***Pulse Transit Time Measuring Device***

BME 200/300
University of Wisconsin – Madison
October 25, 2006

*Team*:
Jonathan Baran – Team Leader
Karen Chen –BSAC
William Stanford – BWIG
Mark Yarmarkovich – Communicator

*Client*:
Christopher G. Green, M.D.
Pediatric Pulmonology
University of Wisconsin Hospital

*Advisor*:
Wally Block
Department of Biomedical Engineering
University of Wisconsin – Madison

## Abstract

*Sleep apnea is a disorder affecting many children, causing them to awake from sleep to unblock their airways. Pulse transit time, the amount of time it takes for the pressure wave from the heart to travel to the finger, indirectly measures the presence of the disorder and its severity. Currently, only the tests conducted in sleep labs can test for sleep apnea and record its affects on the body. The work on this project will be in improving a device created by a previous engineering group that measures pulse transit time. The ultimate goal of the project is to create a device that can be used at home by the incorporation of a data logger that can record electrical signals from ECG and plethysmograph leads, improving the signal to noise ration by including an instrumentation amplifier and by modifying existing software to better detected the peaks of the stored voltage data.*

## Table of Contents

## Background Information

### ***Sleep Apnea***

Sleep apnea is a sleep disorder where the patient has a pauses breathing during sleep. Typical breathing rates occur anywhere from 10-20 breathes per minute. During sleep apnea, the tongue blocks the airway and a 10-30 second pause in breathing occurs, causing the sufferer to miss one to two breathes[i]. This problem could occur repeatedly during a night of sleep, which lowers the sleeping quality. Patients would experience symptoms such as snoring and restless sleeps. Moreover, they would have poor day performance, loss of concentration, anxiety and other negative health effects. Researchers have shown that people who are over the age of 40 are at higher risks of sleep apnea. However, it can affect people at any ages.

There are three different forms of sleep apnea – central sleep apnea, obstructive sleep apnea and mixed sleep apnea. Central sleep apnea causes pauses in breathing by the lack of effort in breathing. This is due to the failure of neurons in sending signals to indicate inhalation. In effect, the level of oxygen delivered to tissues decreases and oxygen is not available for cellular respiration. On the other hand, obstructive sleep apnea is where the air path inside the throat is blocked by an object, such as the tongue. As the muscles relax during sleep, the tongue can block the airway (as seen in Figure 1), which causes the patient to enter a lighter sleep stage or possibly cause the patient to awaken. Most patients suffering from obstructive apnea have trouble getting into a deep sleep state. Even though the light sleep time may be numerous, it is still not as effective as deep sleep. Finally, mixed apnea is the combination of central and obstructive sleep apnea. While obstructive sleep apnea takes place during sleep, central sleep apnea is often developed. Patients experience problems breathing and constantly wake up from sleep because of long-term obstructive apnea[i].

The pauses in breathing during sleep result from a decrease in the oxygen concentration of the blood. Furthermore, the level of carbon dioxide increases. The worst-case scenario is when the oxygen level becomes so low that it causes brain damage, and eventually leads to death.

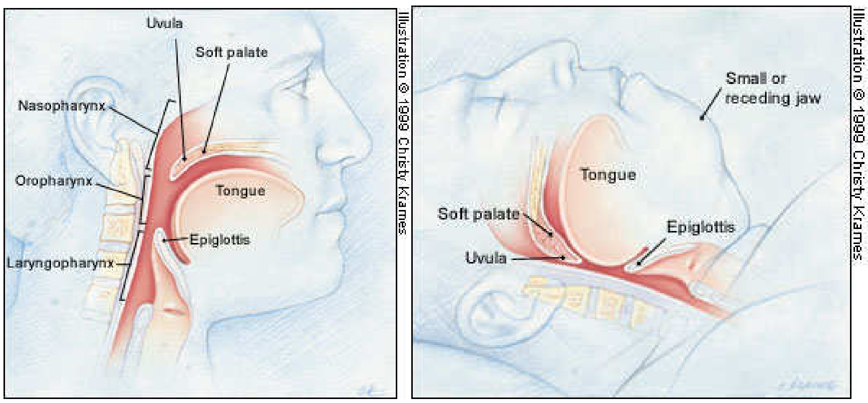



**Figure 1:** The left figure shows a normal trachea opening. The right figure shows that of a person who has multiple obstructions.

***Pulse Transit Time***

Pulse transit time (PTT) is a noninvasive method of measuring respiratory changes in children with breathing sleep disorders. PTT is the measure of the time it takes for the pulse pressure wave to go from the heart to the periphery[ii]. It is measured by use of both an ECG machine and a pulse oximeter. An ECG machine generates a curve based on the depolarization of the heart while the oximeter measures the pressure wave, or pulse, at the tip of the finger[iii]. A value for pulse transit time is given by calculating the difference in time between the peak of the R wave from the ECG and the peak of the pressure wave from the oximeter (Figure 2).

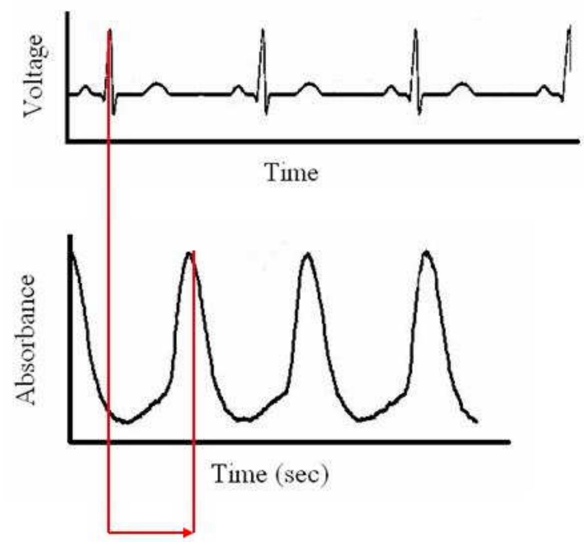



**Figure 2:** Calculation of pulse transit time from ECG and oximeter graphs.

Normal PTT range from 250-350 milliseconds[iv]; a significant variation in this time can help identify sleep apnea in two ways. First, as blood pressure decreases the arterial wall stiffness decreases. As stiffness decreases, it causes the pulse to take a longer time to reach the finger, causing on increase in PTT. This increases helps to diagnose sleep apnea. Second, the increase of blood pressure as the obstruction clears increases arterial wall. The increases in stiffness increases blood pressure, causing pulse transit time to decrease. Decreases in PTT in patients with sleep apnea can range from 15 to 50 milliseconds. Any decrease over 50 milliseconds is anatomically impossible. This decrease in PTT can help diagnose the severity of the apnea[v].

## Motivation

Some consequences of prolonged sleep apnea are hyperactivity, poor daytime performance, loss of concentration and other negative health effects. This sleep disorder can occur in both adults and children. To detect this disorder, patients need to participate in sleep studies at sleep centers. This is a rather expensive study to operate on a day-to-day basis, and thus its utilization is limited. Furthermore, sleep centers are not specifically for sleep apnea studies. Thus, a small, portable instrument that is able to conduct sleep studies at home would largely benefit the patients. This medical device must be able to detect PTT. Ideally, this device should

be as small as a flash drive that needs a low power supply. In addition, the number of wires should be reduced to allow more room for the patient to move. This allows easy access and operation, and the patient can self-detect sleep apnea in earlier stages.

## Client Requirements

The design must be able to be used with children. Our client works in pediatrics and wants to use the device to assess sleep apnea in children. This requires the device to be small, as to not cause discomfort for the child while sleeping. A miniaturizing of the device will allow families to conduct the tests at home. The device will be able to store many nights worth of data, which means the affects of sleep apnea can be analyzed on a day-to-day basis to determine if the condition varies nightly. Having the test conducted at home will also make the patient feel more comfortable.

The client has also specified three things he would like improved. The most important improvement is the incorporation of flash memory into the device. This memory will allow the collected data to be stored onto a memory stick that could be used to upload the data at a different time, eliminating the need for a bedside laptop computer. Next, he wants an improved LabView software program. The current software cannot account for when the recording device misses a beat due to patient movement. This inability causes the cascading of all data after the skipped beat. The new program should be able to edit out the skip beat and continue to graph regular beats as if no skipped beat occurred. Last is the improving and printing of the current circuit. Research shows that the existing circuit in the previous device needs several improvements in order to reduce the noise from surrounding frequencies[vi]. Once the circuit is improved, the printing will be an easy step and lead to a miniaturizing of the device. Since this step is relatively simple, compared with the others, the client would like more thought and effort to be given to the flash memory and the LabView program.

## Problem Statement

The current instruments used in the measurement of pulse transit time are inefficient for home use. An existing product with working ECG and pulse wave circuits along with software to analyze the data has been provided. The primary goal will be to optimize the existing setup for use at home. This will be performed by miniaturizing the circuit, increasing the signal to noise ratio, and improving the already existing software. These tasks will attempt to be rectified by numerous design additions.

### Previous Setup

### ***Electrocardiogram***

Willem Eitnthoven invented the ECG to measure heartbeats in 1901, for which he was awarded the Nobel Prize in Medicine in 1924. With each heartbeat, the cardiac tissue releases ions that depolarize the tissue. This creates a voltage of about 1 mV that can be measured with various leads attached at the body. Six ECG leads may be placed

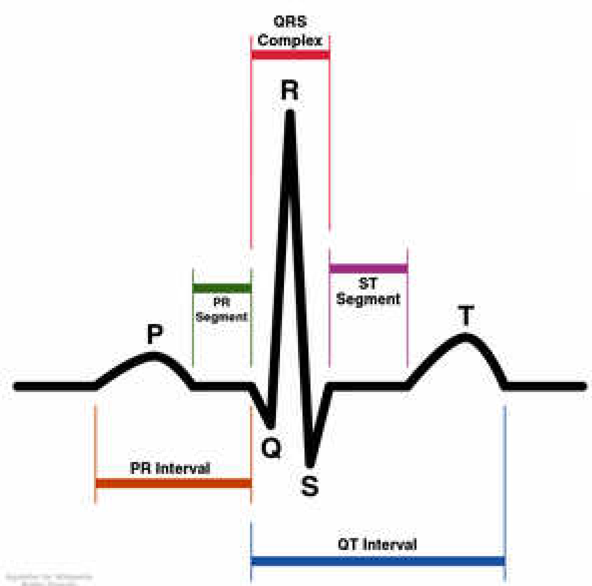



**Figure 3**: The Electrocardiogram

around the body in 30-degree intervals to measure the activity of the various sections of the heart based on the spatial readings. These readings form discrete waves represent the various stages of a heartbeat as seen in Figure 3. The P wave represents the current that causes the atrial contraction in which both the left and right atria contract simultaneously. The QRS complex represents the contraction of the left and right ventricles, a much stronger contraction of greater muscle mass. This results in a much greater reading on the ECG spanning a very short duration of time. Repolarization can be seen by observing the T wave that dips below the neutral voltage in re-establishing an action potential.

The ECG measures the small 1mV voltage generated at the peak of the depolarization. This small voltage is easily distorted from interfering electric fields from surrounding electronics or muscles. The signal fed into the circuit is filtered through a three-stage cascade: a buffer (Figure 4), a differential amplifier (Figure 5), and a bandpass filter (Figure 6). Resistors are used to ensure that minimal current is drawn from the body. Operational amplifiers are used to combine the signals from two leads and amplify the output voltage of this stage. The next stage involves a differential amplifier that uses the Common Mode Rejection Ratio (CMRR), which is a measure of the tendency of a device to reject common signal from multiple inputs. The bandpass stage is used to obtain desired frequencies by eliminating common interfering frequencies. The bandpass frequency is set to filter out frequencies that are out of the 160 mHz to 58.9 Hz range. This helps reduce the distortion of 60Hz noise from other devices and frequencies generated by muscle movement. This bandpass does filter some of the frequencies generated by

the heart; however, the QRS complex, which is of primary interest in measuring pulse transit time, is passed through the circuit at about 5-15Hz. The circuit in its entirety is seen in Figure 7.

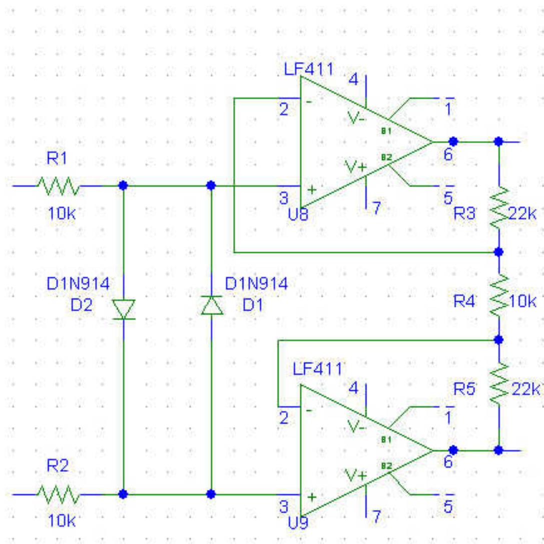



**Figure 4:** <u>The buffer stage of the ECG circuit</u>. The patient would be connected to $v_a$ and $v_b$ inputs. R1 and R2 add extra input impedance. The op amps provide high input impedance as to not load the patient and draw a lot of current that would distort the ECG signal. Diodes, D1 and D2, are used to minimize damage to the circuit from transient voltages. The gain for each buffer in this figure is 22k/10k=2.2 V/V.

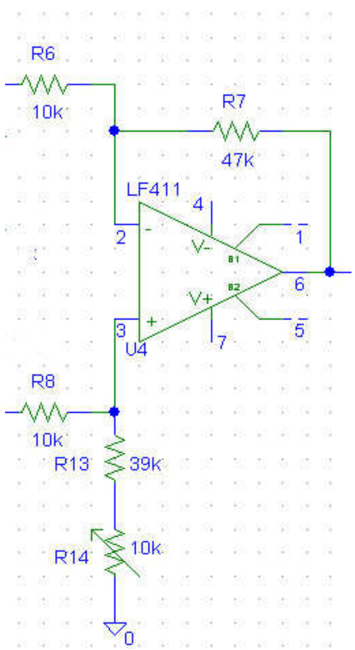



**Figure 5:** <u>The differential amplification stage of the ECG circuit design</u>. The potentiometer located at R14 can be adjusted to match R7 so that the common mode gain is minimal. This will improve the common mode rejection ratio (CMRR). The gain of this stage is 47k/10k = 4.7 V/V. Nodes *c* and *d* are the inputs from the output of the buffer stage. A voltage source of ±15 V/200 mA is used to power the op amp.

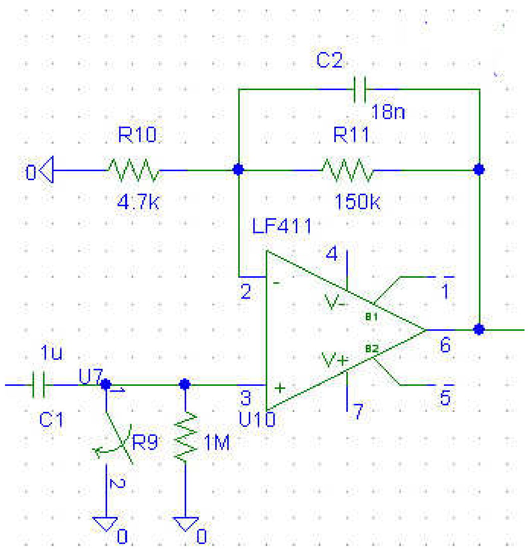
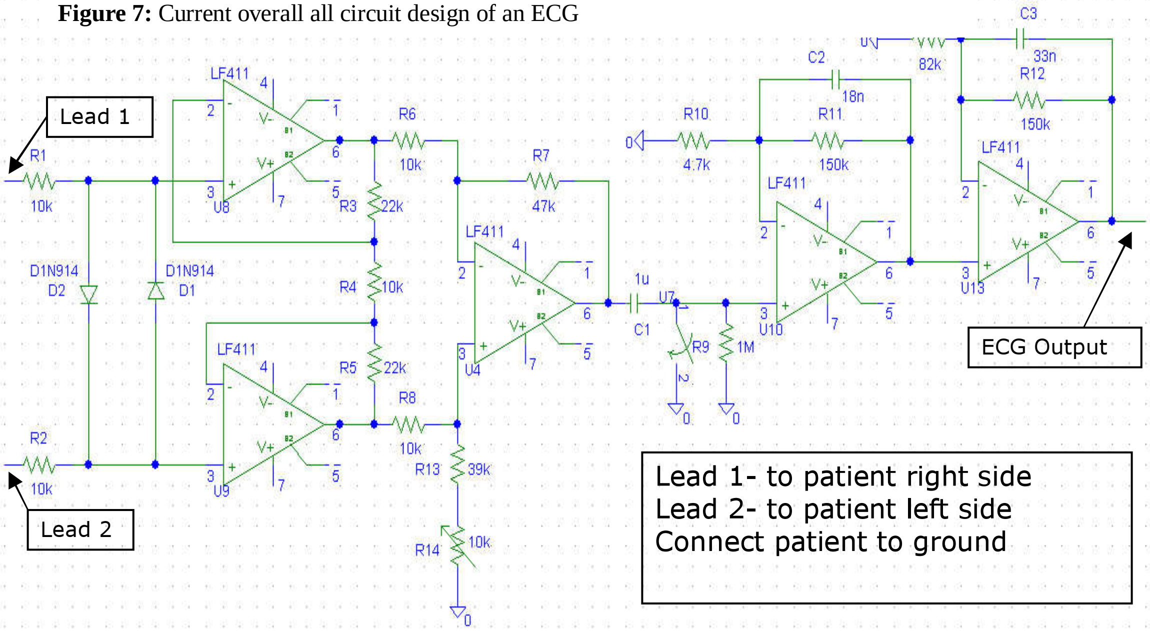



**Figure 6:** <u>The bandpass filter for the ECG circuit.</u> C1 and R9 make a highpass filter with a cutoff frequency of 160 mHz. C2 and R11 make a low pass filter with a cutoff frequency of 58.95 Hz. Node e is connected to the output of the differential amplifier. A voltage source of ±15 V/200 mA is used to power the op amp.

**Figure 7:** Current overall all circuit design of an ECG




## ***Plethysmograph***

A plethysmograph is a device designed to measure the changes of volume in a tissue or organ. For the fingertip, it is used as a non-invasive measure of the amount of blood arriving at the fingertip due to the pressure wave created by the contraction of the heart. The device's LED emits light at two wavelengths: 650nm and 805nm. The light waves pass through the finger and the remaining light is picked up by a phototransistor. A common finger plethysmograph is

manufactured by Nellcor. The configuration of the serial connection to the circuit is shown in Figure 8.

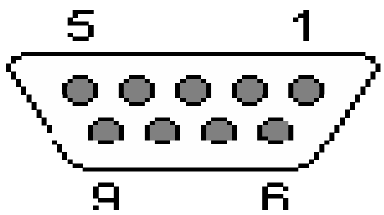



**Figure 8:**
Wire connections:
Pin 2 = Red (ground for LED)
Pin 3 = Black (power supply for LED)
Pin 5 = Orange (ground for phototransistor)
Pin 9 = Brown (circuit input for phototransistor)

A 1-4 mV signal is fed into the circuit, requiring several stages of operational amplifiers to process the signal. All op amps are powered by a +/- 15 V/200mA source, which also power the LED. The signal first passes through a bandpass filter to eliminate DC offset. Another filter is designed to filter common 60 Hz noise, similarly to the ECG circuit. Another bandpass filter is used to further reduce the noise. In the next stage, the signal is amplified to the required voltage of 4-8 V as required for measurement by the software. The final stage involves cutting out low frequencies produced by the motion of the finger. The four stages of the circuit are seen in Figure 9-12.

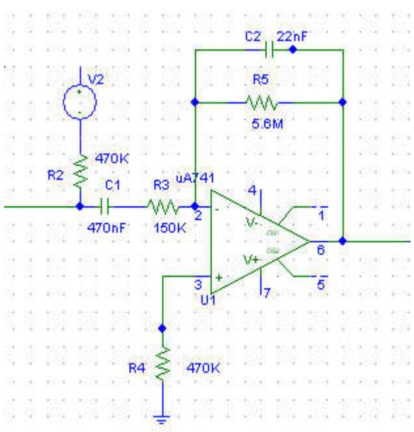



**Figure 9:** The first stage in the fingerplethysmograph circuit. It is an inverting bandpass filter with calculations. R2 and C1 create a HPF. C2 and R5 create a LPF. R5 and R3 provide gain. A voltage source of ±15 V/200 mA powers the op amp. The input of the circuit is connected to the finger probe.

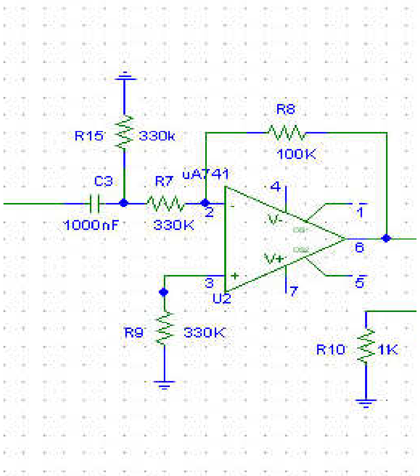



**Figure 10:** <u>The second stage in the finger plethysmograph</u>. It consists of a HPF, with R15 and C3 to setting the cutoff frequency at 0.4288Hz. A voltage source of ±15 V/200 mA powers the op amp. The input of the circuit is connected to the first bandpass filter.

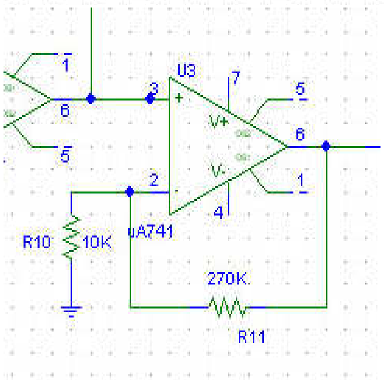



**Figure 11:** <u>The third stage of the finger plethysmograph</u>. R10 and R11 provide a gain of 28 V/V. A voltage source of ±15 V/200 mA powers the op amp. The input of the circuit is connected to the output of the second stage of the circuit.

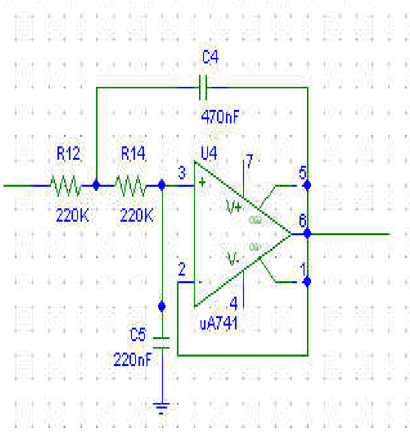



**Figure 12:** <u>The final stage of the finger plethysmogaph circuit</u>. The diagram is a sallen-key low pass filter, with a 2.32 Hz cutoff frequency. A voltage source of ±15 V/200 mA powers the op amp. The input of the circuit is connected to the output of the second stage of the circuit.

## ***<u>Existing LabVIEW</u>***

LabVIEW is a platform and development environment for a visual programming language from National Instruments. This graphical programming software lessens the complexity of software programming and coding. The language used by LabVIEW is called “G”, or the data flow language[vii]. To collect data, the interface is connected with different function-

nodes by wires. The wires gather data, inputs the data into LabVIEW, and the software records the data as waveforms. It is important that the user connects the nodes and any other outlets before opening the software. This will allow the components to be detected before the program starts running. In addition, there are three components, or virtual instruments, in LabVIEW. This is because its appearance and operation imitates a physical instrument, such as an oscilloscope. These instruments include block diagram, front panel and connector panel.

The former BME students have created a LabVIEW setup to calculate PTT and display it to the user (as seen in Figure 13). It also includes sampled ECG and plethysmograph waves. The program was programmed in LabVIEW 7.1. The configuration of the program is indicated as the following:

- BoardNum = the DAQ's board number from InstaCal installation
- LowChan = 0
- HighChan = 1
- Count = 1000 = (number of seconds you would like to calculate PTT for) / [(# of channels) * (rate) ]
- Rate = 100 = (# of channels) * (highest desired sampling rate)
- Range = +/- 10 V
- Cont/Sngl = SINGLE

With the configuration properly entered, the program is initiated by clicking the RUN button. The ECG and plethysmograph waves are input to the program using VIs from Measurement Computing's Universal Library for LabVIEW[viii]. Next, the frequency is set to 0.05 Hz. Meanwhile, the threshold level is used to detect the peaks and frequencies of the QRS complex and plethysmograph waves for each heartbeat. For the interpretation of the data, the difference between the QRS complex peak and plethysmograph wave peak is taken, and it is calculated to be PTT.

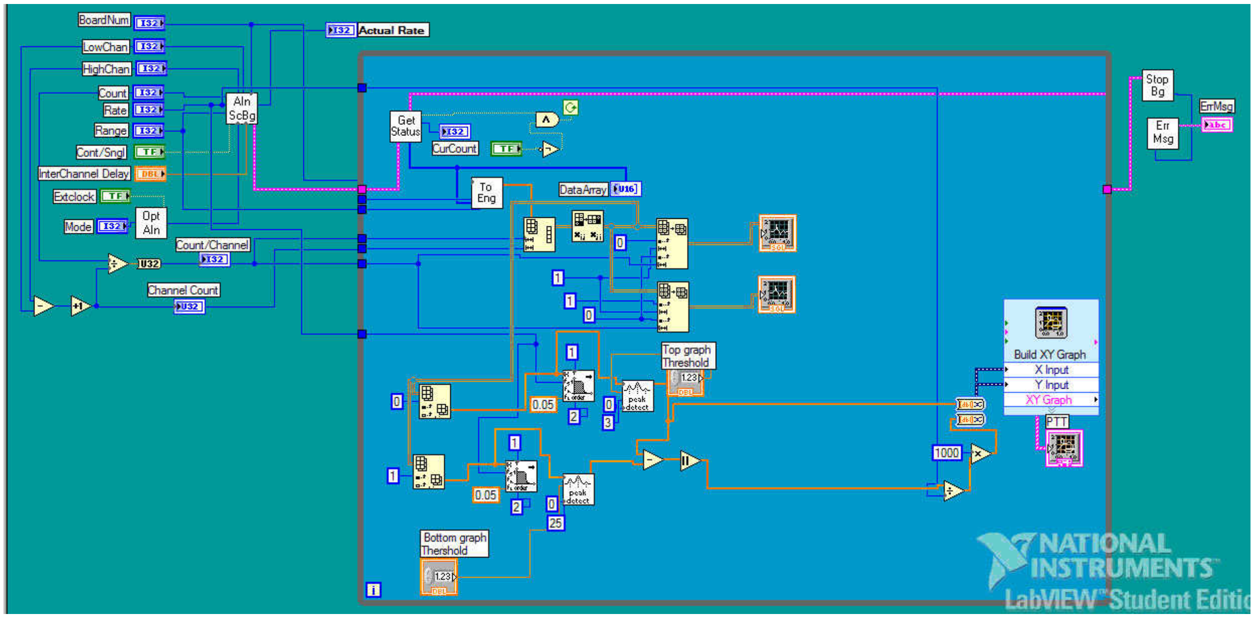



**Figure 13:** Block Diagram – on of the LabVIEW VIs.

## New Setup

## ECG Circuit

In order to provide the best signal output, the circuits must be optimized to provide the best signal possible. The ECG circuit was an area of particular interest as additional types of circuits can be used to provide a better signal to noise ratio. This increase in signal to noise ratio can be performed with the use of an instrumentation amplifier.

Two problems typically arise when dealing with the ECG circuits. One being the electrical potentials on the skin of a subject can be constantly changing, varying as much as 2 volts over time. This two volt signal is very small in comparison with the 1 mV signal which constitutes the electrical activity of the heart. Also, 60 Hz noise will be present anywhere around a subject. This signal is emitted from most electrical devices.

Since both of these signals are found in the environment, this signal will be common to both of the inputs. Therefore, in order to rid the output of both of these common noise signals, a high CMRR is needed. By rejecting this common signal to both, the true underlying signal can be found. In order to ensure that no more than 1% of the signal consists of this common noise, a CMRR of 100 decibels is desirable. Since this type of precision is hard to find in most standard operational amplifiers, an instrumentation amplifier can be used. The instrumentation amplifier, which is used in the new ECG circuit, is the Analog Devices AD624AD. The AD624AD was used because it could be made to have a gain of 1000, which is needed for the very weak ECG signals. It could also produce a CMRR of over 100 decibels. The instrumentation amplifier thus has the ability to be a very precise differential amplifier, or an amplifier which finds the difference of the two inputs and amplifies the signal. The use of the instrumentation amplifier will also provide the added benefit of taking up a smaller physical space compared with the previous five operation amplifier setup.

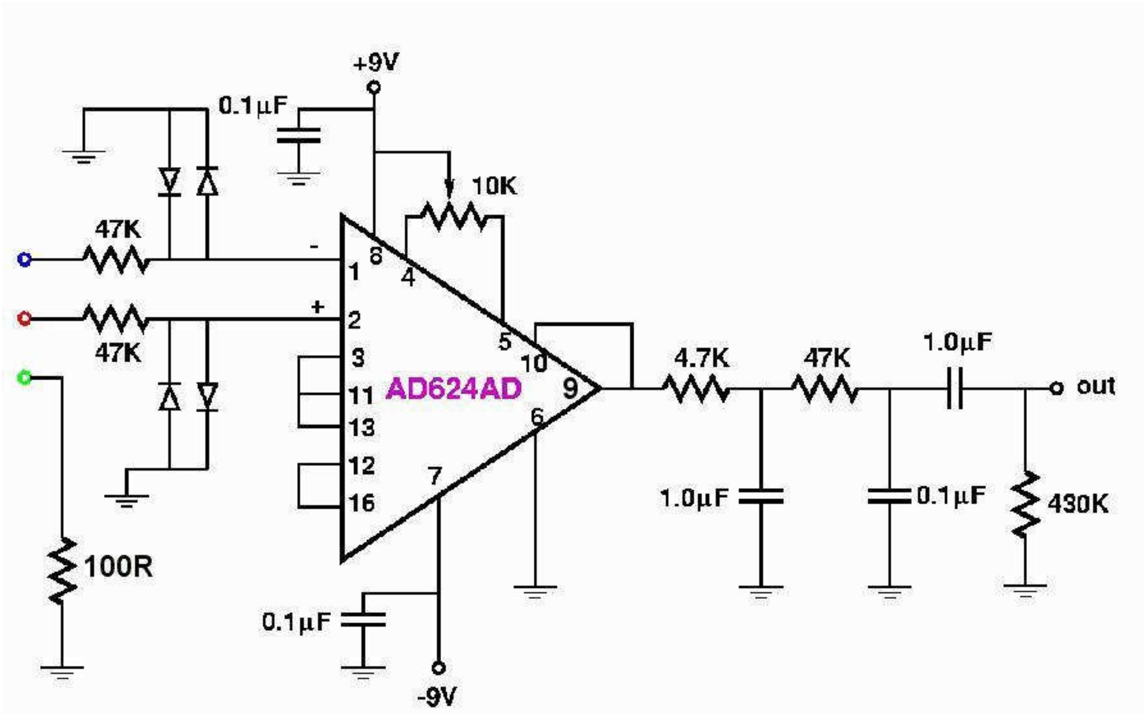



**Figure 14: The circuit diagram of the new circuit breadboard[ix].**

As can be seen from Figure 14, the Analog Devices AD624AD is at the heart of our new ECG. The two 47K resistors and the diodes are used for the safety of the subject. The 47K resistors increase the impedance in the case the signal was sent out of instrumentation amplifier. To further ensure the safety of the patients, the diodes were used to ground the output of current. To ensure the instrumentation amplifier has an output gain of 1000, pins 3,11,12,13, and 16 must be shorted as shown.

Once the output from the instrumentation is amplified, certain frequencies must be attenuated to decrease noise. Thus, a bandpass filter was created. Since the majority of the signal needed for the purposes of this setup resides within 30 Hz, a low pass filter is designed to attenuate signals greater than 33.9 Hz. The first low pass filter was designed using a 4.7K resister and a 1.0µF capacitor and the second was designed using a 47K

resistor and a 0.1µF capacitor. This can be calculated through the use of the following equation:

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

The addition of one low pass filter would mean the attenuation of 70.9% of the signal above 33.9 Hz. However, with the addition of a second low pass filter with a frequency response of 33.9 Hz, 50% of the signal greater than 33.9 Hz will be attenuated from the signal. Since the frequencies higher than 33.9 Hz are attenuated, the remaining 60 Hz noise will be removed from the sample. Since the DC offset is present in the signal, a high bandpass filter is added to the circuit. The DC offset accounts for variation in the baseline of the ECG signal. So the high bandpass filter will be used to attenuate signal lower than 0.37 Hz. The high bandpass filter was designed with a 1.0µF capacitor and a 430K resistor.

**New Design Setup**

The old setup had to be setup with a laptop at all times. This was a problem for the client as the device was not as portable, and problems may have arose because of a laptops sliding off a table while recording a patient during sleep. It would be beneficial to rid the setup of a laptop, which would make the setup much more portable compared with the previous design.

The addition of flash memory will allow the removal of the laptop computer from the setup. In order to implement a flash memory collector into the setup, three specific devices must be added to the circuit. First, an analog to digital converter must be added to allow the output voltages from the ECG and the finger plethysmogram circuit to be converted into binary, which can be read by the computer. The second element that needs

to be added is a microcontroller. The microcontroller is the “brains” of the operation. The microcontroller must be programmed to write the voltage outputs from the circuits to the memory storage device. Finally, a flash memory interface will be added to the design. This is how the flash memory will attach to the circuit and is the element that will transfer the data to the memory card.

In order to get the actual signal that is being output from the circuit, certain specifications of the system need to be met. The analog to digital converter is an important piece of the setup and has some requirements that need to be met in order to work with the circuits. The resolution of the ADC determines the accuracy that can be achieved with the converter. This is important because it will allow our client to best determine PTT. The use of a 12-bit ACD will allow for resolution of 1.2mV to be detected. Since a 12-bit converter will have $2^{12} = 4096$ “states”, then the output ranges of the circuits are 0-5 volts and the resolution will be equal to (5-0)/4096=1.22. This will be more than sufficient for the client to use in his data collection. The system must also have at least 3 IO slots. The inputs needs are: input from ECG circuit, input from finger plethysmogram circuit, and output to the flash memory. The sample rate needed to obtain an ECG signal is around 25 Hz and rate for the finger plethysmogram is around 7 Hz.  To avoid anti-aliasing, the Nyquist theorem states that the sampling frequency must be at least twice the maximum value. Therefore, the ECG is the limiting factor for the sampling rate and an absolute minimum of 50 Hz is required to get a signal. To error on the side of caution, a final sampling rate of 100 Hz was chosen.  In addition, the amount of memory needed for storage needed to be addressed. Since two channels running with a sample rate of 100 Hz for 8 hours (one nights sleep) uses approximately 10 MB of data, a

memory card of at least 32 MB will be needed to ensure adequate memory storage is available. In addition, since the ECG signal output can fall within +/- 1 V, a resolution of approximately 10 mV is needed. Moreover, the system will need to run on a battery source, so portability issues are maintained. Finally, the system must be small and compact, the final goal of the project is to have a compact design, and so this must be taken into consideration.

Since all of the specifications were determined, the best way to go about adding the memory card to the circuit needed to be determined. After doing preliminary research, it was found that creating this new device would require much more knowledge than originally thought. The decision to look elsewhere was made. After more research, a data logger was decided on. A data logger is an instrument that allows voltages to be inputted and stored onto a memory card.

After matching all of the specifications with the product, the Dataq 710-ULS was chosen. This product meets all of the specifications for our system. The datasheet for the Dataq 710-ULS is attached in the appendix.

## New software

### *Pulse Transit Time Calculating Algorithm*

The data acquired by the logger can be imported to the computer from the flash memory card and saved into a text file. A separate text file is created for both the ECG and the finger plethysmograph containing the corresponding voltages and times. The program used in calculating the pulse transit time is designed to determine the times corresponding to each R wave peak and the peak of the pressure waves, and subtracting these values to give the pulse transit time. The program is given 5 parameters in the command prompt: -e for the ECG file path, -f for the finger plethysmograph file path, -o

for the file to which the data is to be output, -te for the threshold value of the ECG, -tf for the threshold value for the finger plethysmograph data. An example parameter path could be: –e C:/ecg/ecg.txt –f C:/ecg/finger.txt –o C:/ecg/output.txt –te 4 –tf 0.04. The program stores these values in an array for each of the two data sets. The program proceeds to search through the points to determine the maximum peak values. A "while" loop is used to read the data until the end of the text file is reached. Running through the data, the program determines the points within the range of the threshold value as entered by the user. "For" loops are used to cycle through all of the points and compare them to the other data to determine if they qualify as a peak. This is done for both of the data sets and stored in separate arrays. The times from each array are then subtracted from one another and the corresponding pulse transit times are output into the designated file using a "for" loop to cycle through the data for the number of peaks entered.

```
#include <string.h>
#include <iostream>
#include <fstream>
#include <sstream>
#include <vector>


int main(int argc, char* argv[])
{

  const char* inputEKGFileName = NULL;
  const char* inputFingerFileName = NULL;
  const char* outputFileName = NULL;
  double thresoldEKG = 0;
  double thresoldFinger = 0;

  int i;
  for (i = 1; i < argc; i++) {
    if (strcmp(argv[i], "-e") == 0) {
      inputEKGFileName = argv[++i];
    }
```

```
    else if (strcmp(argv[i], "-f") == 0) {
      inputFingerFileName = argv[++i];
    }
    else if (strcmp(argv[i], "-o") == 0) {
      outputFileName = argv[++i];
    }
    else if (strcmp(argv[i], "-te") == 0) {
      std::stringstream ss;
      ss << argv[++i];
      ss >> thresoldEKG;
    }
    else if (strcmp(argv[i], "-tf") == 0) {
      std::stringstream ss;
      ss << argv[++i];
      ss >> thresoldFinger;
    }
    else {
      std::cerr << "usage: " << argv[0] << "\n"
                << " [-h (for help)]\n"
                << " -e (EKG input file name, no
spaces!)\n"
                << " -f (Finger input file name, no
spaces!)\n"
                << " -o (input file name, no spaces!)\n"
                << " -te (thresold value fo EKG)\n"
                << " -tf (thresold value finger)\n"
                << std::endl;
      return 0;
    }
  }
  std::ifstream infileEKG(inputEKGFileName);
  std::ifstream infileFinger(inputFingerFileName);
  std::ofstream outfile(outputFileName);

  double threshValue = -1000;
  double minTimeInterval = 0.25;

  // PROCESS EKG DATA

  // read two columns of data from input EKG file, should
be space separated
  double t,v;
  std::vector<double> timeEKG;
  std::vector<double> voltsEKG;
  while(infileEKG >> t >> v) {
    timeEKG.push_back(t);
    voltsEKG.push_back(v);
```

```
  }


  // threshold volts array
  std::vector<double> voltsEKGThresh;
  for( i=0; i<voltsEKG.size(); i++) {
    if (voltsEKG[i] < thresoldEKG) {
      voltsEKGThresh.push_back(threshValue);
    }
    else {
      voltsEKGThresh.push_back(voltsEKG[i]);
    }
  }

  // find local max in threshold array
  std::vector<double> peakTimesEKG;
  for( i=1; i<voltsEKGThresh.size()-1; i++) {
    if (voltsEKGThresh[i] != threshValue &&
      voltsEKGThresh[i] >= voltsEKGThresh[i-1] &&
      voltsEKGThresh[i] >= voltsEKGThresh[i+1]) {
      if (peakTimesEKG.size() == 0) {
        peakTimesEKG.push_back(timeEKG[i]);
      }
      else if (timeEKG[i] -
peakTimesEKG[peakTimesEKG.size()-1] > minTimeInterval ) {
        peakTimesEKG.push_back(timeEKG[i]);
      }
    }
  }


  // PROCESS FINGER DATA

  // read two columns of data from input finger file,
should be space separated
  std::vector<double> timeFinger;
  std::vector<double> voltsFinger;
  while(infileFinger >> t >> v) {
    timeFinger.push_back(t);
    voltsFinger.push_back(v);
  }


  // threshold volts array
  std::vector<double> voltsFingerThresh;
  for( i=0; i<voltsFinger.size(); i++) {
    if (voltsFinger[i] < thresoldFinger) {
```

```
      voltsFingerThresh.push_back(threshValue);
    }
    else {
      voltsFingerThresh.push_back(voltsFinger[i]);
    }
  }

  // find local max in thresholded array
  std::vector<double> peakTimesFinger;
  for( i=1; i<voltsFingerThresh.size()-1; i++) {
    if (voltsFingerThresh[i] != threshValue &&
      voltsFingerThresh[i] >= voltsFingerThresh[i-1] &&
      voltsFingerThresh[i] >= voltsFingerThresh[i+1]) {
      if (peakTimesFinger.size() == 0) {
        peakTimesFinger.push_back(timeFinger[i]);
      }
      else if (timeFinger[i] -
peakTimesFinger[peakTimesFinger.size()-1] > minTimeInterval
) {
        peakTimesFinger.push_back(timeFinger[i]);
      }
    }
  }

  // output diff between peak times
  int n = peakTimesEKG.size();
  if (peakTimesFinger.size() < n) {
    n = peakTimesFinger.size();
  }

  for( i=0; i<n; i++) {
    outfile << peakTimesEKG[i] << " , "<<
peakTimesFinger[i]<< " , " << peakTimesEKG[i]-
peakTimesFinger[i] << std::endl;
  }

  infileEKG.close();
  infileFinger.close();
  outfile.close();
  return 0;
}
```

## Testing

The previous and new ECG circuits are compared against each other in the following diagrams (Figure 16 and 17 respectively).

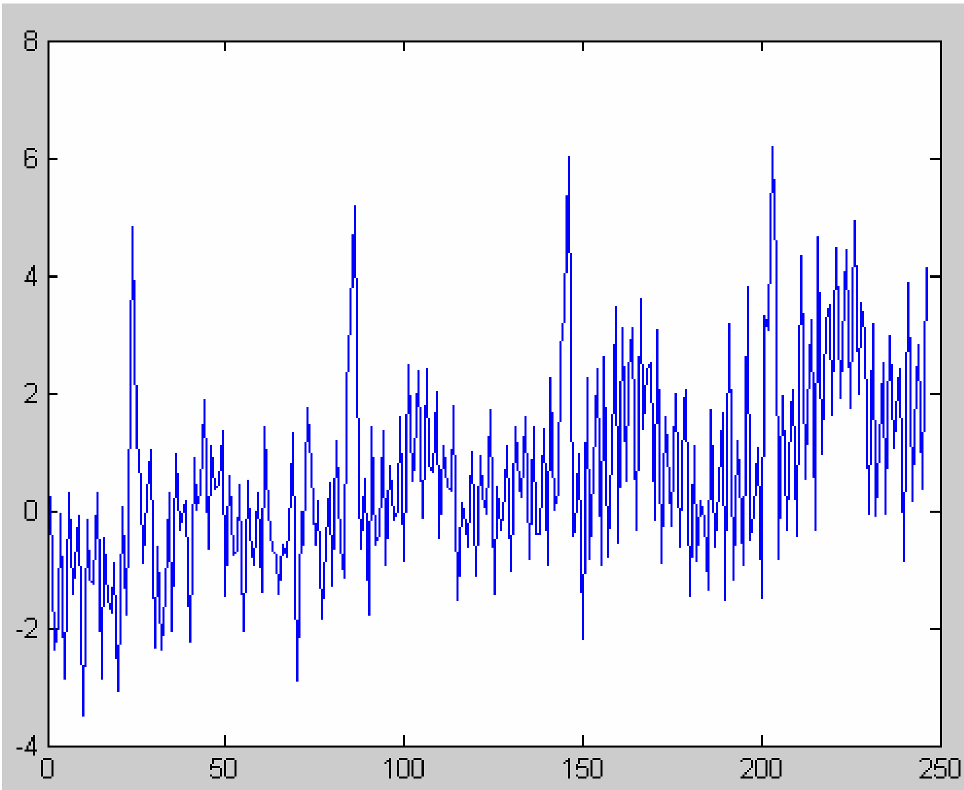



**Figure 16: Electrocardiogram signal from old circuit.**

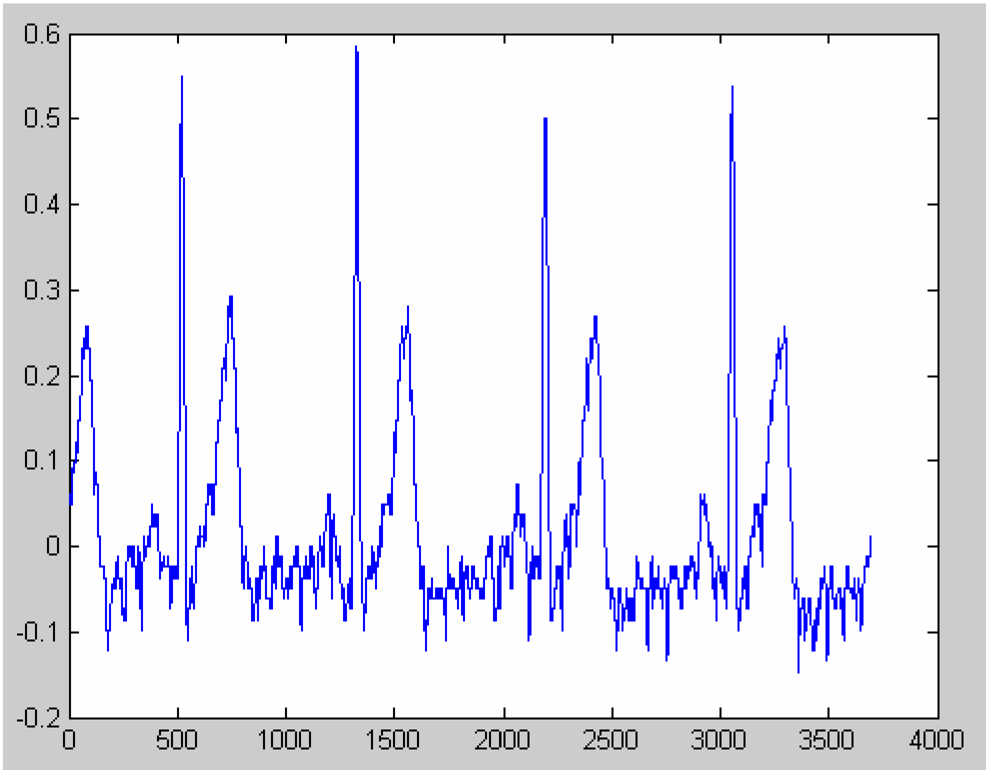



**Figure 17: Electrocardiogram from the new ECG circuit recorded by the data logger**

As can be seen from the two graphs, the previous ECG signal has a large amount of noise present around the 60 Hz frequency. This is due to the use of the low pass filter that has a frequency response of only 58.9 Hz and thus not attenuated the 60 Hz frequencies.. The new ECG signal, however, does show a much cleaner signal compared with the previous signal due to the use of the bandpass filter that reduces the noise and stabilizes the signal.

## Future Work

The future work of this project is split into three categories: the circuit, the overall setup and the software. To further reduce the noise of the circuit and its overall size, it would be beneficial to obtain a printed circuit. A printed circuit was not made in this phase of the project so that further research could be made into the circuit to make sure it truly is the best design. If

the new design is the best, then a printed circuit can be made. A printed circuit would reduce the interference from wire crossings that exist on a breadboard. Printing the circuit would also compact the circuit so that it could better fit into a small device.

Once the circuit has been printed, the next step will to be to combine the ECG and plethysmograph circuits, the data logger, and the power supply into one unit. This unit should be as small as possible so that it causes the least discomfort to the patient. The design must allow easy access to the batteries for when they need changing and to the memory card slot so there is no hassle in removing the card. The power supply is another area that could be looked to increase efficiency. The current device runs off of several 9V batteries that take up a lot of space in the device and are quite costly. Future work should be focused on determining if another type of battery can be used or if there is another configuration design for the batteries. Rechargeable batteries could also be looked into and designed to be similar to the ones used in cell phones.

Moreover, the software needs to continue to be improved. The current software only records in a previously defined period. If the heart rate increases or decreases during sleep, then the period between R waves can change; the software does not know how to recognize this. The client specified this would be an important feature to look at. An algorithm also needs to be developed to further reduce the noise of the ECG signal. Modifications to the circuit can only reduce the noise to a certain level and cannot account for the noise generated from movement of the patient. If the signals experience a few disturbances during the night, it may be possible for the software to recognize the long pause between recognizable R waves and be able to reset itself once a true R wave is found. The client is willing to sacrifice a few data points for a cleaner data set.

The improvements made to this design over the semester have made the miniaturization of the device possible. The incorporation of a data logger enhances its portability and its safety. The new ECG circuit decreases the noise in the signal and combined with the new software, allows for a more accurate measure of PTT. Future work on this project will bring it closer to

being ready for clinical trial and to the ultimate goal of the project: incorporation into sleep studies.

October 25, 2006
Updated: December 11, 2006

Jonathan Baran
Mark Yarmarkovich
William Stanford
Karen Chen

# Pulse Transit Time Device PDS

## Problem Statement

The current instruments used in the measurement of pulse transit time are inefficient for home use. An existing product with working ECG and pulse wave circuits along with software to analyze the data has been provided. The primary goal will be to optimize the existing setup for use at home. This will be performed by miniaturizing the circuit, increasing the signal to noise ratio, and improving the already existing software. These tasks will attempt to be rectified by numerous design additions.

## Client Requirements

- Optimized for use with children
- Integrate with previously designed system
- Write program to deal with skipped heart beat
- Integrate flash memory to make device more compact
- Improve signal to noise ratio

## Design Requirements

### 1. Physical and Operational Characteristics

a. *Performance requirements*: The probe needs to be small enough to stay fixed to children's fingers. More over, the probe needs to stay attached to the finger during sleep. The monitoring system needs to be non-evasive and not interfere with sleep patterns. The monitoring

software needs to record the signals from the finger probe and ECG leads while filtering out skipped beats.

b. *Safety*: Because the device will be used while children are sleeping, the chords must not present a strangling hazard. Also circuits must be made to minimize electrical hazards.

c. *Shelf life*: The device must be operated by batteries to allow it to be compact. The adhesive needs to be reusable.

d. *Accuracy and Reliability*: The device must employ a circuit design the takes into account both gain and CMRR.  A gain of 1000 is need to amplify the ECG signal and a CMRR of 100 decibels is also needed

e. *Life in Service*: The device should be able to be used for multiple uses by a single patient and then repeated for many more patients.

f. *Operating Environment*: Ideally, the device will be able to be used in the patient's home while they are sleeping. If optimal minimization is achieved, it could be worn on the waste.

g. *Size*: The device needs to be small enough to fit on a bed size table, or optimally, attached to the waste of the patient to allow the chords to be non-evasive.

h. *Weight*: The device needs to be as light as possible so it will not be felt while sleeping.

i. *Materials*: Plastic would be used for the casing of the device.

**2. Product Characteristics**

a. *Quantity*: For the time being, only one device is necessary. If the device is successful, multiple units may be made in the future.

b. *Target Product Cost*: The device should be as inexpensive as possible so not to burden families when the device is taken home.

**3. Miscellaneous**

a. *Standards and Specifications*: The device must be meet FDA medical device specifications.

b. *Customer*: Christopher Green, M.D. Dept. of Pediatrics UW Medical School

c. *Patient-related concerns*: The patient's personal information will not be stored in the device. No sterilization of the device will be needed in-between uses aside from the application of new EKG patches at every use.

d. *Competition*: Probe devices with LED are on the market but there are no devices that measure blood flow and pulse transit time.

# Appendix

## Appendix Figure 1: Datasheet for the Dataq 710-ULS[x]

# DI-710 Specifications

### Analog Inputs

| | |
|---|---|
| **Number of Channels:** | 16 |
| **Channel Configuration:** | 16 single-ended; 8 differential; programmable per channel |

**Measurement range, Accuracy, and Resolution**

| | Gain | Range | Accuracy | Resolution |
|---|---|---|---|---|
| PGL models: | 1 | ±10V | ±.25%FSR | ±1.22mV |
| | 10 | ±1V | ±.25%FSR | ±122µV |
| | 100 | ±100mV | ±.25%FSR | ±12.2µV |
| | 1000 | ±10mV | ±.25%FSR | ±1.22µV |
| PGH models: | 1 | ±10V | ±.25%FSR | ±1.22mV |
| | 2 | ±5V | ±.25%FSR | ±610µV |
| | 4 | ±2.5V | ±.25%FSR | ±305µV |
| | 8 | ±1.25V | ±.25%FSR | ±153µV |

| | |
|---|---|
| **Input Impedance, single-ended:** | 1MΩ |
| **Input impedance, differential:** | 1MΩ each input to common |
| **Input bias current:** | 10µA for a 10V input, single channel |
| **Input offset voltage:** | Auto-zero |
| **Input offset current:** | 2nA (single channel) |
| **Max. normal mode voltage:** | 30V DC or peak AC |
| **Max. common mode voltage:** | 30V DC or peak AC |
| **Common mode rejection:** | 80db, gain=1, 1KΩ unbalance |
| **Channel-to-channel crosstalk rejection:** | -75db @ 100Ω unbalance |
| **Gain temperature coefficient:** | 50 ppm/°C |
| **Offset temperature coefficient:** | 0.25µV/°C |
| **Digital filtering:** | Standard: Conditional over-sampling<br>Stand-alone: None |

### A/D Characteristics

| | |
|---|---|
| **Type:** | Successive approximation |
| **Resolution:** | 14-bit |
| **Monotonicity:** | ±2 LSB |
| **Conversion Time:** | 69µs |

### Scanning Characteristics

| | |
|---|---|
| **Max. throughput sample rate:*** | Standard: 4,800 Hz<br>Stand-alone: 14,400 Hz** |

*When acquiring more than one channel at a gain of 100 max throughput is 7200 Hz; When acuiring more than one channel at a gain of 1000 max throughput is 900 Hz.
**Assumes SD memory latencies of 80 milliseconds or less.

| | |
|---|---|
| **Min. throughput sample rate:** | Standard: 0.0034 Hz<br>Stand-alone: 0.0017 Hz |
| **Max. scan list size:** | 17 entries |
| **Sample buffer size:** | 2kb |

### Controls (Stand-alone models)

| | |
|---|---|
| **Single push-button:** | Manual control Record and Standby |

### Calibration

| | |
|---|---|
| **Calibration cycle:** | One year |
| **Calibration method:** | Automated Calibration Software, provided. |

### Digital I/O

| | |
|---|---|
| **Bits:** | 8 bidirectional bits |
| **Configuration:** | Each bit is programmable as Input or Output |
| **Output voltage levels:** | Min. "1" 3V @ 2.5mA sourcing<br>Max. "0" 0.4V @ 2.5mA sinking |
| **Output current:** | Max. source, -2.5 mA; Max. sink, 2.5mA |
| **Input voltage levels:** | Min. required "1" 2V; Max allowed "0" 0.8V |

### Ethernet Interface

| | |
|---|---|
| **Type:** | 10/100Base-T |
| **Connector:** | RJ-45 |
| **Protocol:** | TCP/IP |
| **Server Type:** | DHCP |

### Removable Memory (Stand-alone models)

| | |
|---|---|
| **Type:** | SD (Recommended: SanDisk SD Card) |
| **Capacity:** | 16 Mb to 1 Gb |

### Real Time Clock (Stand-alone models)

| | |
|---|---|
| **Type:** | Date, hour, minute, second |
| **Resolution:** | 1 second |
| **Accuracy:** | 20 ppm |

### Indicators

| | |
|---|---|
| **Stand-alone models:** | Three-color LED indicating Record, Standby, and Error conditions |
| **Standard models:** | Power LED |

### Transfer Rate to PC

| | |
|---|---|
| **Real Time:** | up to 4,800 samples per second |
| **From Memory Card:** | up to 2,400 samples per second (Ethernet only) |

### General

| | |
|---|---|
| **Panel indicators:** | Mode LED |
| **Panel Controls:** | Control push button (Stand-alone models) |
| **Panel Slots:** | Accepts SD-type flash memory |
| **Input connectors:** | Two, removable sixteen position terminal blocks |
| **Operating Environment:** | 0°C to 70°C |
| **Enclosure:** | Aluminum base with steel wrap-around. Aluminum end-panels with plastic bezels. |
| **Dimensions:** | $5\frac{7}{16}$"D × $4\frac{1}{8}$"W × $1\frac{1}{2}$"H<br>13.81D × 10.48W × 3.81H cm. |
| **Weight:** | 14 oz. |
| **Power Requirements:** | USB: 9 to 36 VDC, 2 watts max<br>Ethernet: 9 to 36 VDC, 2.5 watts max |

**References**


[i] *Sleep apnea.* (2006). Retrieved October 9, 2006 from http://en.wikipedia.org/wiki/Sleep_apnea

[ii] Smith R, Argod J, Pépin JL, Lévy P. 1999. Pulse Transit Time: an appraisal of potential clinical applications. *Thorax*. 54:452-458.

[iii] Pagani J, Pia M, Calcagnini G, Alterio A, Ambrosio R, Censi F, Ronchetti R. 2003. Pulse Transit Time as Measure of Inspiratory Effort in Children. *Chest*. 124:1487-1493.

[iv] Karas, A., Hondl, B., Olson, M., & Cohen, Z. (2005). *Measurement of pulse transit time.* Unpublished manuscript. Retrieved September 25, 2006,

[v] Katz E, Lutz J, Black C, Marcus C. 2003. Pulse Transit Time as a Measure of Arousal and Respiratory Effort in Children with Sleep-Disordered Breathing. *Pediatr Res*. 53:580-588.

[vi] Carlson, S. (2000, the amateur scientist: Home is where the ECG is. *Scientific American Magazine,* (June) 3. Retrieved October 17, 2006,

[vii] *Labview.* (2006). Retrieved October 1, 2006 from http://en.wikipedia.org/wiki/LabVIEW

[viii] *LabVIEW tutorials.* (2005). Retrieved October 5, 2006 from http://www.upscale.utoronto.ca/GeneralInterest/LabView.html

[ix] *Electrocardiogram (ECG) project for DrDaq.* (2006). Retrieved October 12, 2006 from http://www.picotech.com/applications/ecg.html#cct

[x] Dataq Instruments. (2006). Retrieved on December 11, 2006 from http://www.dataq.com/support/documentation/pdf/datasheets/710ds.pdf

# *Pulse Transit Time Measuring Device*

BME 200/300
University of Wisconsin – Madison
October 25, 2006

*Team*:
Jonathan Baran – Team Leader
Karen Chen –BSAC
William Stanford – BWIG
Mark Yarmarkovich – Communicator

*Client*:
Christopher G. Green, M.D.
Pediatric Pulmonology
University of Wisconsin Hospital

*Advisor*:
Wally Block
Department of Biomedical Engineering
University of Wisconsin – Madison

## Abstract

*Sleep apnea is a disorder affecting many children, causing them to awake from sleep to unblock their airways. Pulse transit time, the amount of time it takes for the pressure wave from the heart to travel to the finger, indirectly measures the presence of the disorder and its severity. Currently, only the tests conducted in sleep labs can test for sleep apnea and record its affects on the body. The work on this project will be in improving a device created by a previous engineering group that measures pulse transit time. The ultimate goal of the project is to create a device that can be used at home by the incorporation of a data logger that can record electrical signals from ECG and plethysmograph leads, improving the signal to noise ration by including an instrumentation amplifier and by modifying existing software to better detected the peaks of the stored voltage data.*

## Table of Contents

## Background Information

### ***Sleep Apnea***

Sleep apnea is a sleep disorder where the patient has a pauses breathing during sleep. Typical breathing rates occur anywhere from 10-20 breathes per minute. During sleep apnea, the tongue blocks the airway and a 10-30 second pause in breathing occurs, causing the sufferer to miss one to two breathes[i]. This problem could occur repeatedly during a night of sleep, which lowers the sleeping quality. Patients would experience symptoms such as snoring and restless sleeps. Moreover, they would have poor day performance, loss of concentration, anxiety and other negative health effects. Researchers have shown that people who are over the age of 40 are at higher risks of sleep apnea. However, it can affect people at any ages.

There are three different forms of sleep apnea – central sleep apnea, obstructive sleep apnea and mixed sleep apnea. Central sleep apnea causes pauses in breathing by the lack of effort in breathing. This is due to the failure of neurons in sending signals to indicate inhalation. In effect, the level of oxygen delivered to tissues decreases and oxygen is not available for cellular respiration. On the other hand, obstructive sleep apnea is where the air path inside the throat is blocked by an object, such as the tongue. As the muscles relax during sleep, the tongue can block the airway (as seen in Figure 1), which causes the patient to enter a lighter sleep stage or possibly cause the patient to awaken. Most patients suffering from obstructive apnea have trouble getting into a deep sleep state. Even though the light sleep time may be numerous, it is still not as effective as deep sleep. Finally, mixed apnea is the combination of central and obstructive sleep apnea. While obstructive sleep apnea takes place during sleep, central sleep apnea is often developed. Patients experience problems breathing and constantly wake up from sleep because of long-term obstructive apnea[i].

The pauses in breathing during sleep result from a decrease in the oxygen concentration of the blood. Furthermore, the level of carbon dioxide increases. The worst-case scenario is when the oxygen level becomes so low that it causes brain damage, and eventually leads to death.

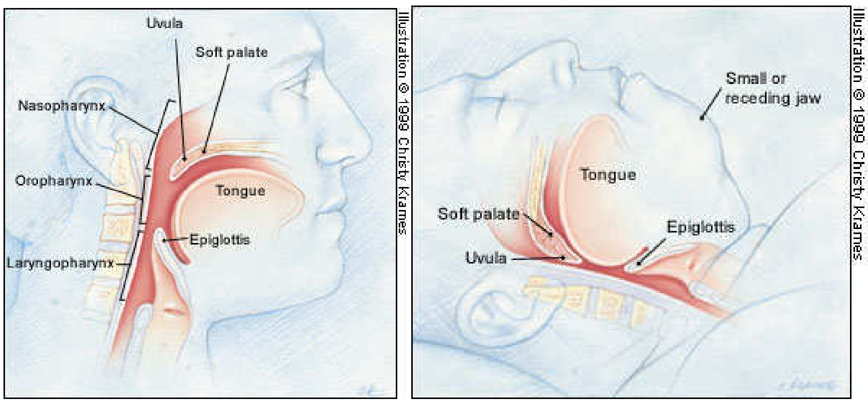



**Figure 1:** The left figure shows a normal trachea opening. The right figure shows that of a person who has multiple obstructions.

## ***Pulse Transit Time***

Pulse transit time (PTT) is a noninvasive method of measuring respiratory changes in children with breathing sleep disorders. PTT is the measure of the time it takes for the pulse pressure wave to go from the heart to the periphery[ii]. It is measured by use of both an ECG machine and a pulse oximeter. An ECG machine generates a curve based on the depolarization of the heart while the oximeter measures the pressure wave, or pulse, at the tip of the finger[iii]. A value for pulse transit time is given by calculating the difference in time between the peak of the R wave from the ECG and the peak of the pressure wave from the oximeter (Figure 2).

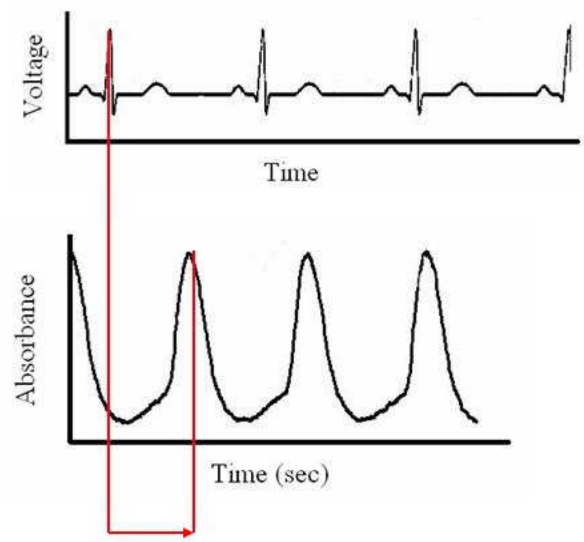



**Figure 2:** Calculation of pulse transit time from ECG and oximeter graphs.

Normal PTT range from 250-350 milliseconds[iv]; a significant variation in this time can help identify sleep apnea in two ways. First, as blood pressure decreases the arterial wall stiffness decreases. As stiffness decreases, it causes the pulse to take a longer time to reach the finger, causing on increase in PTT. This increases helps to diagnose sleep apnea. Second, the increase of blood pressure as the obstruction clears increases arterial wall. The increases in stiffness increases blood pressure, causing pulse transit time to decrease. Decreases in PTT in patients with sleep apnea can range from 15 to 50 milliseconds. Any decrease over 50 milliseconds is anatomically impossible. This decrease in PTT can help diagnose the severity of the apnea[v].

## Motivation

Some consequences of prolonged sleep apnea are hyperactivity, poor daytime performance, loss of concentration and other negative health effects. This sleep disorder can occur in both adults and children. To detect this disorder, patients need to participate in sleep studies at sleep centers. This is a rather expensive study to operate on a day-to-day basis, and thus its utilization is limited. Furthermore, sleep centers are not specifically for sleep apnea studies. Thus, a small, portable instrument that is able to conduct sleep studies at home would largely benefit the patients. This medical device must be able to detect PTT. Ideally, this device should

be as small as a flash drive that needs a low power supply. In addition, the number of wires should be reduced to allow more room for the patient to move. This allows easy access and operation, and the patient can self-detect sleep apnea in earlier stages.

## Client Requirements

The design must be able to be used with children. Our client works in pediatrics and wants to use the device to assess sleep apnea in children. This requires the device to be small, as to not cause discomfort for the child while sleeping. A miniaturizing of the device will allow families to conduct the tests at home.  The device will be able to store many nights worth of data, which means the affects of sleep apnea can be analyzed on a day-to-day basis to determine if the condition varies nightly.  Having the test conducted at home will also make the patient feel more comfortable.

The client has also specified three things he would like improved. The most important improvement is the incorporation of flash memory into the device. This memory will allow the collected data to be stored onto a memory stick that could be used to upload the data at a different time, eliminating the need for a bedside laptop computer. Next, he wants an improved LabView software program. The current software cannot account for when the recording device misses a beat due to patient movement. This inability causes the cascading of all data after the skipped beat. The new program should be able to edit out the skip beat and continue to graph regular beats as if no skipped beat occurred. Last is the improving and printing of the current circuit. Research shows that the existing circuit in the previous device needs several improvements in order to reduce the noise from surrounding frequencies[vi]. Once the circuit is improved, the printing will be an easy step and lead to a miniaturizing of the device. Since this step is relatively simple, compared with the others, the client would like more thought and effort to be given to the flash memory and the LabView program.

## Problem Statement

The current instruments used in the measurement of pulse transit time are inefficient for home use. An existing product with working ECG and pulse wave circuits along with software to analyze the data has been provided. The primary goal will be to optimize the existing setup for use at home. This will be performed by miniaturizing the circuit, increasing the signal to noise ratio, and improving the already existing software. These tasks will attempt to be rectified by numerous design additions.

### Previous Setup

### ***Electrocardiogram***

Willem Eitnthoven invented the ECG to measure heartbeats in 1901, for which he was awarded the Nobel Prize in Medicine in 1924. With each heartbeat, the cardiac tissue releases ions that depolarize the tissue. This creates a voltage of about 1 mV that can be measured with various leads attached at the body. Six ECG leads may be placed

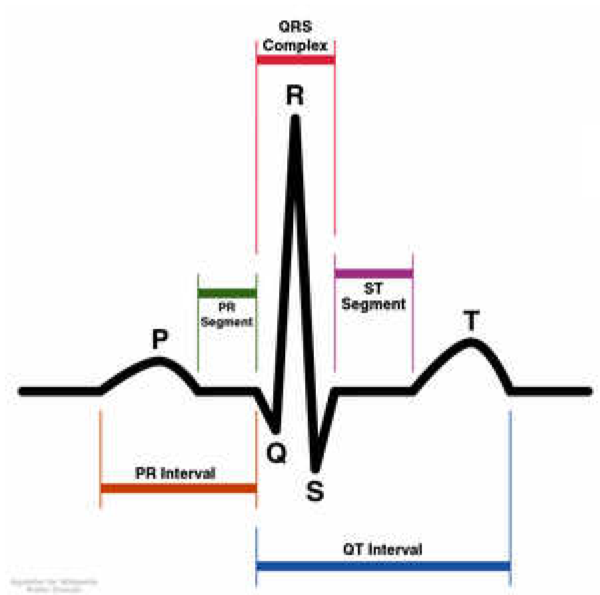



**Figure 3**: The Electrocardiogram

around the body in 30-degree intervals to measure the activity of the various sections of the heart based on the spatial readings. These readings form discrete waves represent the various stages of a heartbeat as seen in Figure 3. The P wave represents the current that causes the atrial contraction in which both the left and right atria contract simultaneously. The QRS complex represents the contraction of the left and right ventricles, a much stronger contraction of greater muscle mass. This results in a much greater reading on the ECG spanning a very short duration of time. Repolarization can be seen by observing the T wave that dips below the neutral voltage in re-establishing an action potential.

The ECG measures the small 1mV voltage generated at the peak of the depolarization. This small voltage is easily distorted from interfering electric fields from surrounding electronics or muscles. The signal fed into the circuit is filtered through a three-stage cascade: a buffer (Figure 4), a differential amplifier (Figure 5), and a bandpass filter (Figure 6). Resistors are used to ensure that minimal current is drawn from the body. Operational amplifiers are used to combine the signals from two leads and amplify the output voltage of this stage. The next stage involves a differential amplifier that uses the Common Mode Rejection Ratio (CMRR), which is a measure of the tendency of a device to reject common signal from multiple inputs. The bandpass stage is used to obtain desired frequencies by eliminating common interfering frequencies. The bandpass frequency is set to filter out frequencies that are out of the 160 mHz to 58.9 Hz range. This helps reduce the distortion of 60Hz noise from other devices and frequencies generated by muscle movement. This bandpass does filter some of the frequencies generated by

the heart; however, the QRS complex, which is of primary interest in measuring pulse transit time, is passed through the circuit at about 5-15Hz. The circuit in its entirety is seen in Figure 7.

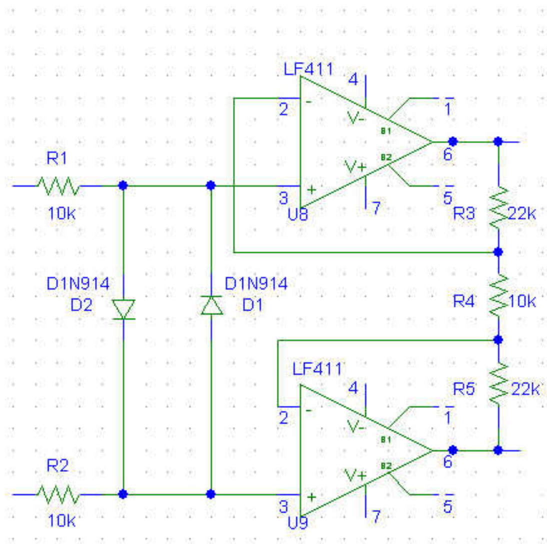



**Figure 4:** The buffer stage of the ECG circuit. The patient would be connected to $v_a$ and $v_b$ inputs. R1 and R2 add extra input impedance. The op amps provide high input impedance as to not load the patient and draw a lot of current that would distort the ECG signal. Diodes, D1 and D2, are used to minimize damage to the circuit from transient voltages. The gain for each buffer in this figure is 22k/10k=2.2 V/V.

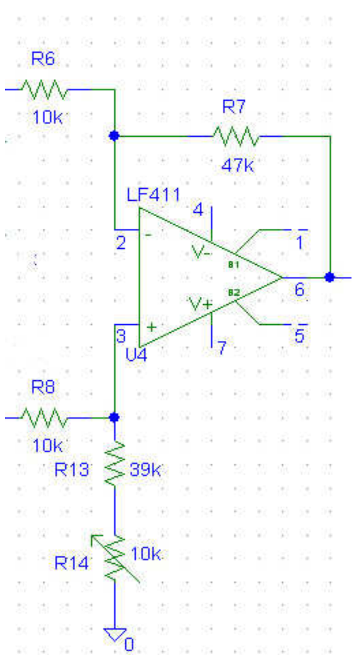



**Figure 5:** The differential amplification stage of the ECG circuit design. The potentiometer located at R14 can be adjusted to match R7 so that the common mode gain is minimal. This will improve the common mode rejection ratio (CMRR). The gain of this stage is 47k/10k = 4.7 V/V. Nodes *c* and *d* are the inputs from the output of the buffer stage. A voltage source of ±15 V/200 mA is used to power the op amp.

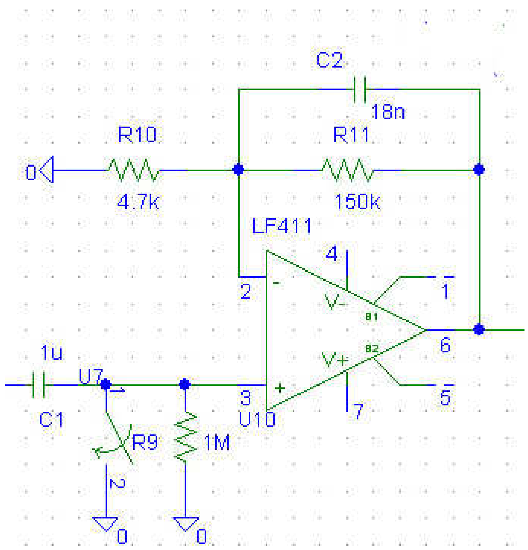
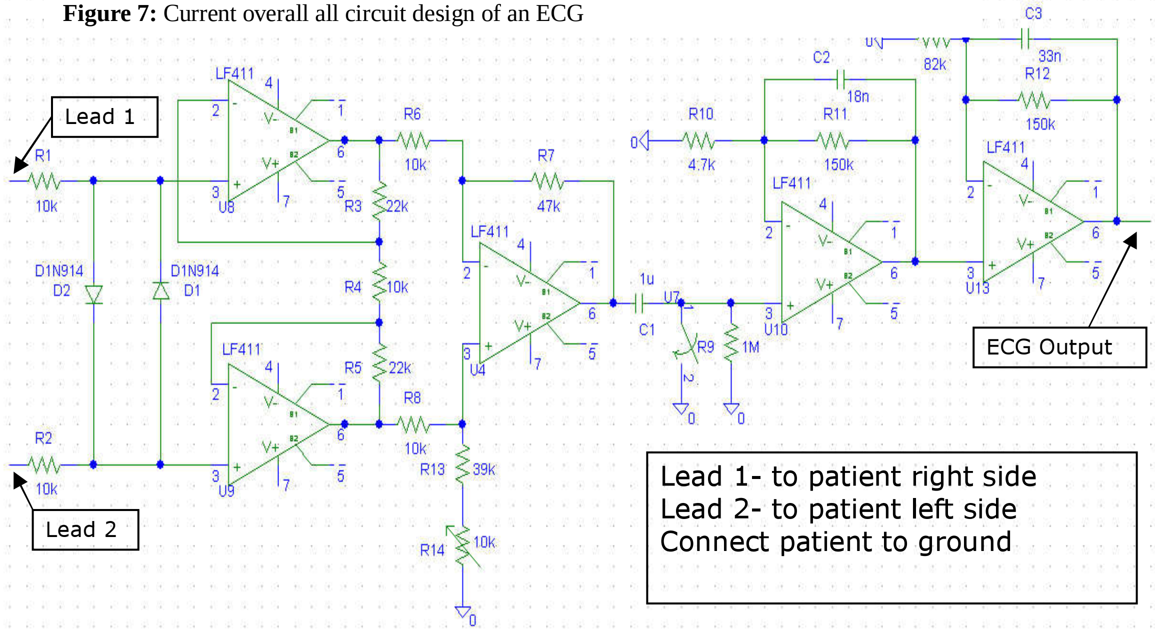



**Figure 6:** The bandpass filter for the ECG circuit. C1 and R9 make a highpass filter with a cutoff frequency of 160 mHz. C2 and R11 make a low pass filter with a cutoff frequency of 58.95 Hz. Node e is connected to the output of the differential amplifier. A voltage source of ±15 V/200 mA is used to power the op amp.

**Figure 7:** Current overall all circuit design of an ECG




## ***Plethysmograph***

A plethysmograph is a device designed to measure the changes of volume in a tissue or organ. For the fingertip, it is used as a non-invasive measure of the amount of blood arriving at the fingertip due to the pressure wave created by the contraction of the heart. The device's LED emits light at two wavelengths: 650nm and 805nm. The light waves pass through the finger and the remaining light is picked up by a phototransistor. A common finger plethysmograph is

manufactured by Nellcor. The configuration of the serial connection to the circuit is shown in Figure 8.

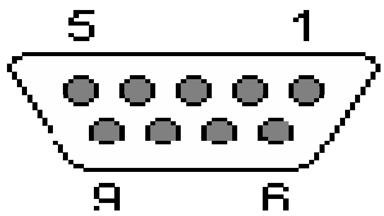



**Figure 8:**
Wire connections:
Pin 2 = Red (ground for LED)
Pin 3 = Black (power supply for LED)
Pin 5 = Orange (ground for phototransistor)
Pin 9 = Brown (circuit input for phototransistor)

A 1-4 mV signal is fed into the circuit, requiring several stages of operational amplifiers to process the signal. All op amps are powered by a +/- 15 V/200mA source, which also power the LED. The signal first passes through a bandpass filter to eliminate DC offset. Another filter is designed to filter common 60 Hz noise, similarly to the ECG circuit. Another bandpass filter is used to further reduce the noise. In the next stage, the signal is amplified to the required voltage of 4-8 V as required for measurement by the software. The final stage involves cutting out low frequencies produced by the motion of the finger. The four stages of the circuit are seen in Figure 9-12.

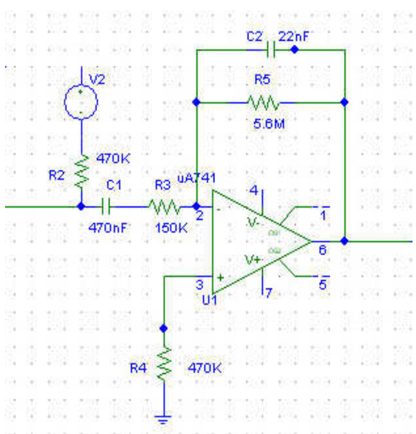



**Figure 9:** The first stage in the fingerplethysmograph circuit. It is an inverting bandpass filter with calculations. R2 and C1 create a HPF. C2 and R5 create a LPF. R5 and R3 provide gain. A voltage source of ±15 V/200 mA powers the op amp. The input of the circuit is connected to the finger probe.

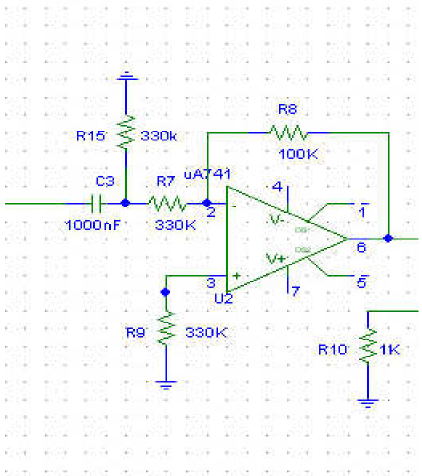



**Figure 10:** <u>The second stage in the finger plethysmograph</u>. It consists of a HPF, with R15 and C3 to setting the cutoff frequency at 0.4288Hz. A voltage source of ±15 V/200 mA powers the op amp. The input of the circuit is connected to the first bandpass filter.

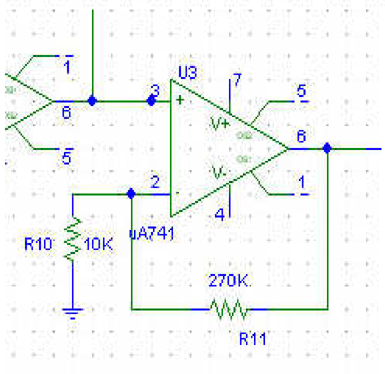



**Figure 11:** <u>The third stage of the finger plethysmograph</u>. R10 and R11 provide a gain of 28 V/V. A voltage source of ±15 V/200 mA powers the op amp. The input of the circuit is connected to the output of the second stage of the circuit.

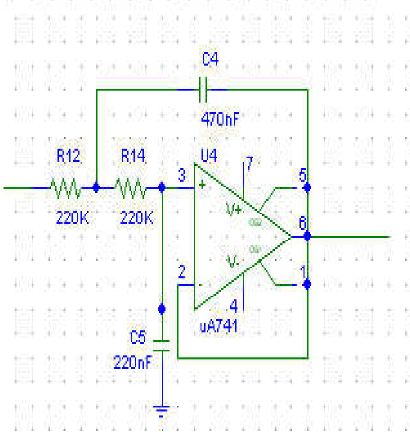



**Figure 12:** <u>The final stage of the finger plethysmogaph circuit</u>. The diagram is a sallen-key low pass filter, with a 2.32 Hz cutoff frequency. A voltage source of ±15 V/200 mA powers the op amp. The input of the circuit is connected to the output of the second stage of the circuit.

***<u>Existing LabVIEW</u>***

LabVIEW is a platform and development environment for a visual programming language from National Instruments. This graphical programming software lessens the complexity of software programming and coding. The language used by LabVIEW is called "G", or the data flow language[vii]. To collect data, the interface is connected with different function-

nodes by wires. The wires gather data, inputs the data into LabVIEW, and the software records the data as waveforms. It is important that the user connects the nodes and any other outlets before opening the software. This will allow the components to be detected before the program starts running. In addition, there are three components, or virtual instruments, in LabVIEW. This is because its appearance and operation imitates a physical instrument, such as an oscilloscope. These instruments include block diagram, front panel and connector panel.

The former BME students have created a LabVIEW setup to calculate PTT and display it to the user (as seen in Figure 13). It also includes sampled ECG and plethysmograph waves. The program was programmed in LabVIEW 7.1. The configuration of the program is indicated as the following:

- BoardNum = the DAQ's board number from InstaCal installation
- LowChan = 0
- HighChan = 1
- Count = 1000 = (number of seconds you would like to calculate PTT for) / [(# of channels) * (rate) ]
- Rate = 100 = (# of channels) * (highest desired sampling rate)
- Range = +/- 10 V
- Cont/Sngl = SINGLE

With the configuration properly entered, the program is initiated by clicking the RUN button. The ECG and plethysmograph waves are input to the program using VIs from Measurement Computing's Universal Library for LabVIEW[viii]. Next, the frequency is set to 0.05 Hz. Meanwhile, the threshold level is used to detect the peaks and frequencies of the QRS complex and plethysmograph waves for each heartbeat. For the interpretation of the data, the difference between the QRS complex peak and plethysmograph wave peak is taken, and it is calculated to be PTT.

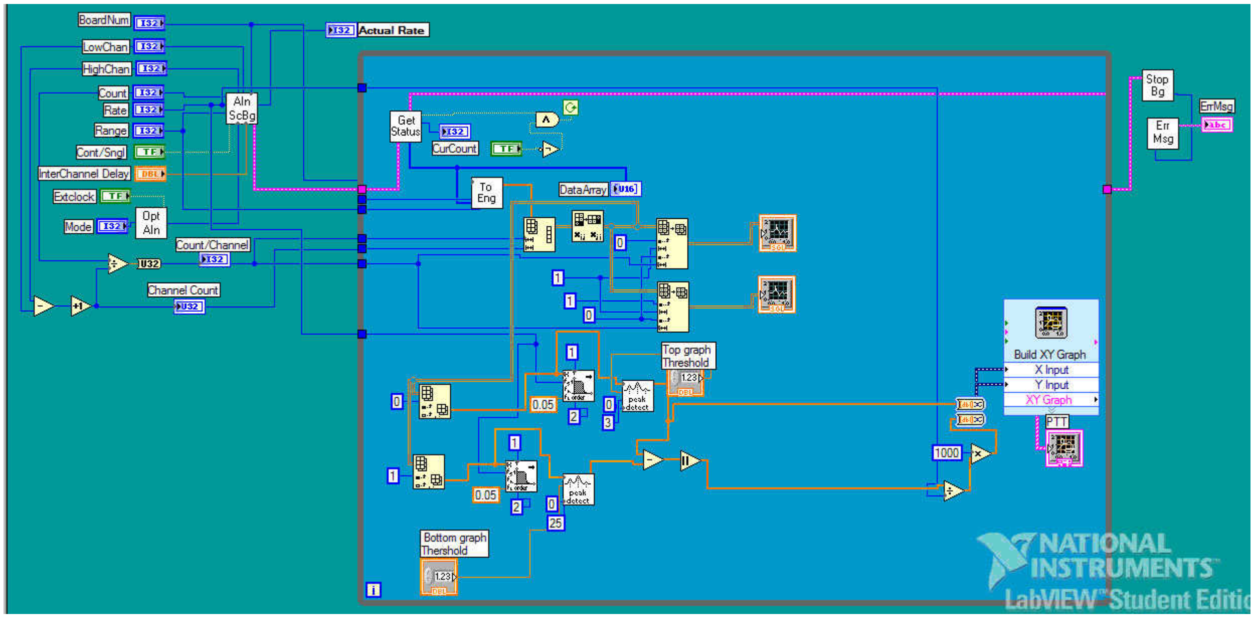



**Figure 13:** Block Diagram – on of the LabVIEW VIs.

## New Setup

## ECG Circuit

In order to provide the best signal output, the circuits must be optimized to provide the best signal possible. The ECG circuit was an area of particular interest as additional types of circuits can be used to provide a better signal to noise ratio. This increase in signal to noise ratio can be performed with the use of an instrumentation amplifier.

Two problems typically arise when dealing with the ECG circuits. One being the electrical potentials on the skin of a subject can be constantly changing, varying as much as 2 volts over time. This two volt signal is very small in comparison with the 1 mV signal which constitutes the electrical activity of the heart. Also, 60 Hz noise will be present anywhere around a subject. This signal is emitted from most electrical devices.

Since both of these signals are found in the environment, this signal will be common to both of the inputs. Therefore, in order to rid the output of both of these common noise signals, a high CMRR is needed. By rejecting this common signal to both, the true underlying signal can be found. In order to ensure that no more than 1% of the signal consists of this common noise, a CMRR of 100 decibels is desirable. Since this type of precision is hard to find in most standard operational amplifiers, an instrumentation amplifier can be used. The instrumentation amplifier, which is used in the new ECG circuit, is the Analog Devices AD624AD. The AD624AD was used because it could be made to have a gain of 1000, which is needed for the very weak ECG signals. It could also produce a CMRR of over 100 decibels. The instrumentation amplifier thus has the ability to be a very precise differential amplifier, or an amplifier which finds the difference of the two inputs and amplifies the signal. The use of the instrumentation amplifier will also provide the added benefit of taking up a smaller physical space compared with the previous five operation amplifier setup.

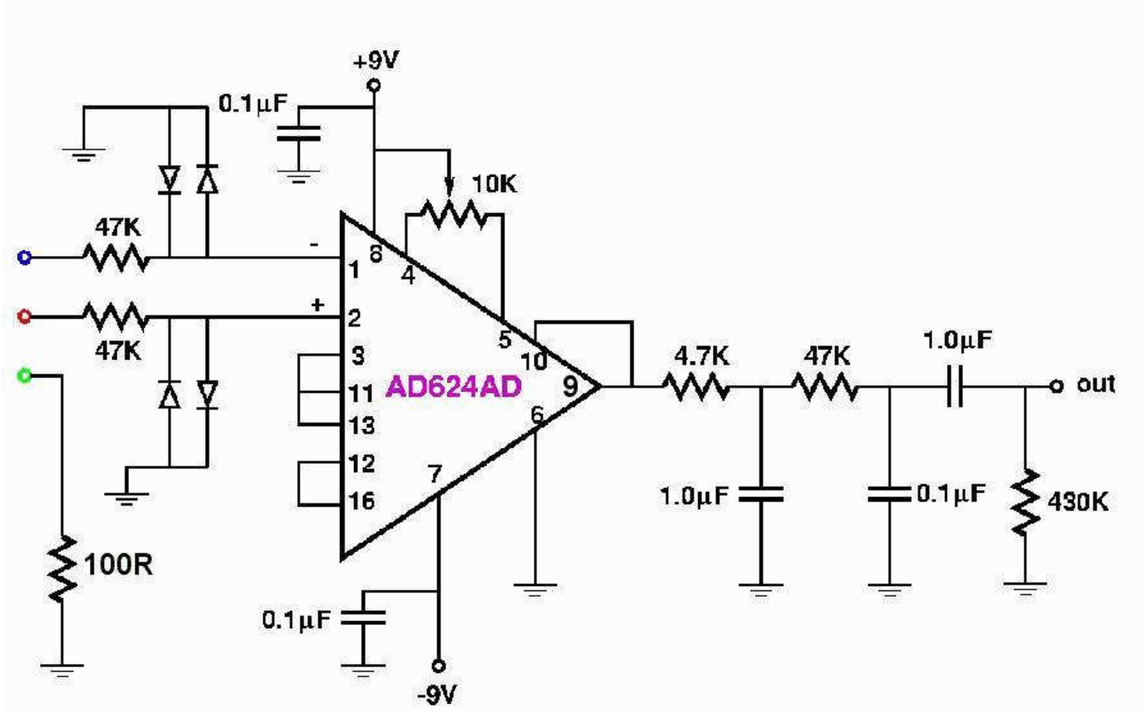



**Figure 14: The circuit diagram of the new circuit breadboard[ix].**

As can be seen from Figure 14, the Analog Devices AD624AD is at the heart of our new ECG. The two 47K resistors and the diodes are used for the safety of the subject. The 47K resistors increase the impedance in the case the signal was sent out of instrumentation amplifier. To further ensure the safety of the patients, the diodes were used to ground the output of current. To ensure the instrumentation amplifier has an output gain of 1000, pins 3,11,12,13, and 16 must be shorted as shown.

Once the output from the instrumentation is amplified, certain frequencies must be attenuated to decrease noise. Thus, a bandpass filter was created. Since the majority of the signal needed for the purposes of this setup resides within 30 Hz, a low pass filter is designed to attenuate signals greater than 33.9 Hz. The first low pass filter was designed using a 4.7K resister and a 1.0µF capacitor and the second was designed using a 47K

resistor and a 0.1µF capacitor. This can be calculated through the use of the following equation:

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

The addition of one low pass filter would mean the attenuation of 70.9% of the signal above 33.9 Hz. However, with the addition of a second low pass filter with a frequency response of 33.9 Hz, 50% of the signal greater than 33.9 Hz will be attenuated from the signal. Since the frequencies higher than 33.9 Hz are attenuated, the remaining 60 Hz noise will be removed from the sample. Since the DC offset is present in the signal, a high bandpass filter is added to the circuit. The DC offset accounts for variation in the baseline of the ECG signal. So the high bandpass filter will be used to attenuate signal lower than 0.37 Hz. The high bandpass filter was designed with a 1.0µF capacitor and a 430K resistor.

**New Design Setup**

The old setup had to be setup with a laptop at all times. This was a problem for the client as the device was not as portable, and problems may have arose because of a laptops sliding off a table while recording a patient during sleep. It would be beneficial to rid the setup of a laptop, which would make the setup much more portable compared with the previous design.

The addition of flash memory will allow the removal of the laptop computer from the setup. In order to implement a flash memory collector into the setup, three specific devices must be added to the circuit. First, an analog to digital converter must be added to allow the output voltages from the ECG and the finger plethysmogram circuit to be converted into binary, which can be read by the computer. The second element that needs

to be added is a microcontroller. The microcontroller is the "brains" of the operation. The microcontroller must be programmed to write the voltage outputs from the circuits to the memory storage device. Finally, a flash memory interface will be added to the design. This is how the flash memory will attach to the circuit and is the element that will transfer the data to the memory card.

In order to get the actual signal that is being output from the circuit, certain specifications of the system need to be met. The analog to digital converter is an important piece of the setup and has some requirements that need to be met in order to work with the circuits. The resolution of the ADC determines the accuracy that can be achieved with the converter. This is important because it will allow our client to best determine PTT. The use of a 12-bit ACD will allow for resolution of 1.2mV to be detected. Since a 12-bit converter will have $2^{12} = 4096$ "states", then the output ranges of the circuits are 0-5 volts and the resolution will be equal to (5-0)/4096=1.22. This will be more than sufficient for the client to use in his data collection. The system must also have at least 3 IO slots. The inputs needs are: input from ECG circuit, input from finger plethysmogram circuit, and output to the flash memory. The sample rate needed to obtain an ECG signal is around 25 Hz and rate for the finger plethysmogram is around 7 Hz. To avoid anti-aliasing, the Nyquist theorem states that the sampling frequency must be at least twice the maximum value. Therefore, the ECG is the limiting factor for the sampling rate and an absolute minimum of 50 Hz is required to get a signal. To error on the side of caution, a final sampling rate of 100 Hz was chosen. In addition, the amount of memory needed for storage needed to be addressed. Since two channels running with a sample rate of 100 Hz for 8 hours (one nights sleep) uses approximately 10 MB of data, a

memory card of at least 32 MB will be needed to ensure adequate memory storage is available. In addition, since the ECG signal output can fall within +/- 1 V, a resolution of approximately 10 mV is needed. Moreover, the system will need to run on a battery source, so portability issues are maintained. Finally, the system must be small and compact, the final goal of the project is to have a compact design, and so this must be taken into consideration.

Since all of the specifications were determined, the best way to go about adding the memory card to the circuit needed to be determined. After doing preliminary research, it was found that creating this new device would require much more knowledge than originally thought. The decision to look elsewhere was made. After more research, a data logger was decided on. A data logger is an instrument that allows voltages to be inputted and stored onto a memory card.

After matching all of the specifications with the product, the Dataq 710-ULS was chosen. This product meets all of the specifications for our system. The datasheet for the Dataq 710-ULS is attached in the appendix**.**

## New software

### *Pulse Transit Time Calculating Algorithm*

The data acquired by the logger can be imported to the computer from the flash memory card and saved into a text file. A separate text file is created for both the ECG and the finger plethysmograph containing the corresponding voltages and times. The program used in calculating the pulse transit time is designed to determine the times corresponding to each R wave peak and the peak of the pressure waves, and subtracting these values to give the pulse transit time. The program is given 5 parameters in the command prompt: -e for the ECG file path, -f for the finger plethysmograph file path, -o

for the file to which the data is to be output, -te for the threshold value of the ECG, -tf for the threshold value for the finger plethysmograph data. An example parameter path could be: –e C:/ecg/ecg.txt –f C:/ecg/finger.txt –o C:/ecg/output.txt –te 4 –tf 0.04. The program stores these values in an array for each of the two data sets. The program proceeds to search through the points to determine the maximum peak values. A “while” loop is used to read the data until the end of the text file is reached. Running through the data, the program determines the points within the range of the threshold value as entered by the user. “For” loops are used to cycle through all of the points and compare them to the other data to determine if they qualify as a peak. This is done for both of the data sets and stored in separate arrays. The times from each array are then subtracted from one another and the corresponding pulse transit times are output into the designated file using a “for” loop to cycle through the data for the number of peaks entered.

```
#include <string.h>
#include <iostream>
#include <fstream>
#include <sstream>
#include <vector>


int main(int argc, char* argv[])
{

  const char* inputEKGFileName = NULL;
  const char* inputFingerFileName = NULL;
  const char* outputFileName = NULL;
  double thresoldEKG = 0;
  double thresoldFinger = 0;

  int i;
  for (i = 1; i < argc; i++) {
    if (strcmp(argv[i], "-e") == 0) {
      inputEKGFileName = argv[++i];
    }
```

```
    else if (strcmp(argv[i], "-f") == 0) {
      inputFingerFileName = argv[++i];
    }
    else if (strcmp(argv[i], "-o") == 0) {
      outputFileName = argv[++i];
    }
    else if (strcmp(argv[i], "-te") == 0) {
      std::stringstream ss;
      ss << argv[++i];
      ss >> thresoldEKG;
    }
    else if (strcmp(argv[i], "-tf") == 0) {
      std::stringstream ss;
      ss << argv[++i];
      ss >> thresoldFinger;
    }
    else {
      std::cerr << "usage: " << argv[0] << "\n"
                << " [-h (for help)]\n"
                << " -e (EKG input file name, no
spaces!)\n"
                << " -f (Finger input file name, no
spaces!)\n"
                << " -o (input file name, no spaces!)\n"
                << " -te (thresold value fo EKG)\n"
                << " -tf (thresold value finger)\n"
                << std::endl;
      return 0;
    }
  }
  std::ifstream infileEKG(inputEKGFileName);
  std::ifstream infileFinger(inputFingerFileName);
  std::ofstream outfile(outputFileName);

  double threshValue = -1000;
  double minTimeInterval = 0.25;

  // PROCESS EKG DATA

  // read two columns of data from input EKG file, should
be space separated
  double t,v;
  std::vector<double> timeEKG;
  std::vector<double> voltsEKG;
  while(infileEKG >> t >> v) {
    timeEKG.push_back(t);
    voltsEKG.push_back(v);
```

```
  }


  // threshold volts array
  std::vector<double> voltsEKGThresh;
  for( i=0; i<voltsEKG.size(); i++) {
    if (voltsEKG[i] < thresoldEKG) {
      voltsEKGThresh.push_back(threshValue);
    }
    else {
      voltsEKGThresh.push_back(voltsEKG[i]);
    }
  }

  // find local max in threshold array
  std::vector<double> peakTimesEKG;
  for( i=1; i<voltsEKGThresh.size()-1; i++) {
    if (voltsEKGThresh[i] != threshValue &&
      voltsEKGThresh[i] >= voltsEKGThresh[i-1] &&
      voltsEKGThresh[i] >= voltsEKGThresh[i+1]) {
      if (peakTimesEKG.size() == 0) {
        peakTimesEKG.push_back(timeEKG[i]);
      }
      else if (timeEKG[i] -
peakTimesEKG[peakTimesEKG.size()-1] > minTimeInterval ) {
        peakTimesEKG.push_back(timeEKG[i]);
      }
    }
  }


  // PROCESS FINGER DATA

  // read two columns of data from input finger file,
should be space separated
  std::vector<double> timeFinger;
  std::vector<double> voltsFinger;
  while(infileFinger >> t >> v) {
    timeFinger.push_back(t);
    voltsFinger.push_back(v);
  }


  // threshold volts array
  std::vector<double> voltsFingerThresh;
  for( i=0; i<voltsFinger.size(); i++) {
    if (voltsFinger[i] < thresoldFinger) {
```

```
      voltsFingerThresh.push_back(threshValue);
    }
    else {
      voltsFingerThresh.push_back(voltsFinger[i]);
    }
  }

  // find local max in thresholded array
  std::vector<double> peakTimesFinger;
  for( i=1; i<voltsFingerThresh.size()-1; i++) {
    if (voltsFingerThresh[i] != threshValue &&
      voltsFingerThresh[i] >= voltsFingerThresh[i-1] &&
      voltsFingerThresh[i] >= voltsFingerThresh[i+1]) {
      if (peakTimesFinger.size() == 0) {
        peakTimesFinger.push_back(timeFinger[i]);
      }
      else if (timeFinger[i] -
peakTimesFinger[peakTimesFinger.size()-1] > minTimeInterval
) {
        peakTimesFinger.push_back(timeFinger[i]);
      }
    }
  }

  // output diff between peak times
  int n = peakTimesEKG.size();
  if (peakTimesFinger.size() < n) {
    n = peakTimesFinger.size();
  }

  for( i=0; i<n; i++) {
    outfile << peakTimesEKG[i] << " , "<<
peakTimesFinger[i]<< " , " << peakTimesEKG[i]-
peakTimesFinger[i] << std::endl;
  }

  infileEKG.close();
  infileFinger.close();
  outfile.close();
  return 0;
}
```

**Testing**

The previous and new ECG circuits are compared against each other in the following diagrams (Figure 16 and 17 respectively).

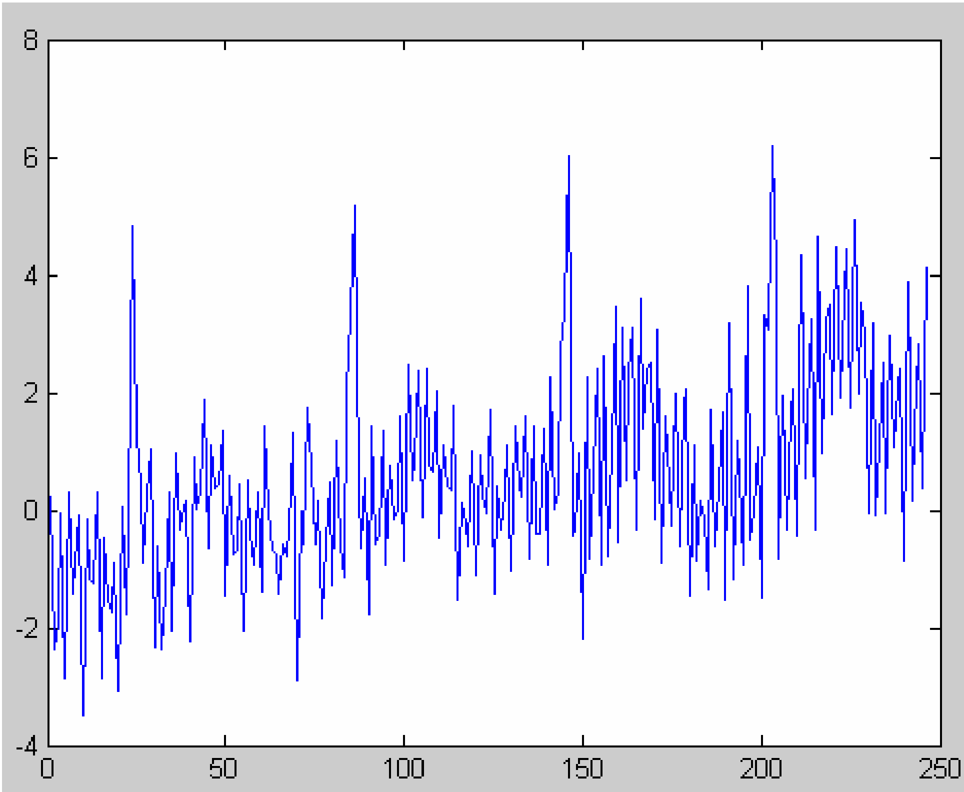



**Figure 16: Electrocardiogram signal from old circuit.**

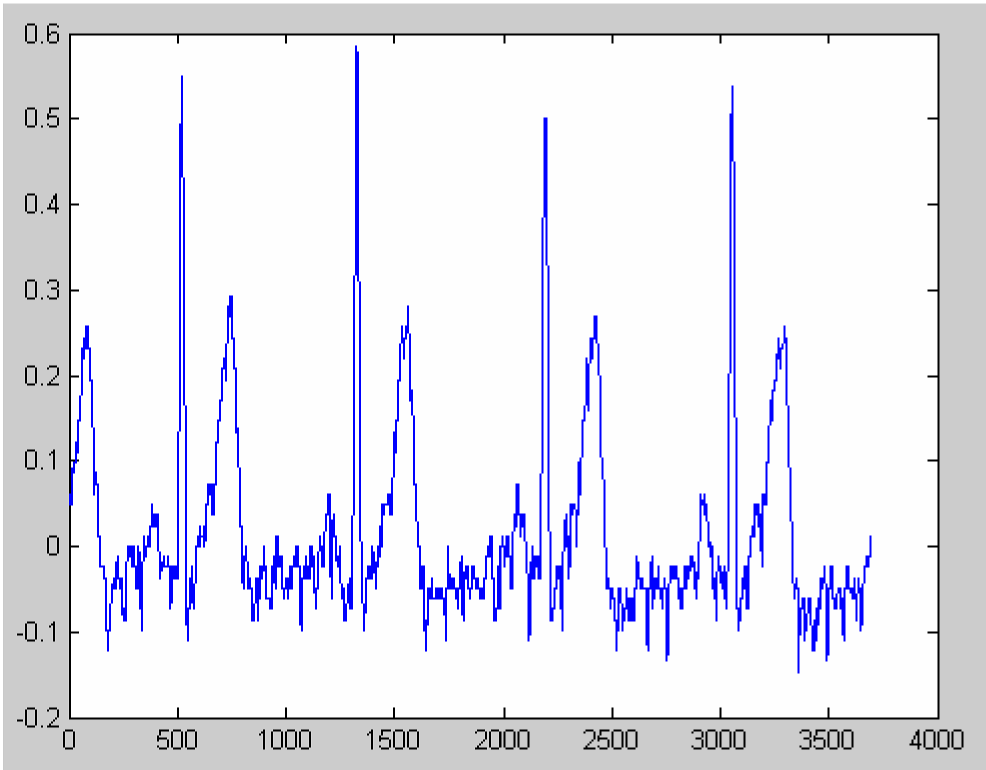



**Figure 17: Electrocardiogram from the new ECG circuit recorded by the data logger**

As can be seen from the two graphs, the previous ECG signal has a large amount of noise present around the 60 Hz frequency. This is due to the use of the low pass filter that has a frequency response of only 58.9 Hz and thus not attenuated the 60 Hz frequencies.. The new ECG signal, however, does show a much cleaner signal compared with the previous signal due to the use of the bandpass filter that reduces the noise and stabilizes the signal.

## Future Work

The future work of this project is split into three categories: the circuit, the overall setup and the software. To further reduce the noise of the circuit and its overall size, it would be beneficial to obtain a printed circuit. A printed circuit was not made in this phase of the project so that further research could be made into the circuit to make sure it truly is the best design. If

the new design is the best, then a printed circuit can be made. A printed circuit would reduce the interference from wire crossings that exist on a breadboard. Printing the circuit would also compact the circuit so that it could better fit into a small device.

Once the circuit has been printed, the next step will to be to combine the ECG and plethysmograph circuits, the data logger, and the power supply into one unit. This unit should be as small as possible so that it causes the least discomfort to the patient. The design must allow easy access to the batteries for when they need changing and to the memory card slot so there is no hassle in removing the card. The power supply is another area that could be looked to increase efficiency. The current device runs off of several 9V batteries that take up a lot of space in the device and are quite costly. Future work should be focused on determining if another type of battery can be used or if there is another configuration design for the batteries. Rechargeable batteries could also be looked into and designed to be similar to the ones used in cell phones.

Moreover, the software needs to continue to be improved. The current software only records in a previously defined period. If the heart rate increases or decreases during sleep, then the period between R waves can change; the software does not know how to recognize this. The client specified this would be an important feature to look at. An algorithm also needs to be developed to further reduce the noise of the ECG signal. Modifications to the circuit can only reduce the noise to a certain level and cannot account for the noise generated from movement of the patient. If the signals experience a few disturbances during the night, it may be possible for the software to recognize the long pause between recognizable R waves and be able to reset itself once a true R wave is found. The client is willing to sacrifice a few data points for a cleaner data set.

The improvements made to this design over the semester have made the miniaturization of the device possible. The incorporation of a data logger enhances its portability and its safety. The new ECG circuit decreases the noise in the signal and combined with the new software, allows for a more accurate measure of PTT. Future work on this project will bring it closer to

being ready for clinical trial and to the ultimate goal of the project: incorporation into sleep studies.

October 25, 2006
Updated: December 11, 2006

Jonathan Baran
Mark Yarmarkovich
William Stanford
Karen Chen

# Pulse Transit Time Device PDS

## Problem Statement

The current instruments used in the measurement of pulse transit time are inefficient for home use. An existing product with working ECG and pulse wave circuits along with software to analyze the data has been provided. The primary goal will be to optimize the existing setup for use at home. This will be performed by miniaturizing the circuit, increasing the signal to noise ratio, and improving the already existing software. These tasks will attempt to be rectified by numerous design additions.

## Client Requirements

- Optimized for use with children
- Integrate with previously designed system
- Write program to deal with skipped heart beat
- Integrate flash memory to make device more compact
- Improve signal to noise ratio

## Design Requirements

### 1. Physical and Operational Characteristics

a. *Performance requirements*: The probe needs to be small enough to stay fixed to children's fingers. More over, the probe needs to stay attached to the finger during sleep. The monitoring system needs to be non-evasive and not interfere with sleep patterns. The monitoring

software needs to record the signals from the finger probe and ECG leads while filtering out skipped beats.

b. *Safety*: Because the device will be used while children are sleeping, the chords must not present a strangling hazard. Also circuits must be made to minimize electrical hazards.

c. *Shelf life*: The device must be operated by batteries to allow it to be compact. The adhesive needs to be reusable.

d. *Accuracy and Reliability*: The device must employ a circuit design the takes into account both gain and CMRR. A gain of 1000 is need to amplify the ECG signal and a CMRR of 100 decibels is also needed

e. *Life in Service*: The device should be able to be used for multiple uses by a single patient and then repeated for many more patients.

f. *Operating Environment*: Ideally, the device will be able to be used in the patient's home while they are sleeping. If optimal minimization is achieved, it could be worn on the waste.

g. *Size*: The device needs to be small enough to fit on a bed size table, or optimally, attached to the waste of the patient to allow the chords to be non-evasive.

h. *Weight*: The device needs to be as light as possible so it will not be felt while sleeping.

i. *Materials*: Plastic would be used for the casing of the device.

**2. Product Characteristics**

a. *Quantity*: For the time being, only one device is necessary. If the device is successful, multiple units may be made in the future.

b. *Target Product Cost*: The device should be as inexpensive as possible so not to burden families when the device is taken home.

**3. Miscellaneous**

a. *Standards and Specifications*: The device must be meet FDA medical device specifications.

b. *Customer*: Christopher Green, M.D. Dept. of Pediatrics UW Medical School

c. *Patient-related concerns*: The patient's personal information will not be stored in the device. No sterilization of the device will be needed in-between uses aside from the application of new EKG patches at every use.

d. *Competition*: Probe devices with LED are on the market but there are no devices that measure blood flow and pulse transit time.

# Appendix

## Appendix Figure 1: Datasheet for the Dataq 710-ULS[x]

# DI-710 Specifications

### Analog Inputs

**Number of Channels:** 16

**Channel Configuration:** 16 single-ended; 8 differential; programmable per channel

**Measurement range, Accuracy, and Resolution**

| | Gain | Range | Accuracy | Resolution |
|---|---|---|---|---|
| PGL models: | 1 | ±10V | ±.25%FSR | ±1.22mV |
| | 10 | ±1V | ±.25%FSR | ±122µV |
| | 100 | ±100mV | ±.25%FSR | ±12.2µV |
| | 1000 | ±10mV | ±.25%FSR | ±1.22µV |
| PGH models: | 1 | ±10V | ±.25%FSR | ±1.22mV |
| | 2 | ±5V | ±.25%FSR | ±610µV |
| | 4 | ±2.5V | ±.25%FSR | ±305µV |
| | 8 | ±1.25V | ±.25%FSR | ±153µV |

**Input Impedance, single-ended:** 1MΩ

**Input impedance, differential:** 1MΩ each input to common

**Input bias current:** 10µA for a 10V input, single channel

**Input offset voltage:** Auto-zero

**Input offset current:** 2nA (single channel)

**Max. normal mode voltage:** 30V DC or peak AC

**Max. common mode voltage:** 30V DC or peak AC

**Common mode rejection:** 80db, gain=1, 1KΩ unbalance

**Channel-to-channel crosstalk rejection:** -75db @ 100Ω unbalance

**Gain temperature coefficient:** 50 ppm/°C

**Offset temperature coefficient:** 0.25µV/°C

**Digital filtering:** Standard: Conditional over-sampling
Stand-alone: None

### A/D Characteristics

**Type:** Successive approximation

**Resolution:** 14-bit

**Monotonicity:** ±2 LSB

**Conversion Time:** 69µs

### Scanning Characteristics

**Max. throughput sample rate:*** Standard: 4,800 Hz
Stand-alone: 14,400 Hz**

*When acquiring more than one channel at a gain of 100 max throughput is 7200 Hz; When acuiring more than one channel at a gain of 1000 max throughput is 900 Hz.
**Assumes SD memory latencies of 80 milliseconds or less.

**Min. throughput sample rate:** Standard: 0.0034 Hz
Stand-alone: 0.0017 Hz

**Max. scan list size:** 17 entries

**Sample buffer size:** 2kb

### Controls (Stand-alone models)

**Single push-button:** Manual control Record and Standby

### Calibration

**Calibration cycle:** One year

**Calibration method:** Automated Calibration Software, provided.

### Digital I/O

**Bits:** 8 bidirectional bits

**Configuration:** Each bit is programmable as Input or Output

**Output voltage levels:** Min. "1" 3V @ 2.5mA sourcing
Max. "0" 0.4V @ 2.5mA sinking

**Output current:** Max. source, -2.5 mA; Max. sink, 2.5mA

**Input voltage levels:** Min. required "1" 2V; Max allowed "0" 0.8V

### Ethernet Interface

**Type:** 10/100Base-T

**Connector:** RJ-45

**Protocol:** TCP/IP

**Server Type:** DHCP

### Removable Memory (Stand-alone models)

**Type:** SD (Recommended: SanDisk SD Card)

**Capacity:** 16 Mb to 1 Gb

### Real Time Clock (Stand-alone models)

**Type:** Date, hour, minute, second

**Resolution:** 1 second

**Accuracy:** 20 ppm

### Indicators

**Stand-alone models:** Three-color LED indicating Record, Standby, and Error conditions

**Standard models:** Power LED

### Transfer Rate to PC

**Real Time:** up to 4,800 samples per second

**From Memory Card:** up to 2,400 samples per second (Ethernet only)

### General

**Panel indicators:** Mode LED

**Panel Controls:** Control push button (Stand-alone models)

**Panel Slots:** Accepts SD-type flash memory

**Input connectors:** Two, removable sixteen position terminal blocks

**Operating Environment:** 0°C to 70°C

**Enclosure:** Aluminum base with steel wrap-around. Aluminum end-panels with plastic bezels.

**Dimensions:** $5\frac{7}{16}$"D × $4\frac{1}{8}$"W × $1\frac{1}{2}$"H
13.81D × 10.48W × 3.81H cm.

**Weight:** 14 oz.

**Power Requirements:** USB: 9 to 36 VDC, 2 watts max
Ethernet: 9 to 36 VDC, 2.5 watts max

**References**


[i] *Sleep apnea.* (2006). Retrieved October 9, 2006 from http://en.wikipedia.org/wiki/Sleep_apnea

[ii] Smith R, Argod J, Pépin JL, Lévy P. 1999. Pulse Transit Time: an appraisal of potential clinical applications. *Thorax*. 54:452-458.

[iii] Pagani J, Pia M, Calcagnini G, Alterio A, Ambrosio R, Censi F, Ronchetti R. 2003. Pulse Transit Time as Measure of Inspiratory Effort in Children. *Chest*. 124:1487-1493.

[iv] Karas, A., Hondl, B., Olson, M., & Cohen, Z. (2005). *Measurement of pulse transit time.* Unpublished manuscript. Retrieved September 25, 2006,

[v] Katz E, Lutz J, Black C, Marcus C. 2003. Pulse Transit Time as a Measure of Arousal and Respiratory Effort in Children with Sleep-Disordered Breathing. *Pediatr Res*. 53:580-588.

[vi] Carlson, S. (2000, the amateur scientist: Home is where the ECG is. *Scientific American Magazine,* (June) 3. Retrieved October 17, 2006,

[vii] *Labview.* (2006). Retrieved October 1, 2006 from http://en.wikipedia.org/wiki/LabVIEW

[viii] *LabVIEW tutorials.* (2005). Retrieved October 5, 2006 from http://www.upscale.utoronto.ca/GeneralInterest/LabView.html

[ix] *Electrocardiogram (ECG) project for DrDaq.* (2006). Retrieved October 12, 2006 from http://www.picotech.com/applications/ecg.html#cct

[x] Dataq Instruments. (2006). Retrieved on December 11, 2006 from http://www.dataq.com/support/documentation/pdf/datasheets/710ds.pdf

# *Pulse Transit Time Measuring Device*

BME 200/300
University of Wisconsin – Madison
October 25, 2006

*Team*:
Jonathan Baran – Team Leader
Karen Chen –BSAC
William Stanford – BWIG
Mark Yarmarkovich – Communicator

*Client*:
Christopher G. Green, M.D.
Pediatric Pulmonology
University of Wisconsin Hospital

*Advisor*:
Wally Block
Department of Biomedical Engineering
University of Wisconsin – Madison

## Abstract

*Sleep apnea is a disorder affecting many children, causing them to awake from sleep to unblock their airways. Pulse transit time, the amount of time it takes for the pressure wave from the heart to travel to the finger, indirectly measures the presence of the disorder and its severity. Currently, only the tests conducted in sleep labs can test for sleep apnea and record its affects on the body. The work on this project will be in improving a device created by a previous engineering group that measures pulse transit time. The ultimate goal of the project is to create a device that can be used at home by the incorporation of a data logger that can record electrical signals from ECG and plethysmograph leads, improving the signal to noise ration by including an instrumentation amplifier and by modifying existing software to better detected the peaks of the stored voltage data.*

# Table of Contents

## Background Information

***Sleep Apnea***

Sleep apnea is a sleep disorder where the patient has a pauses breathing during sleep. Typical breathing rates occur anywhere from 10-20 breathes per minute. During sleep apnea, the tongue blocks the airway and a 10-30 second pause in breathing occurs, causing the sufferer to miss one to two breathes[i]. This problem could occur repeatedly during a night of sleep, which lowers the sleeping quality. Patients would experience symptoms such as snoring and restless sleeps. Moreover, they would have poor day performance, loss of concentration, anxiety and other negative health effects. Researchers have shown that people who are over the age of 40 are at higher risks of sleep apnea. However, it can affect people at any ages.

There are three different forms of sleep apnea – central sleep apnea, obstructive sleep apnea and mixed sleep apnea. Central sleep apnea causes pauses in breathing by the lack of effort in breathing. This is due to the failure of neurons in sending signals to indicate inhalation. In effect, the level of oxygen delivered to tissues decreases and oxygen is not available for cellular respiration. On the other hand, obstructive sleep apnea is where the air path inside the throat is blocked by an object, such as the tongue. As the muscles relax during sleep, the tongue can block the airway (as seen in Figure 1), which causes the patient to enter a lighter sleep stage or possibly cause the patient to awaken. Most patients suffering from obstructive apnea have trouble getting into a deep sleep state. Even though the light sleep time may be numerous, it is still not as effective as deep sleep. Finally, mixed apnea is the combination of central and obstructive sleep apnea. While obstructive sleep apnea takes place during sleep, central sleep apnea is often developed. Patients experience problems breathing and constantly wake up from sleep because of long-term obstructive apnea[i].

The pauses in breathing during sleep result from a decrease in the oxygen concentration of the blood. Furthermore, the level of carbon dioxide increases. The worst-case scenario is when the oxygen level becomes so low that it causes brain damage, and eventually leads to death.

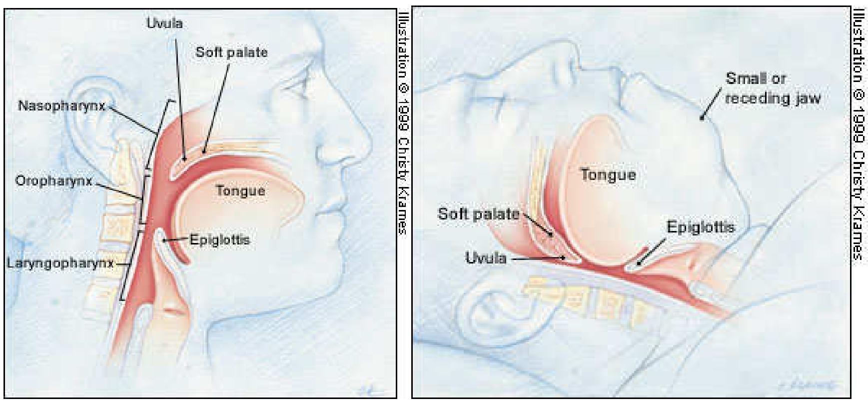



**Figure 1:** The left figure shows a normal trachea opening. The right figure shows that of a person who has multiple obstructions.

## ***Pulse Transit Time***

Pulse transit time (PTT) is a noninvasive method of measuring respiratory changes in children with breathing sleep disorders. PTT is the measure of the time it takes for the pulse pressure wave to go from the heart to the periphery[ii]. It is measured by use of both an ECG machine and a pulse oximeter. An ECG machine generates a curve based on the depolarization of the heart while the oximeter measures the pressure wave, or pulse, at the tip of the finger[iii]. A value for pulse transit time is given by calculating the difference in time between the peak of the R wave from the ECG and the peak of the pressure wave from the oximeter (Figure 2).

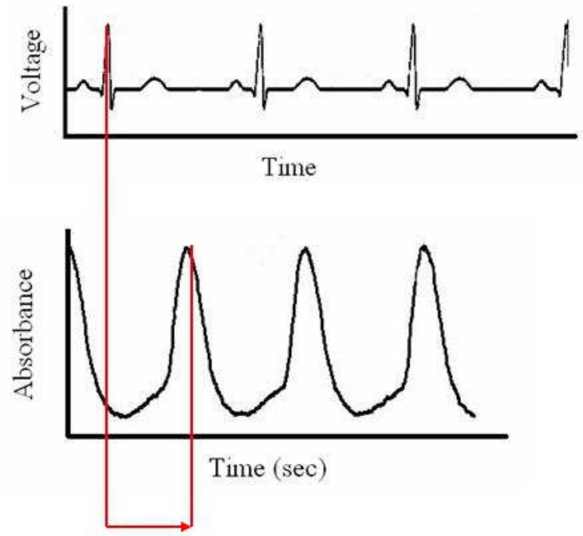



**Figure 2:** Calculation of pulse transit time from ECG and oximeter graphs.

Normal PTT range from 250-350 milliseconds[iv]; a significant variation in this time can help identify sleep apnea in two ways. First, as blood pressure decreases the arterial wall stiffness decreases. As stiffness decreases, it causes the pulse to take a longer time to reach the finger, causing on increase in PTT. This increases helps to diagnose sleep apnea. Second, the increase of blood pressure as the obstruction clears increases arterial wall. The increases in stiffness increases blood pressure, causing pulse transit time to decrease. Decreases in PTT in patients with sleep apnea can range from 15 to 50 milliseconds. Any decrease over 50 milliseconds is anatomically impossible. This decrease in PTT can help diagnose the severity of the apnea[v].

## Motivation

Some consequences of prolonged sleep apnea are hyperactivity, poor daytime performance, loss of concentration and other negative health effects. This sleep disorder can occur in both adults and children. To detect this disorder, patients need to participate in sleep studies at sleep centers. This is a rather expensive study to operate on a day-to-day basis, and thus its utilization is limited. Furthermore, sleep centers are not specifically for sleep apnea studies. Thus, a small, portable instrument that is able to conduct sleep studies at home would largely benefit the patients. This medical device must be able to detect PTT. Ideally, this device should

be as small as a flash drive that needs a low power supply. In addition, the number of wires should be reduced to allow more room for the patient to move. This allows easy access and operation, and the patient can self-detect sleep apnea in earlier stages.

## Client Requirements

The design must be able to be used with children. Our client works in pediatrics and wants to use the device to assess sleep apnea in children. This requires the device to be small, as to not cause discomfort for the child while sleeping. A miniaturizing of the device will allow families to conduct the tests at home. The device will be able to store many nights worth of data, which means the affects of sleep apnea can be analyzed on a day-to-day basis to determine if the condition varies nightly. Having the test conducted at home will also make the patient feel more comfortable.

The client has also specified three things he would like improved. The most important improvement is the incorporation of flash memory into the device. This memory will allow the collected data to be stored onto a memory stick that could be used to upload the data at a different time, eliminating the need for a bedside laptop computer. Next, he wants an improved LabView software program. The current software cannot account for when the recording device misses a beat due to patient movement. This inability causes the cascading of all data after the skipped beat. The new program should be able to edit out the skip beat and continue to graph regular beats as if no skipped beat occurred. Last is the improving and printing of the current circuit. Research shows that the existing circuit in the previous device needs several improvements in order to reduce the noise from surrounding frequencies[vi]. Once the circuit is improved, the printing will be an easy step and lead to a miniaturizing of the device. Since this step is relatively simple, compared with the others, the client would like more thought and effort to be given to the flash memory and the LabView program.

## Problem Statement

The current instruments used in the measurement of pulse transit time are inefficient for home use. An existing product with working ECG and pulse wave circuits along with software to analyze the data has been provided. The primary goal will be to optimize the existing setup for use at home. This will be performed by miniaturizing the circuit, increasing the signal to noise ratio, and improving the already existing software. These tasks will attempt to be rectified by numerous design additions.

## Previous Setup

### ***<u>Electrocardiogram</u>***

Willem Eitnthoven invented the ECG to measure heartbeats in 1901, for which he was awarded the Nobel Prize in Medicine in 1924. With each heartbeat, the cardiac tissue releases ions that depolarize the tissue. This creates a voltage of about 1 mV that can be measured with various leads attached at the body. Six ECG leads may be placed

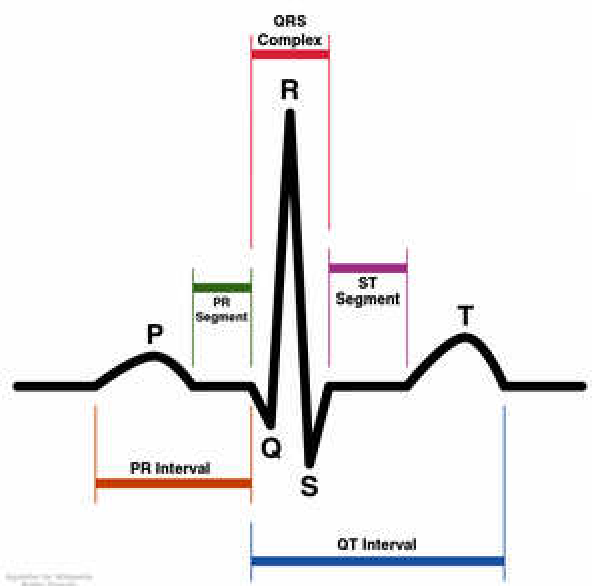



**Figure 3**: The Electrocardiogram

around the body in 30-degree intervals to measure the activity of the various sections of the heart based on the spatial readings. These readings form discrete waves represent the various stages of a heartbeat as seen in Figure 3. The P wave represents the current that causes the atrial contraction in which both the left and right atria contract simultaneously. The QRS complex represents the contraction of the left and right ventricles, a much stronger contraction of greater muscle mass. This results in a much greater reading on the ECG spanning a very short duration of time. Repolarization can be seen by observing the T wave that dips below the neutral voltage in re-establishing an action potential.

The ECG measures the small 1mV voltage generated at the peak of the depolarization. This small voltage is easily distorted from interfering electric fields from surrounding electronics or muscles. The signal fed into the circuit is filtered through a three-stage cascade: a buffer (Figure 4), a differential amplifier (Figure 5), and a bandpass filter (Figure 6). Resistors are used to ensure that minimal current is drawn from the body. Operational amplifiers are used to combine the signals from two leads and amplify the output voltage of this stage. The next stage involves a differential amplifier that uses the Common Mode Rejection Ratio (CMRR), which is a measure of the tendency of a device to reject common signal from multiple inputs. The bandpass stage is used to obtain desired frequencies by eliminating common interfering frequencies. The bandpass frequency is set to filter out frequencies that are out of the 160 mHz to 58.9 Hz range. This helps reduce the distortion of 60Hz noise from other devices and frequencies generated by muscle movement. This bandpass does filter some of the frequencies generated by

the heart; however, the QRS complex, which is of primary interest in measuring pulse transit time, is passed through the circuit at about 5-15Hz. The circuit in its entirety is seen in Figure 7.

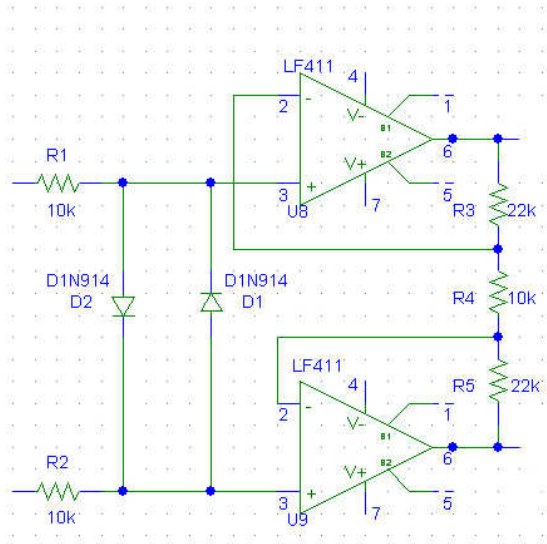



**Figure 4:** <u>The buffer stage of the ECG circuit</u>. The patient would be connected to $v_a$ and $v_b$ inputs. R1 and R2 add extra input impedance. The op amps provide high input impedance as to not load the patient and draw a lot of current that would distort the ECG signal. Diodes, D1 and D2, are used to minimize damage to the circuit from transient voltages. The gain for each buffer in this figure is 22k/10k=2.2 V/V.

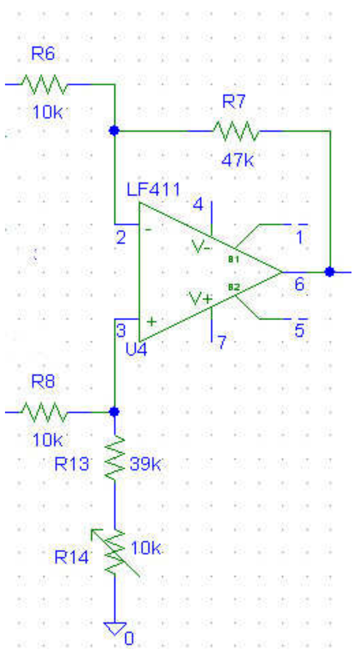



**Figure 5:** <u>The differential amplification stage of the ECG circuit design</u>. The potentiometer located at R14 can be adjusted to match R7 so that the common mode gain is minimal. This will improve the common mode rejection ratio (CMRR). The gain of this stage is 47k/10k = 4.7 V/V. Nodes *c* and *d* are the inputs from the output of the buffer stage. A voltage source of ±15 V/200 mA is used to power the op amp.

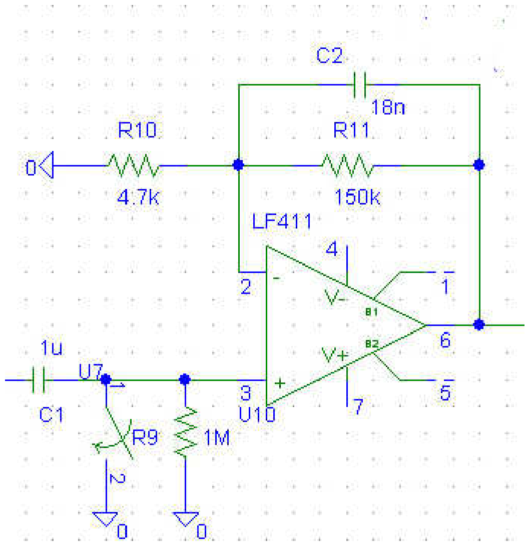
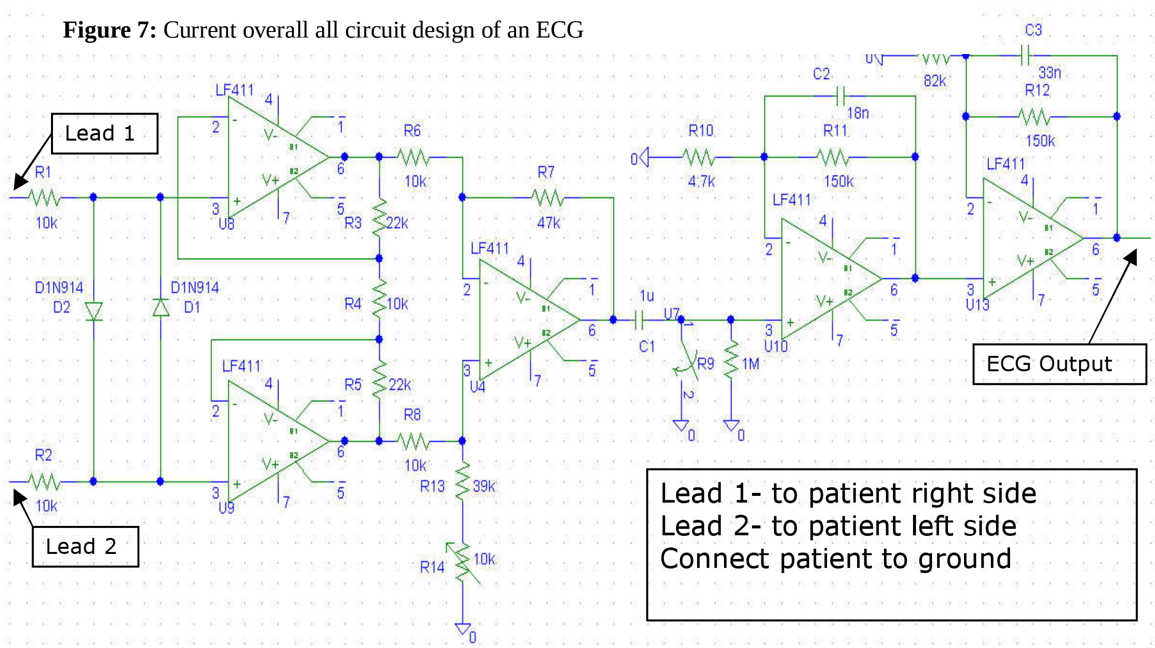



**Figure 6:** <u>The bandpass filter for the ECG circuit.</u> C1 and R9 make a highpass filter with a cutoff frequency of 160 mHz. C2 and R11 make a low pass filter with a cutoff frequency of 58.95 Hz. Node e is connected to the output of the differential amplifier. A voltage source of ±15 V/200 mA is used to power the op amp.

**Figure 7:** Current overall all circuit design of an ECG




## *Plethysmograph*

A plethysmograph is a device designed to measure the changes of volume in a tissue or organ. For the fingertip, it is used as a non-invasive measure of the amount of blood arriving at the fingertip due to the pressure wave created by the contraction of the heart. The device's LED emits light at two wavelengths: 650nm and 805nm. The light waves pass through the finger and the remaining light is picked up by a phototransistor. A common finger plethysmograph is

manufactured by Nellcor. The configuration of the serial connection to the circuit is shown in Figure 8.

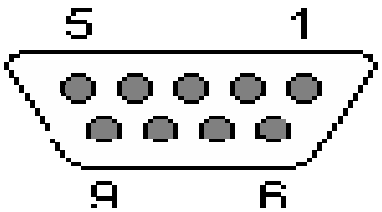



**Figure 8:**

Wire connections:

Pin 2 = Red (ground for LED)

Pin 3 = Black (power supply for LED)

Pin 5 = Orange (ground for phototransistor)

Pin 9 = Brown (circuit input for phototransistor)

A 1-4 mV signal is fed into the circuit, requiring several stages of operational amplifiers to process the signal. All op amps are powered by a +/- 15 V/200mA source, which also power the LED. The signal first passes through a bandpass filter to eliminate DC offset. Another filter is designed to filter common 60 Hz noise, similarly to the ECG circuit. Another bandpass filter is used to further reduce the noise. In the next stage, the signal is amplified to the required voltage of 4-8 V as required for measurement by the software. The final stage involves cutting out low frequencies produced by the motion of the finger. The four stages of the circuit are seen in Figure 9-12.

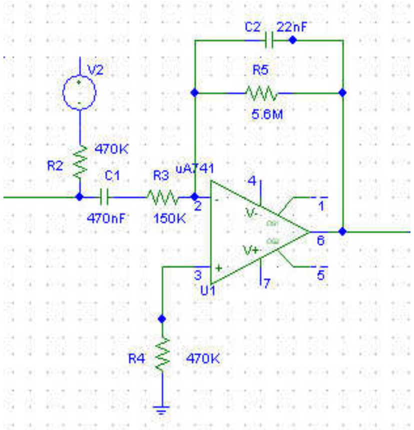



**Figure 9:** The first stage in the fingerplethysmograph circuit. It is an inverting bandpass filter with calculations. R2 and C1 create a HPF. C2 and R5 create a LPF. R5 and R3 provide gain. A voltage source of ±15 V/200 mA powers the op amp. The input of the circuit is connected to the finger probe.

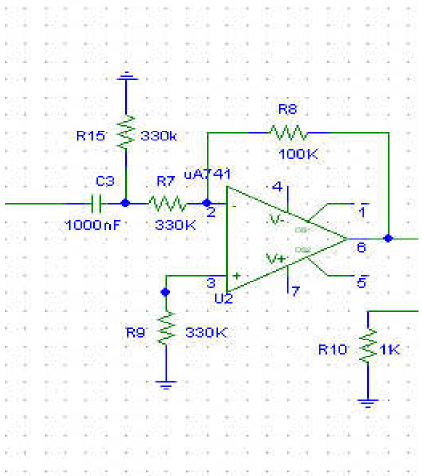



**Figure 10:** The second stage in the finger plethysmograph. It consists of a HPF, with R15 and C3 to setting the cutoff frequency at 0.4288Hz. A voltage source of ±15 V/200 mA powers the op amp. The input of the circuit is connected to the first bandpass filter.

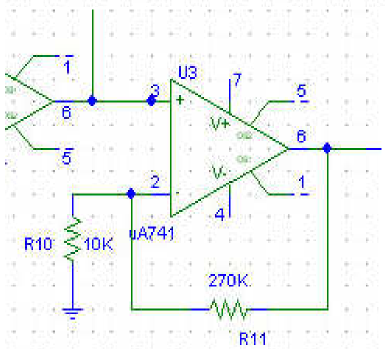



**Figure 11:** The third stage of the finger plethysmograph. R10 and R11 provide a gain of 28 V/V. A voltage source of ±15 V/200 mA powers the op amp. The input of the circuit is connected to the output of the second stage of the circuit.

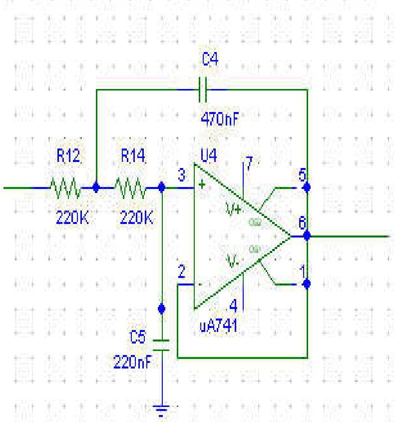



**Figure 12:** The final stage of the finger plethysmogaph circuit. The diagram is a sallen-key low pass filter, with a 2.32 Hz cutoff frequency. A voltage source of ±15 V/200 mA powers the op amp. The input of the circuit is connected to the output of the second stage of the circuit.

## ***Existing LabVIEW***

LabVIEW is a platform and development environment for a visual programming language from National Instruments. This graphical programming software lessens the complexity of software programming and coding. The language used by LabVIEW is called "G", or the data flow language[vii]. To collect data, the interface is connected with different function-

nodes by wires. The wires gather data, inputs the data into LabVIEW, and the software records the data as waveforms. It is important that the user connects the nodes and any other outlets before opening the software. This will allow the components to be detected before the program starts running. In addition, there are three components, or virtual instruments, in LabVIEW. This is because its appearance and operation imitates a physical instrument, such as an oscilloscope. These instruments include block diagram, front panel and connector panel.

The former BME students have created a LabVIEW setup to calculate PTT and display it to the user (as seen in Figure 13). It also includes sampled ECG and plethysmograph waves. The program was programmed in LabVIEW 7.1. The configuration of the program is indicated as the following:

- BoardNum = the DAQ's board number from InstaCal installation
- LowChan = 0
- HighChan = 1
- Count = 1000 = (number of seconds you would like to calculate PTT for) / [(# of channels) * (rate) ]
- Rate = 100 = (# of channels) * (highest desired sampling rate)
- Range = +/- 10 V
- Cont/Sngl = SINGLE

With the configuration properly entered, the program is initiated by clicking the RUN button. The ECG and plethysmograph waves are input to the program using VIs from Measurement Computing's Universal Library for LabVIEW[viii]. Next, the frequency is set to 0.05 Hz. Meanwhile, the threshold level is used to detect the peaks and frequencies of the QRS complex and plethysmograph waves for each heartbeat. For the interpretation of the data, the difference between the QRS complex peak and plethysmograph wave peak is taken, and it is calculated to be PTT.

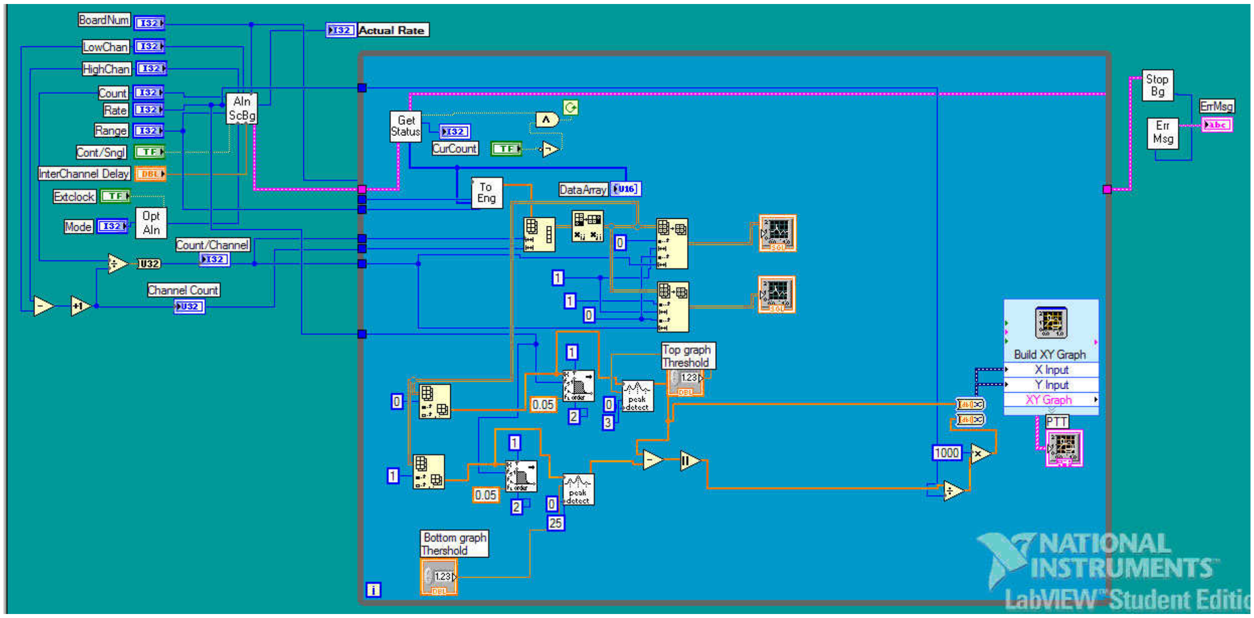



**Figure 13:** Block Diagram – on of the LabVIEW VIs.

## New Setup

## ECG Circuit

In order to provide the best signal output, the circuits must be optimized to provide the best signal possible. The ECG circuit was an area of particular interest as additional types of circuits can be used to provide a better signal to noise ratio. This increase in signal to noise ratio can be performed with the use of an instrumentation amplifier.

Two problems typically arise when dealing with the ECG circuits. One being the electrical potentials on the skin of a subject can be constantly changing, varying as much as 2 volts over time. This two volt signal is very small in comparison with the 1 mV signal which constitutes the electrical activity of the heart. Also, 60 Hz noise will be present anywhere around a subject. This signal is emitted from most electrical devices.

Since both of these signals are found in the environment, this signal will be common to both of the inputs. Therefore, in order to rid the output of both of these common noise signals, a high CMRR is needed. By rejecting this common signal to both, the true underlying signal can be found. In order to ensure that no more than 1% of the signal consists of this common noise, a CMRR of 100 decibels is desirable. Since this type of precision is hard to find in most standard operational amplifiers, an instrumentation amplifier can be used. The instrumentation amplifier, which is used in the new ECG circuit, is the Analog Devices AD624AD. The AD624AD was used because it could be made to have a gain of 1000, which is needed for the very weak ECG signals. It could also produce a CMRR of over 100 decibels. The instrumentation amplifier thus has the ability to be a very precise differential amplifier, or an amplifier which finds the difference of the two inputs and amplifies the signal. The use of the instrumentation amplifier will also provide the added benefit of taking up a smaller physical space compared with the previous five operation amplifier setup.

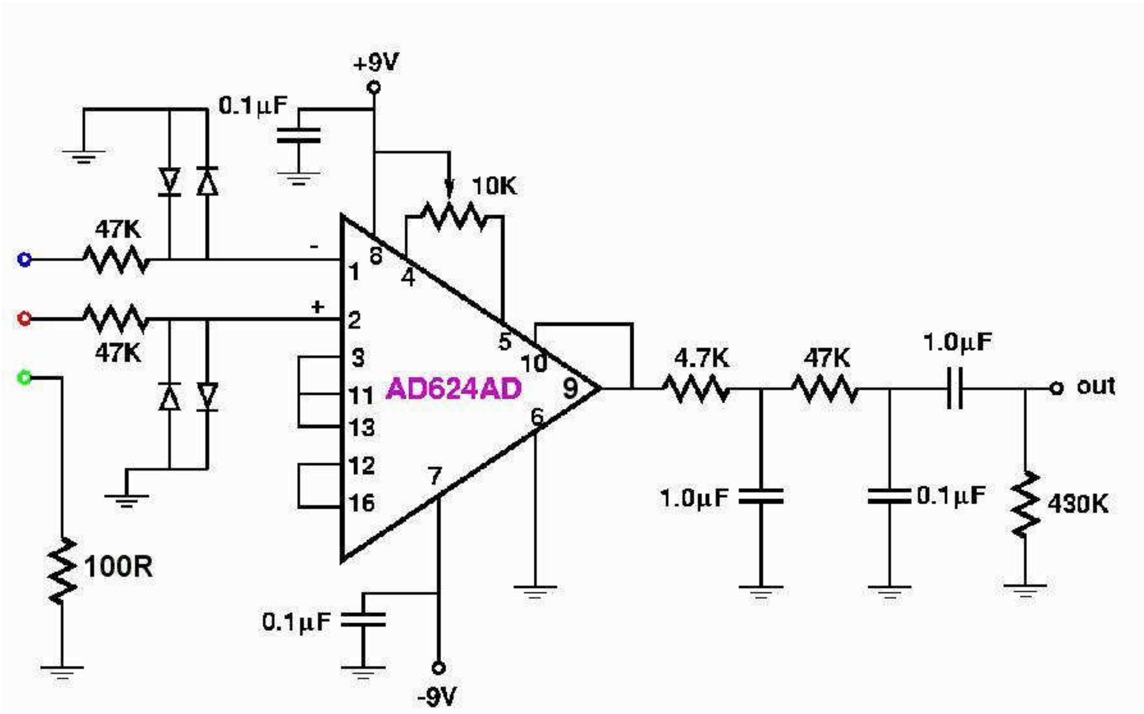



**Figure 14: The circuit diagram of the new circuit breadboard[ix].**

As can be seen from Figure 14, the Analog Devices AD624AD is at the heart of our new ECG. The two 47K resistors and the diodes are used for the safety of the subject. The 47K resistors increase the impedance in the case the signal was sent out of instrumentation amplifier. To further ensure the safety of the patients, the diodes were used to ground the output of current. To ensure the instrumentation amplifier has an output gain of 1000, pins 3,11,12,13, and 16 must be shorted as shown.

Once the output from the instrumentation is amplified, certain frequencies must be attenuated to decrease noise. Thus, a bandpass filter was created. Since the majority of the signal needed for the purposes of this setup resides within 30 Hz, a low pass filter is designed to attenuate signals greater than 33.9 Hz. The first low pass filter was designed using a 4.7K resister and a 1.0µF capacitor and the second was designed using a 47K

resistor and a 0.1µF capacitor. This can be calculated through the use of the following equation:

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

The addition of one low pass filter would mean the attenuation of 70.9% of the signal above 33.9 Hz. However, with the addition of a second low pass filter with a frequency response of 33.9 Hz, 50% of the signal greater than 33.9 Hz will be attenuated from the signal. Since the frequencies higher than 33.9 Hz are attenuated, the remaining 60 Hz noise will be removed from the sample. Since the DC offset is present in the signal, a high bandpass filter is added to the circuit. The DC offset accounts for variation in the baseline of the ECG signal. So the high bandpass filter will be used to attenuate signal lower than 0.37 Hz. The high bandpass filter was designed with a 1.0µF capacitor and a 430K resistor.

**New Design Setup**

The old setup had to be setup with a laptop at all times. This was a problem for the client as the device was not as portable, and problems may have arose because of a laptops sliding off a table while recording a patient during sleep. It would be beneficial to rid the setup of a laptop, which would make the setup much more portable compared with the previous design.

The addition of flash memory will allow the removal of the laptop computer from the setup. In order to implement a flash memory collector into the setup, three specific devices must be added to the circuit. First, an analog to digital converter must be added to allow the output voltages from the ECG and the finger plethysmogram circuit to be converted into binary, which can be read by the computer. The second element that needs

to be added is a microcontroller. The microcontroller is the “brains” of the operation. The microcontroller must be programmed to write the voltage outputs from the circuits to the memory storage device. Finally, a flash memory interface will be added to the design. This is how the flash memory will attach to the circuit and is the element that will transfer the data to the memory card.

In order to get the actual signal that is being output from the circuit, certain specifications of the system need to be met. The analog to digital converter is an important piece of the setup and has some requirements that need to be met in order to work with the circuits. The resolution of the ADC determines the accuracy that can be achieved with the converter. This is important because it will allow our client to best determine PTT. The use of a 12-bit ACD will allow for resolution of 1.2mV to be detected. Since a 12-bit converter will have $2^{12} = 4096$ “states”, then the output ranges of the circuits are 0-5 volts and the resolution will be equal to (5-0)/4096=1.22. This will be more than sufficient for the client to use in his data collection. The system must also have at least 3 IO slots. The inputs needs are: input from ECG circuit, input from finger plethysmogram circuit, and output to the flash memory. The sample rate needed to obtain an ECG signal is around 25 Hz and rate for the finger plethysmogram is around 7 Hz. To avoid anti-aliasing, the Nyquist theorem states that the sampling frequency must be at least twice the maximum value. Therefore, the ECG is the limiting factor for the sampling rate and an absolute minimum of 50 Hz is required to get a signal. To error on the side of caution, a final sampling rate of 100 Hz was chosen. In addition, the amount of memory needed for storage needed to be addressed. Since two channels running with a sample rate of 100 Hz for 8 hours (one nights sleep) uses approximately 10 MB of data, a

memory card of at least 32 MB will be needed to ensure adequate memory storage is available. In addition, since the ECG signal output can fall within +/- 1 V, a resolution of approximately 10 mV is needed. Moreover, the system will need to run on a battery source, so portability issues are maintained. Finally, the system must be small and compact, the final goal of the project is to have a compact design, and so this must be taken into consideration.

Since all of the specifications were determined, the best way to go about adding the memory card to the circuit needed to be determined. After doing preliminary research, it was found that creating this new device would require much more knowledge than originally thought. The decision to look elsewhere was made. After more research, a data logger was decided on. A data logger is an instrument that allows voltages to be inputted and stored onto a memory card.

After matching all of the specifications with the product, the Dataq 710-ULS was chosen. This product meets all of the specifications for our system. The datasheet for the Dataq 710-ULS is attached in the appendix**.**

## New software

### *Pulse Transit Time Calculating Algorithm*

The data acquired by the logger can be imported to the computer from the flash memory card and saved into a text file. A separate text file is created for both the ECG and the finger plethysmograph containing the corresponding voltages and times. The program used in calculating the pulse transit time is designed to determine the times corresponding to each R wave peak and the peak of the pressure waves, and subtracting these values to give the pulse transit time. The program is given 5 parameters in the command prompt: -e for the ECG file path, -f for the finger plethysmograph file path, -o

for the file to which the data is to be output, -te for the threshold value of the ECG, -tf for the threshold value for the finger plethysmograph data. An example parameter path could be: –e C:/ecg/ecg.txt –f C:/ecg/finger.txt –o C:/ecg/output.txt –te 4 –tf 0.04. The program stores these values in an array for each of the two data sets. The program proceeds to search through the points to determine the maximum peak values. A “while” loop is used to read the data until the end of the text file is reached. Running through the data, the program determines the points within the range of the threshold value as entered by the user. “For” loops are used to cycle through all of the points and compare them to the other data to determine if they qualify as a peak. This is done for both of the data sets and stored in separate arrays. The times from each array are then subtracted from one another and the corresponding pulse transit times are output into the designated file using a “for” loop to cycle through the data for the number of peaks entered.

```
#include <string.h>
#include <iostream>
#include <fstream>
#include <sstream>
#include <vector>


int main(int argc, char* argv[])
{

  const char* inputEKGFileName = NULL;
  const char* inputFingerFileName = NULL;
  const char* outputFileName = NULL;
  double thresoldEKG = 0;
  double thresoldFinger = 0;

  int i;
  for (i = 1; i < argc; i++) {
    if (strcmp(argv[i], "-e") == 0) {
      inputEKGFileName = argv[++i];
    }
```

```
    else if (strcmp(argv[i], "-f") == 0) {
      inputFingerFileName = argv[++i];
    }
    else if (strcmp(argv[i], "-o") == 0) {
      outputFileName = argv[++i];
    }
    else if (strcmp(argv[i], "-te") == 0) {
      std::stringstream ss;
      ss << argv[++i];
      ss >> thresoldEKG;
    }
    else if (strcmp(argv[i], "-tf") == 0) {
      std::stringstream ss;
      ss << argv[++i];
      ss >> thresoldFinger;
    }
    else {
      std::cerr << "usage: " << argv[0] << "\n"
                << " [-h (for help)]\n"
                << " -e (EKG input file name, no
spaces!)\n"
                << " -f (Finger input file name, no
spaces!)\n"
                << " -o (input file name, no spaces!)\n"
                << " -te (thresold value fo EKG)\n"
                << " -tf (thresold value finger)\n"
                << std::endl;
      return 0;
    }
  }
  std::ifstream infileEKG(inputEKGFileName);
  std::ifstream infileFinger(inputFingerFileName);
  std::ofstream outfile(outputFileName);

  double threshValue = -1000;
  double minTimeInterval = 0.25;

  // PROCESS EKG DATA

  // read two columns of data from input EKG file, should
be space separated
  double t,v;
  std::vector<double> timeEKG;
  std::vector<double> voltsEKG;
  while(infileEKG >> t >> v) {
    timeEKG.push_back(t);
    voltsEKG.push_back(v);
```

```
  }


  // threshold volts array
  std::vector<double> voltsEKGThresh;
  for( i=0; i<voltsEKG.size(); i++) {
    if (voltsEKG[i] < thresoldEKG) {
      voltsEKGThresh.push_back(threshValue);
    }
    else {
      voltsEKGThresh.push_back(voltsEKG[i]);
    }
  }

  // find local max in threshold array
  std::vector<double> peakTimesEKG;
  for( i=1; i<voltsEKGThresh.size()-1; i++) {
    if (voltsEKGThresh[i] != threshValue &&
      voltsEKGThresh[i] >= voltsEKGThresh[i-1] &&
      voltsEKGThresh[i] >= voltsEKGThresh[i+1]) {
      if (peakTimesEKG.size() == 0) {
        peakTimesEKG.push_back(timeEKG[i]);
      }
      else if (timeEKG[i] -
peakTimesEKG[peakTimesEKG.size()-1] > minTimeInterval ) {
        peakTimesEKG.push_back(timeEKG[i]);
      }
    }
  }


  // PROCESS FINGER DATA

  // read two columns of data from input finger file,
should be space separated
  std::vector<double> timeFinger;
  std::vector<double> voltsFinger;
  while(infileFinger >> t >> v) {
    timeFinger.push_back(t);
    voltsFinger.push_back(v);
  }


  // threshold volts array
  std::vector<double> voltsFingerThresh;
  for( i=0; i<voltsFinger.size(); i++) {
    if (voltsFinger[i] < thresoldFinger) {
```

```
      voltsFingerThresh.push_back(threshValue);
    }
    else {
      voltsFingerThresh.push_back(voltsFinger[i]);
    }
  }

  // find local max in thresholded array
  std::vector<double> peakTimesFinger;
  for( i=1; i<voltsFingerThresh.size()-1; i++) {
    if (voltsFingerThresh[i] != threshValue &&
      voltsFingerThresh[i] >= voltsFingerThresh[i-1] &&
      voltsFingerThresh[i] >= voltsFingerThresh[i+1]) {
      if (peakTimesFinger.size() == 0) {
        peakTimesFinger.push_back(timeFinger[i]);
      }
      else if (timeFinger[i] -
peakTimesFinger[peakTimesFinger.size()-1] > minTimeInterval
) {
        peakTimesFinger.push_back(timeFinger[i]);
      }
    }
  }

  // output diff between peak times
  int n = peakTimesEKG.size();
  if (peakTimesFinger.size() < n) {
    n = peakTimesFinger.size();
  }

  for( i=0; i<n; i++) {
    outfile << peakTimesEKG[i] << " , "<<
peakTimesFinger[i]<< " , " << peakTimesEKG[i]-
peakTimesFinger[i] << std::endl;
  }

  infileEKG.close();
  infileFinger.close();
  outfile.close();
  return 0;
}
```

## Testing

The previous and new ECG circuits are compared against each other in the following diagrams (Figure 16 and 17 respectively).

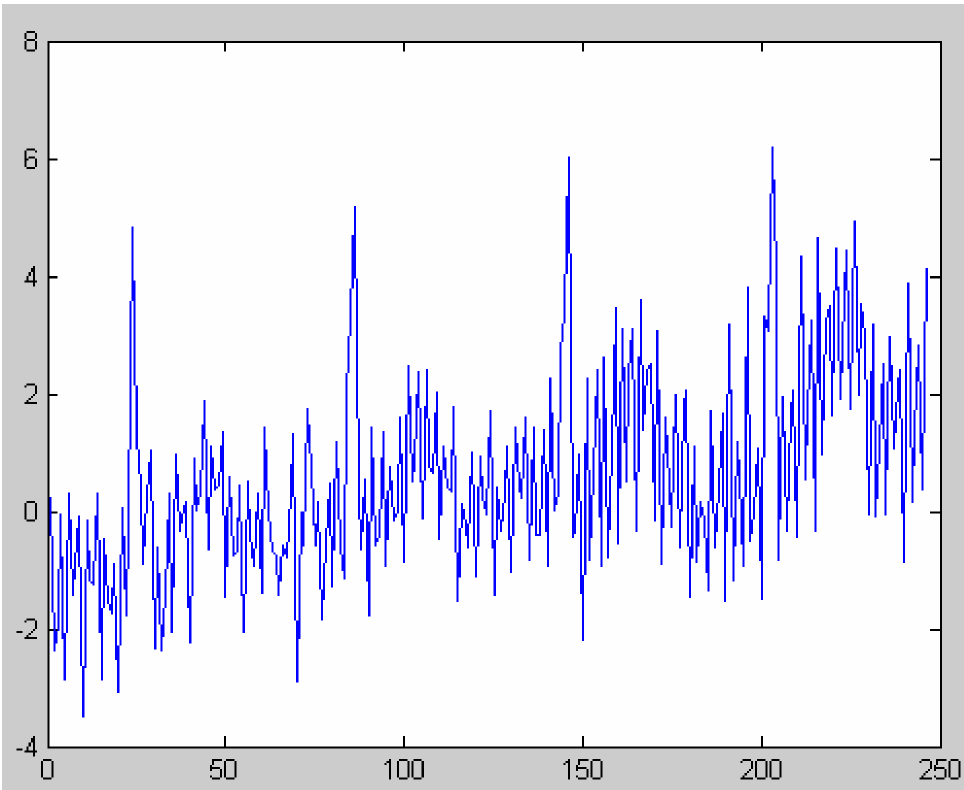



**Figure 16: Electrocardiogram signal from old circuit.**

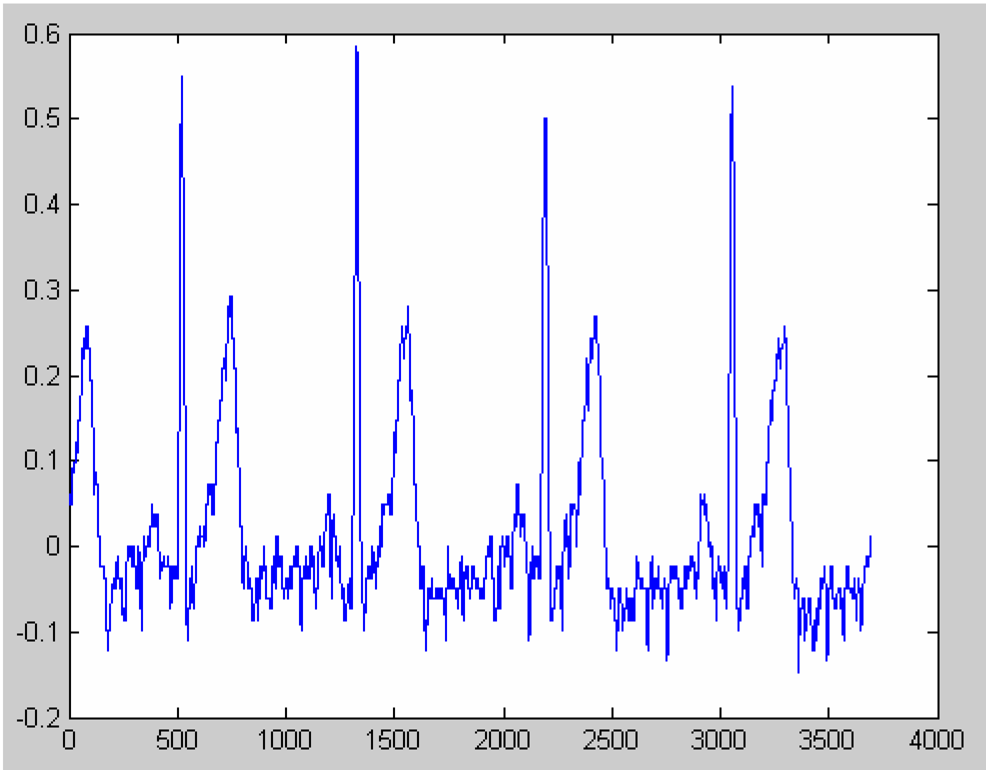



**Figure 17: Electrocardiogram from the new ECG circuit recorded by the data logger**

As can be seen from the two graphs, the previous ECG signal has a large amount of noise present around the 60 Hz frequency. This is due to the use of the low pass filter that has a frequency response of only 58.9 Hz and thus not attenuated the 60 Hz frequencies.. The new ECG signal, however, does show a much cleaner signal compared with the previous signal due to the use of the bandpass filter that reduces the noise and stabilizes the signal.

## Future Work

The future work of this project is split into three categories: the circuit, the overall setup and the software. To further reduce the noise of the circuit and its overall size, it would be beneficial to obtain a printed circuit. A printed circuit was not made in this phase of the project so that further research could be made into the circuit to make sure it truly is the best design. If

the new design is the best, then a printed circuit can be made. A printed circuit would reduce the interference from wire crossings that exist on a breadboard. Printing the circuit would also compact the circuit so that it could better fit into a small device.

Once the circuit has been printed, the next step will to be to combine the ECG and plethysmograph circuits, the data logger, and the power supply into one unit. This unit should be as small as possible so that it causes the least discomfort to the patient. The design must allow easy access to the batteries for when they need changing and to the memory card slot so there is no hassle in removing the card. The power supply is another area that could be looked to increase efficiency. The current device runs off of several 9V batteries that take up a lot of space in the device and are quite costly. Future work should be focused on determining if another type of battery can be used or if there is another configuration design for the batteries. Rechargeable batteries could also be looked into and designed to be similar to the ones used in cell phones.

Moreover, the software needs to continue to be improved. The current software only records in a previously defined period. If the heart rate increases or decreases during sleep, then the period between R waves can change; the software does not know how to recognize this. The client specified this would be an important feature to look at. An algorithm also needs to be developed to further reduce the noise of the ECG signal. Modifications to the circuit can only reduce the noise to a certain level and cannot account for the noise generated from movement of the patient. If the signals experience a few disturbances during the night, it may be possible for the software to recognize the long pause between recognizable R waves and be able to reset itself once a true R wave is found. The client is willing to sacrifice a few data points for a cleaner data set.

The improvements made to this design over the semester have made the miniaturization of the device possible. The incorporation of a data logger enhances its portability and its safety. The new ECG circuit decreases the noise in the signal and combined with the new software, allows for a more accurate measure of PTT. Future work on this project will bring it closer to

being ready for clinical trial and to the ultimate goal of the project: incorporation into sleep studies.

October 25, 2006
Updated: December 11, 2006

Jonathan Baran
Mark Yarmarkovich
William Stanford
Karen Chen

# Pulse Transit Time Device PDS

## Problem Statement

The current instruments used in the measurement of pulse transit time are inefficient for home use. An existing product with working ECG and pulse wave circuits along with software to analyze the data has been provided. The primary goal will be to optimize the existing setup for use at home. This will be performed by miniaturizing the circuit, increasing the signal to noise ratio, and improving the already existing software. These tasks will attempt to be rectified by numerous design additions.

## Client Requirements

- Optimized for use with children
- Integrate with previously designed system
- Write program to deal with skipped heart beat
- Integrate flash memory to make device more compact
- Improve signal to noise ratio

## Design Requirements

### 1. Physical and Operational Characteristics

a. *Performance requirements*: The probe needs to be small enough to stay fixed to children's fingers. More over, the probe needs to stay attached to the finger during sleep. The monitoring system needs to be non-evasive and not interfere with sleep patterns. The monitoring

software needs to record the signals from the finger probe and ECG leads while filtering out skipped beats.

b. *Safety*: Because the device will be used while children are sleeping, the chords must not present a strangling hazard. Also circuits must be made to minimize electrical hazards.

c. *Shelf life*: The device must be operated by batteries to allow it to be compact. The adhesive needs to be reusable.

d. *Accuracy and Reliability*: The device must employ a circuit design the takes into account both gain and CMRR. A gain of 1000 is need to amplify the ECG signal and a CMRR of 100 decibels is also needed

e. *Life in Service*: The device should be able to be used for multiple uses by a single patient and then repeated for many more patients.

f. *Operating Environment*: Ideally, the device will be able to be used in the patient's home while they are sleeping. If optimal minimization is achieved, it could be worn on the waste.

g. *Size*: The device needs to be small enough to fit on a bed size table, or optimally, attached to the waste of the patient to allow the chords to be non-evasive.

h. *Weight*: The device needs to be as light as possible so it will not be felt while sleeping.

i. *Materials*: Plastic would be used for the casing of the device.

**2. Product Characteristics**

a. *Quantity*: For the time being, only one device is necessary. If the device is successful, multiple units may be made in the future.

b. *Target Product Cost*: The device should be as inexpensive as possible so not to burden families when the device is taken home.

**3. Miscellaneous**

a. *Standards and Specifications*: The device must be meet FDA medical device specifications.

b. *Customer*: Christopher Green, M.D. Dept. of Pediatrics UW Medical School

c. *Patient-related concerns*: The patient's personal information will not be stored in the device. No sterilization of the device will be needed in-between uses aside from the application of new EKG patches at every use.

d. *Competition*: Probe devices with LED are on the market but there are no devices that measure blood flow and pulse transit time.

## Appendix

### Appendix Figure 1: Datasheet for the Dataq 710-ULS[x]

# DI-710 Specifications

## Analog Inputs

**Number of Channels:** 16

**Channel Configuration:** 16 single-ended; 8 differential; programmable per channel

**Measurement range, Accuracy, and Resolution**

| | Gain | Range | Accuracy | Resolution |
|---|---|---|---|---|
| PGL models: | 1 | ±10V | ±.25%FSR | ±1.22mV |
| | 10 | ±1V | ±.25%FSR | ±122µV |
| | 100 | ±100mV | ±.25%FSR | ±12.2µV |
| | 1000 | ±10mV | ±.25%FSR | ±1.22µV |
| PGH models: | 1 | ±10V | ±.25%FSR | ±1.22mV |
| | 2 | ±5V | ±.25%FSR | ±610µV |
| | 4 | ±2.5V | ±.25%FSR | ±305µV |
| | 8 | ±1.25V | ±.25%FSR | ±153µV |

**Input Impedance, single-ended:** 1MΩ

**Input impedance, differential:** 1MΩ each input to common

**Input bias current:** 10µA for a 10V input, single channel

**Input offset voltage:** Auto-zero

**Input offset current:** 2nA (single channel)

**Max. normal mode voltage:** 30V DC or peak AC

**Max. common mode voltage:** 30V DC or peak AC

**Common mode rejection:** 80db, gain=1, 1KΩ unbalance

**Channel-to-channel crosstalk rejection:** -75db @ 100Ω unbalance

**Gain temperature coefficient:** 50 ppm/°C

**Offset temperature coefficient:** 0.25µV/°C

**Digital filtering:** Standard: Conditional over-sampling
Stand-alone: None

## A/D Characteristics

**Type:** Successive approximation

**Resolution:** 14-bit

**Monotonicity:** ±2 LSB

**Conversion Time:** 69µs

## Scanning Characteristics

**Max. throughput sample rate:*** Standard: 4,800 Hz
Stand-alone: 14,400 Hz**

*When acquiring more than one channel at a gain of 100 max throughput is 7200 Hz; When acuiring more than one channel at a gain of 1000 max throughput is 900 Hz.
**Assumes SD memory latencies of 80 milliseconds or less.

**Min. throughput sample rate:** Standard: 0.0034 Hz
Stand-alone: 0.0017 Hz

**Max. scan list size:** 17 entries

**Sample buffer size:** 2kb

## Controls (Stand-alone models)

**Single push-button:** Manual control Record and Standby

## Calibration

**Calibration cycle:** One year

**Calibration method:** Automated Calibration Software, provided.

## Digital I/O

**Bits:** 8 bidirectional bits

**Configuration:** Each bit is programmable as Input or Output

**Output voltage levels:** Min. “1” 3V @ 2.5mA sourcing
Max. “0” 0.4V @ 2.5mA sinking

**Output current:** Max. source, -2.5 mA; Max. sink, 2.5mA

**Input voltage levels:** Min. required “1” 2V; Max allowed “0” 0.8V

## Ethernet Interface

**Type:** 10/100Base-T

**Connector:** RJ-45

**Protocol:** TCP/IP

**Server Type:** DHCP

## Removable Memory (Stand-alone models)

**Type:** SD (Recommended: SanDisk SD Card)

**Capacity:** 16 Mb to 1 Gb

## Real Time Clock (Stand-alone models)

**Type:** Date, hour, minute, second

**Resolution:** 1 second

**Accuracy:** 20 ppm

## Indicators

**Stand-alone models:** Three-color LED indicating Record, Standby, and Error conditions

**Standard models:** Power LED

## Transfer Rate to PC

**Real Time:** up to 4,800 samples per second

**From Memory Card:** up to 2,400 samples per second (Ethernet only)

## General

**Panel indicators:** Mode LED

**Panel Controls:** Control push button (Stand-alone models)

**Panel Slots:** Accepts SD-type flash memory

**Input connectors:** Two, removable sixteen position terminal blocks

**Operating Environment:** 0°C to 70°C

**Enclosure:** Aluminum base with steel wrap-around. Aluminum end-panels with plastic bezels.

**Dimensions:** $5^{7}/_{16}$"D × $4^{1}/_{8}$"W × $1^{1}/_{2}$"H
13.81D × 10.48W × 3.81H cm.

**Weight:** 14 oz.

**Power Requirements:** USB: 9 to 36 VDC, 2 watts max
Ethernet: 9 to 36 VDC, 2.5 watts max

**References**

[i] *Sleep apnea.* (2006). Retrieved October 9, 2006 from http://en.wikipedia.org/wiki/Sleep_apnea

[ii] Smith R, Argod J, Pépin JL, Lévy P. 1999. Pulse Transit Time: an appraisal of potential clinical applications. *Thorax*. 54:452-458.

[iii] Pagani J, Pia M, Calcagnini G, Alterio A, Ambrosio R, Censi F, Ronchetti R. 2003. Pulse Transit Time as Measure of Inspiratory Effort in Children. *Chest*. 124:1487-1493.

[iv] Karas, A., Hondl, B., Olson, M., & Cohen, Z. (2005). *Measurement of pulse transit time.* Unpublished manuscript. Retrieved September 25, 2006,

[v] Katz E, Lutz J, Black C, Marcus C. 2003. Pulse Transit Time as a Measure of Arousal and Respiratory Effort in Children with Sleep-Disordered Breathing. *Pediatr Res*. 53:580-588.

[vi] Carlson, S. (2000, the amateur scientist: Home is where the ECG is. *Scientific American Magazine,* (June) 3. Retrieved October 17, 2006,

[vii] *Labview.* (2006). Retrieved October 1, 2006 from http://en.wikipedia.org/wiki/LabVIEW

[viii] *LabVIEW tutorials.* (2005). Retrieved October 5, 2006 from http://www.upscale.utoronto.ca/GeneralInterest/LabView.html

[ix] *Electrocardiogram (ECG) project for DrDaq.* (2006). Retrieved October 12, 2006 from http://www.picotech.com/applications/ecg.html#cct

[x] Dataq Instruments. (2006). Retrieved on December 11, 2006 from http://www.dataq.com/support/documentation/pdf/datasheets/710ds.pdf