<u>Gas Pressure Meter</u>

Engineering World Health

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BME 400

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<u>Abstract</u>

Generally, gas pressures created by anesthesia and ventilation machines are measured with an internal pressure sensor in order to avoid unsafe conditions for patients. However, in developing countries this sensor may not be functional and an external device may be required to monitor pressure. Engineering World Health (EWH), an organization that donates medical equipment to underserved areas of the world, is in need of a device like this. Once background research was completed, a pressure transducer design was selected, consisting of a transduction circuit, a microcontroller, and a LED display. Testing showed that the pressure sensor was precise and linear. The final device was found to measure pressure to within 6% of its true value.

Background Information

Engineering World Health (EWH) is an organization of engineers, scientists, doctors, students, and other people who are interested in making a difference in medical care in disadvantaged areas of the world. The organization supplies appropriate medical technology to third world countries in order to provide adequate medical care for people living in these regions. (1)

EWH has a program which sets specifications for certain medical devices that would be useful in these disadvantaged areas and then accepts design ideas for these devices. If the designs meet the requirements and are approved, EWH provides up to $150 for prototype production. Eventually, if the device is mass-produced for use in a third world country, the designers can accompany EWH members to distribute the device. (1)

The Engineering World Health website (http://ewh.org) has provided specifications for the gas pressure meter project, which will be used with anesthesia and ventilation machines. The most limiting factor of the design process for this device is the cost, which tends to be true for most of the EWH projects. Normally pressure is measured with a meter found within anesthesia and ventilation machines, but oftentimes equipment in developing countries breaks and is rendered useless because the resources are not available to repair the small malfunction. Therefore, an external attachment for the machines may be all that is necessary to make the equipment functional again and to ensure that the pressures are safe for the patient.

Physiological Considerations

The purpose of a ventilator is to allow a person to breathe when they are unable to do so on their own. Although breathing may seem automatic and instinctive, the muscles controlling the diaphragm and the intercostals muscles are under voluntary control. Under certain conditions the body cannot control the breathing function.

When a patient is hooked up to a ventilator machine, the air pressure that is being pushed into the lungs must be carefully monitored to prevent damage to lung tissue. The lungs are filled with tiny sacs called alveoli; these sacs could sustain damage if the pressure exiting of the ventilation machine is too great. The pressure simply needs to be great enough to provide enough oxygen to the lungs in order to properly oxygenate the blood. Any extra pressure could be detrimental to the health of the patient.

Physiologically, the pressure in the lungs fluctuates subtly during normal breathing (2). The pressure in the alveoli during inhalation and exhalation is 759 mmHg and 761 mmHg, respectively. Thus, the pressure going into the lungs must be measured very precisely in order to prevent any tissue damage.

Current Technology

There are many forms of pressure measurement in use today in various applications. Pressure is most commonly known as the relationship between the amount of force and the area on which it is exerted. This is the basis for the equation Pressure = Force / Area. Most pressure measurement devices incorporate this concept in some way, and make measurements with some reference to a zero point (usually atmospheric pressure). There are also two types of pressure considered – static pressure and dynamic pressure. Static pressure is exerted in all directions. Dynamic pressure is produced by the flow of gas or liquid and is dependent on this movement. These two types of pressure also differ in the way that they are measured. Since static pressure is constant in all directions, the measurement device may be positioned anywhere along the line. Dynamic pressures are measured in a more complex fashion, using references to other pressures.

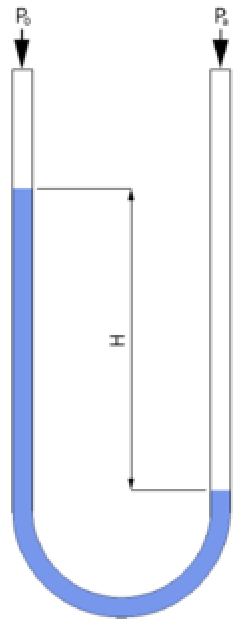

Figure 1 - Diagram of a monometer. Pa represents the unknown pressure, and Po represents the reference pressure. (Wikipedia.org, n.d.)

However, since dynamic pressures are less important in this design (because the existing pressure sensors in ventilation machines only measure static pressures), they will not be considered in depth. Pressure measurement is governed by Bernoille's equation, P $= 1/2\rho v^2 + \rho gh + c$, where P is the pressure, ρ is the density, v is the velocity, h is elevation, c is a constant, and g is gravitational acceleration.

The static component of pressure is represented by the constant c, whereas the dynamic pressure has the velocity component and is represented by $1/2*\rho v^2$. In this application, the velocity of the fluid is negligible, so the main concern is measuring the constant c in the pressure equation.

One of the most basic forms of pressure measurement is the monometer. It utilizes a U-shaped column filled with some liquid, typically mercury or water (Figure 1). The pressure displaces the liquid until the liquid's weight compensates and reaches equilibrium. The height of the liquid in one side of the column is used as a reference and compared to the height in the other side, using the difference as a measure of the corresponding change in pressure. This is a preferred method for analog analysis due to its simplicity and inexpensiveness.

Another common form of measurement is called the bourdon gauge. It involves a coiled tube connected to a port that is exposed to a changing pressure (Figures 2 and 3). As the pressure increases, the tube straightens out, rotating a gear train connected to a dial. The stronger pressure extends the tube farther and the dial is rotated. This is another gauge that is primarily used in analog pressure analysis.

The most common pressure detector in digital analysis is a strain gauge. An example is the foil strain gauge, which works as follows: as the foil is deformed by pressure, its resistance changes. Using a Wheatstone bridge, this resistance can be measured and converted into a voltage. With a gauge factor and the change in voltage, corresponding pressures can be output in a digital readout. Our particular device uses a patented Motorola sensor die with a piezoresistive gauge. It also has built in temperature compensation up to 40°C (~104°F).

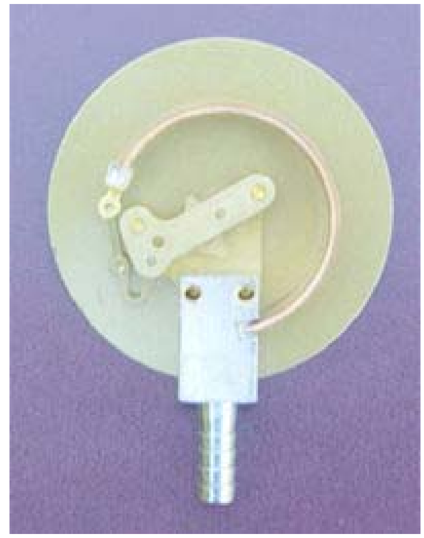

Figure 2- Backside view of a Bourdon Gauge. (Wikipedia.org, n.d.)

<u>Problem Statement</u>

Our goal is to design and build a gas pressure meter that meets the specifications of the Engineering World Health design program in order to improve medical care in developing countries. The gas pressure meter will be used to measure pressures at the outlet of anesthesia and ventilation machines, and to display these pressures through a digital readout. It will be able to measure oxygen, medical air, and carbon dioxide. It is very important that the pressure is measured precisely, accurately, and easily to ensure proper medical treatment of adults and children. (3)

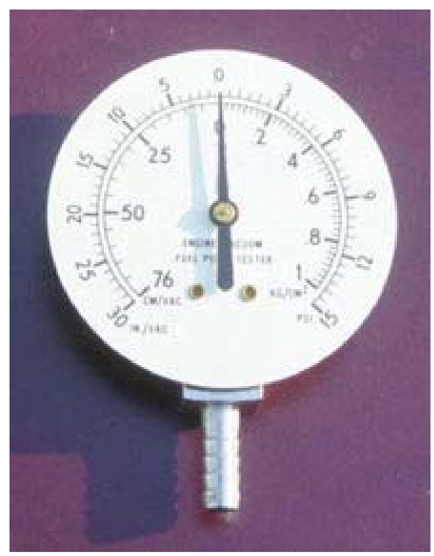

Figure 3- Frontside view of a Bourdon Gauge. (Wikipedia.org, n.d.)

Ethical Considerations

Ethics is an important concern for this project. We have to consider several issues, such as the relationship between cost and accuracy, the concept of what is "good enough" given that they have no other options, and the fact that a life could be depending on its functionality. Some of the issues, such as cost versus accuracy, were already decided for our group by our sponsors. Our responsibilities lie in the quality and impartiality of our testing and our honesty in our final report. We need to be more concerned about the safety of the people using our device than the grades we get from a poorly tested but apparently successful project. Though Engineering World Health or the FDA of the country will have a more final say on the usability of our device, we still need to be as rigorous as possible.

Constraints

This device must abide by several constraints in order to be a plausible solution to the problem of the lack of functional machines. Several of these constraints were explicitly defined by EWH, and the rest were specified by the group through research.

With regards to functionality, EWH has specified that the device must be able to measure pressure to within at least 10% of its true value, while 1% is optimal. The device needs to have a continuous readout, implying that the measurement system must also be continuous. The readout must be digital, which will improve the accuracy of the device by removing error of the operator. Dimensions must be 4 inches by 4 inches by 1 inch for a device with only one segment, or 1 inch by 4 inches by 1 inch for a device with several segments. EWH has specified that the device will be used to measure medical gasses, such as O_2, CO_2, and medical air, and the range of pressures that the device will measure should be between 0 mmHg and 75 mmHg. (4)

The connectors for the device must be universally compatible with anesthesia and ventilation machines, which will require coupling that is generally available in a hospital setting or that is included with the machines. The device must also be reusable, although not necessarily autoclave compatible. The final specification is the cost; in quantities of 500, the device needs to cost less than $5, including packaging, but not including the cost of manufacturing.

Several other safety requirements were not specified by EWH, but should still be taken into consideration when designing the device. The materials that are used must be nontoxic and should not shed any debris. Also, since the device will be directly linked to the airflow going to the patient, the design of the device should not block the airways in the case of a malfunction. Weight was not specified, and will probably not be a significant constraint on the design.

Last Semester's Work

Last semester a device employing a strain gauge to transduce the input pressure into an output voltage was chosen. The basic principle of the strain gauge is the Wheatstone bridge shown in Figure 4. Using the known resistances of three resistors, an unknown voltage can be calculated over the fourth resistor, labeled R_c. Often, a material that changes in impedance due to some sort of change in shape is used for R_C. Typically this is a piezoresistive material (5). These devices are often inexpensive and will be useful in our design.

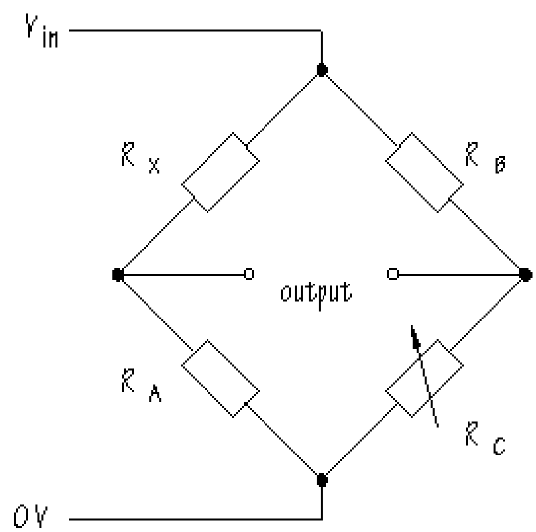

Figure 4 – Depiction of a wheatstone bridge. Using three known resistors a change in voltage can be determined off of a fourth resistor based on the mechanical stimulus it receives. (Westminster School Intranet, n.d.)

One advantage of this design is that it is highly accurate. This provides the best chance of reaching the goal of measuring to within 1% of the true value of pressure input. Also, this design is relatively inexpensive because the pressure transducers are already commercially available. The main disadvantage of this design is that it has many circuitry components, which increases the chance of failure in any single component of the circuit. Last semester we designed, built, and tested a "proof-of-concept" prototype gas pressure meter, consisting of a circuit that converted pressure input into voltage output using a strain gauge. Figure 5 shows the schematic of our pressure transduction circuit from last semester. In our design, the current first passed through a buffer to prevent a high source resistance from being loaded by the skin, which has a low-impedance (5). Once the voltage was established through the strain gauge, the current was sent through an inverting amplifier. To test the accuracy of our circuit we used known pressure inputs and compared these values to the voltage output of our device. The measurements from the prototype pressure gauge were compared to those obtained using a KAL84 pressure calibrator, which has an accuracy of within 0.1 mmHg. By trial and error, we found that 30kΩ resistors showed the

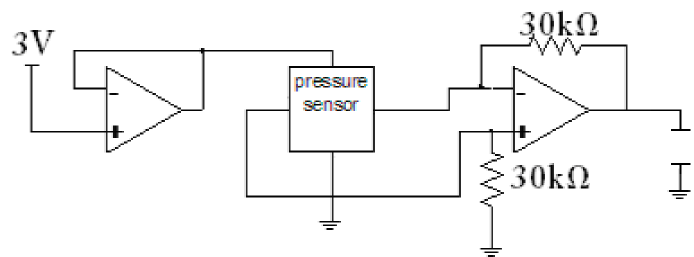

Figure 5 – Preliminary circuit design used in the first iteration of the prototype. A buffer reduces loading on the strain gauge. The final stage is an inverting amplifier with two 30k resistors.

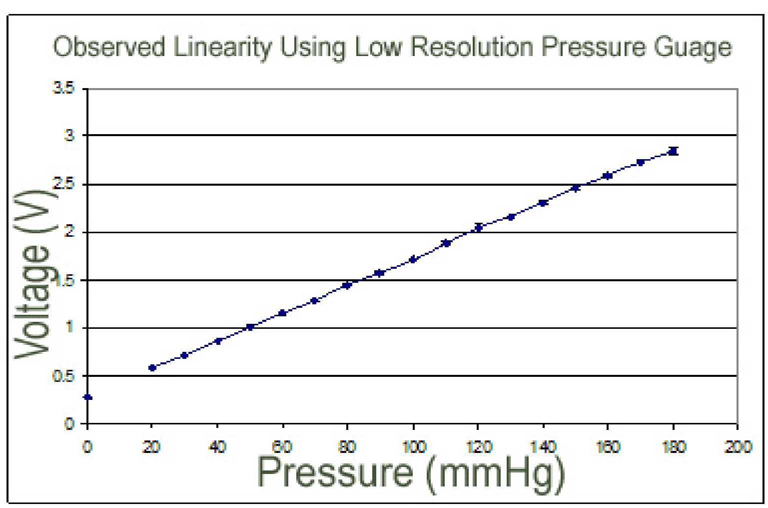

Figure 6 – Voltage output vs. pressure input as transduced by the first generation prototype displaying linearity from 0 to 200mmHg.

most consistently accurate voltage readings (the lowest standard deviation) of all resistor pairs tested with the inverting amplifier. The source and indicator were borrowed from Professor Naomi Chesler's lab. Figure 6 shows the linear relationship between the pressure input and the voltage output, acquired through preliminary testing. Because the relationship is linear, the algorithm necessary to convert this voltage to the displayed pressure value, in terms of mmHg, on the two LEDs will be a first degree equation, and therefore will require only basic computing techniques.

<u>This Semester's Work</u>

One of the main goals for this semester was to submit a proposal and receive money from Engineering World Health to build and test a prototype of our gas pressure meter design. We submitted this proposal halfway through the semester and EWH approved the design by providing $120 for the project.

The current design contains a circuit, similar to that used last semester with several additional components. We began by adding an additional gain amplifier following the differential amplifier in order to improve the sensitivity of the output. We then added a microcontroller with built-in analog to digital conversion to power and control two 7-segment LEDs. The final circuit design will be assembled on a printed circuit board (PCB), which will be one of the focuses of

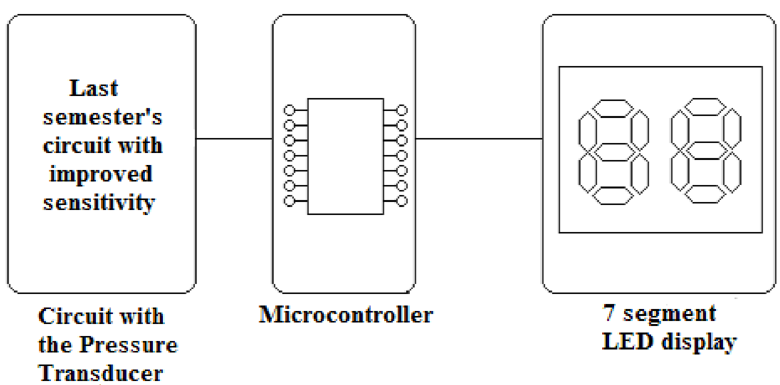

Figure 7- Block diagram of the final system. A pressure source will enter the system through the circuit appearing in Figure 1. The voltage exiting the circuit will enter the microcontroller which will power the LED display, showing a corresponding pressure input in the form of mmHg.

next semester's work. The system schematic can be seen in Figure 7. The device is kit-able, which was one of the requirements put forth by Engineering World Health.

Due to the many changes to last semester's circuit, subsequent testing was performed in order to ensure that the device maintained high performance levels. The relationship between the pressure input and voltage output was tested to determine whether the relationship was linear as it was last semester. This will also be discussed in the Testing section of the paper. However, it is important to note that preliminary testing revealed that the range of voltage output values was only 0.2 Volts and therefore that the device was susceptible to large calibration variances due to interference from surrounding electrical equipment.   The addition of a gain for a range of output voltages covering 10 Volts reduced this effect. This will also be discussed in the Testing section of the paper.

Another facet of the design that has been addressed this semester is the connection of the device to the ventilation or anesthesia machine. Since the device is required to measure only static pressure, a T-connector was chosen to connect the device to the machines. This type of connector is commonly found in hospitals and is compatible with ventilation and anesthesia machines. This reduces the cost of the device because the part will be locally available. As shown in Figure 8, one normal ventilation path, the other to the pressure transducer. Since the diameter of

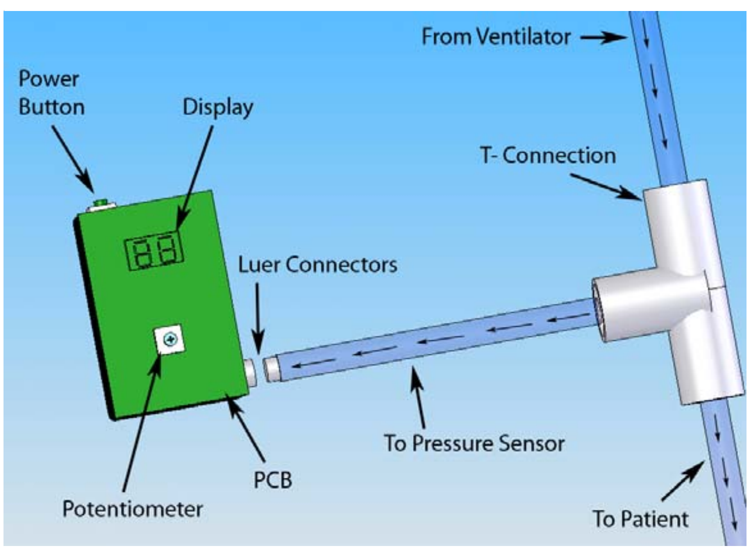

Figure 8- Schematic of proposed design, showing couplings, directions of air flow, and attached pressure sensor device.

the output tube is different than the pressure sensor input, a rubber sleeve can be used to connect the tubes.

One concern addressed during the semester was the possibility of significant pressure losses due to friction because of the small diameter IV sized tubing to which the pressure sensor is connected. In order to investigate whether this would have an effect on the design, the pressure loss was calculated using basic fluid mechanics principles. Assuming that the device was in laminar flow (because of the T connection) and assuming that there was one foot of IV sized tubing between the ventilator and the device (which is a generous estimate) it was found that the pressure loss due to friction would be 0.015mmHg. This pressure loss would be essentially negligible for the pressure calculations needed. The equation used in order to determine this value was the equation for a laminar flow pressure drop: $\Delta P = \frac{32 l \bar{v} \mu}{D^2}$ (6). In this equation, l is the length of the tubing, D is the diameter of the tubing, μ is the viscosity of the fluid and v is the average velocity of the fluid. All of the variables except v could be taken from a reference book (6) but the average velocity had to be estimated based on tidal volume and average breaths at rest.

Calibration of the device in a hospital setting will be another topic addressed in the next semester, as it is necessary in order to provide an accurate pressure measurement, and therefore a safe environment for patients. In order to accommodate calibration of the instrument's offset voltage, a potentiometer was added to the circuit. Adjustments can be made

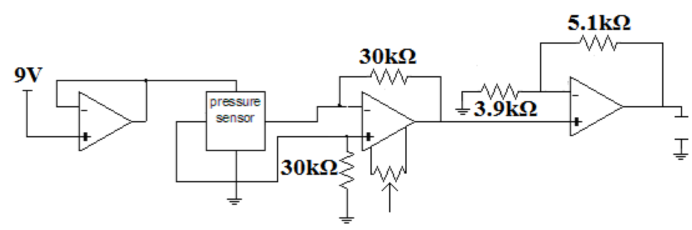

Figure 9 – Final circuit design of this semester, including potentiometer and the non-inverting amplifier.

using a small instrument, and the point of zero pressure can be set to read zero on the digital display. It will also be necessary to add a gain potentiometer next semester in order to ensure that the voltages throughout the range of input pressures are accurate. Figure 9 shows this semester's final circuit design including the potentiometer and the non-inverting amplifier additions. One of the focal points of this semester's work was the addition of the microcontroller, used to control the digital display of the device. Originally, there were two options for the type of microcontroller that could be used: a Peripheral Integrated Circuit (PIC) by Microchip or a Mixed Signal Microcontroller by Texas Instruments. However, we were able to get assistance from another student with programming experience on Atmel processors. Furthermore, he had access to programming pads for preliminary testing. The Atmel Tiny26 chip was chosen as the microcontroller. The chip contained 2 kilobytes of programmable memory, 10 bit A/D converting, and 16 programmable I/O pins (20 pins total). It was programmed using

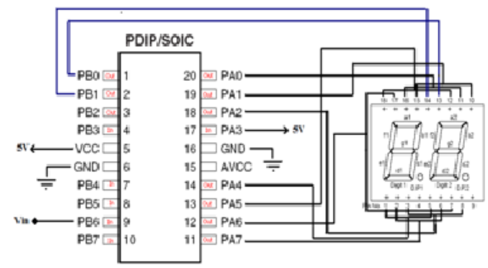

Figure 10 – Atmel Tiny26 microcontroller schematic showing pins connected to LED display.

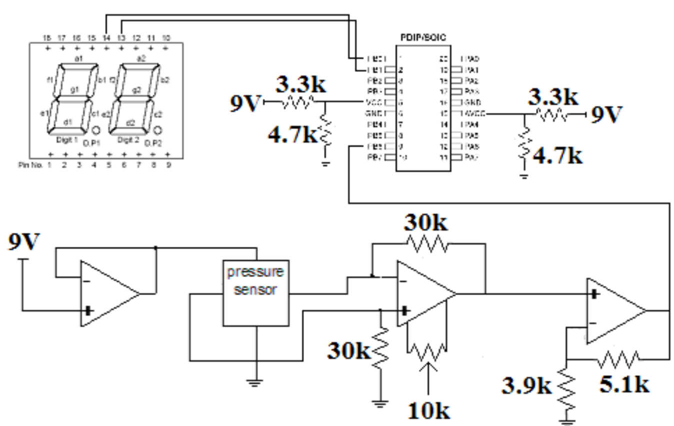

Figure 11 – Full system schematic of final prototype of the semester, containing full circuit, microcontroller, and digital display.

CodeVisonAVR, a C language based compiler, and an STK500 programming pad which was purchased by the BME department. Voltage was sampled at a rate of 15 times per second and then multiplied by a ratio value that was derived from prior testing. The value was then displayed as 2-digit values corresponding to the pressure input in terms of mmHg on the LEDs. Figure 10 shows the final configuration of the microcontroller with the LED. A full documentation of the code can be seen in the Appendix. Figure 11 shows the entire schematic, containing the full circuit, the microcontroller, and the LED display.

Budget

Since the cost of the device has been the main concern in this project, the budget has been a very important part of the design process. We estimated the budget necessary for this device based on production of three units and of 500 units. Three units were chosen because PCBs must be ordered a minimum of three at a time. The PCB unit cost dropped considerably when the quantity was increased to 500 units. Also, part of the price calculated in the three-unit cost consists of shipping charges. Those costs would be essentially negligible when parts are ordered in mass quantity. A detailed budget can be seen in the Appendix.

Testing

Several different tests were conducted in various stages of the project this semester. First, the relationship between initial pressure input and voltage output of the circuit needed to be determined. This was done using a KAL84 pressure calibrator from Professor Chesler's lab. The relationship was linear and the results mimicked those of last semester.

After the sensitivity of the circuit was improved, it was necessary to test the relationship between input pressure and voltage output again. The test was conducted in the same manner as the

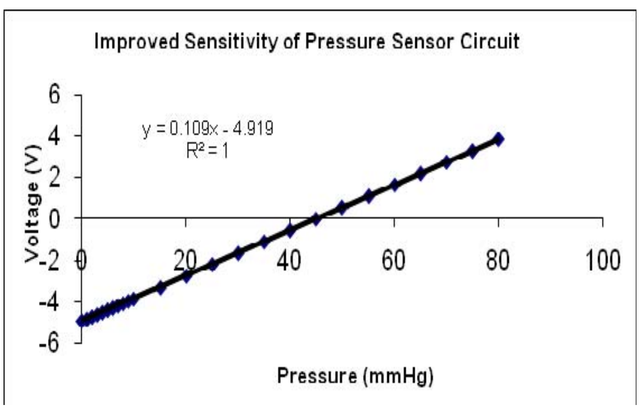

Figure 12 – Results of testing after the sensitivity of the circuit was increased for an output voltage range of 10 Volts (-5 to 5 V).

first. The results of this test are shown in Figure 12.

We also wanted to test a variety of gas compositions in our device to find out if the density of the gases resulted in different voltage outputs of the circuit. This testing was conducted at the UW Hospital with the help of Willie Backes, the respiratory therapy equipment manager. Medical air, which is composed of 21% O_2 and 79% N_2, and 100% O_2 were tested, and the results are shown in Figure 13. Clearly, the composition and density of the gas has negligible effect on the output of the circuit, since the two data sets are almost identical.

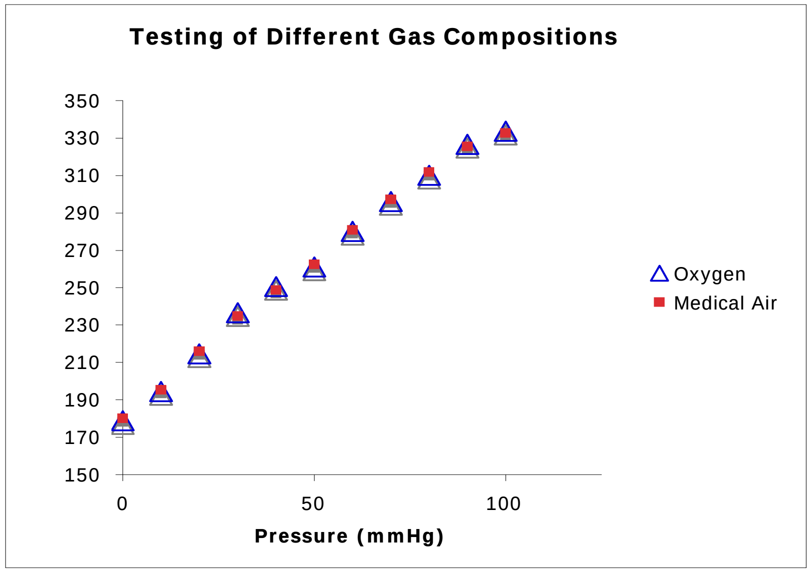

Figure 13 – Results of testing pressure-transduction circuit with two different gasses, oxygen and medical air.

The final step of testing took place after the microcontroller was programmed and the digital display showed values corresponding to input pressures. This consisted of inputting a known pressure and recording the numerical readout of the digital display. The results can be

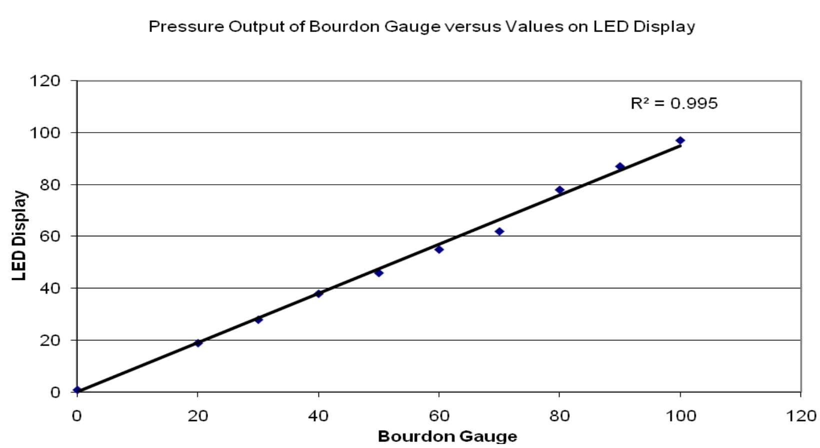

Figure 14- Pressure output of Bourdon gauge vs. LED display

seen in Figure 14. They show that the digital display and input pressure correspond in mmHg.

Future Work

Next semester we plan to focus on transferring the circuit to a printed circuit board (PCB) following our design. However, even before that, we will incorporate the addition of a second calibrating potentiometer. We will also need to address battery power, specifically the type and quantity. We hope to complete testing regarding their life expectancy as well. We will also continue testing the current prototype along with constructing and testing at least one additional prototype before transferring to a PCB. Testing will be continued, using anesthesia and ventilation machines. Making the device as small and compact as possible in order to maximize the ease of use will also be a main focus of next semester. Overall, we would like to see our device mass-produced for distribution in a developing country.

References

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Appendix

Microprocessor Code

```
#include "PressureDefine.h"
#include <delay.h>

void LEDDisplay(unsigned char Number)
{
    int Tens, Ones;

    Tens = (Number) / 10;
    Ones = Number - 10 * Tens;

    /********
      Ones
    *********/
    ONESON;
    TENSOFF;


    if (Ones == 0)
            {
            PORTA = ZERO;
            }
    if (Ones == 1)
            {
            PORTA = ONE;
            }
    if (Ones == 2)
            {
            PORTA = TWO;
            }
    if (Ones == 3)
            {
            PORTA = THREE;
            }
    if (Ones == 4)
            {
            PORTA = FOUR;
            }
    if (Ones == 5)
            {
            PORTA = FIVE;
            }
    if (Ones == 6)
            {
            PORTA = SIX;
            }
    if (Ones == 7)
            {
            PORTA = SEVEN;
            }
    if (Ones == 8)
            {
```

```
        PORTA = EIGHT;
        }
if (Ones == 9)
        {
        PORTA = NINE;
        }
delay_ms(2);

 /********
   Tens
 ********/
 ONESOFF;
 TENSON;


 if (Tens == 0)
        {
        PORTA = ZERO;
        }
 if (Tens == 0 && Number < 10)
        {
        PORTA = 0xFF;
        }
 if (Tens == 1)
        {
        PORTA = ONE;
        }
 if (Tens == 2)
        {
        PORTA = TWO;
        }
 if (Tens == 3)
        {
        PORTA = THREE;
        }
 if (Tens == 4)
        {
        PORTA = FOUR;
        }
 if (Tens == 5)
        {
        PORTA = FIVE;
        }
 if (Tens == 6)
        {
        PORTA = SIX;
        }
 if (Tens == 7)
        {
        PORTA = SEVEN;
        }
 if (Tens == 8)
        {
        PORTA = EIGHT;
        }
 if (Tens == 9)
```

```
            {
            PORTA = NINE;
            }

       delay_ms(2);


return;
}

------------------------------------------------------------------------------------------------------------------

#include <tiny26.h>
#include "PressureDefine.h"
#include "PressureFunctions.h"

unsigned char ADCResult;
float Voltage;
int Pressure;


#define ADC_VREF_TYPE 0x60

// Read the 8 most significant bits
// of the AD conversion result
unsigned char read_adc(unsigned char adc_input)
{
ADMUX=adc_input|ADC_VREF_TYPE;
// Start the AD conversion
ADCSR|=0x40;
// Wait for the AD conversion to complete
while ((ADCSR & 0x10)==0);
ADCSR|=0x10;
return ADCH;
}


// Timer 1 output compare A interrupt service routine
interrupt [TIM1_CMP1A] void timer1_compa_isr(void)
{
       ADCResult = read_adc(9);
       Voltage = (ADCResult)*0.0196;
       Pressure = 24.27 * Voltage; //24.27 derived from testing
}



// Declare your global variables here


void main(void)
{
// Declare your local variables here

// Input/Output Ports initialization
// Port A initialization
```

```
// Func7=In Func6=In Func5=In Func4=In Func3=In Func2=In Func1=In Func0=In
// State7=T State6=T State5=T State4=T State3=T State2=T State1=T State0=T
PORTA=0x00;
DDRA=0b11110111;

// Port B initialization
// Func7=In Func6=In Func5=In Func4=In Func3=In Func2=In Func1=In Func0=In
// State7=T State6=T State5=T State4=T State3=T State2=T State1=T State0=T
PORTB=0x00;
DDRB=0b00000111;

// Timer/Counter 0 initialization
// Clock source: System Clock
// Clock value: Timer 0 Stopped
TCCR0=0x00;
TCNT0=0x00;

// Timer/Counter 1 initialization
// Clock source: System Clock
// Clock value: 0.977 kHz
// Mode: CTC top=OCR1C
// OC1A output: Disconnected
// OC1B output: Disconnected
// Timer 1 Overflow Interrupt: Off
// Compare A Match Interrupt: On
// Compare B Match Interrupt: Off
PLLCSR=0x00;
TCCR1A=0x00;
TCCR1B=0x8E;
TCNT1=0x00;
OCR1A=0x00;
OCR1B=0x00;
OCR1C=0xC3;

// External Interrupt(s) initialization
// INT0: Off
// Interrupt on any change on pins PA3, PA6, PA7 and PB4-7: Off
// Interrupt on any change on pins PB0-3: Off
GIMSK=0x00;
MCUCR=0x00;

// Timer(s)/Counter(s) Interrupt(s) initialization
TIMSK=0x40;

// Universal Serial Interface initialization
// Mode: Disabled
// Clock source: Register & Counter=no clk.
// USI Counter Overflow Interrupt: Off
USICR=0x00;

// Analog Comparator initialization
// Analog Comparator: Off
ACSR=0x80;

// ADC initialization
// ADC Clock frequency: 115.200 kHz
```

```
// ADC Voltage Reference: AREF pin
// Only the 8 most significant bits of
// the AD conversion result are used
ADMUX=ADC_VREF_TYPE;
ADCSR=0x87;

// Global enable interrupts
#asm("sei")


while (1)
    {
    LEDDisplay(Pressure);
    };
}
```

Budget

Part Description	Seller	Part Number	Unit Cost (per 3)	Total Cost (Unit x3)	Unit cost (per 500)	Total Cost (Unit x500)
Pressure Sensor	Electronic Goldmine	G15473	$ 3.91	$ 11.73	$ 1.51	$ 755.00
Battery Holder	All Electronics	BST-3	$ 4.00	$ 12.00	$ 0.32	$ 160.00
PCB	Express PCB		$ 17.00	$ 51.00	$ 1.60	$ 800.00
Display	Electronic Goldmine	G4031	$ 2.66	$ 7.98	$ 0.24	$ 120.00
Microcontroller			$ 3.00	$ 9.00	$ 1.31	$ 655.00
Resistors (four per unit)	Digikey	Varies Based on Size	$ 0.80	$ 2.40	$ 0.12	$ 60.00
Operational Amplifier (3 per unit)	Allied Electronics	LM358ADE4	$ 2.43	$ 7.29	$ 0.33	$ 165.00
Potentiometer	All Electronics	TPR-10K	$ 1.60	$ 4.80	$ 0.16	$ 80.00
Luer Connectors (male and female)	Cole Parmer	EW-45500-00	$ 0.84	$ 2.52	$ 0.51	$ 255.00
Button	All Electronics	PB-151	$ 0.40	$ 1.20	$ 0.25	$ 125.00
Total			**$ 36.64**	**$ 109.92**	**$ 6.35**	**$ 3,175.00**