front-end

The chip's measurement properties can be adjusted by switching its integration capacitor values built (with switches), and also by adjusting the integration time through software as programmed into the microcontroller driving the chip. With integration capacitor values adjustable from 12.5 to 1000 pF, the integrator can accept currents ranging from -0.2 to 1000 pC. Combined with varying integration time, we have an almost infinite dynamic range for measurement.

RANGE2	RANGE1	RANGE0	C_F (pF, typ)	INPUT RANGE (pC, typ)
0	0	0	External 12.5 to 250	Up to 1000
0	0	1	12.5	–0.2 to 50
0	1	0	25	–0.4 to 100
0	1	1	37.5	–0.6 to 150
1	0	0	50	–0.8 to 200
1	0	1	62.5	–0.1 to 250
1	1	0	75	–1.2 to 300
1	1	1	87.5	–1.4 to 350

TABLE I. Range Selection of the DDC112.

Figure 6: DDC112 Integration Capacitor Ranges

The flexibility of this chip is very promising for the purposes of this project because the goal is to gather a variety of data, or even unexpected types of information. For most medical research usage, 20-bit is excessive. However, due to the nature of the substance, overlapping signals need to be separated, and minute changes in the signal are preserved using a higher resolution detector. It should be noted, however, that this might also contribute to the difficulties related to noise later on.

Software

Smartphone Applications

The original goal was to integrate the optical glucometer into a smartphone as an attachment. Offloading the primary processing tasks to the smartphone's CPU and displaying glucose info almost instantaneously. To achieve this, a sufficiently powerful hardware is necessary, along with a stable and robust platform for development. Last, but not least, the development platform needs to be accommodating.

For this reason, the iPhone platform was chosen over Android, WebOS (Palm Pre) or BlackBerry. The iPhone OS has a well-established SDK which is geared towards consumer and professional applications development. In the past 18 months, hundreds of thousands of apps have been developed, and a good number has found its medical niche. Current applications exist which can track drug consumption, drug administration (for medical professionals), treatment dosages, medical imaging viewing tools, bioinstrumentation apps, etc. The platform has proven itself reliable, secure and robust enough to handle medical applications.

For glucose monitoring purposes, integration with a smartphone also opens up the possibility for cloud-based tracking. The meter can update a cloud-based database for constant monitoring or a warning system for physicians over the Internet.

Driver Development

The DDC112 needs to be driven by a microcontroller. The hardware aspects necessary for sensing and measurement are enclosed in a single chip. However, the software is completely external. Every high-level aspect of the chip's operation can be controlled externally from the microcontroller. This high-level approach lends simplicity to programming the chip. All we are concerned about is the integration time and data extraction from the 10-bit A/D Converter. The data extraction method will be discussed in the next sections.

All of the driver development is done using plain text C code with CCS compiler, and programmed into the 16F877A with MPLAB IDE's ICD2 programmer. The 16F877A is mounted on the PIC DEM 2 Plus demo board. There are many peripheral components on

the demo board that are not necessary for our purposes. However, the platform was chosen for simplicity and expedience. In reality, a simple Arduino Duemilanove would probably do the job just as well.

METHODOLOGY

Hardware Design

Preliminary Design and Testing – The Pulse Oximeter

To familiarize myself with the challenges in acquiring a light signal, a simple prototyping board (breadboard) implementation of a pulse oximeter was constructed. The primary components include a red and infra-red source transmitted through the tip of a finger into a photodiode. This photodiode is then connected to a trans-impedance amplifier, and then into a low-pass filter and a low-gain pre-amp before being passed into the analog inputs of the LabView DAQ board.

It took heavy filtering and excessive tweaking before LabView displayed a usable photoplethysmograph. However, taking into consideration that this prototype was a very rudimentary setup on a breadboard, it revealed the sort of challenges one will encounter when working with small signals in regards to noise and artifacts.

A breadboard does not provide a very secure connection or conduction between the components, nor does it provide a very good isolation to surrounding electro-magnetic interference (emi). The board is essentially a large piece of metal covered in plastic. Add all the wire-jumpers into this and we have a very noisy system, which traps a lot of emi. Clearly a more elegant and EM-isolated solution needs to be created.

SMT Hardware Prototype

This is where SMD and properly designed and fabricated PCB come in. The small size of SMDs and the single-chip solution provides a very clean environment for signal measurement and conversion. By having the Integrator circuit, Digital Filter and Analog/Digital Converter on the same chip, EM noise is minimized; there are less stray, exposed conductors to act as antennas. More importantly, with delicate soldering, the

components now benefit from secure and reliable connection. Stray capacitances or inductances between the sockets can be minimized.

Schematic

As shown below, the circuitry is relatively simple. The main feature of the interface board is in the DDC112. The surrounding circuitry is designed to provide a stable reference voltage, as well as analog and digital supplies to power the chip. The switching system for the source and detector modules can be found on the upper left regions of the schematic. By implementing up to 12 switches, each controlling an independent source or detector, a wide variety of source-detector combinations can be configured. This allows for maximal experimentation possibility.

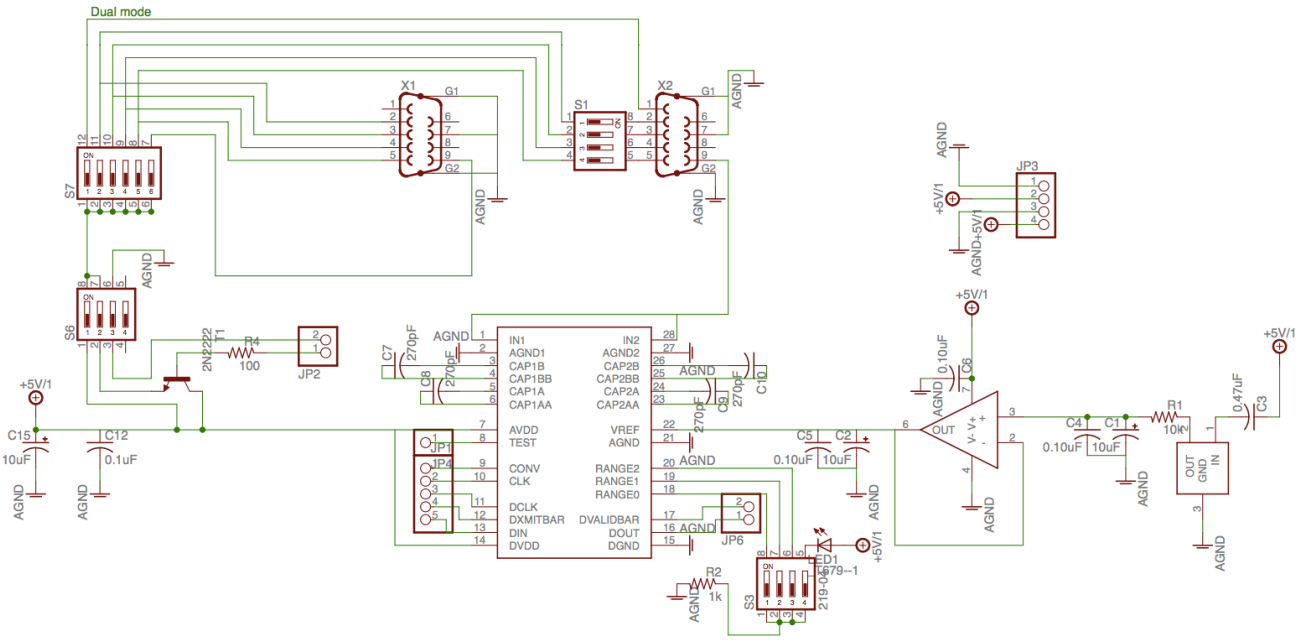


Figure 7: Board Schematic - See Appendix for larger size.

The DDC112 itself has most of its I/O pins broken out into header pins to be jumped over onto a microcontroller board. Any microcontroller of at least 8-bit capacity can be used to

drive the chip. Although a 32-bit microcontroller can allow for simpler simpler code, it is not necessary and is excessive resource for this purpose. An 8-bit microcontroller can be implemented without losing appreciable data rate under the right clock frequency and sampling rate.

In this exercise, a PIC DEM 2 Plus demo board was chosen. It operates on a PIC16F877A microcontroller and is equipped with all the necessary I/O pin-outs necessary to conveniently connect the demo board to the Glucometer interface board. The primary connections are:

- CLK
    - System clock to command the general operations of the DDC112
- DCLK
    - Separate clock to drive the serial data extraction process
    - Integration and Digital conversion are two separate and independent processes in the chip
- CONV
    - Integration command.
    - Controls Integration time
        - Affects Dynamic Range
- DOUT
    - Serial Data Out
    - 20-bit data size
    - Shifted out of a 40-bit register, one bit per clock
    - Controlled by DXMIT_BAR
- DVALID_BAR
    - DDC112 flag for Data Ready for extraction
    - Automatically goes low when Any data is ready to be extracted
    - Dependent on CONV and CLK
- DXMIT_BAR
    - DDC112 input to trigger Data Transmit
    - Enables DOUT

PCB Layout

The PCB design (and schematic) layout was done with EAGLE CADSoft. Custom parts had to be created for the new DDC112 chip, which is not yet commonly used. However, the SOIC-28 package is typical. Also, due to the sensitivity of the analog inputs, the input paths on the board must be guarded by a ground plane at all times. This is good practice for most PCB layout design. One side of the board is designated plane for a 5V supply and the other side is a ground plane. The voltage planes simplified many of the routing problems and provided a stable base and easy access to ground.

Care was also taken in routing the paths properly to avoid any shorts, or charge collection. Orthogonal angles are avoided at all costs and almost all of the copper traces on the board travel in 45-degree angles – standard practice in the industry. Due to cost considerations, size limitations were taken into account. This complicated the layout process because a more convoluted path must be taken in order to minimize size. This is also a valuable experience gained in the project, and standard practice in the industry.

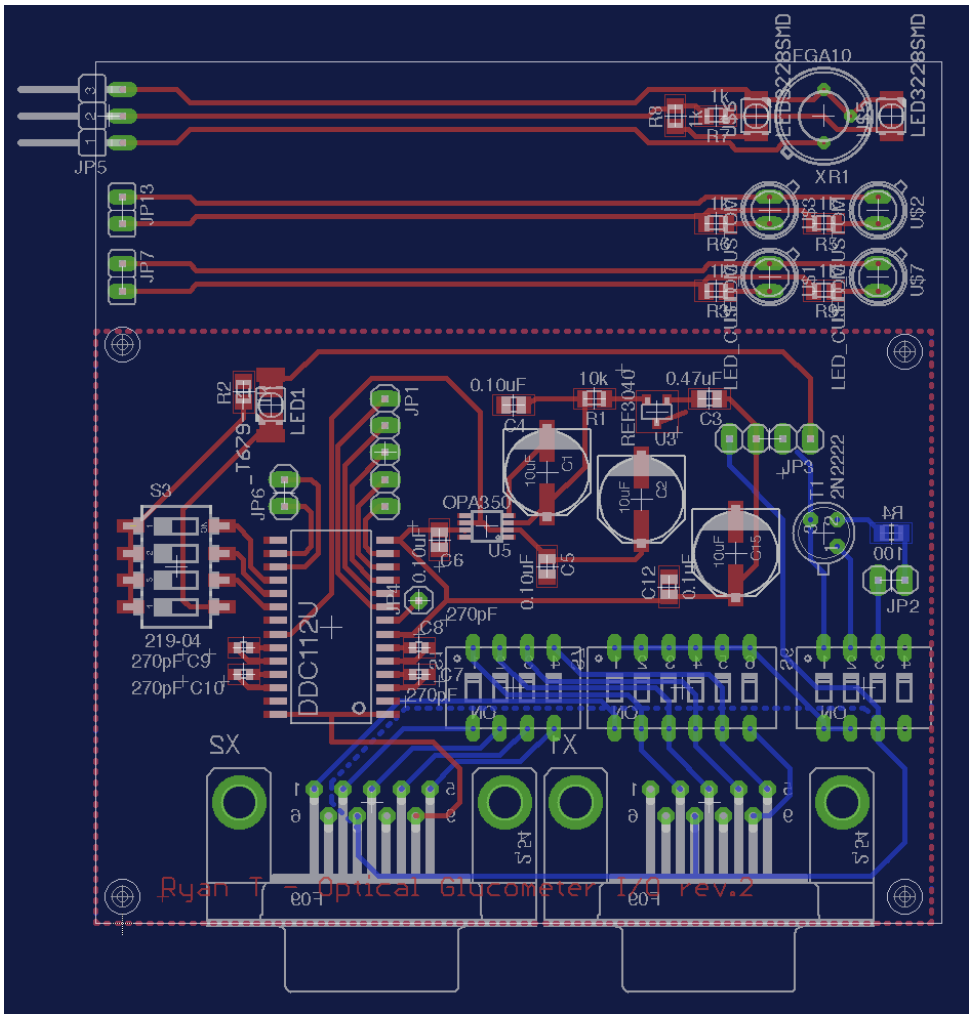


Figure 8: Board Layout on EAGLE - See Appendix for larger size.

Fabrication

To avoid any problems, the PCBs were sent out to a fabrication house in Calgary. The service provided speedy board fabrication, and accurate copper placement. The chemical silk-screening process is much more accurate and achieve smaller trace-width and trace-gaps compared to direct drill milling procedures.

A professional fabrication house also provided other services such as through-hole plating, which was a crucial component in the PCB design. It allowed for a more fluid integration between the top and bottom layers of the board (reduced the number of vias/interlayer connections).

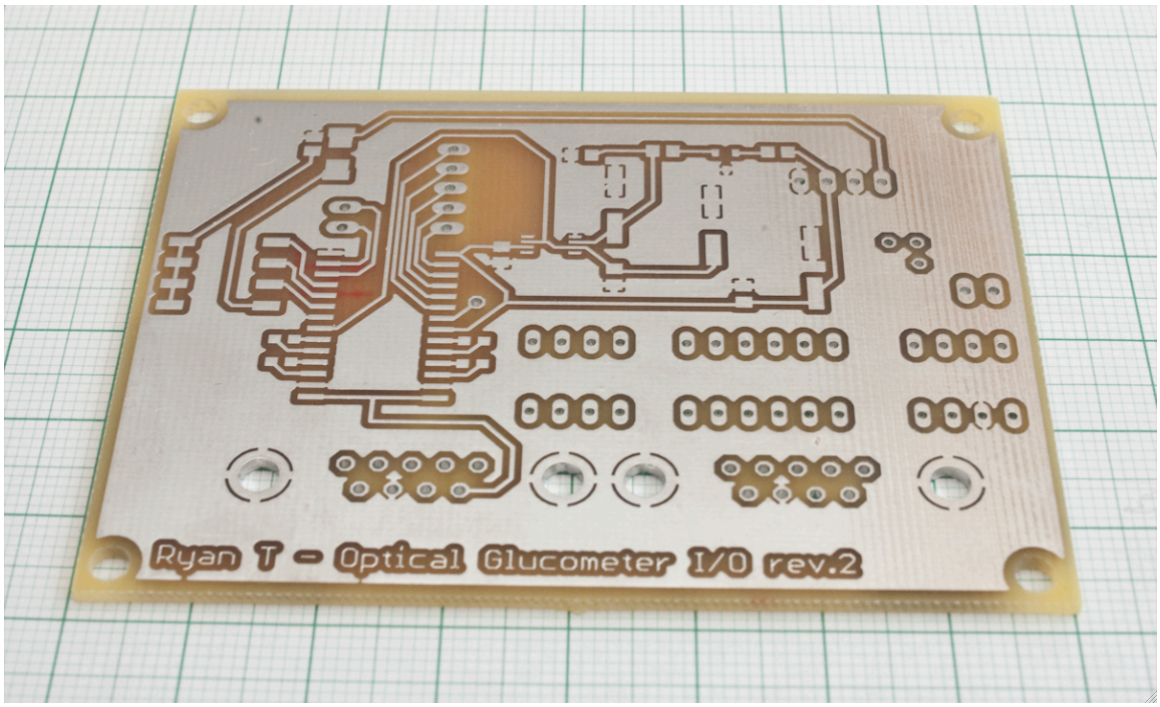


Figure 9: PCB Top side

Note that the Sensor input trace from the left-side DB-9 connector is guarded by a ground plane all the way to the input pins 1 and 28 of the DDC112 chip.

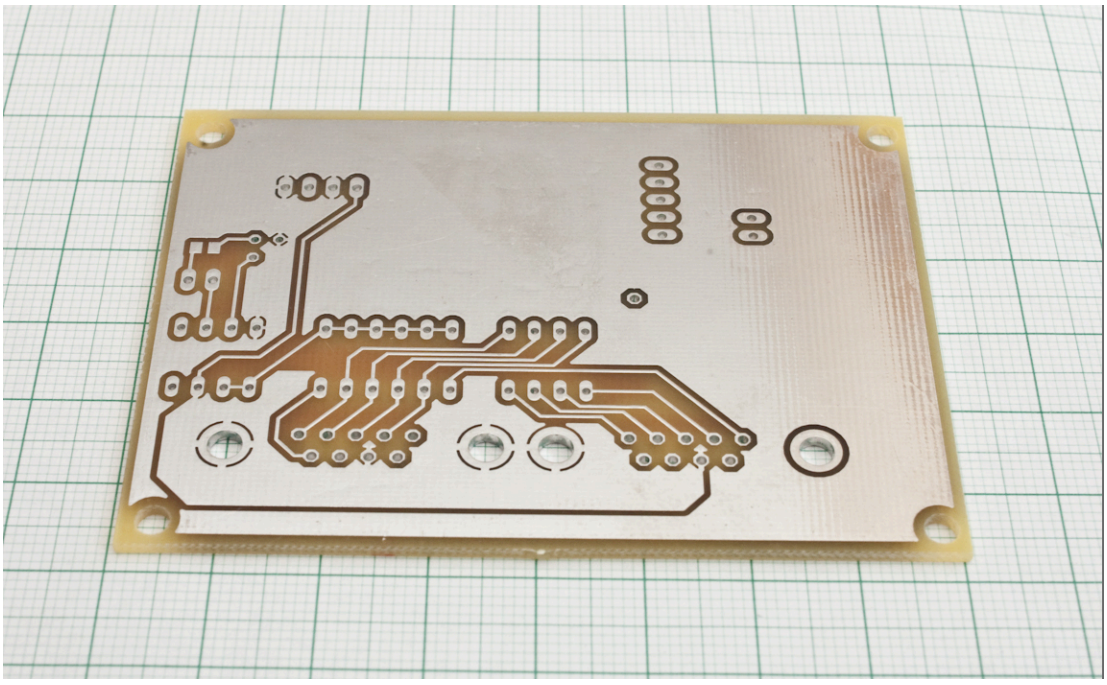


Figure 10: PCB Bottom side

The bottom side has 5V plane all over, except for a small region where all the inputs are, a ground plane on both sides adds shielding.

Application Considerations

Modular source and detector assemblies allow for a large variety of applications.

Case 1 – Dual-mode (Integrated Source/Detector)

- The source/detector unit is rested against material to be measured or tissue (finger tip, inner cheek, etc)

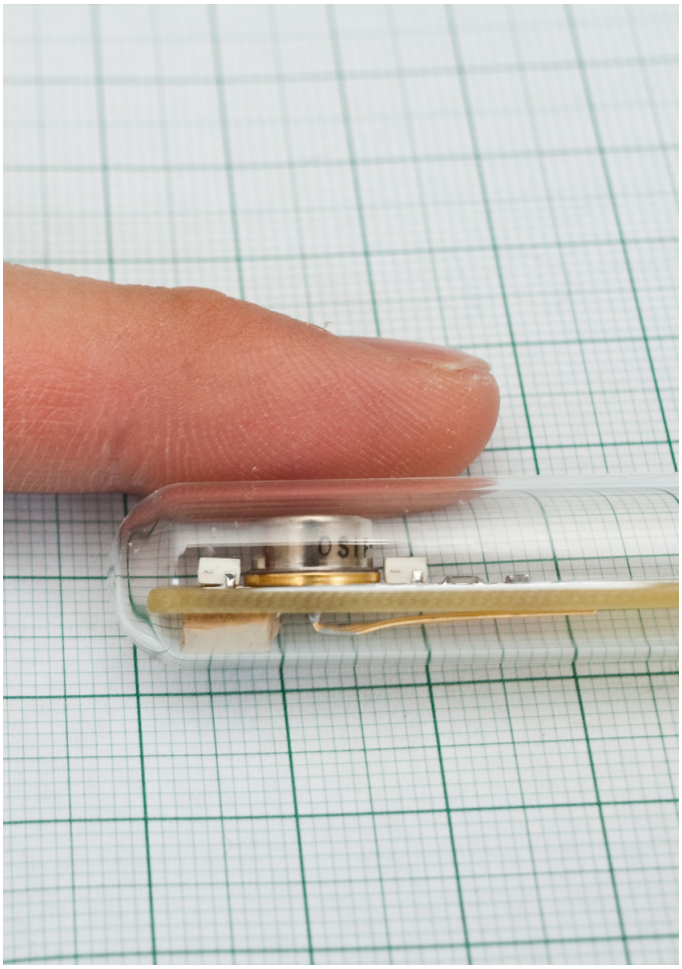


Figure 11: Dual-Mode against finger tip.

Case 2 – Dedicated Source and Detector Units

- Separate modules for the source and detector
- Source may be placed next to the detector

  - Tranflectance
  - Absorption
  - Reflectance
- Source may be placed across tissue/substance
  - Transmission
  - Absorption
  - Scattering

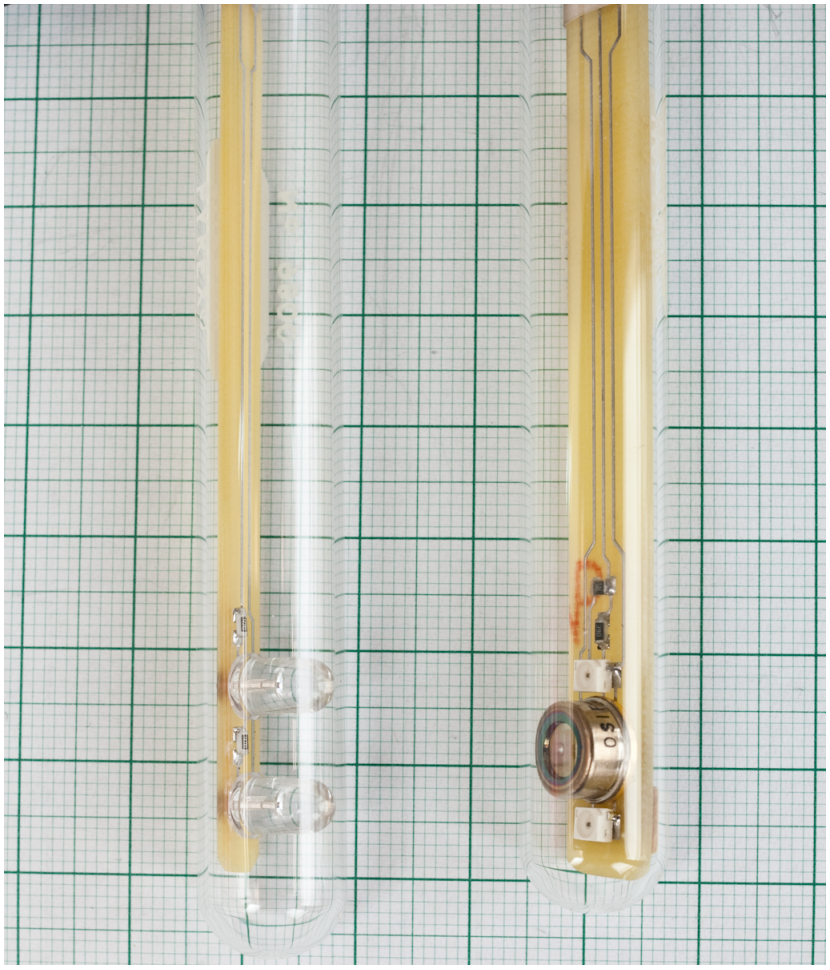


Figure 12: Dedicated modes across medium

Case 3 – Free Detector Operation

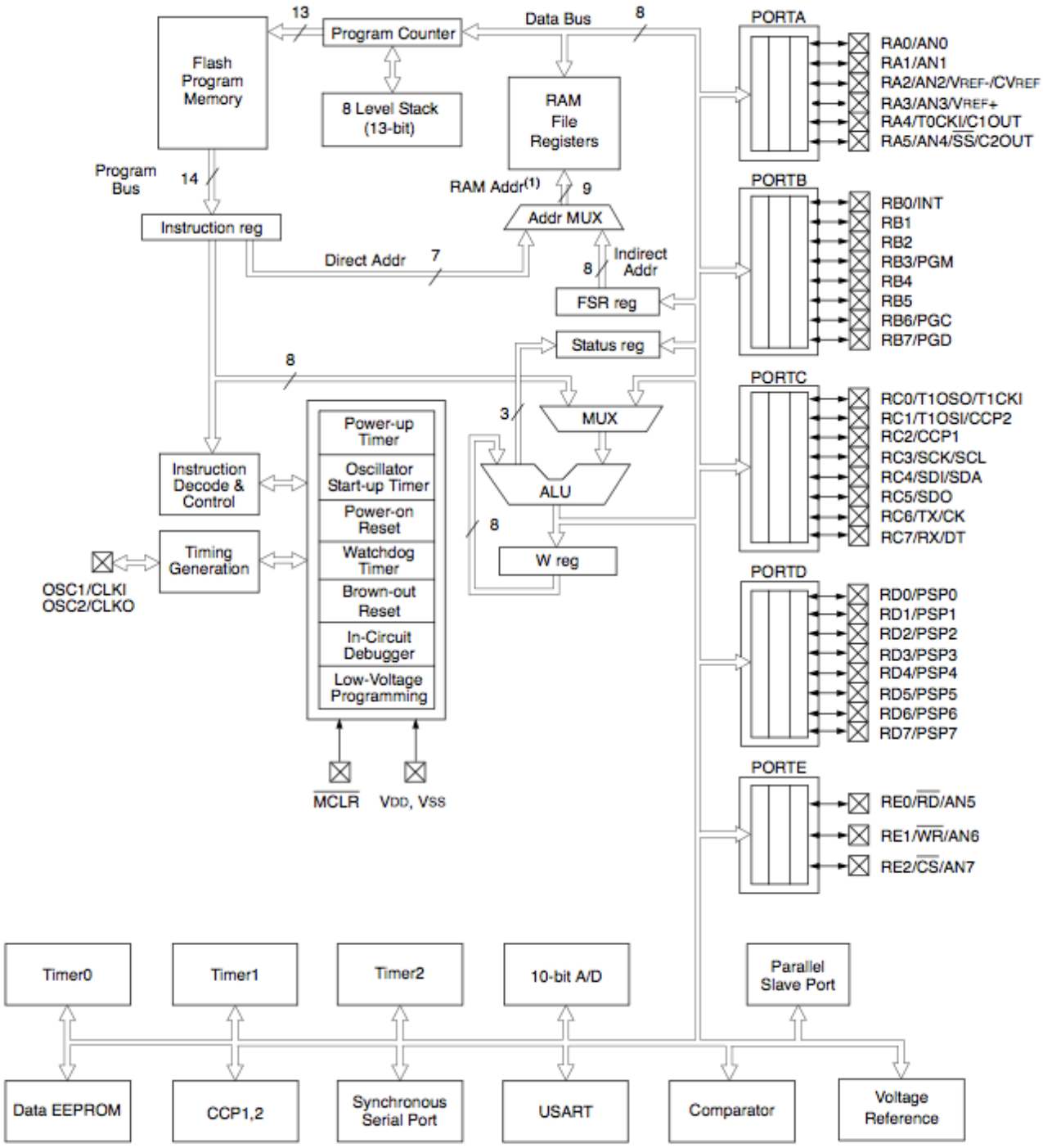
- Use of detector only
- Measure light transmission through air, vacuum or other medium

- Separate source unit
    - Laser module
    - Other light sources

Software

The driver code for the DDC112 is written in C and can be programmed into most 8-bit microcontroller. Our application utilizes the PIC16F877A, which are widely available and affordable. The PIC has a built-in USART interface allowing for data streaming through serial port into a PC for analysis in MATLAB or LabView (USB connection could also be implemented with an FTDI FT232, but it was not deemed necessary at this time, and will be included as possible future improvements).

The PIC DEM 2 Plus will be using a 10MHz clock as its primary oscillator, which will also be fed into the DDC112. A secondary DCLK will be generated by fractioning the primary oscillator (inside the PIC). 10MHz was chosen because it is fast enough to allow for some programming room of movement. Integration time and capacitance values can be calculated based on reference numbers as described in the DDC112 Data Sheet. 10MHz was also chosen to remain consistent with the known operational parameters of the DDC112. These parameters include the time-delays of the different digital and analog activities within the chip. Although the numbers are only within nanoseconds ranges, it was deemed simpler to stay within the known parameters.



Device	Program Flash	Data Memory	Data EEPROM
PIC16F874A	4K words	192 Bytes	128 Bytes
PIC16F877A	8K words	368 Bytes	256 Bytes

Figure 13: PIC16F877A

Data Extraction Procedure

The DDC112 is a dual-input 20-bit analog current to digital converter. The CONV signal can be configured to put the chip into continuous conversion mode where two integrating sides will always be processing/converting data – allowing real-time measurements.

The conversion results are put into a 40-bit long shift register. From here, the data can be extracted 1(one) bit/clock (DCLK) after DXMIT_BAR is activated. The simplest method to extract the data would be to extract the information in chunks of 8 bits because the PIC operates in terms of 8-bit words. After extracting three 8-bit chunks, the data is then re-assembled into a 32-bit integer (int32), with zeros padded to the front of the Most Significant Bit (MSB).

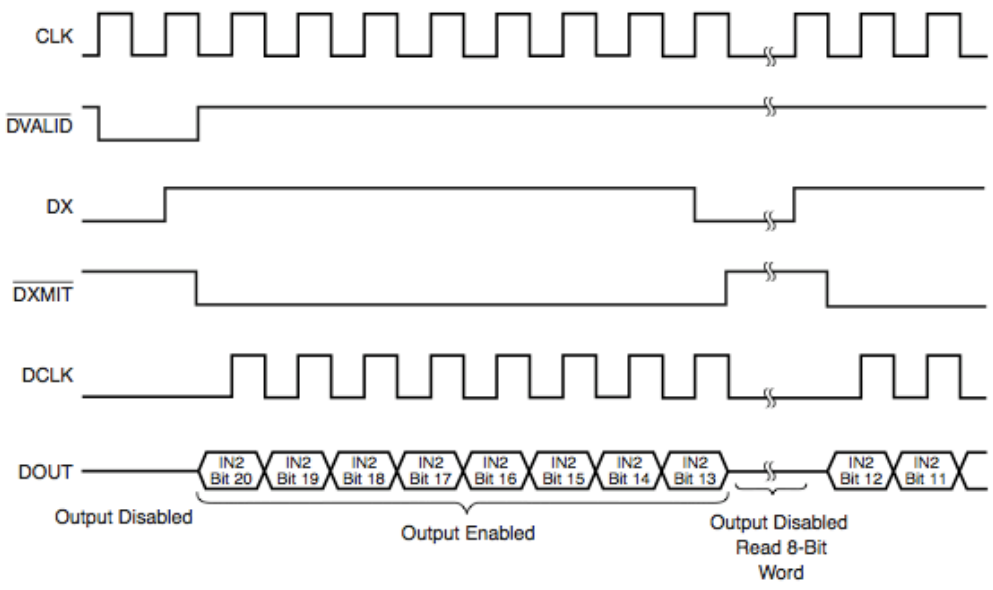


Figure 14: DDC112 Data Extraction

The 32-bit integer can then be sent over serial port to a computer terminal where it can be streamed directly into MATLAB, LabView or any other processing/analysis software.

Figure 15: 20 bit number inside 32 bit integer

Results & Discussions

The Results

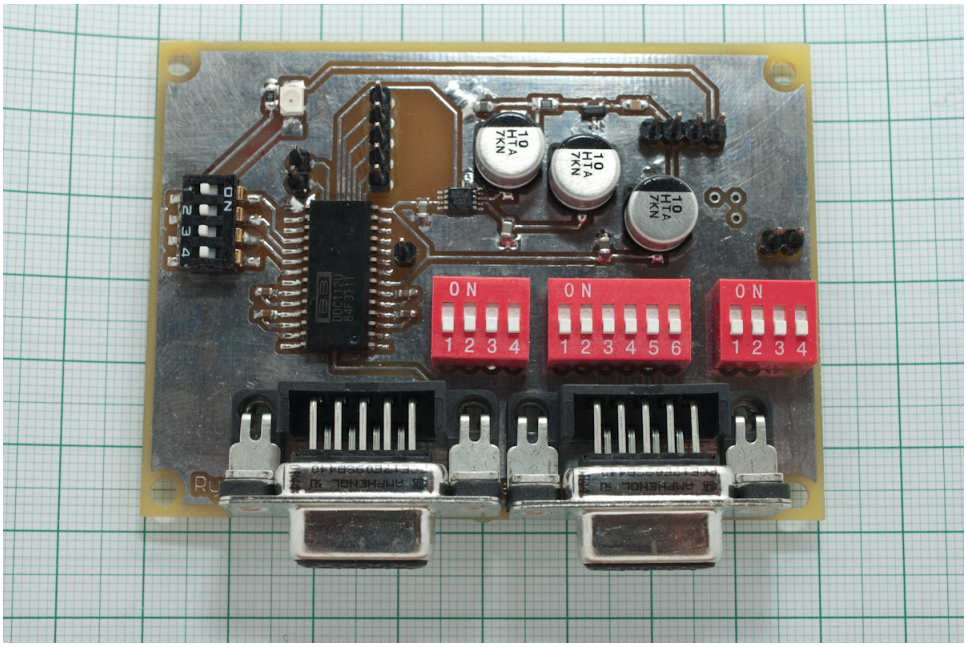


Figure 16: Completed Glucometer Interface Board

Due to time constraints, lack of manpower and overstretching of my attention and resources over the course of Level 4, the software component of the project was not completed in time. There were some difficulties in getting the CONV timing code to properly trigger the integration sequence. The digital extraction sequence was successfully completed. This was tested by setting values inside the PIC and extracting it as an int32 constructed using the aforementioned method.

Preliminary experimentation results would indicate that, given a successful signal acquisition using a very rudimentary circuit on the Pulse Oximeter demo (Figure 17, Figure 18), the circuit designed in this project would have no problems acquiring the same signal. It works on nearly identical principles of converting a photodiode current into a digital value. The transimpedance amplifier was replaced with the integrating

amplifier. The LPF and pre-amp stage was replaced with the built-in digital filtering circuit. And finally, the A/D Conversion done by the DAQ interface board for LabView is now done on the chip, as well.

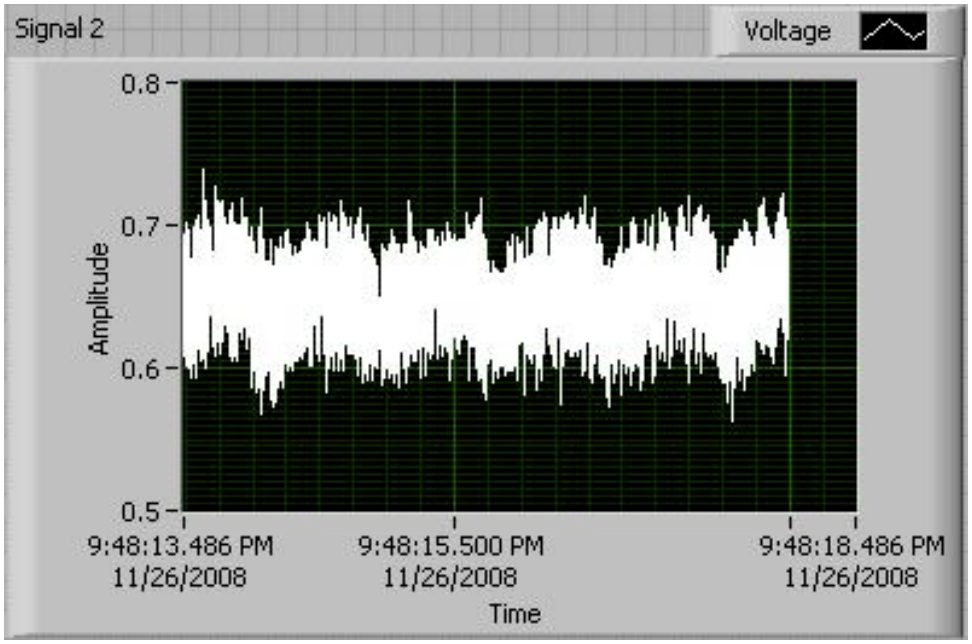


Figure 17: Unfiltered signal from LabView DAQ

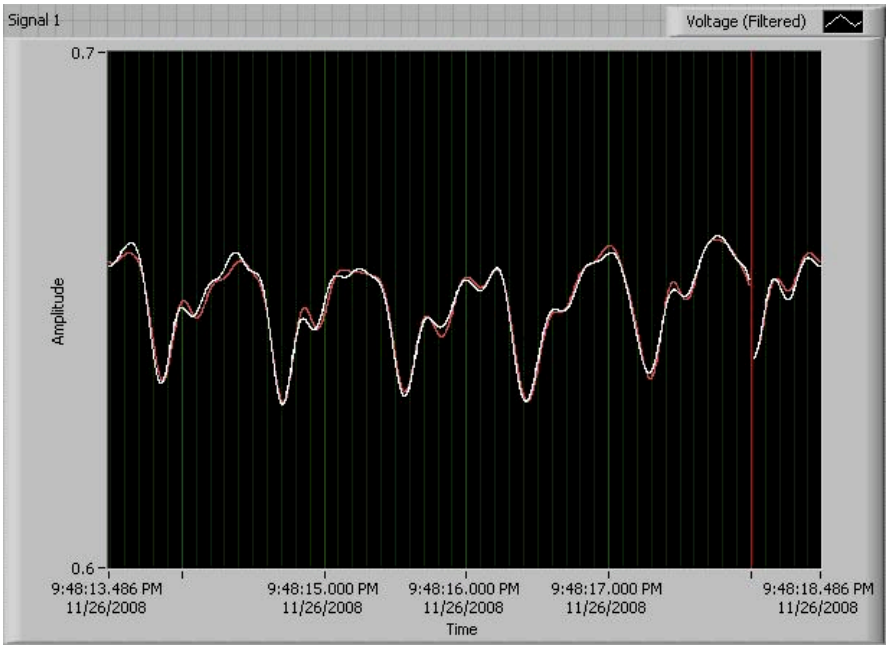


Figure 18: Processed signal from rudimentary Pulse Oximeter.

Costs

The bulk of the cost in this project fell into the sensor assembly. The THOR FGA10 NIR sensor was specially designed to detect 700-1800nm, light and was significantly much more expensive to acquire than the typical 600-1100nm photodiodes.

Bill of Materials

Part Name	Digikey Part #	Cost
DDC112		$24.90
OPA350		$3.23
REF3040		$2.16
10uF capacitor		
0.1mF capacitor		
270pF capacitor		
1k resistor		
100k resistor		
10k resistor		
DB- 9 Serial port connector (female)		
DB-9 Serial cable (male)		
4 button switch (through-hole)		
4 button switch (SMT)		
6 button switch (through hole)		
Red LED (660nm)		
IR LED (940nm)		$1.69
NIR LED (1300nm)		$8.00
NIR LED (1450nm)		$8.00
NIR LED (1550nm)		$8.00
PIN Photodiode (600-1100nm)		$8.00
InGaAs Photodiode (700-1800nm)		$180
PCB		$70
Test Tubes		
Corks		

Future Improvements

The system can be improved from a few different approaches.

- The interface between the glucometer board to the PC can be updated with an FTDI FT232 chip to remove the need of a serial-to-USB adapter.
- The glass enclosures of the source and detector modules can be secured with better material, such as silicone or liquid rubber dip to ensure a waterproof seal between the cable and the internals of the glass tube.
- Separate 5V supplies for Analog and Digital V_{DD} for better isolation.
- Integration of the microcontroller onto the board
    - Eliminating the need for a secondary driver board.
    - Combined with the FTDI solution, the system then becomes a very compact USB-enabled data-gathering platform.

Encountered Challenges

Moving from the literary review stage onto the design phase was a leap. There was too many options, and open variables to consider. The sheer amount of options and possible approaches can be counter-productive to development. This is where the importance of working in a team reveals itself. Even in a team of two people, the collaboration of the minds and the ability to counter each other's arguments is invaluable. The other member act as a sort of reality check when it comes to decision-making. Absurd options can be discarded quickly, and less time is wasted due to the brain-stall of one person's inability to choose.

Secondary physical challenges were in designing the PCB and schematic diagram for the first time. Schematic layout was not very difficult. However, it became apparent very quickly that simplicity in schematic phase in crucial in order to at least give a 'fighting-chance' when it comes to routing the PCB layout.

The task of routing a PCB is more of an art than science. There are many ways to route PCB traces properly, and a designer can spend a lot of time getting lost in the puzzle. Even after a design or layout is finished, there is always room for improvement. There is always some way of reducing the board footprint by moving components to a different area. Tossing components to the opposite side and taking advantage of the multi-layered nature (working in 3D) opens up many different possibilities in conduction.

It was also a challenge to decide on the proper type of sensor based on the application. In the end, the most versatile option was chosen. It was an interesting process to determine the best approach to encase the sensor and source modules. Ease of use, repeatability, and cleanliness were all part of the factors taken into account when choosing the glass tube enclosure. The challenge was in finding a readily available solution.

When it comes to specialized design, typical sources do not always have the components required. It took some searching through the net in order to get the parts required, or the services provider that meet our budget and criteria.

CONCLUSIONS & RECOMMENDATIONS

There is still work to do before optical glucose measurements can be done in vivo. The project set out to integrate optical glucose testing device into common mobile devices; however, the technology does not actually exist yet. There are no publicized solutions to optical glucose measurement in vivo at this time. There are patents granted and companies claiming that they are on the verge of a commercial product, but there is not working demonstration at the moment.

Therefore, one can conclude that any and all claim of success was simply marketing strategy to secure support and funding for research. In which case, there is potential need for a reliable and accurate experimentation platform to collect data on. The high resolution and low-noise potentials of the solution attempted in this project still holds.

The only downfall was in perfecting the software driver. With additional effort under a more reasonable roadmap (or timeline, the interface can be perfected, and data collection can begin. Various experiments can then be conducted to further learn of glucose behaviour in blood.

Beyond glucose, the interface can also be used for various other bioinstrumentation applications, or even as a lab test bench for most things IR or NIR absorbance related.

APPENDIX A

Pulse Oximetry Experiments

Intro

Pulse oximeters work on the principle of utilizing the two different wavelengths of light, red and infrared in order to determine the oxygen saturation and the pulse rate of a patient.

A pulse oximeter makes use of two light emitting diodes (LEDs); a red one at 600-750 nm and an infrared one at 850-1000 nm [17]. The light is typically transmitted through either the finger or the ear, and sensed by a photo detector, which generate a photo current. The photocurrent generated is then converted into a voltage signal, which in turn is amplified and further processed to obtain the desired measurements.

The amount of absorption of infrared and red light vary based on the oxygen saturation level of hemoglobin in arterial blood. The oxygenated hemoglobin molecules (HbO_2) absorb more infrared light; while the reduced hemoglobin molecules (Hb) absorb more red light [18], as can be seen in the spectrum shown in the figure below.

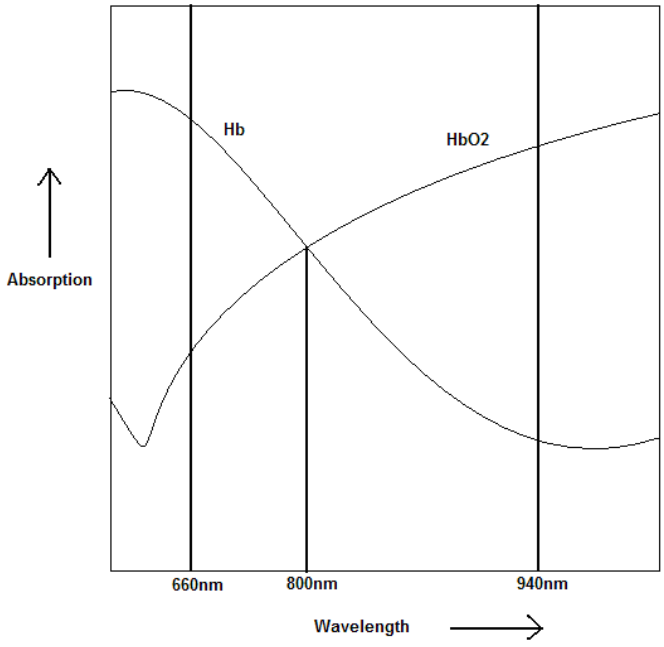


Figure 19: Absorption spectrum of HbO_2 and Hb over a wavelength of 500-1000nm

The ratio of the amount of red light absorbed to the amount of infrared light absorbed can be compared against a lookup table in order to determine the oxygen saturation in hemoglobin. By differentiating the various causes for absorption at the measuring site such as the skin, tissue and venous blood from the arterial blood, one can also determine the pulse rate of the patient. During the contraction of the heart, there is a momentary increase in oxygenated hemoglobin in arterial blood across the measuring site which in turn causes more infrared light absorption.

The pulse wave that indicates the heart rate is known the photoplethysmographic signal (PPG)[18], and this signal is mainly due to the changes in the volume of arterial blood in the measured site.

Hardware Design

The pulse oximeter block diagram below presents a high level view of the design.

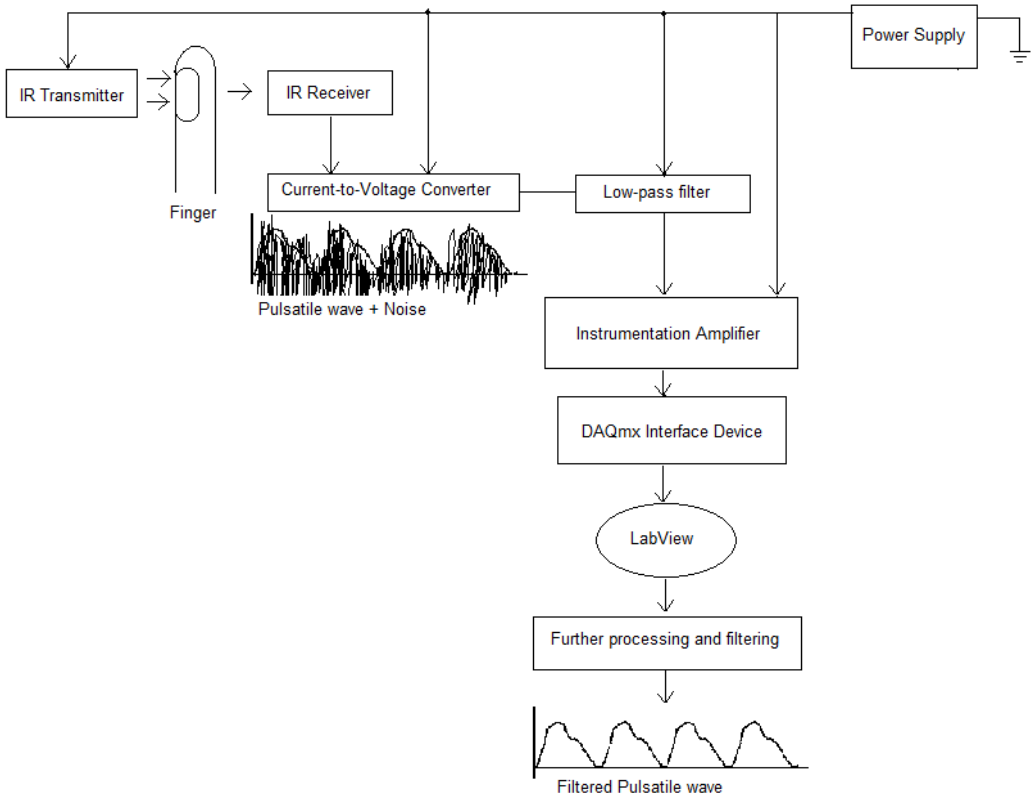


Figure 20: Block Diagram of the Pulse Oximeter

The following components were used in designing the pulse oximeter:

Infrared transmitter and receiver

An infrared LED was used for transmitting light of wavelength 940nm through the finger, some of which was absorbed and the remaining IR light reaches the receiver photodiode. The photodiode converts the light into a current that is fed into the current-to-voltage converter for further processing. Depending on the oxygenation level of hemoglobin at any given instant, the amount of absorbed light varies.

Current to voltage converter circuit

In order to amplify the incoming signal, the photocurrent generated at the photodiode must be converted into a voltage signal. The current-to-voltage conversion was accomplished using an op-amp as shown in the figure below.

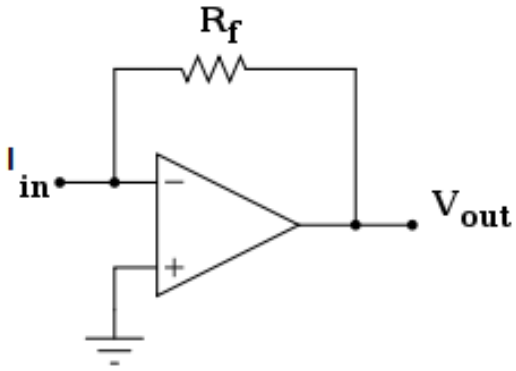


Figure 21: Current to Voltage Converter

R_f = 1000 ohms

Low pass filter circuit (Cutoff frequency : 5Hz)

In order to filter out the high frequency noises introduced through the powers lines, motion artifacts and other sources, a low pass filter was required. The low pass filter used is as shown in the figure below.

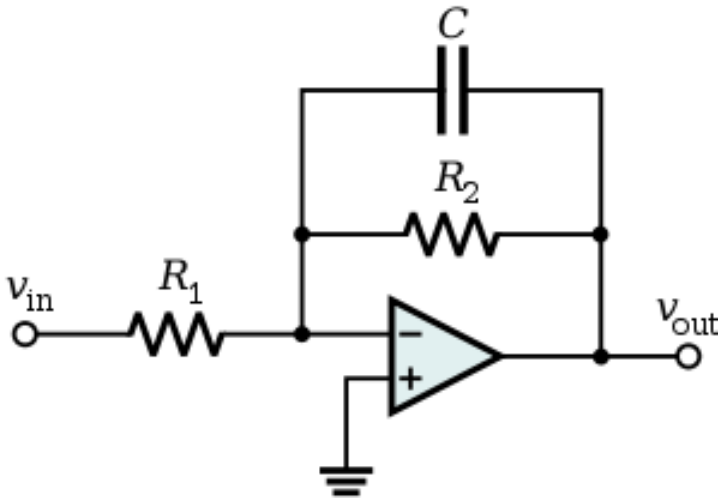


Figure 22: Active Low-Pass Filter

R1 and R2 were resistors of equal value, so that the gain of the circuit may be kept at unity, until the next stage (differential amplifier).

Using the following equation chose the capacitance and resistance values:

$f_{cutoff} = ½* (pi)*R_2C$, where $f_{cutoff} = 5$ Hz, $R_2 = 1$ Mohms, C = 3.2 nF

Instrumentation amplifier circuit

The below differential amplifier removes the DC offset present in the signal, and also amplifies the signal by a factor of 5. R_G was set to 1Mohms

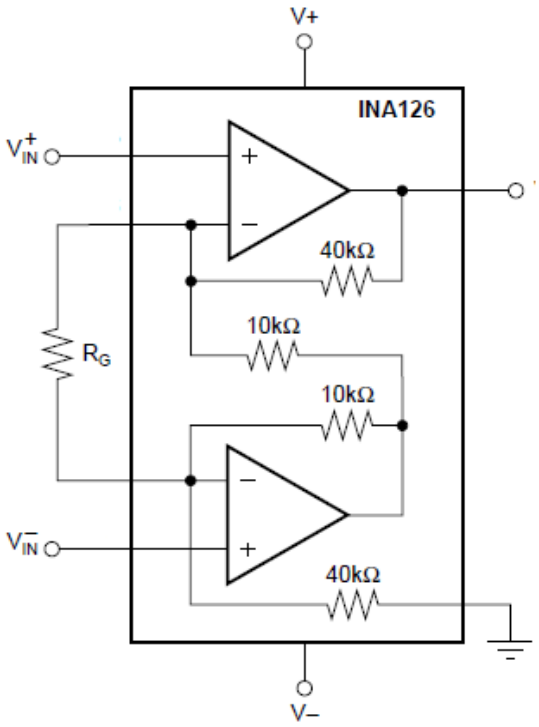


Figure 23: Instrumentation Amplifier

Band pass filter (0.5 – 5Hz)

The BPF is used primarily to remove any DC component in the incoming signal and also filter out the high frequency noise contributed due to various sources. This filter was

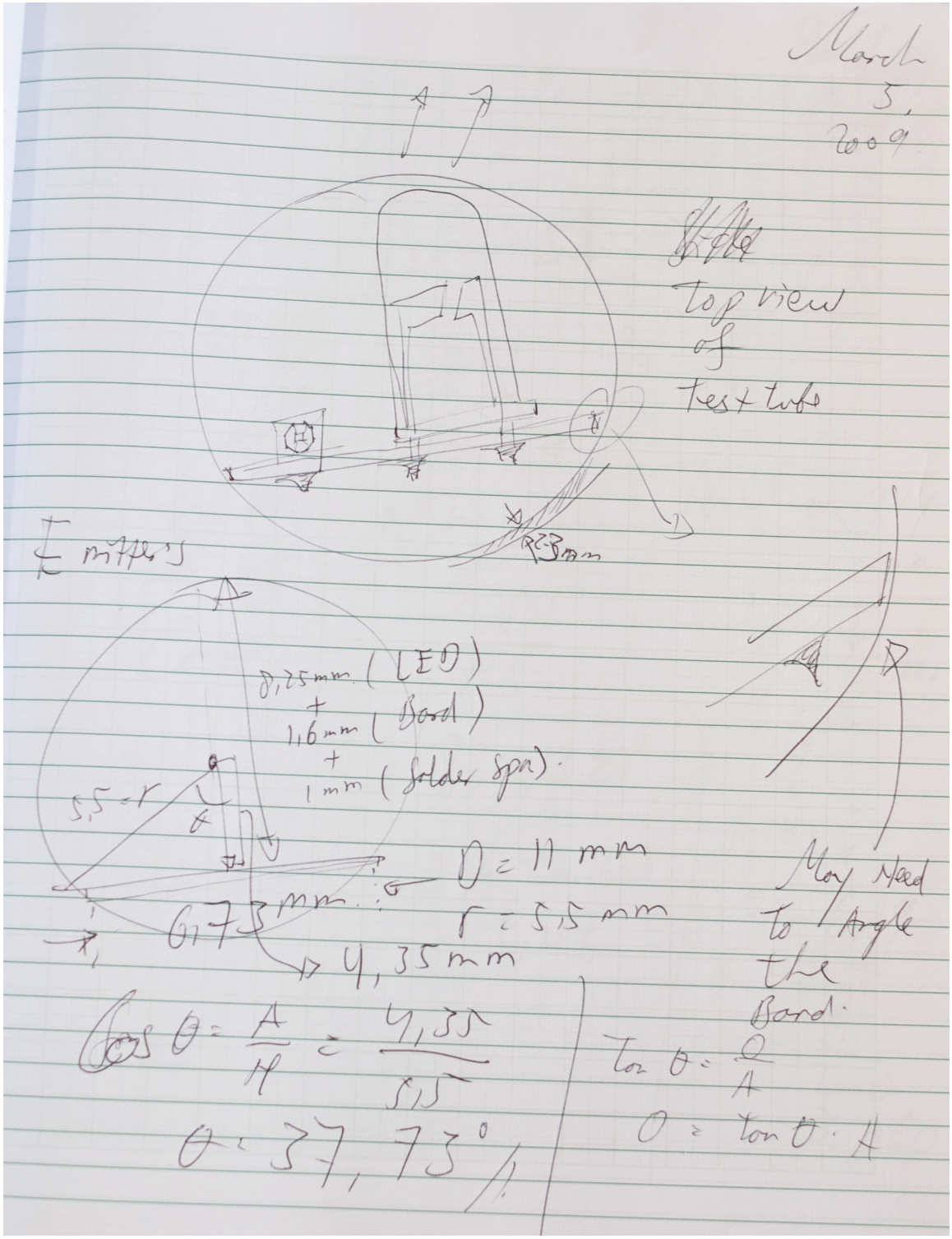
simulated using LabView. The order of the filter was set to 10, ensuring a sharp cutoff at the desired frequency range limits.

Results

Refer to Figure 17: Unfiltered signal from LabView DAQ and Figure 18: Processed signal from rudimentary Pulse Oximeter.

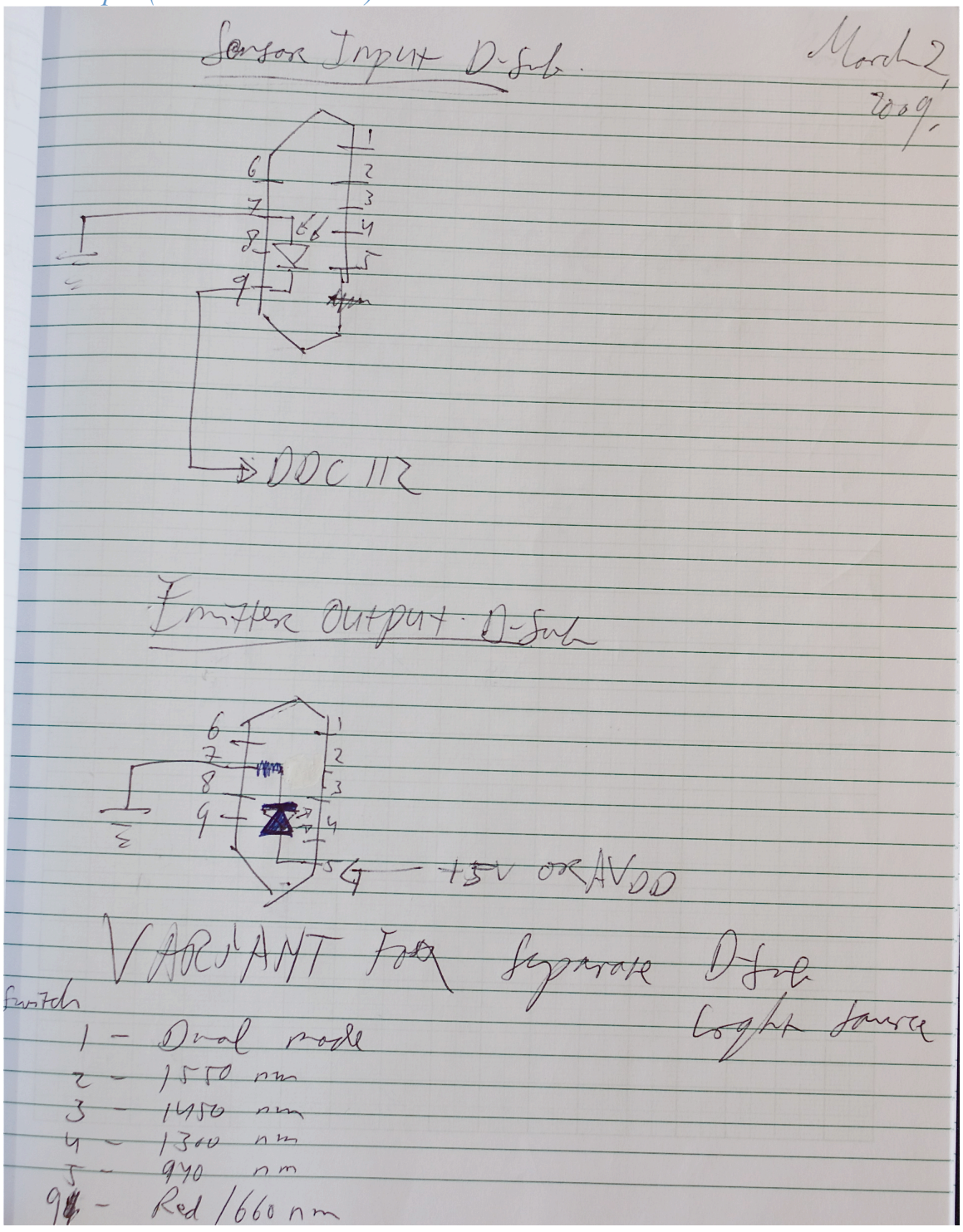
APPENDIX B

Test Tube Clearance calculations



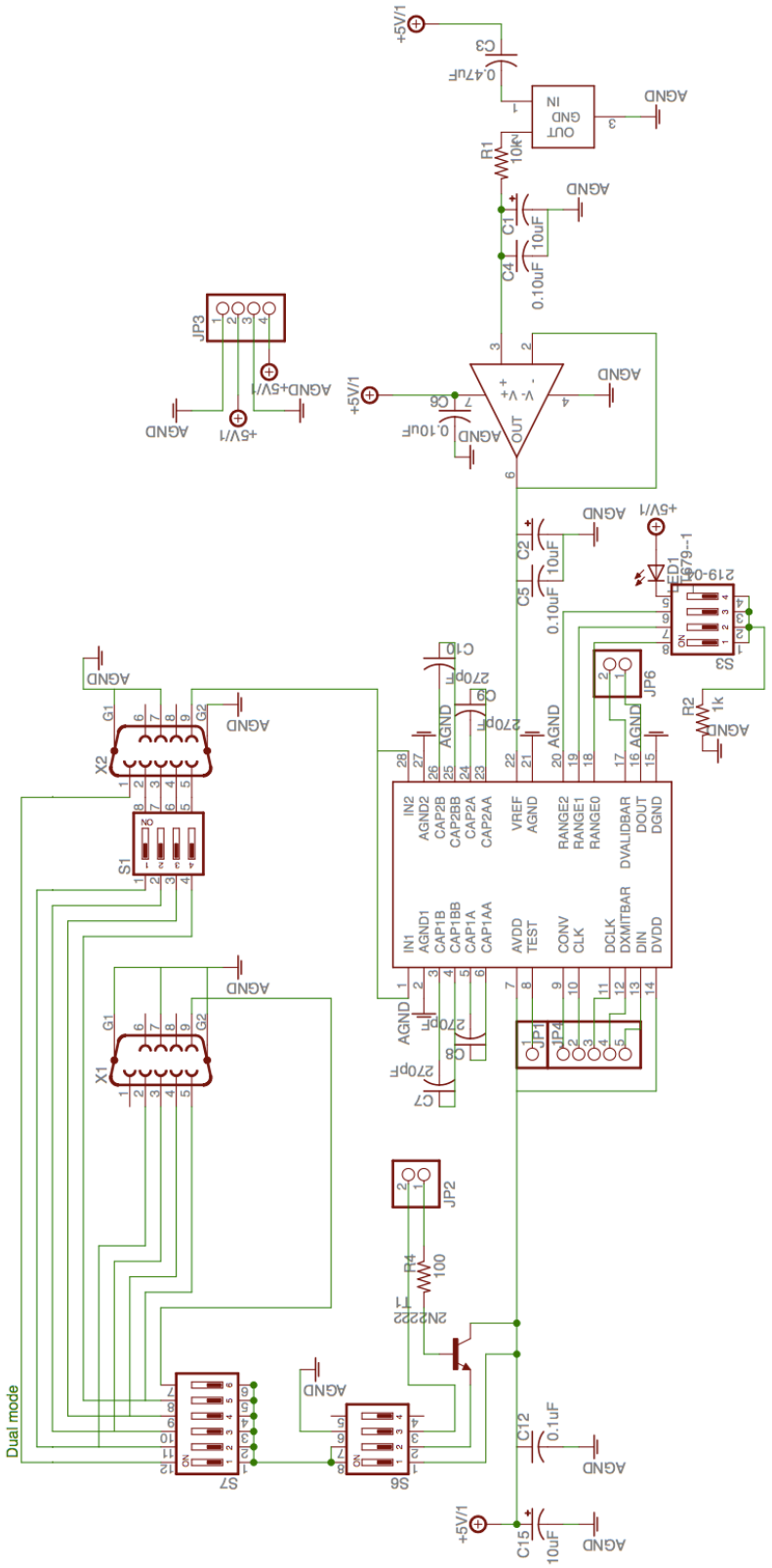
APPENDIX C

D-Sub 9-pin (Serial Connection) Source and Detector Pin-out



APPENDIX D

Schematic



APPENDIX E

PCB Layout

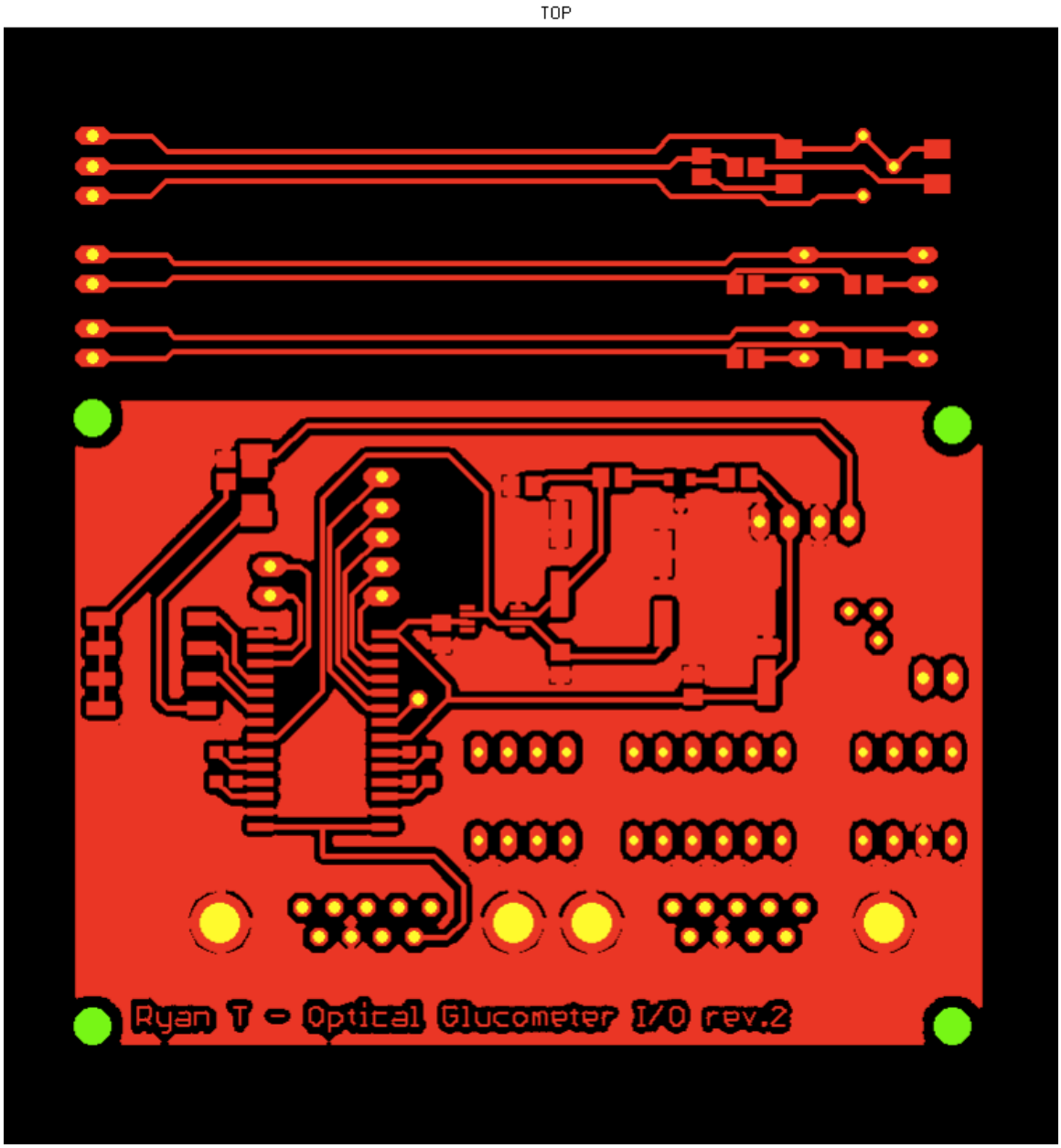


Figure 24: PCB Top

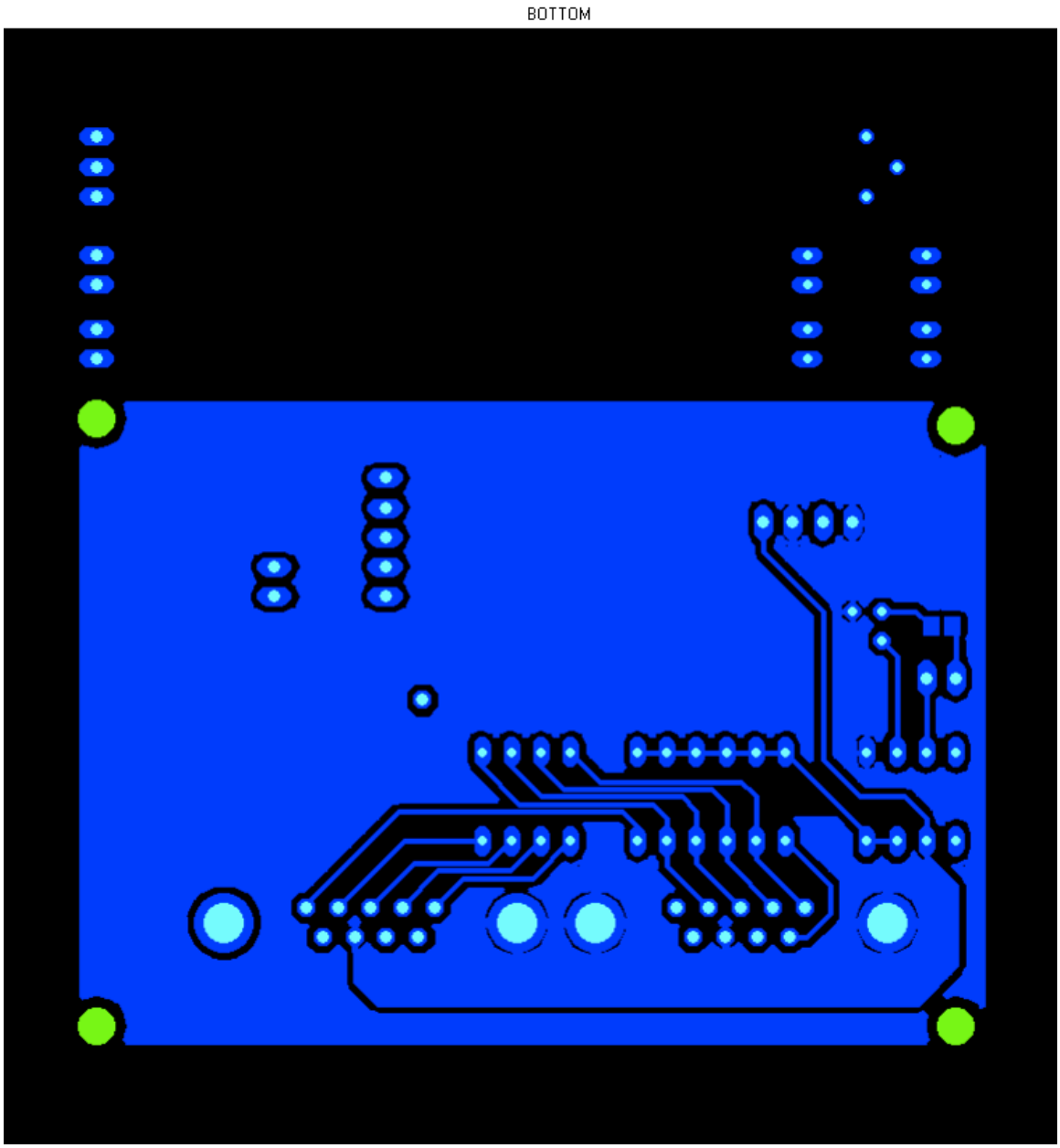


Figure 25: PCB Bottom

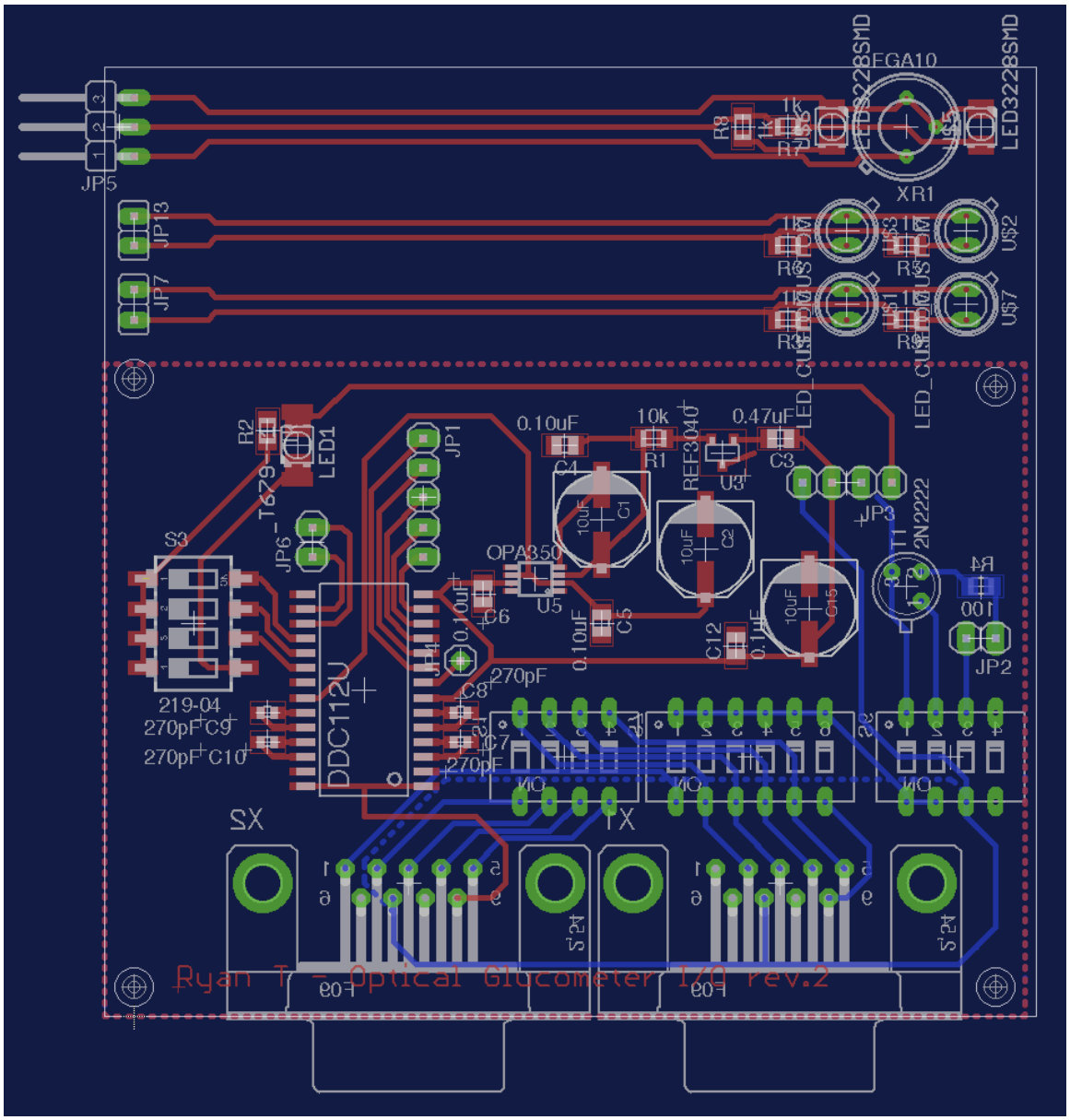


Figure 26: PCB with parts

APPENDIX F

Code

```
#include <16F877A.H>
#device *=16                    // This allows auto variables
over location 0xFF
#fuses HS,NOWDT,NOPROTECT,NOLVP
#use delay(clock=4000000)
#use rs232(baud=9600, xmit=PIN_C6, rcv=PIN_C7)

//#bit t1_overflow=0x0C.0


#define CLK pin_C0
#define DCLK pin_C1
#define DXMITBAR pin_C2
#define CONV pin_C3
#define DATAIN pin_C4
#define DVALIDBAR pin_B0

#define MASK 0b00000001
#define CLEARMASK 0b00000000


//int8 Tint; / /in microseconds
int32 value;
int8 temp[3];
int8 beep;


//#int_rb
void extractData(){
      int8 i,j;
      temp[0] = 0;
      temp[1] = 0;
      temp[2] = 0;
      value = 0;
      beep = 43;
```

```
      printf("%d \r\n",beep);

      if (DVALIDBAR == 0){
            output_low(DCLK);

            output_low(DXMITBAR);
            // For the first two bytes

            for (i = 1; i < 2; ++i){
                  for (j=1; j < 8; ++i){
                        output_high(DCLK);

                        if(DATAIN == 1){
                              temp[i-1] << 1;
                              temp[i-1] = temp[i-1] | MASK;
                        }else if (DATAIN == 0){
                              temp[i-1] << 1;
                              temp[i-1] = temp[i-1] | CLEARMASK;
                        }

                        output_low(DCLK);
                  }

            // For the last 4 bits - going into the 3rd byte

            for (j=1; j < 4; ++i){
                        output_high(DCLK);
                        if(DATAIN == 1){
                              temp[2] << 1;
                              temp[2] = temp[i-1] | MASK;
                        }else if (DATAIN == 0){
                              temp[2] << 1;
                        }
                        output_low(DCLK);
                  }
            }
      }

      value = make32(temp[2], temp[1], temp[0]);
      printf("%LU \r\n",value);
}

#INT_RTCC
```

```
void convertInput(){
      int8 Tint = 300;
      beep = 44;
      output_high(CLK);
      printf("%d \r\n",beep);
      if (input(CONV) == 0){
            // Init Process (toggling before Start Integration)
            output_high(CONV);
            delay_us(50);
            // Integration of side A
            delay_us(Tint);
            output_low(CONV);
            // Integration of side B
            delay_us(Tint);
            output_high(CONV);

      } else if(input(CONV) == 1) {
            // Init Process (toggling before Start Integration)
            output_low(CONV);
            delay_us(50);
            // Integration of side B
            delay_us(Tint);
            output_high(CONV);
            // Integration of side A
            delay_us(Tint);
            output_low(CONV);
      }
      output_low(CLK);
      extractData();
}

void main() {

      delay_ms(100);
      beep = 42;

      set_timer0(0);;
      setup_counters( RTCC_INTERNAL, RTCC_DIV_2 );
      //setup_timer_1( T1_INTERNAL | T1_DIV_BY_2 | T1_CLK_OUT );

      enable_interrupts(INT_RTCC);
      //enable_interrupts(INT_TIMER1);
      enable_interrupts(INT_RB);
      enable_interrupts(GLOBAL);
```

```
      //convert_input();
      do {
            //set_timer1(0 );
      while(input(DVALIDBAR)== 1){
            beep = 42;
            delay_us(20);
      }

      extractData();
      //value = 0;
      // Assemble the 32bit number (Excess space padded with
zeros)
      //value = make32(temp[2], temp[1], temp[0]);
      //printf("%LU \r\n",value);
      delay_ms(100);
      } while(1);
}
```

BIBLIOGRAPHY

[1] E. Hirt a, Yu. Feldman a, Z. Ali c, L. Heinemann A. Caduff, "First human experiments with a novel non-invasive, non-optical continuous glucose monitoring system," *Biosensors and Bioelectronics*, vol. 19, pp. 209-217, 2003.

[2] F. Dewarrat, M. Talary, G. Stalder, L. Heinemann, Yu. Feldman A. Caduff, "Non-invasive glucose monitoring in patients with diabetes: A novel system based on impedance spectroscopy," *Biosensors and Bioelectronics*, vol. 22, pp. 598-604, 2006.

[3] Alberto Maran, Giovanni Pacini Andrea Tura, "Non-invasive glucose monitoring: Assessment of technologies and devices according to quantitative criteria," *Diabetes Research and Clinical Practice*, vol. 77, pp. 16-40, 2007.

[4] Jason J. Burmeister and Mark A. Arnold, "Spectroscopic Considerations for Noninvasive Blood Glucose Measurements with Near Infrared Spectroscopy," *Department of Chemistry and Optical Science and Technology Center Advanced Technology Laboratories*, vol. 12, no. 2, April 1998.

[5] Benhard Wolf Carlos Eduardo Ferrante do Amaral, "Current Development in non-Invasive Glucose Monitoring," *Medical Engineering & Physics*, vol. 30, pp. 547-549, 2008.

[6] J.E. Crooks, *The Spectrum in Chemistry*, King's College. University of London Chemistry Department, Ed. London: Academic Press, 1978.

[7] Giuseppe Palleschi. Glenn Lubrano George G. Guilbault, "Non-Invasive Biosensors in Clinical Analysis," *Biosensors & Bioelectronics*, vol. 10, pp. 379-392, 1995.

[8] Y. Ozaki, S. Kawata, H.M. Heise H.W. Siesler, *Near-Infrared Spectroscopy - Principles, Instruments, Applications*, Y. Ozaki, S. Kawata, H.M. Heise H.W. Siesler, Ed. Germany: Wiley-VCH, 2002.

[9] JR., Lois Weyer Jerry Workman, *Practical Guide to Interpretive Near-infrared Spectroscopy*. US: CRC Press - Taylor & Francis Group, 2008.

[10] Faeiza Hussain, Nicholas D. Evans, Nabihah Sachedina John C. Pickup, "In vivo glucose monitoring: the clinical reality and the promise," *Biosensors and Bioelectronics*, vol. 20, pp. 1897-1902, 2005.

[11] T. Isaksson, T.Naes, A, Tandberg K.I. Hildurn, Ed., *Near Infrared Spectroscopy - Bridging the Gap between Data Analysis and NIR Applications*. Great Britain: Ellis

Horwood Limited, 1992.

[12] Philippa Solomon Koji Nakanishi, *Infrared Absorption Spectroscopy*. San Fransisco, US: Holden-Day Inc, 1977.

[13] Huang Fu-Rong, Chen Zhe Luo Yun-Han, "Detection Limit of Glucose Concentration with Near-Infrared Absorption and Scattering Spectroscopy," *Department of Optoelectronics Engineering*, 2008.

[14] J.A. Tamada a, R.O. Potts a, L. Jovanovic b, S. Garg c, Cygnus Research Team M.J. Tierney a, "Clinical evaluation of the GlucoWatch® biographer: a continual, non-invasive glucose monitor for patients with diabetes," *Biosensors & Bioelectronics*, vol. 16, pp. 621-629, 2003.

[16] Ph.D.* and V. M. Chenault, Ph.D., MT(ASCP) R. W. Waynant, "Overview of Non-Invasive Fluid Flucose Mesurement Using Optical Techniques to Maintain Glucose Control in Diabetes Mellitus," *IEEE LEOS Newsletter*, vol. 12, no. 2, April 1998.

[17] Steven J. Barker, 2002. Principles of Pulse Oximetry Technology. Retrieved October 10, 2008 from www.oximeter.org

[18] Ricardo Gutierrez-Osuna, 2004. Notes on Pulse Oximetry. Retrieved form Texas A&M University, Department of Computer Science. http://courses.cs.tamu.edu/rgutier/cpsc483_s04/pulse_oximetry_notes.pdf

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