DESIGN OF A PRESSURE AND FLOW MEASURING EQUIPMENT FOR MEDICAL USE

by

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ABSTRACT

DESIGN OF A PRESSURE AND FLOW MEASURING EQUIPMENT FOR MEDICAL USE

Accuracy and reliability of measurements play a very important role in healthcare regardless of the field which may be therapy, diagnosis or life support. Medical equipment, on the other hand, is subject to failure due to mechanical damage, user abuse, component failure, aging or some other reasons and may cause undesired or irreversible results. Therefore, periodic inspections of medical equipment are essential to ensure safe and reliable use of medical equipment.

In this thesis, blood pressure and gas flow measurement in medical fields have been focused on and a "Pressure and Flow Measuring Equipment", has been developed. This prototype instrument is, presently, capable of testing non-invasive blood pressure measuring apparatus and bedside oxygen flow meters with a high level of accuracy, hence providing a valuable inspection tool for preventive maintenance.

The accuracy of the instrument has been tested against calibrated reference test equipment. The results obtained show that the prototype instrument developed is able to measure pressure in 100% agreement with the reference while it can measure gas-flow within 95% confidence interval.

Keywords: Blood Pressure, Gas-flow in Medicine, Pressure Measurement, Flow Measurement, Preventive Maintenance in Medicine.

ÖZET

TIBBİ KULLANIM AMAÇLI OLARAK BİR BASINÇ VE AKIŞ ÖLÇER TASARIMI

Sağlık ile ilgili tüm ölçümlerde, kesinlik ve güvenilirlik, ölçümün teşhis, tedavi ya da hayat desteği verici olmasına bakılmaksızın büyük önem taşımaktadır. Medikal cihazlar; mekanik arızalar, kullanıcı hatası, parça hatası ya da eskime vs. gibi nedenlerle bozulabilir, istenmeyen ve geri dönüşü olmayan hatalara neden olabilir. Bu nedenle medikal cihazların programlı olarak kontrolü ölçümlerin güvenilirliğini, cihazların emniyetini ve muhtemel arızaların önlenmesini sağlar.

Bu çalışmada, kan basıncı ve gaz akışı ölçümüne odaklanılmış olup, "Basınç ve Akış Ölçer" adı verilen bir sistem geliştirilmiştir. Bu prototip cihaz, indirekt kan basıncı ölçen aletlerin ve hastabaşı akış ölçerlerinin testinde kullanılmak üzere tasarlanmıştır.

Sistemin doğruluğu referans olarak alınan kalibre edilmiş ekipmanlar ile test edilmiş, elde edilen sonuçlar tasarlanan cihazın basınç ölçümünü referans cihazla %100 uyumlu, akış ölçümünü de %95 güvenilirlik sınırları içinde okuduğunu göstermiştir.

Anahtar Sözcükler: Kan Basıncı, Tıpta Gaz Akışı, Basınç Ölçümü, Akış Ölçümü, Tıpta Koruyucu Bakım.

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LIST OF SYMBOLS

A	Area
C	Discharge coefficient
d	Diameter
D	Diameter of pipe
F	Force
P	Pressure
Q	Mass flow rate
β	Diameter ratio
Δp	Differential pressure
ρ	Mass density

LIST OF ABBREVIATIONS

ADC	Analog to Digital Conversion/Converter
CCU	Critical Care Unit
CVP	Central Venous Pressure
FA	Flow Analyzer
GPIO	General Purpose Input Output
ICU	Intensive Care Unit
I/O	Input/Output
LCD	Liquid Crystal Display
PCB	Printed Circuit Board
PM	Preventive Maintenance
PSoC®	Programmable System on Chip
ROM	Read Only Memory
RS	Register Select
R/W	Read/Write

1. INTRODUCTION

1.1 Motivation and Objectives

Medical equipment is defined as any fixed or portable non-drug item or apparatus used for the diagnosis, treatment, monitoring, and direct care of patients [1]. Research and development phase, clinical testing, approval and production of a medical equipment imply a complex, expensive and lengthy process.

Although extra precautions are taken, medical equipments can occasionally fail with the causes of failure including wear, mechanical damage, user abuse, component failure, and aging. Signs of these failures may not always be apparent to clinical staff. The breakdown of a medical instrument in service may cause undesired and irreversible results since it may be used in life-support situations, either contributing directly to therapy or indirectly by providing information on the patient's health status. Scheduled inspections help to ensure the safety and efficacy of medical equipment and help to prevent failure appearing in service, and to predict future failures.

An equipment inspection procedure, called preventive maintenance (PM), should identify whether equipment is functioning properly in order to prevent patient injuries caused by the use of malfunctioning equipment. As used here, an injury is broadly defined as any adverse patient event related to the equipment, whether it is caused by directly physical actions (e.g. electrical shock, explosion) [2] or by technical failure or misusage of the device (e.g. measurement error, energy delivery error, operational error, technical or other applications problems, inappropriate installation, usage and application) [2, 3].

The primary objective of the maintenance program is to avert predictable and preventable device failure. Most of these failures are time-related. The most common one is calibration drift and termed "systematic" failure [4]. The other time-related failures may be count as wear and tear of components, accumulation of dust, and etc. On the other hand, some failure is not predictable and may be caused by user abuse, manufacturing defects, design flaws, and etc. For example, a device may fail due to some random occurrence the day immediately following a full maintenance inspection. So, these unpredictable random failures may not be preventable with periodic inspections.

Accuracy, which is defined as the closeness of the agreement between the result of a measurement and the true value of the measurand [5], is the common objective of maintenance programs and used to demonstrate quality and validity. It is a measure of the total error regardless of the type or source of the error, and usually expressed as percentage. Heart rate measurement, breath rate delivery, infusion pump rate, blood pressure measurement, or gas flow measurement constitute important examples of typical measurements in medicine required accuracy.

In this thesis, pressure and gas flow measurement have been focused on and an instrumentation system, called "Pressure and Flow Meter", has been developed. Since pressure and gas flow measurements are very important in medical applications, the accuracy of each should be assured as well as their reliability. The objective of this thesis is to design a prototype instrument to test non-invasive blood pressure measuring apparatus and bedside flow meters with a high level of accuracy in order to provide a secure preventive maintenance.

1.2 Outline of the Thesis

Chapter 1 introduces the motivation and objectives behind this study and presents a description and outline of the thesis. In Chapter 2, a brief overview of pressure and flow measurements in medical fields is given. The Pressure and Flow Analyzer developed in the thesis is described in Chapter 3. Chapter 4 gives and discusses the measurement results obtained at the end of development cycle. The conclusions and recommendations for future work are presented in Chapter 5.

2. PRESSURE AND FLOW MEASUREMENTS IN MEDICAL FIELDS

In this chapter, we will briefly explain the principles of pressure and flow measurements in clinical use. Since the prototype analyzer has been intended to test non-invasive blood pressure measuring apparatus and bedside oxygen flow meters, physiological and physical principles behind these instruments are also discussed.

2.1 Pressure Measurement

Pressure is defined as the force per unit area which may be exerted by a fluid (liquid or gas) [6]. Only the component of force normal to the surface needs to be considered for the determination of pressure and the basic pressure formula is given by [7]

$$P = \frac{F}{A} \tag{2.1}$$

where P is the pressure in newtons per square meter (N/m^2) or pascals (Pa); F is the applied force in newtons (N); and A is the area in square meters (m^2). It should be noted that pressure is not defined as a vector quantity and is, therefore, non-directional.

Three types of pressure measurement are commonly performed:

- *Absolute Pressure* represents the pressure difference between the point of measurement and a perfect vacuum where pressure is zero.
- *Gage Pressure* represents the pressure difference between the point of measurement and the ambient.
- *Differential Pressure* represents the pressure difference between two points, one of which is chosen to be the reference.

The air forming the atmosphere exerts a pressure on the surface of the earth which is termed as atmospheric pressure. This pressure at sea level is usually called 1 atmosphere, or 1 atm.

Gage pressures are usually given in mmHg above or below atmospheric pressure. The zero reference in gage pressure measurements, therefore, is a pressure of 1 atm (= 760 mmHg). Gage pressures below 1 atm are said to be measured on a *vacuum gage*. Most gage pressure measurement apparatus, including most of the electronic instruments, will measure both positive pressures (above 1 atm) and negative pressures, i.e. vacuum gages. Even though real atmospheric pressure varies from one place to another and moment to moment, zero adjustment can be established at each measurement by setting the zero scale with the pressure-meter open to atmospheric pressure.

2.1.1 Pressure Sensors

Since pressure is defined as the force per unit area, the most direct way of measuring pressure is to isolate an area on an elastic mechanical element for the force to act on. The deformation of the sensing element produces displacements and strains that can be precisely sensed to give a calibrated measurement of the pressure. This forms the basis for essentially all commercially available pressure sensors today [7].

It is, therefore, possible to define a pressure sensor as a device which converts the mechanical energy by an applied pressure into electrical energy. The magnitude of the electrical signal, which is the output of the sensor, is proportional to the magnitude of the applied pressure.

The operation of pressure sensors require an electrical excitation signal. This means that a pressure transducer has two inputs, i.e. the excitation signal and the applied pressure, and one output, i.e. an electrical signal dependent on both its inputs. The sensitivity of a sensor is defined as the change in its output signal caused by a unit change in each of its input signals and is usually expressed in units of μV output

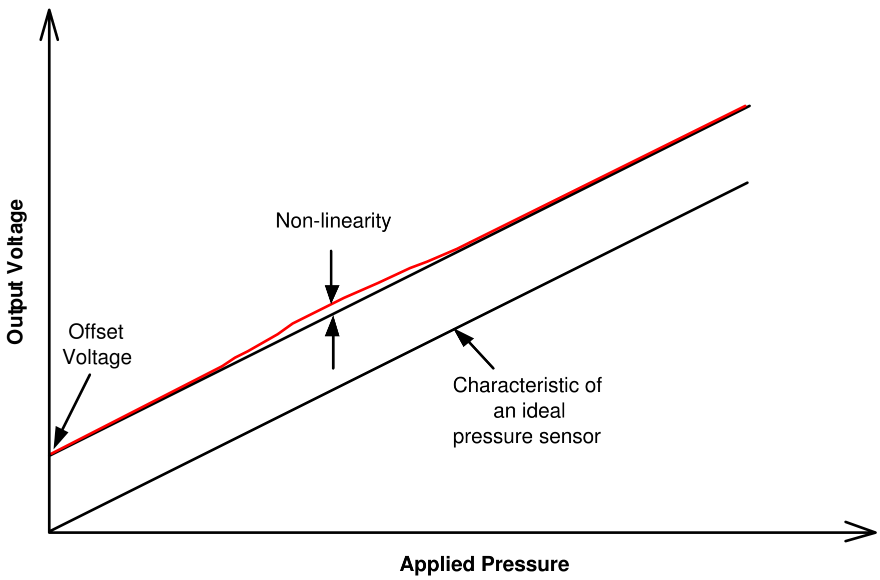


Figure 2.1 Characteristic of a pressure sensor

signal per V of excitation per cmHg of applied pressure.

A typical pressure sensor characteristic curve is illustrated in Figure 2.1. An ideal sensor would have a linear response to changes in excitation voltage or applied pressure and an output signal of 0 with no applied pressure as shown in the figure. In practice, all sensors have some non-zero output with no applied pressure. This output at zero pressure is known as the offset voltage. As long as this offset is constant, it will introduce no error in measurement because the signal conditioning circuits can generally be employed to produce a zero output with an input equal to the sensor-offset voltage [8].

2.1.2 Physiological Pressures

The measurement of physiological fluid pressures is of interest to both biomedical researchers and clinicians. The most common pressure measurement is *arterial blood pressure*, which is almost routinely monitored by electronic instruments in Intensive Care Unit (ICU), Critical Care Unit (CCU) and other critical care areas. Also of interest are *the central venous pressure* (CVP), *intracardiac blood pressure*, special

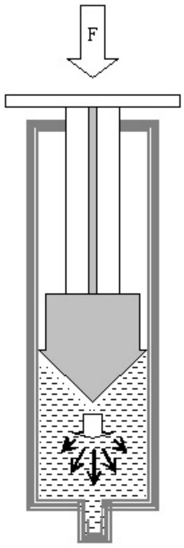


Figure 2.2 Sectioned view of a syringe. A pressure exerted by the plunger is transmitted through the fluid to all parts of the system.

pressures in *the pulmonary artery*, *spinal fluid pressures*, and *intraventricular (brain) pressures* [7].

Pressure in a closed system obeys a physical law known as Pascal's Principle which states that pressure applied to an enclosed fluid is transmitted undiminished to every portion of the fluid and the walls of the containing vessel [9].

For example, if a pressure is applied to the stoppered syringe in Figure 2.2, then the same pressure is felt throughout the interior of the syringe. Changing the pressure applied to the plunger causes the same changes to be reflected at every point inside of the syringe.

Pressure is exerted in the human circulatory system by the force created by the pumping heart transmitted through the fluid (i.e. blood) against the vessel walls. The circulatory system, on the other hand, regulates blood pressure by constricting and dilating vessels, which causes changes in the vessel surface area by changing the vessel diameter. The pressure, as a result, is never constant, and the measurements always assume an average value.

Pressures in the human circulatory system are measured against atmospheric pressure and are gage pressures. Gage pressure is used because it is more easily referenced at atmospheric pressure, and can be easily recalibrated at each use, and the absolute pressure confers no special advantage as to information content.

2.1.2.1 Blood Pressure Measurement. Blood pressure, as one of the physiological variables that can be quite readily measured, is considered a good indicator of the status of the cardiovascular system. A history of blood pressure measurements has saved many people from an untimely death by providing warnings of dangerously high blood pressure (hypertension) in time to provide treatment.

The earliest recorded attempt at the quantitative measurement of arterial blood pressure was performed in 1773 by Stephen Hales, English scientist [8]. He used an open-ended vertical tube inserted directly into an artery in the neck of an unanesthetized horse, observing the height to which blood rose in the tube. From that beginning, many investigators developed and improved instruments for the measurement of blood pressure. By the beginning of the twentieth century, instruments were reported which were capable of accurately measuring the instantaneous blood pressure and of following its variations within each cardiac cycle. However, only since about 1950 have techniques been in use, which allow the measurement of blood pressure with ease and safety at almost any point in the cardiovascular system of a person.

In routine clinical tests, blood pressure is usually measured by means of an indirect method using a sphygmomanometer (from the Greek word, sphygmos, meaning pulse). This method is easy to use and can be automated. It has, however, certain disadvantages in that it does not provide a continuous recording of pressure variations and its practical repetition rate is limited. Furthermore, only systolic and diastolic arterial pressure readings can be obtained, with no indication of the details of the pressure waveform. The indirect method is also somewhat subjective, and often fails when the blood pressure is very low (as would be the case when a patient is in shock).

Methods for direct blood pressure measurement, on the other hand, provide a continuous readout or recording of the blood pressure waveform and are considerably more accurate than the indirect method. They require, however, that a blood vessel be punctured in order to introduce the sensor. This limits their use to those cases in which the condition of the patient warrants invasion of the vascular system [10].

2.2 Flow Measurement

Flow measurement is performed in almost every sector of industry. Whether you are filling up a car with petrol or curious to know how much air is passing through the piping system, a flow meter is required.

Flow can be defined as the amount of a substance, usually a fluid, passing through a point per unit time (e.g., kilogram per second, liter per minute, etc.). Fluid flow measurements, expressed in terms of either volume flow rate or mass flow rate, are used in many applications, such as industrial process control, gas-pipeline systems [11], or ventilation systems in hospitals.

The field of flow measurement spans a large variety of sensing principles and a large number of different flow sensors. Commonly used flow meters fall into one of four categories: rotating-vane, ultrasonic, thermal convection, and differential pressure flow meters.

In this thesis, we have employed the differential pressure type of flow measurement techniques to detect gas-flow rates specific especially to the respiratory system.

2.2.1 Differential Pressure Flow Meters

Bernoulli's principle [6] states that in fluid flow, an increase in velocity occurs simultaneously with decrease in pressure. From the relationship between pressure difference and volume flow rate through a system, measurement of the difference in pressure yields an estimate of flow. Flow meters based on this idea have incorporated several mechanisms to establish the pressure drop and flow. In practice, constriction in a pipeline consists of some devices such as an orifice plate, a nozzle, or a venturi tube. The orifice plate appliance is used in this work.

According to Bernoulli's principle, if an annular restriction (refer as to differential producer) is placed in a pipeline, then the velocity of the fluid through the restriction is increased in order to retain continuity. The increase in velocity at the restriction causes the static pressure to decrease at this section, and a pressure difference is created across the element. The difference between the pressure upstream and pressure downstream of this obstruction is related to the rate of fluid flowing through the restriction and therefore through the pipe. This pressure difference also depends upon the ratio of the area of the construction, and a decrease in this ratio produces a corresponding increase in the pressure difference to be measured.

In order to measure this pressure drop across the obstruction, a differential pressure transducer is needed. Apart from the differential producer, correct selection and installation of the differential pressure transducer plays an important part in determining the accuracy of the flow rate measurement.

2.2.1.1 Orifice Plate. The orifice plate is the simplest and cheapest type of differential pressure flow meters. It is simply a plate with a hole of specified size and position cut in it, which can be clamped between flanges in a pipeline. Through the hole the fluid can pass and so affects a local increase in velocity. (Figure 2.3).

Over 40% of all liquid, gas, and steam measurements made in industry are still accomplished using the orifice plate type of *differential pressure flow meters* [6]. The flow rate of an incompressible fluid through this type of flow meter can be expressed as [12]

$$Q = \frac{C}{\sqrt{1-\beta^4}} \frac{\pi d^2}{4} \sqrt{2\rho \Delta p} \tag{2.2}$$

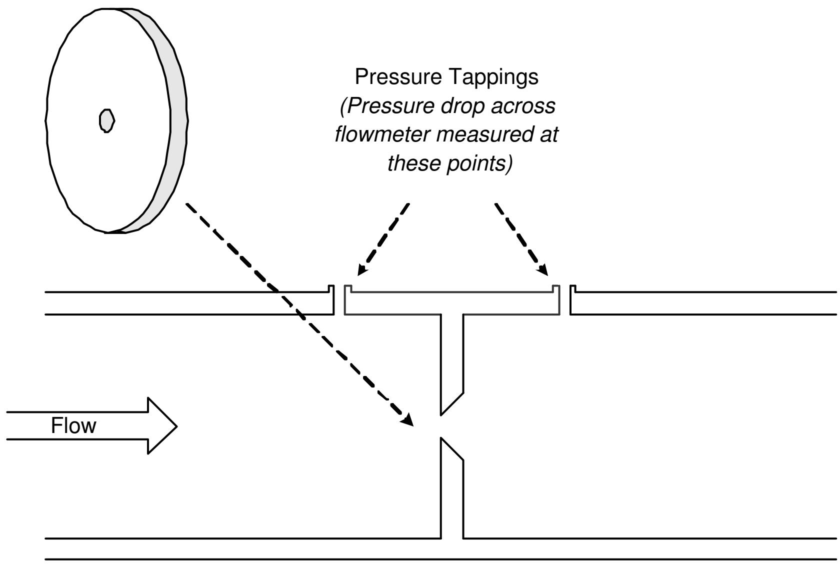


Figure 2.3 Flow through the orifice plate

where Q is the actual mass flow rate in kg/s; C is the discharge coefficient; d is the diameter of orifice in m; β is the diameter ratio which is equal to d/D where D is the diameter of pipe in m; ρ is the mass density of flowing fluid in kg/m^3; and Δp is the differential pressure across the orifice in Pa.

2.2.2 Flow Measurement in the Respiratory System

Respiratory system takes part in the process in the lungs that are involved in the exchange of gases between the blood and the atmosphere. The process by which the air enters the lungs is called *ventilation* and the method by which the blood fills the blood vessels in the lungs is called *perfusion* [13]. Measurements of ventilation and perfusion related variables enables physician to perform clinically relevant tasks; assess the functional status of the respiratory system (lungs, airways, and chest wall) and intervene in its function.

Even though a number of variables are included in the gas transport and the mechanics of the respiratory system, only a very limited subset can be measured directly. This includes volume flow of gas through the mouth and nose.

When the lungs change volume during breathing, a mass of gas is transported through the airway opening by convective flow. The volume-flow rate and the time integral of volume flow rate are used to estimate rate of change of lung volume and changes of lung volume, respectively. Volume flow rate equals the mass flow rate divided by the density of the gas at the measurement site. The instruments, used the measure the volume flow rate, are referred to as volume flow meters.

Even though breathing movements are cyclic by nature and involve alternating gas flow, some tests of pulmonary function require the measurement of flow in only one direction.

2.2.2.1 Requirements for Respiratory Gas Flow Measurements. In respiratory experiments, especially those involving measurement of breathed gas, the usual practice is to have the entire flow stream pass through or into the instrument. Any pressure imposed at the airway during measurements must be withstood by the sensor without damage, distortion, or leakage. Also the device should not obstruct breathing or produce a back pressure during flow that might affect respiratory performance.

As with any instrument, the gas flow measuring sensor must have a stable baseline (reference output) and sensitivity so that measurements are accurate. However, changes in composition and temperature of gas can affect the calibration factors of various flow meters. The measurement procedure, also, must not alter inspired air by adding excessive heat or toxic substances.

3. DESIGN OF THE PRESSURE AND FLOW METER

In this chapter, we begin with the general layout of the proposed instrument, then we proceed further including all aspects of the design. The design at the component level is discussed in detail. In addition, calibration tasks of the critical components and performance obtained during the calibrations are presented in related sections.

3.1 General Layout of the Pressure and Flow Meter

The block diagram of the Pressure and Flow Meter is shown in Figure 3.1 and the definition of each block is briefly given below.

1. **Sensors:** Two pressure sensors convert applied pressure to an electrical signal. They provide a differential voltage in the milli-volt ranges proportional to the input pressure.

2. **Amplification:** Amplification is necessary to bring the sensors' output to an appropriate level for signal conditioning while performing the function of interfacing to micro-controller. Amplification is also used to amplify sensors' differential outputs for flow measurement.

3. **Offset Adjustment:** Because the pressure sensors have non-zero outputs with no applied pressure, offset adjustment is needed to obtain zero-volt output at zero pressure. This has been readily achieved by multi-turn potentiometers.

4. **Micro-Controller and Control Buttons:** Sensed output voltages must be processed before being displayed, so it is converted to a form that the operator can perceive. Analog to Digital Conversion (ADC), mathematical operations, unit conversions, control mechanisms and display routines are performed in this phase by a software implementation. Control mechanism has been implemented by two push-button switches. They change the mode of the micro-controller,

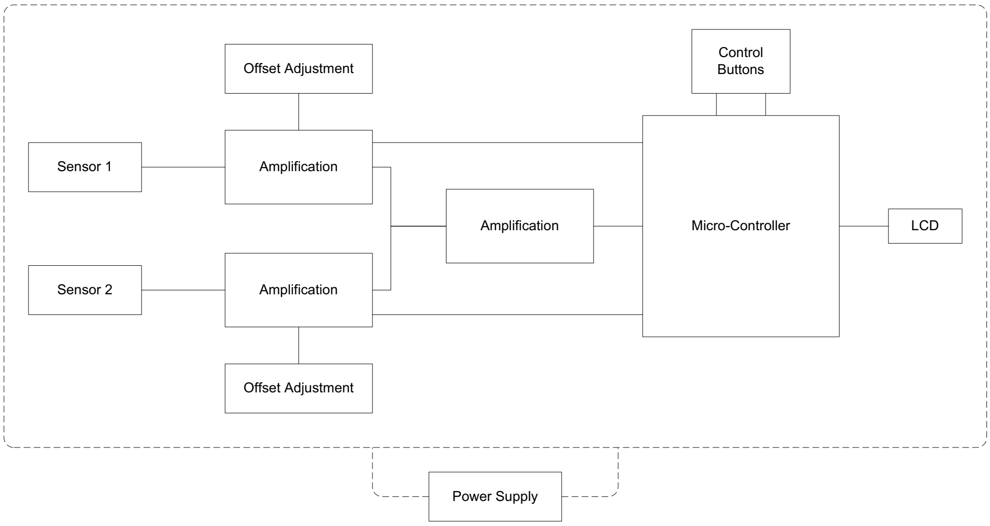


Figure 3.1 Block diagram of the Pressure and Flow Meter

and decide the operations, conversions and display configuration according to measurement type and material used.

5. **LCD:** A simple 2-line, 16-character LCD module with 14-pin has been used to display the measurement results.

6. **Power Supply:** Power supply has been designed to provide appropriate functioning of all components in the equipment.

3.2 Pressure Sensor

Since pressure measurement is the main part of this work, the system has to start with a pressure sensor. Making a decision about which pressure sensor should be used is a demanding process since there are so many trademarks with many different types of sensors in the market. First of all the requirements should be specified clearly.

In our design, measurement type should be the gage pressure measurement. Gage pressure is measured relative to the local ambient pressure and defined as the difference between the measured pressure and atmospheric pressure. In other words it can be said that, it is a form of differential pressure measurement in which atmospheric pressure is used as the reference. Blood pressure is a significant example of this type of pressure measurement. Because gage pressure is measured relative to local atmospheric pressure, it can be either positive or negative. Negative gage pressures are defined as vacuum.

$\pm$750 mmHg is enough pressure range for a pressure sensor used in testing a medical equipment. Pressure range for most pressure sensors is usually presented in psi unit. 750 mmHg is equivalent to 14.5 psi (1 mmHg = 1.934 x 10^{-2}psi). So the actual range can be chosen as $\pm$15 psi and a sensor covering this range can be readily found in the market.

Honeywell® XPX pressure sensor series satisfy our needs, which involve piezoresistive pressure sensors that use silicon micro machined chips mounted on a ceramic and they are protected with a plastic cap. They provide pressure measurement in differential, gage and absolute type with millivolt output. They can be safely used in medical applications, in applications requiring small size and in vacuum measurement. They are also favorable for flow measurement since they accommodate many gases that are used in medical applications.

XPX15GFS from Honeywell® XPX series is a good choice with its measurement type, pressure range and tube arrangement, and it is also suited for printed circuit board (PCB) mounting. Its tube dimensions match the tube dimensions of a typical sphygmomanometer. Technical drawing with mounting dimensions are shown in Figure 3.2.

As mentioned above, XPX15GFS is a silicon micro machined pressure sensor. This refers to a class of pressure sensors that employ integrated circuit batch processing techniques to realize a thinned-out diaphragm sensing element on a silicon chip. Over the past 20 years, silicon micro machined pressure sensors have gradually replaced their mechanical counterparts and have captured over 80% of the pressure sensor market. There are several unique advantages that silicon offers. Silicon is an ideal mechanical

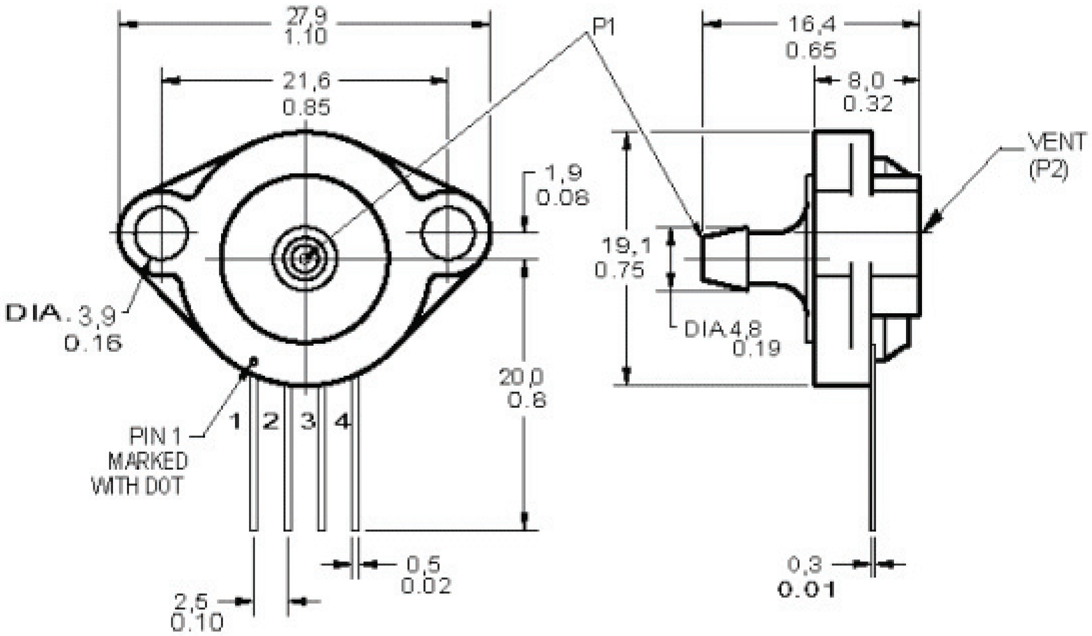


Figure 3.2 Technical drawing of the pressure transducer

material that does not display any hysteresis or yield and is elastic up to the fracture limit. Silicon has been widely used in integrated circuit manufacturing for which reliable batch for fabrication technology and high-precision dimension control techniques have been well developed. A typical silicon wafer yields hundreds of identical pressure sensor chips at very low cost. All these factors contribute to the success of silicon micro machined pressure sensors [6].

3.2.1 Pressure Sensing Principles

The sensing element of the pressure sensor consists of a four nearly identical piezoresistors buried in the surface of thin circular silicon diaphragm. A pressure or force causes the thin diaphragm to flex, inducing a stress or strain in the diaphragm and also in the buried resistors. The resistor values will change depending on the amount of strain they undergo, which depends on the amount of pressure or force applied to the diaphragm. Therefore, a change in pressure (mechanical input) is converted to a change in resistance (electrical output). For a pressure or force applied to the diaphragm using a Wheatstone bridge arrangement, the resistors can be theoretically

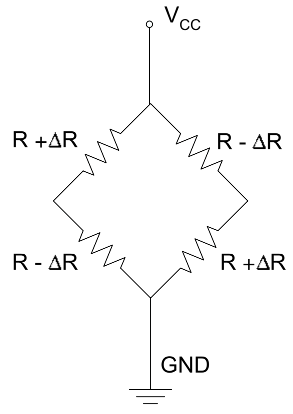


Figure 3.3 Wheatstone Bridge configuration

represented as shown in Figure 3.3. $R \pm \Delta\mathrm{R}$ represents the actual resistor values at the applied pressure or force. R represents the resistor value for the undeflected diaphragm where all four resistors are nearly equal in value. This is the balanced condition at atmospheric pressure. ΔR represents the change in resistance due to an applied pressure or force. All four resistors will change by approximately the same amount. Two of resistors will increase while the other two decrease depending on their orientation with respect to the crystalline direction of the silicon material. The signal voltage generated by the bridge arrangement is proportional to the bias voltage (Vcc) and the amount of pressure or force applied.

3.2.2 Characteristics of Sensors

The output of a sensor is a function of the input. We derived the characteristics of the sensors by applying a known value of the pressure and observing the output voltage. This procedure helped us to calibrate the sensor.

In doing so, the setup shown in Figure 3.4 was utilized. Pressure was applied via an injection syringe to the sensor. This input was also applied to a reference pressure meter (DPM-III Universal Biometer from Bio-Tek Instruments Inc), using a T connector. Pressure changes in the system were observed on the pressure meter in units of mmHg; while the sensor's output voltage changes were observed on a multimeter

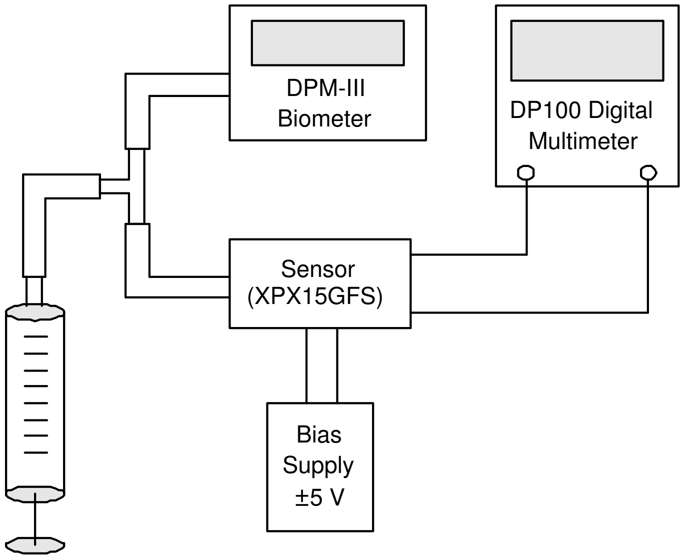


Figure 3.4 Setup for observing sensors' characteristics

(Analogic DP100 Multimeter). Voltage changes versus pressure changes were recorded. This process was performed for both sensors and results are shown in Figure 3.5.

3.3 Signal Conditioning

Signal conditioning is the overall name for circuits, which converts a sensor output to a level suitable for processing [14] and it embraces amplification, offset adjustment, filtering, modulation or demodulation according to the application. A simple signal conditioning circuit should allow the output of the amplifier to be independent of the sensor used, providing interchangeability and high level output at very low cost.

Output voltage of pressure sensors is generally in millivolt ranges. Interface to micro-controllers, therefore, generally involves gaining up the relatively small output voltage performing a differential to single ended conversion.

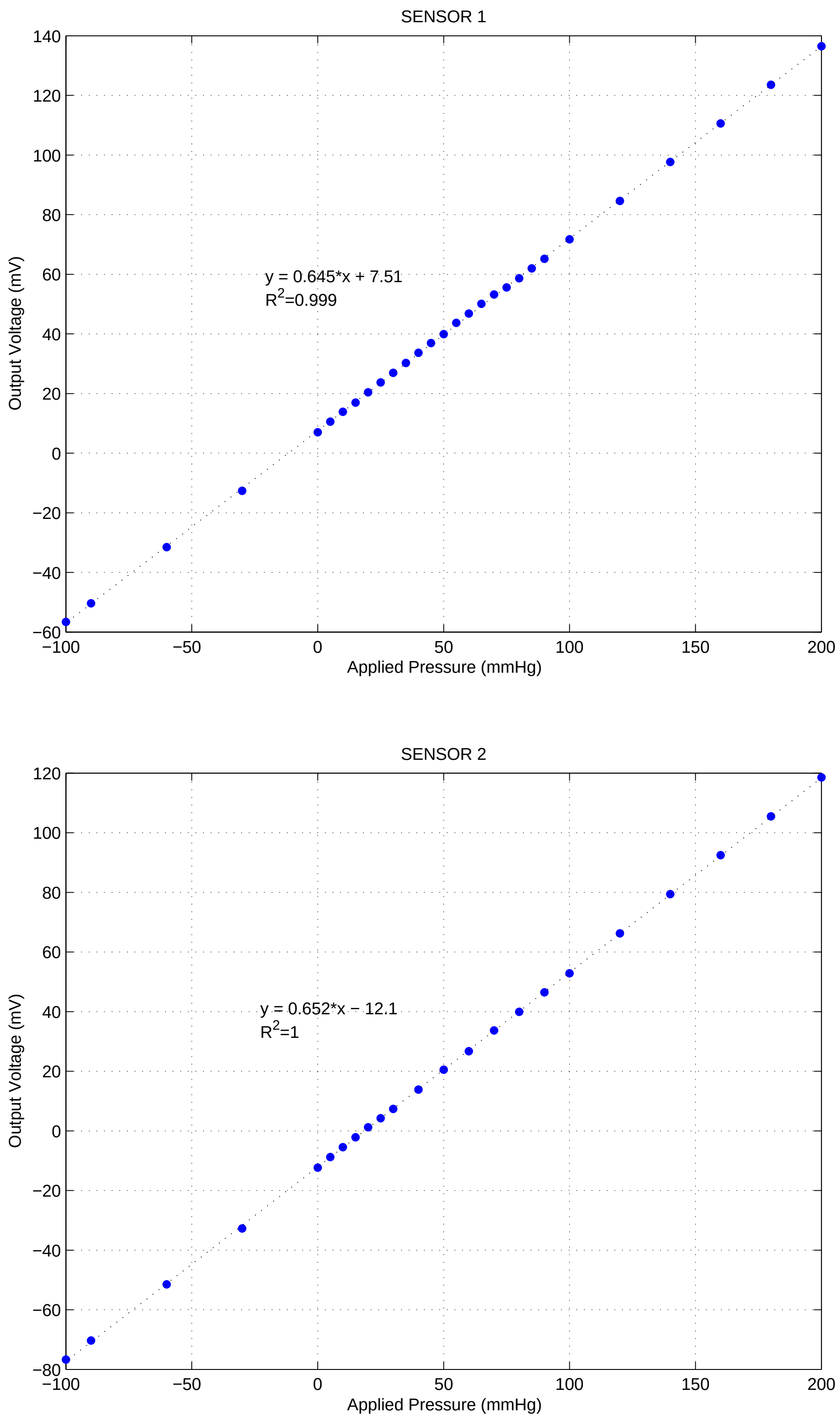


Figure 3.5 Characteristics of Sensor 1 (top) and Sensor 2 (bottom). Data points are given in Table A.1 in Appendix A.

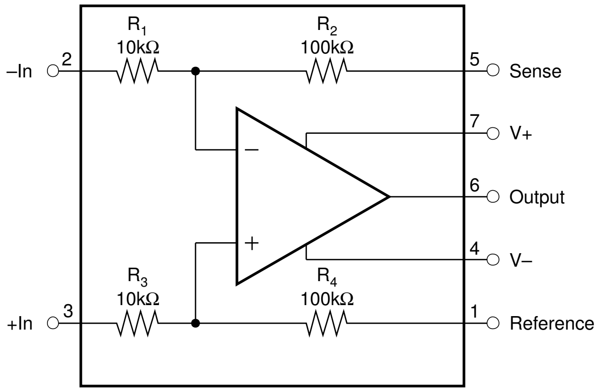


Figure 3.6 Schematic diagram of INA106

3.3.1 Amplification

Whatever else may be required, the signal conditioning will almost always require amplification. When the electrical signals from sensors are very small, i.e. in the order of millivolts or less, it is often difficult to design an appropriate amplifier to tight specifications with basic op-amps and components.

Differential amplifiers are most common interface circuits that are used with pressure sensors. These circuits produce an output voltage that is proportional to the difference between two ground-referenced input voltages.

The Burr Brown INA106, differential amplifier, is chosen due to its accurate gain (±0.025% max. error) and high common mode rejection. It is a monolithic differential amplifier with a fixed gain, 10. This amplifier has relatively less connection pins with its DC power, differential input, offset adjustment, ground and output connections. Representative schematic diagram with the pin connections is shown in Figure 3.6. The resistors are laser trimmed for accurate gain and high common-mode rejection.

The circuit shown in Figure 3.7 is set and connected to the output of two sensors to observe the signal when no pressure is applied. Although we expect the outputs of the

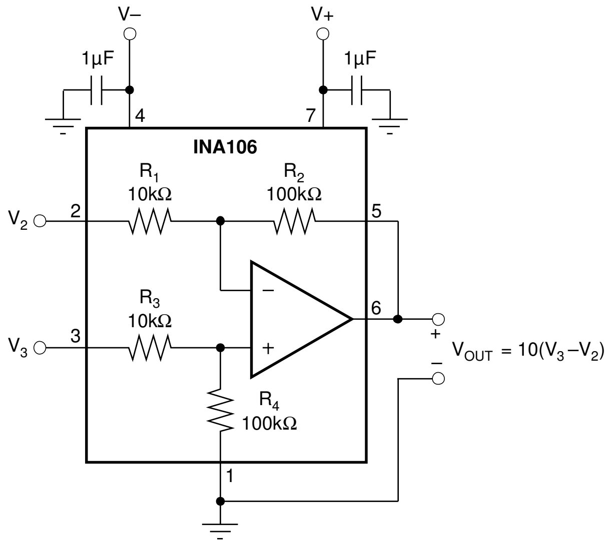


Figure 3.7 Basic power supply and signal connections of INA106

sensor 1 to be 70 mV and that of sensor 2 to be -120 mV (see Figure 3.5), the amplified outputs of these sensors are measured as about 60 mV and -130 mV respectively. This is a simple impedance and high bias current problem, and was averted by using voltage followers (LM310 from National Semiconductor). These components with unity gain and high input impedance adjusted the impedance between the sensors and the amplifiers.

3.3.2 Offset Adjustment

Although an ideal pressure sensor would have zero output signal with no applied pressure, all sensors, in practice, have some dc offset voltages. Figure 3.5 and Table 3.1 show the offset values of the sensors.

Because INA106 provides a gain of 10, these offset values go up to approximately 70 mV and -120 mV at the output of the amplifier. So these voltage values must be subtracted from the output of the amplifiers. To do so, Pin 1, the output reference terminal of INA106, is used. The voltage applied to this pin is summed with the output signal.

Table 3.1
Sensor's Offset Voltages at Zero Applied Pressure

	Applied Pressure (mmHg)	Output Voltage (mV)
Sensor 1	0	7.001
Sensor 2	0	-12.351

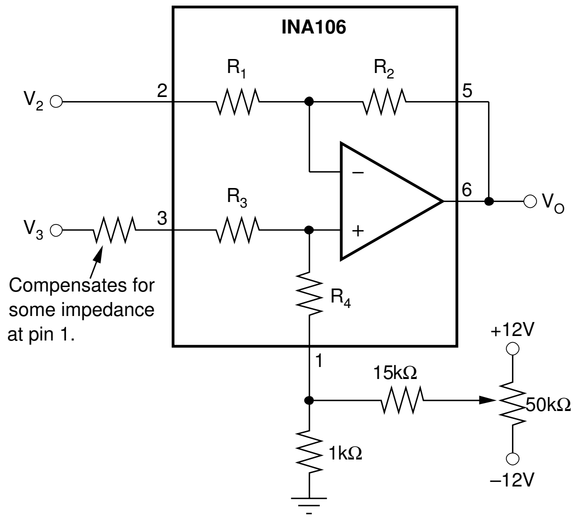


Figure 3.8 Basic circuit for offset adjustment with INA106

Figure 3.8 shows the potentiometer circuit accompanying the amplifier to achieve offset adjustments. Because INA 106 operates using a power supply with ±12 V, same voltage values are applied to a 50 kΩ potentiometer in order to reduce cost.

After zero-balance adjustment of the sensor is achieved, same process done at the beginning can be performed again to observe the output of the amplifier. This process is to apply a known level of pressure input while observing the output. Results are shown in Figure 3.9.

In this phase of the design, temperature effect on the behavior of the sensor and the signal conditioning circuit was also tested. The circuit was operated in different temperatures and offset values were recorded accordingly. No significant variation on the characteristic of the sensors was observed. Results are given in Table 3.2.

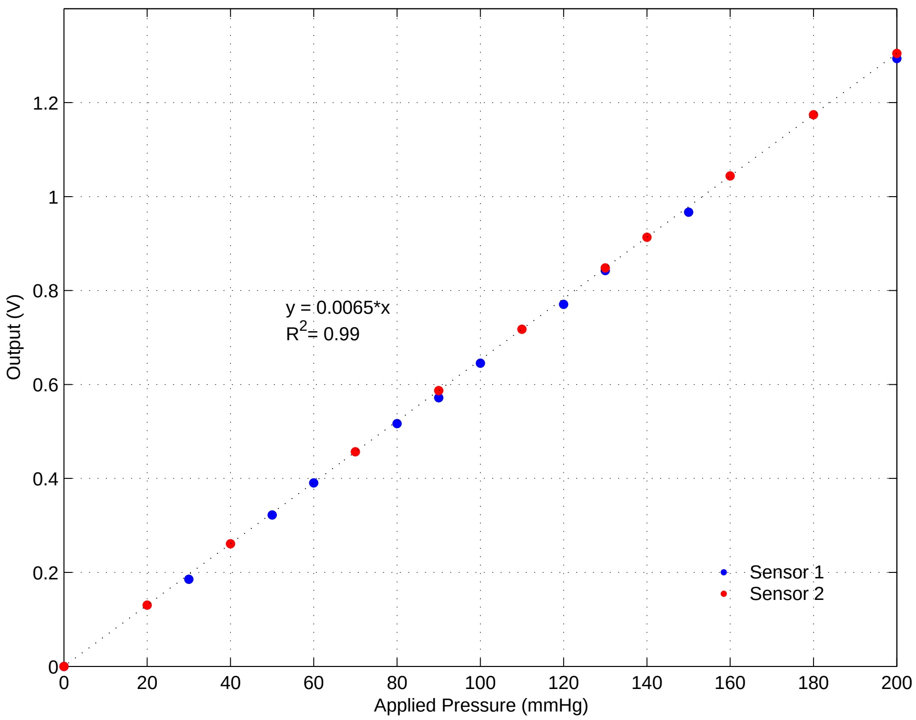


Figure 3.9 Observed output with signal conditioning. Data points are given in Table A.2

Table 3.2
Temperature effect on sensor's offset voltages at atmospheric pressure

	Sensor 1	Sensor 2
Temperature (^{0}C)	Output Voltage (mV)	Output Voltage (mV)
26	-12.3	7.2
27	-12.3	7.2
28	-12.3	7.2
30	-11.9	7.4
31	-11.9	7.4
33	-11.9	7.4
35	-11.8	7.5
37	-11.8	7.5
38	-11.8	7.5

3.4 Flow Characteristics

According to what we described in Chapter 2, flow measurement by means of differential pressure can be achieved by an annular restriction placed in the flow line.

The two basic elements of this procedure are the pressure sensor to measure differential pressure and the obstruction to cause a pressure drop, called the airway resistor. We used two XPX15GFS's from Honeywell as a differential pressure sensor and designed an airway resistor to provide an obstruction for flowing gas to create a drop in pressure.

3.4.1 Airway Resistor

An orifice plate, with a circular cut, constricts the area through which the fluid can pass, causing a resistance to flow. We used this type of apparatus for flow measurement and named it as the airway resistor. Polyamide was chosen as the material of the airway resistor for its stiffness and toughness and its being heat and chemical resistant over a wide temperature range. Its strength, heat and chemical resistance are so good that these materials often replace glass and metals in many industrial applications. They also weight less compared to metal tubes. The schematic view of the airway resistor is illustrated in Figure 3.10 and its technical drawing is presented in Figure B.1 in Appendix B.

Figure 3.10 Actual view of the airway resistor

3.4.2 Flow Measurements

The airway resistor was positioned in series in a patient gas delivering circuit which was a bedside patient oxygen flow meter in this work. A biomedical test equipment, Certifier® Flow Analyzer (FA) Test System from TSI, was also connected as a reference to the set-up illustrated in Figure 3.11. Gas flow rate was changed and

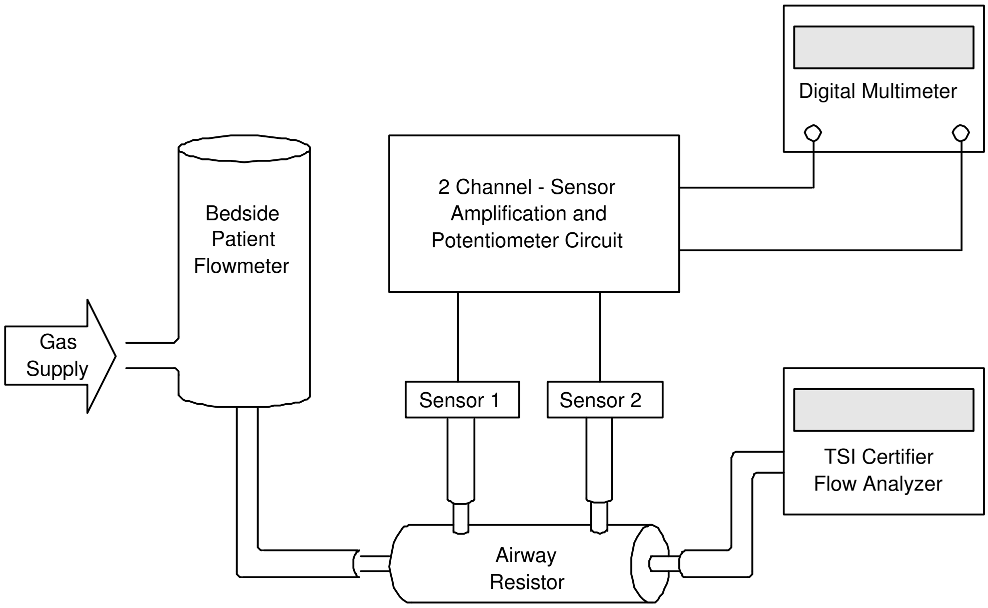
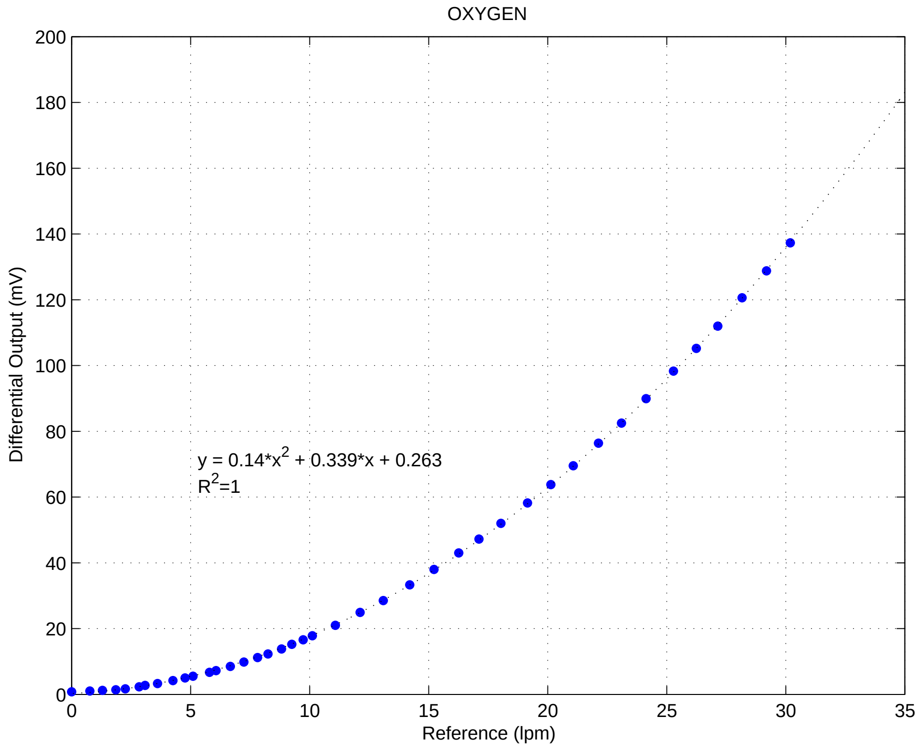


Figure 3.11 Layout for flow measurement

controlled by a valve mounted on the bedside patient flow meter. The reference flow values in lpm was monitored on the Flow Analyzer while corresponding output voltage for each flow value was measured and recorded. These output voltages are related to differential pressure measured on the amplified outputs of Sensor 1 and Sensor 2. This procedure was repeated several times in backward and forward direction to test repeatability and hysteresis of the system. Results are presented in Figure 3.12 and Figure 3.13.

As shown in Figure 3.13, the forward and the backward curves for air flow do not overlap exactly. This is because the air supply used was poor-suited for low flow, and resulted in difficulties in adjusting the desired flow rate, and is not because of the system's response. It can be, therefore, said that the system has not any significant hysteresis for both oxygen and air flow.



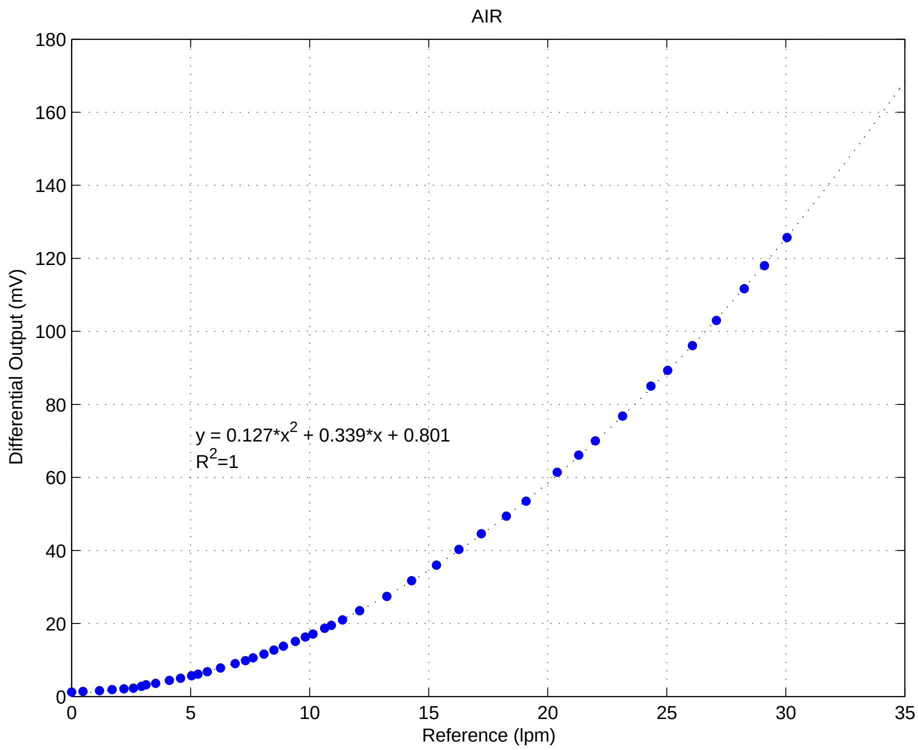


Figure 3.12 Differential measurements through the airway resistor for oxygen (top) and air (bottom) flow. Data points are given in Table A.3 in Appendix A.

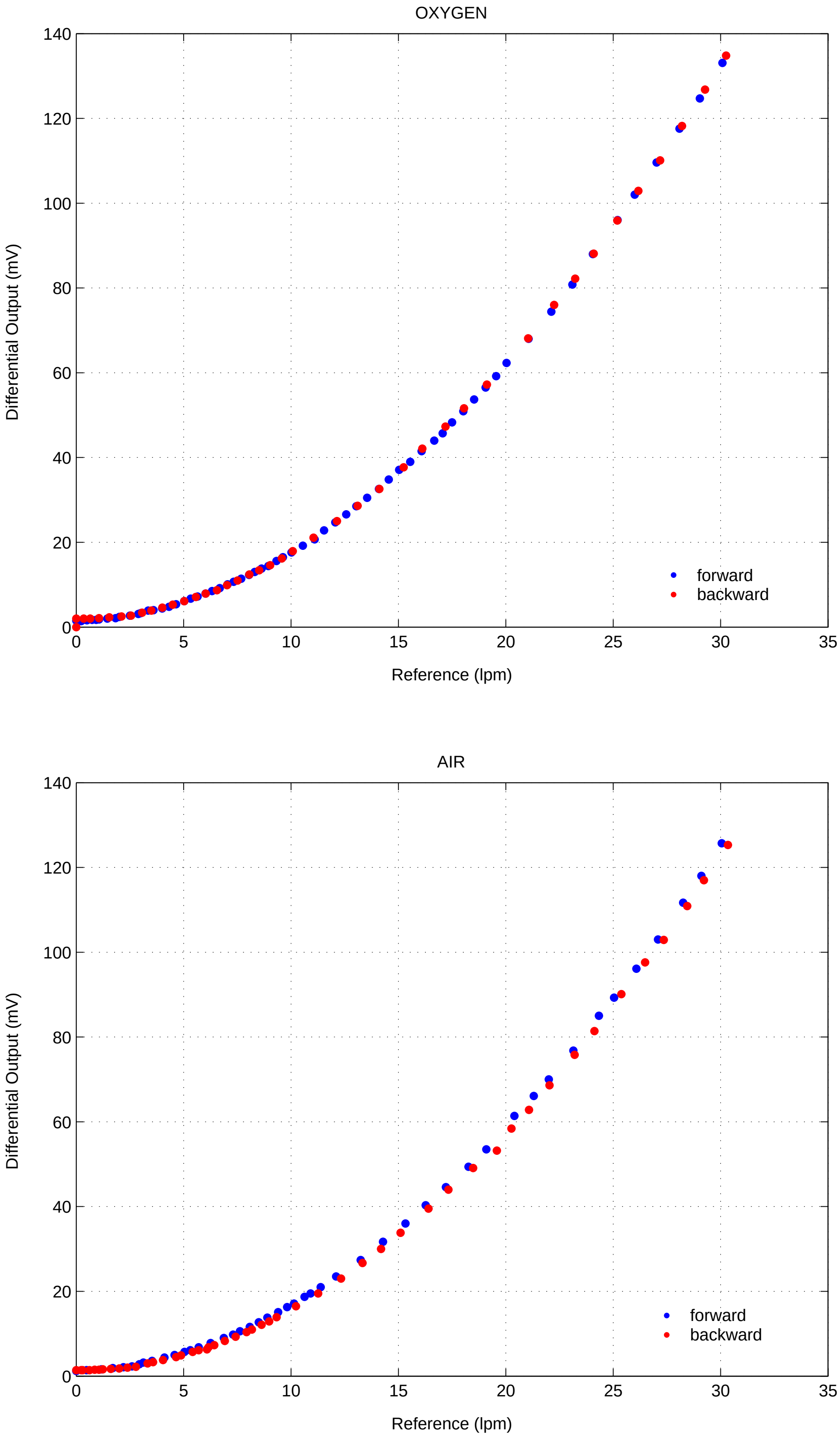


Figure 3.13 Hysteresis results for oxygen (top) and air (bottom) flow

3.5 Processing Acquired Data and Displaying

3.5.1 Microcontroller

Determination of which micro-controller to be used is a demanding phase in the design process, since there are too many parameters to be considered and many kinds of micro-controllers are available in the market. In order to test almost all pressure measuring equipment in clinical use, our pressure analyzer should cover the range of ± 750 mmHg. Moreover, such a test equipment should have high resolution and fast sampling rate as much as possible while keeping its cost affordable.

PSoC® series, from Cypress Micro Systems®, have micro-controllers including programmable gain amplifiers, up to 14-bit ADC's, up to 32 KB of flash memory and an 8x8 multiplier with 32-bit accumulator. PSoC® devices cover configurable blocks of analog and digital logic, as well as programmable interconnects:

- *Digital System* is composed of 8 digital blocks, and each block is an 8-bit resource that can be used alone or combined with other blocks. The digital blocks can be connected to any GPIO through a series of global buses that can route any signal to any pin.
- *Analog System* is composed of 12 configurable blocks, each comprised of an op-amp circuit allowing the creation of complex analog signal flows. Analog peripherals are flexible and can be customized to support specific application requirements.

The architecture described above allows us to create customized peripheral configurations that match the requirements of our application, and dynamic definitions of connections between pins and functional blocks allow us to change configurations when we need. Also, user-friendly and flexible software development tools (PSoC Designer® v.4.2) for configuring and programming analog- and digital-peripheral functionality of these micro-controllers have supported our decision.

Since we had planed to use five inputs in our intended design, CY8C27443 from PSoC® CY8C27x43 family, which has up to five I/O ports providing access to digital and analog blocks, was preferred. Three input ports of CY8C27443 were used for Sensor 1 (S1), Sensor 2 (S2), and the sensor's differential output (DIFF INPUT) and the others were allocated for two control buttons (GAS SELECT SW and MODE SELECT SW). *Gas Select Button* sets the micro-controller according to which type of gases (air or oxygen) is being used in the flow measurement, and changes the display properties. *Mode Select Button* decides the type of measurements to be performed and displays the result according to the measurement type by performing the function of mode selection. Figure 3.14 shows the pin connections of a configured CY8C27443 in this work.

Digital and analog blocks configurations of the micro-controller are given in Figure B.2 in Appendix B. We preferred to use two kinds of ADC module in our design. For simultaneous and precise sampling of two sensors' output, we used a dual input incremental ADC module with a resolution of 12 bits, DUALADC_1. On the other hand, ADCINC_1 user module was employed to convert sensor's differential output to digital form with a 10 bit resolution. All input signals comes to ADC modules after passing through programmable gain amplifiers with a gain of 1, PGA's.

3.5.2 LCD Module

A two-line by 16 character chip module was used in the design for display purposes. The interface between micro-controller and the LCD was established by using the LCD Tool Box User Module in PSoC Designer® v. 4.2. This user module is a set of library routines that writes text strings and formatted numbers to a common two- or four-line LCD module and uses a single I/O port to interface to a standard LCD controller. Port 2 of the micro-controller was chosen for LCD drive functions. Allocated pins for LCD is depicted in Figure 3.14.

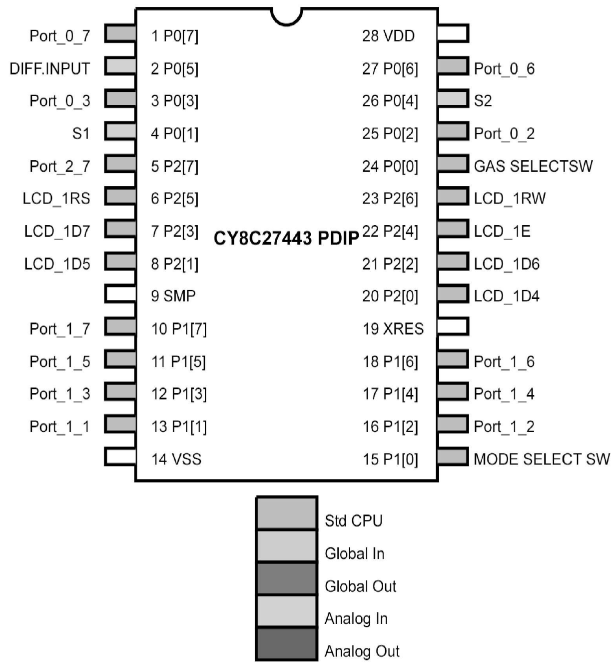


Figure 3.14 Pin connections of a configured CY8C27443

3.5.3 Programming the Micro-Controller

We developed an operation algorithm for the micro-controller to use while it is functioning. This algorithm helps the micro-controller to convert the input signals to an interpretable data. There were two important parameters to be taken into consideration while developing the algorithm:

- Which measurement is going to be performed? (Flow or Pressure Measurement)
- Which type of gases is flowing through the system? (Air or Oxygen)

All mathematical operations and display modifications are performed by micro-controller according to these two parameters. For this purpose, we used two buttons to control our system. *Mode Select Button* executes the decision of which measurement is going to be performed, and modifies the micro-controller's function according to the measurement type. *Gas Select Button* sets the micro-controller's function according to which type of gases is flowing through the system. These two buttons are also used for

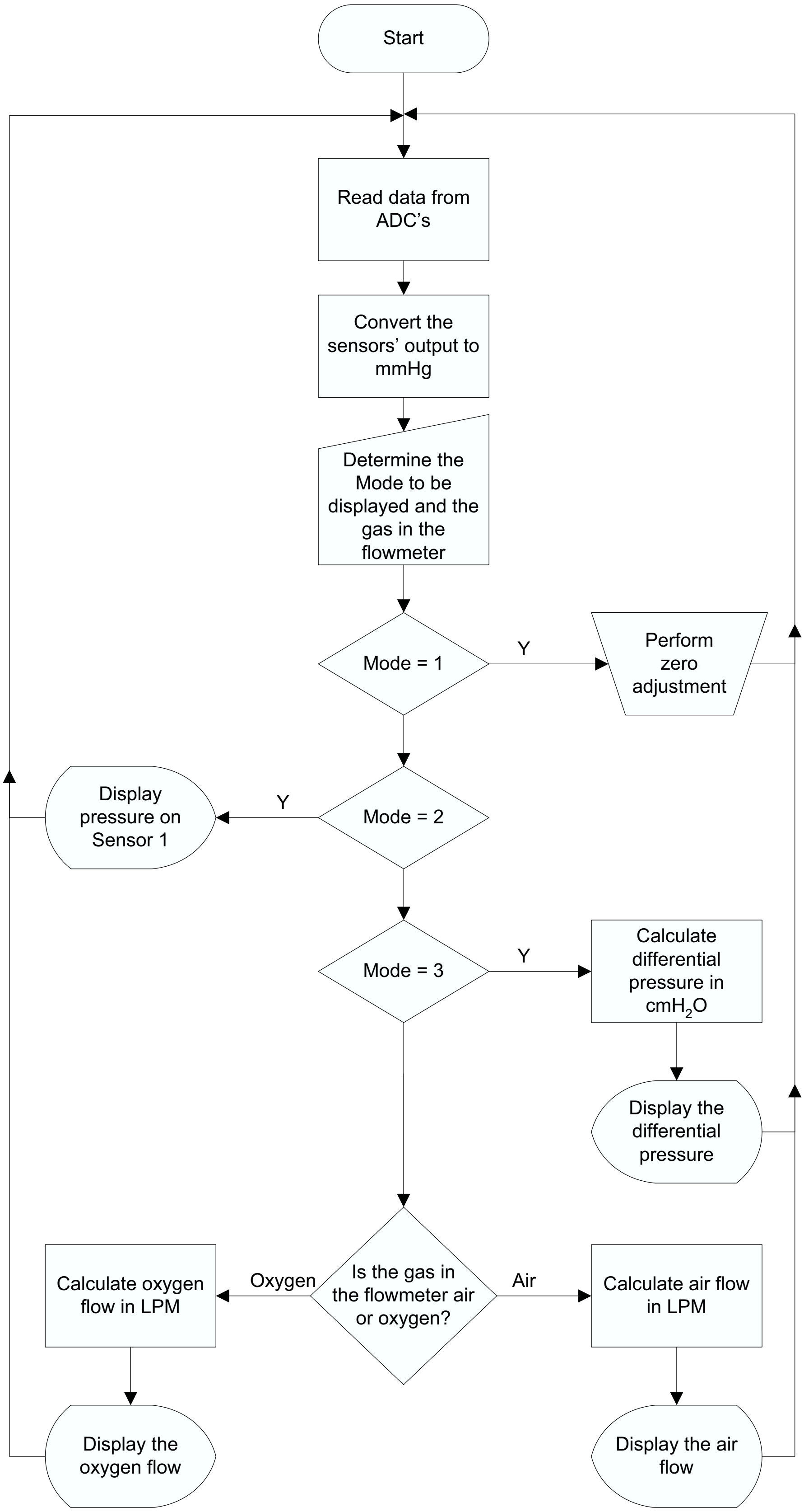


Figure 3.15 Basic flow diagram of the software

display modification purposes.

Flow chart of the developed algorithm and the code listing of the program are given in Figure 3.15 and in Appendix C respectively.

4. RESULTS AND DISCUSSION

In this chapter, we will concentrate on the measurement results obtained at the end of development cycle. While obtaining the readings, same procedures were followed as in deriving the sensors (see Section 3.2.2) and the flow characteristics (see Section 3.4.2). Namely, we applied known values of the input and then observed the output.

4.1 Pressure Measurement

For pressure measurement testing, a simple system like shown in Figure 3.4 was utilized. Pressure was applied via an injection syringe as an input to our pressure and flow analyzer. This input was also applied to a pressure meter (DPM-III Universal Biometer from Bio-Tek Instruments Inc), using a T connector. Pressure changes in the system were observed both on DPM-III Biometer and our instrument in the unit of mmHg. Measured values from DPM-III Biometer were accepted as reference and our instrument output was recorded for each reference value. Results are shown in Figure 4.1.

We had theoretically expected an equation as $y = x$ for pressure measurement. Figure 4.1 represents, however, a graph of $y = 1.002 \times x$ with a good fit (R^2=1). This negligible difference between the slopes might be caused by the disagreement between the measurement precision of our analyzer and DPM-III Biometer. Biometer detects the pressure with 1 mmHg intervals while our analyzer reacts to even 0.377 mmHg change in pressure.

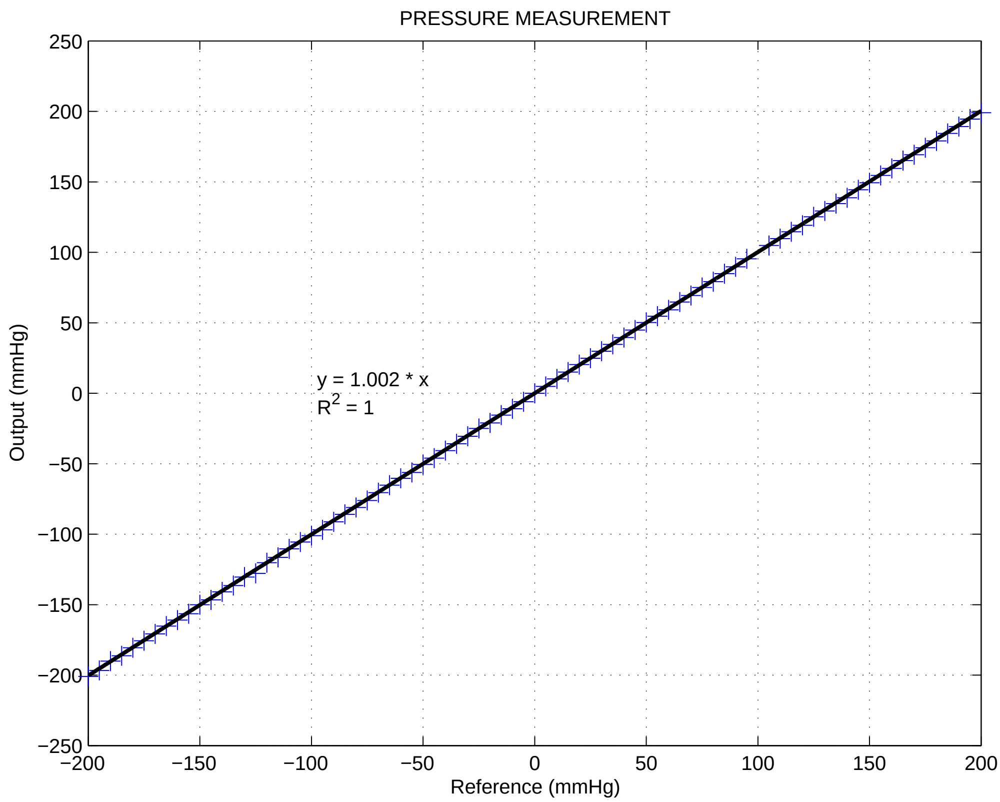


Figure 4.1 Result of pressure measurement. Data points are given in Table A.4.

4.2 Flow Measurement

For flow measurement testing, our airway resistor was positioned in series in a patient gas delivering circuit which was a bedside patient flow meter in this work. Using tubes, two measurement points on the airway resistor were connected to two pressure sensors of our pressure and flow analyzer. A biomedical test equipment, Certifier® Flow Analyzer (FA) Test System from TSI, was also connected as a reference to the set-up illustrated in Figure 3.11. Gas flow rate was changed and controlled by a valve mounted on the bedside patient flow meter. The reference flow values in lpm was monitored on Certifier® Flow Analyzer while the corresponding outputs of our analyzer for each flow value was measured and recorded. This procedure was performed for both oxygen and air flow. Results are presented in Figure 4.2 and Figure 4.3.

Figure 4.2 and Figure 4.3 show that flow measurement results are mostly within 95% confidence interval (red dotted lines on the graphs). Due to the effect of the parabolic relationship between flow and differential pressure (see Figure 3.12), the response of our analyzer gets better for higher flow rate. Although the overall performance of the analyzer is satisfactory, the readings below 5 lpm is not good enough. This might have two primary reasons: AD converter and the airway resistor. Since the airway resistor causes very small pressure drop, thereby the voltage, for values below 5 lpm, these small voltage drops can not be detected by the micro-controller's AD Converter. The deviation for the higher flow values might be, on the other hand, in consequence of Certifier® Flow Analyzer's unstable operation in such ranges.

AD Converter with higher resolution could not be used since the micro-controller's 32 kB-memory had not enough space (approximately 80% full). This problem will be mentioned in Future Development section.

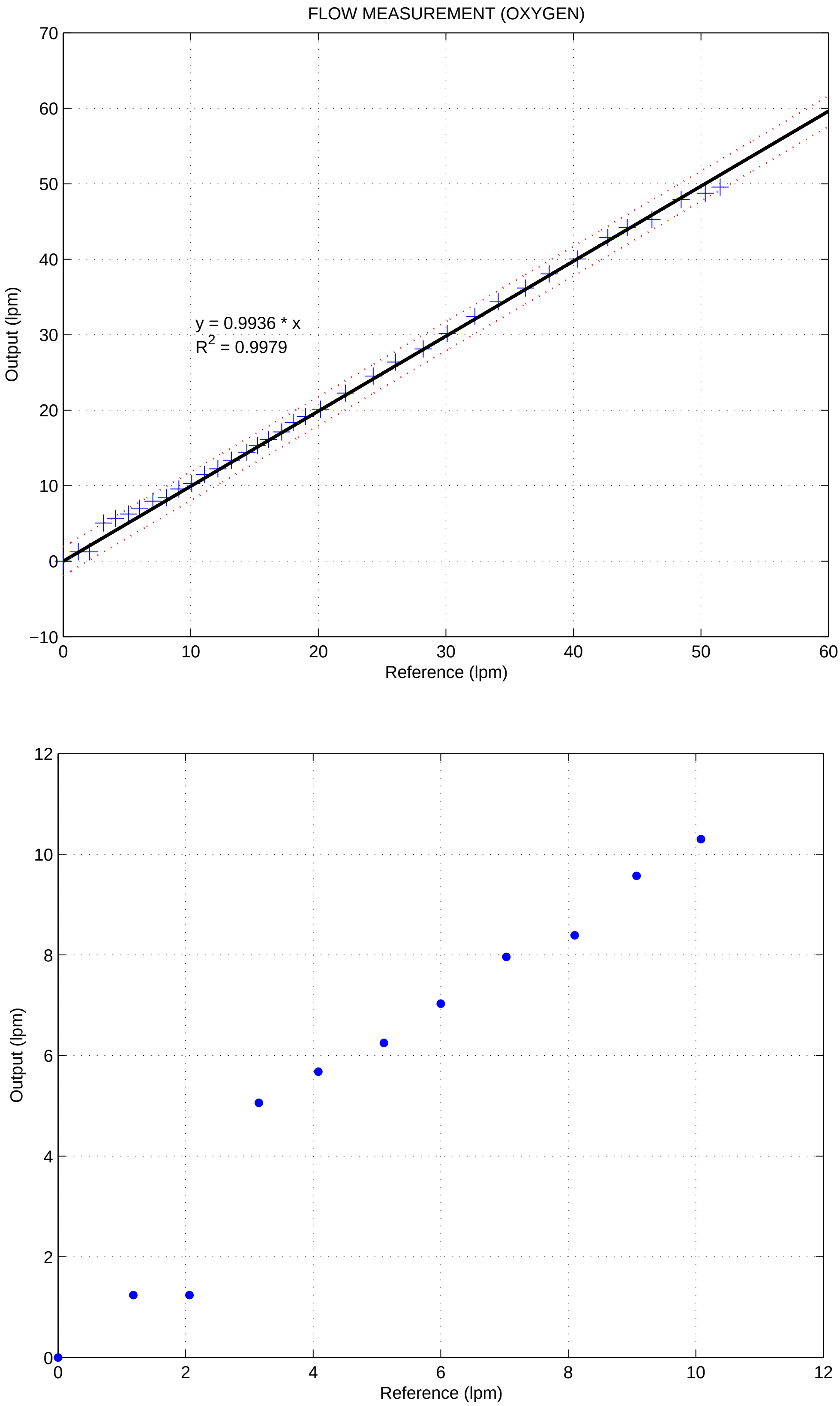


Figure 4.2 Results of measurement for oxygen flow. The problematic range of 0-5 lpm is zoomed in bottom figure. Data points are given in Table A.5.

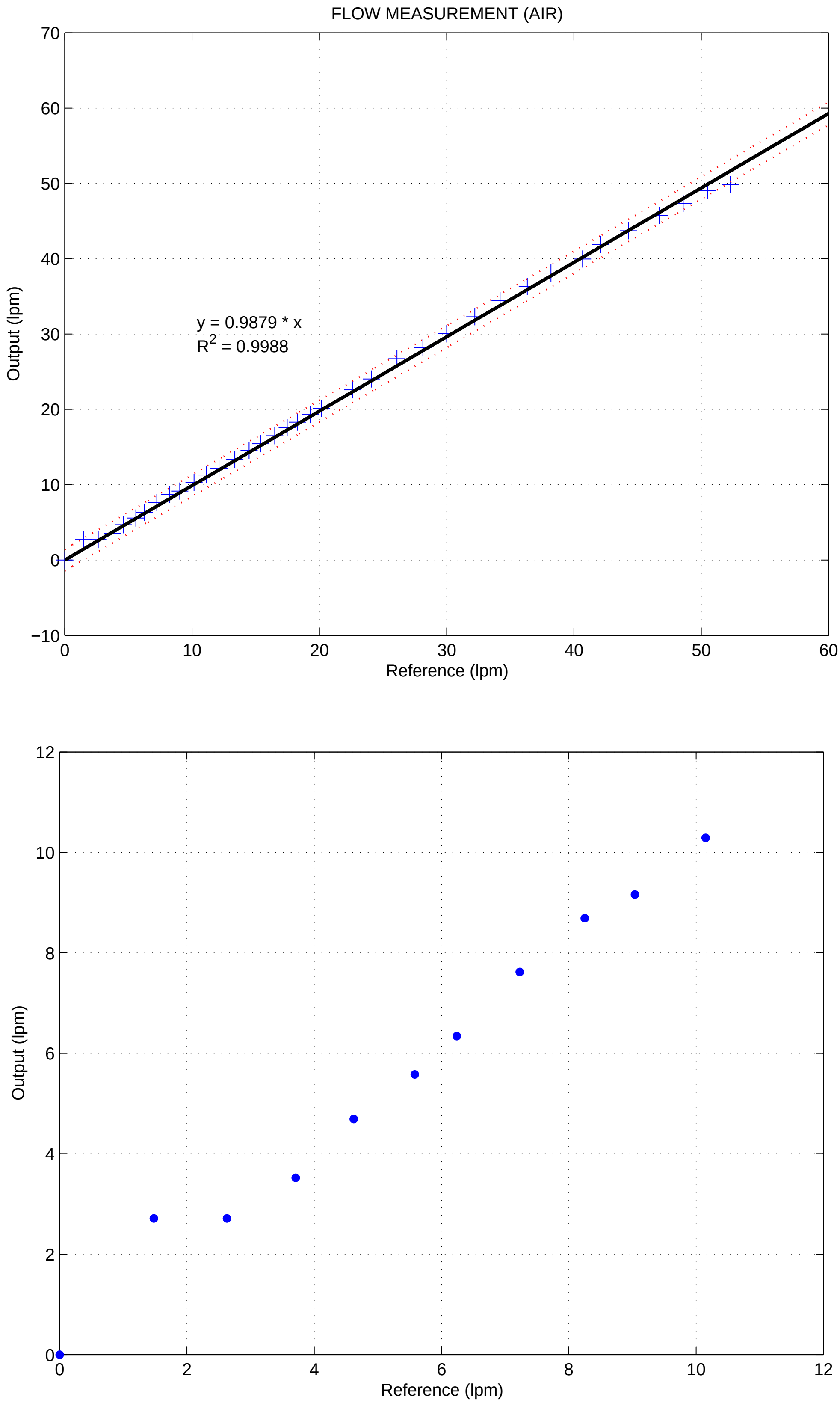


Figure 4.3 Results of measurements for air flow. The problematic range of 0-5 lpm is zoomed in bottom figure. Data points are given in Table A.5.

5. CONCLUSIONS

The measurement of pressure and gas flow form a central part of many physiological measurements. Pressure measurements provide essential information related to the functions of the cardiovascular system, while gas flow measurements play an important role in treatment and diagnostic procedure in the respiratory system. Therefore, medical equipment used in pressure and flow measurements, as all other medical equipment, needs scheduled preventive maintenance against unexpected device failures.

In this thesis, we have developed an instrumentation system which can be used in testing of non-invasive blood pressure and gas flow measuring apparatus. Throughout the development process; we determined the sensors which would be used and derived their characteristics. Then we performed necessary amplification and offset adjustment. In order to be able to measure gas flow rate, we designed and produced an orifice type airway resistor. So, we performed differential pressure measurement and determined the flow characteristics of the resistance for both oxygen and air. These characteristic curves show that there is a parabolic relationship between differential pressure and gas flow through the orifice plate, as expected.

In the other major part of this work, we configured the microcontroller's functional blocks in a way that they match the requirements of our application. Then we developed an algorithm to convert acquired data to interpretable form and coded a program in C to run on the micro-controller.

In the last part, we assembled all system mentioned in a box (See Figure 5.1) and compared pressure results with those obtained from DPM-III Universal Biometer (Bio-Tek Instruments Inc.) and flow results with those obtained from Certifier® Flow Analyzer (FA) Test System (TSI).

Finally, we have concluded that our designed instrument can be used to test non-invasive blood pressure measuring apparatus which has typically a range of 0-300 mmHg, bedside patient flow meters for vacuum, air or oxygen which have typically a range of 0-15 lpm, patient ventilators (with a range of 0-50 lpm), or aspirators (with a range of 0-30 lpm). Instructions for prospective users are given in Appendix D with related figures.

5.1 Cost Analysis

Hardware development cycle brings the need for cost accounting to evaluate the efficiency of the design. Here, we prepared a table listing for all the components we have used in the design process with their names, part numbers and prices (Table 5.1). The total cost for such a system is approximately 240.40 $.

5.2 Future Developments

Future work will focus on further improving in the flow measurement and achieving accurate flow measurement for values below 5 lpm. In this instance, the characterization of the designed airway resistor plays an important role. Reducing the diameter of the orifice in the airway resistor causes an increase in differential pressure for small values, resulting in an improvement in detectability. Number of bits used for AD conversion in flow measurement can also be increased to improve sensitivity and detectability for values below 5 lpm. Since our micro-controller's memory does not have enough space for look-up tables with higher resolution, we can replace the micro-controller used with one having higher ROM capacities from the same series.

Designing a power supply by using batteries will provide convenience in portability. Also, observed shifts in the response of DUALADC can also be corrected by using a different AD converter, or changing micro-controller design specifications.

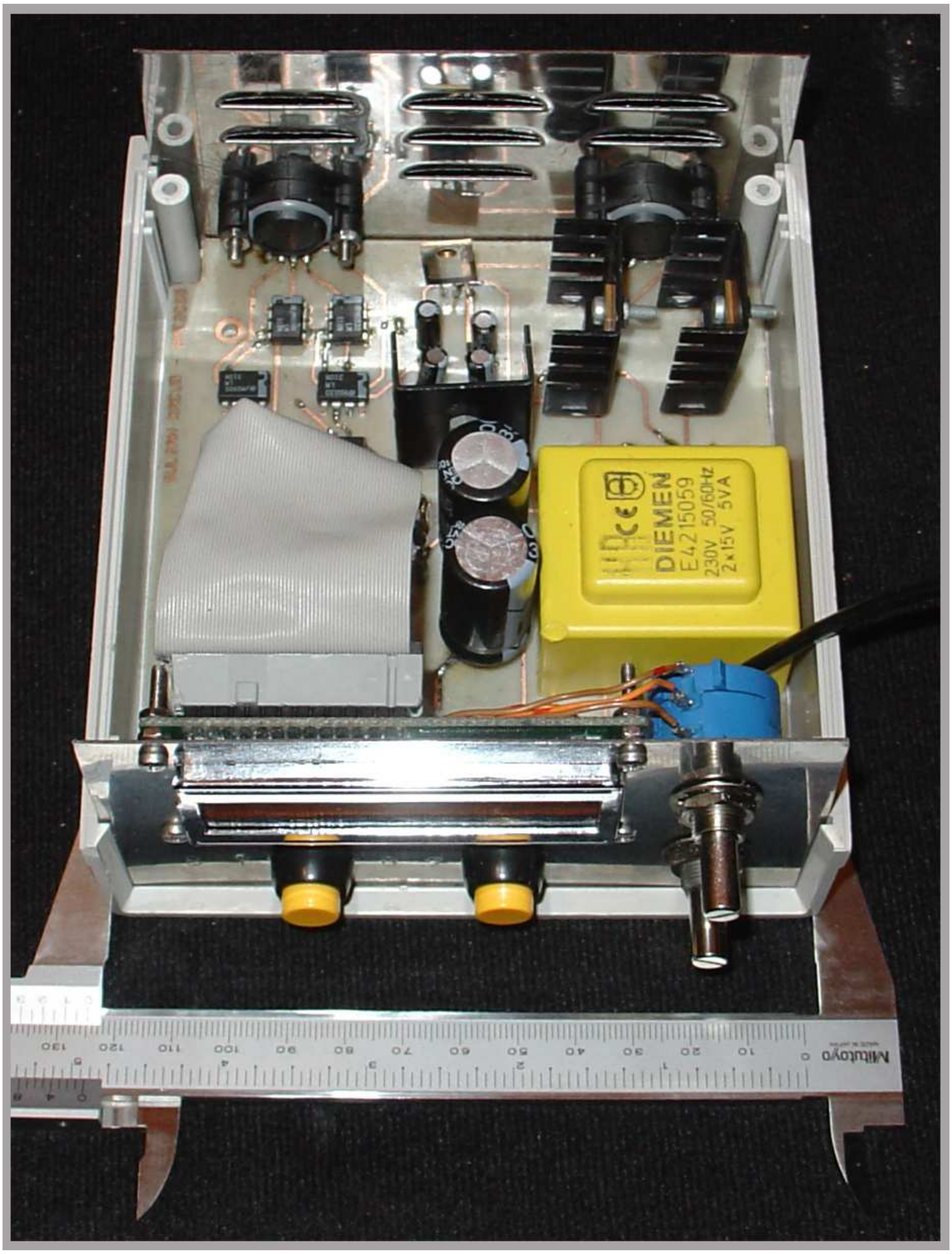


Figure 5.1 Top view of the analyzer (Cover removed)

Table 5.1
Cost of materials (Prices are subject to change.)

Description	**Part # / Value**	**Quantity**	**Unit Price ($)**	**Price ($)**
1.0 Ampere General Purpose Rectifier	1N4001	4	0.25	1.00
LCD Module (2 x 16)		1	10.00	10.00
3-Terminal Negative Fixed Voltage Regulator	MC7912CT	1	0.35	0.35
3-Terminal Negative Fixed Voltage Regulator	MC7905CT	1	0.35	0.35
3-Terminal Positive Fixed Voltage Regulator	MC7812CT	1	0.35	0.35
3-Terminal Positive Fixed Voltage Regulator	MC7805CT	1	0.35	0.35
Airway Resistor		1	55.00	55.00
Box		1	10.00	10.00
Double Side PCB		1	15.00	15.00
Fuse	250 mA	1	0.10	0.10
Fuse Housing		1	1.00	1.00
Header, 14-Pin		1	0.10	0.10
Header, 3-Pin		2	0.10	0.20
Heat Sink		3	0.50	1.50
Multi-turn Potentiometer	50K	2	5.00	10.00
On/Off Switch		1	0.50	0.50
Polarized Capacitor (Radial)	2200uF	2	1.00	2.00
Polarized Capacitor (Radial)	10uF	4	0.40	1.60
Power Cord		1	2.00	2.00
Precision Gain=10 Differential Amplifier	INA106KP	3	20.00	60.00
Pressure Sensor	XPX15GFS	2	24.00	48.00
Programmable System-On-Chip	CY8C26443-24PI	1	3.90	3.90
Resistor	1K	2	0.10	0.20
Resistor	15K	2	0.10	0.20
Resistor	10K	4	0.20	0.80
Resistor	5K	1	0.10	0.10
Sphygmomanometer Tubes (50 cm)		2	1.00	2.00
Switch		2	0.50	1.00
Three-winding Transformer (Non-Ideal)		1	5.00	5.00
Voltage Follower	LM310N	4	1.95	7.80
			Total:	**240.40**

APPENDIX A. DATA

Table A.1
Output Voltage Changes According to the Applied Pressure

	Sensor 1	Sensor 2
Applied Pressure (mmHg)	Output Voltage (mV)	Output Voltage (mV)
-100	-56.599	-76.711
-90	-50.362	-70.336
-60	-31.529	-51.465
-30	-12.624	-32.714
0	7.001	-12.351
5	10.548	-8.766
10	13.876	-5.477
15	16.945	-2.159
20	20.399	1.202
25	23.725	4.259
30	26.931	7.350
35	30.272	10.724
40	33.679	13.867
45	36.927	17.250
50	39.917	20.499
55	43.671	23.800
60	46.807	26.734
65	50.066	30.280
70	53.226	33.679
75	55.553	36.853
80	58.654	39.942
85	61.948	43.324
90	65.174	46.442
100	71.723	52.849
120	84.590	66.263
140	97.640	79.393
160	110.598	92.489
180	123.533	105.483
200	136.484	118.560

Table A.2
Amplified Output

Sensor 1		Sensor 2	
Applied Pressure (mmHg)	Output Voltage (V)	Applied Pressure (mmHg)	Output Voltage (V)
0	0	0	0
30	0.1853	20	0.1305
50	0.3223	40	0.2610
60	0.3906	70	0.4567
80	0.5168	90	0.5872
90	0.5716	110	0.7176
100	0.6454	130	0.8481
120	0.7706	140	0.9134
130	0.8424	160	1.0438
150	0.9668	180	1.1743
200	1.2939	200	1.3048

Table A.3
Differential measurements through the airway resistor for oxygen and air flow

Oxygen Flow		Air flow	
Reference (lpm)	Differantial Output(mV)	Reference (lpm)	Differential Output(mV)
0.00	0.80	0.00	1.20
0.77	1.00	0.48	1.40
1.30	1.20	1.17	1.60
1.86	1.40	1.70	1.90
2.26	1.70	2.20	2.10
2.84	2.30	2.60	2.30
3.09	2.70	2.94	2.80
3.61	3.30	3.13	3.20
4.26	4.20	3.53	3.60
4.77	5.00	4.11	4.40
5.10	5.50	4.58	5.00
5.79	6.70	5.04	5.70
6.07	7.20	5.31	6.10
6.67	8.50	5.70	6.80
7.24	9.80	6.26	7.80
7.81	11.20	6.87	9.00
8.25	12.30	7.30	9.80
8.82	13.80	7.62	10.60
9.25	15.20	8.08	11.60
9.73	16.60	8.50	12.70
10.11	17.80	8.90	13.80
11.08	21.00	9.40	15.10
12.12	24.90	9.82	16.30

Table A.3 continued

Oxygen Flow		Air flow	
Reference (lpm)	Differantial Output(mV)	Reference (lpm)	Differential Output(mV)
13.09	28.50	10.14	17.10
14.20	33.30	10.63	18.70
15.22	38.00	10.91	19.50
16.26	43.00	11.38	21.00
17.11	47.20	12.10	23.50
18.03	52.00	13.24	27.40
19.15	58.20	14.28	31.70
20.13	63.80	15.33	36.00
21.07	69.50	16.27	40.30
22.13	76.40	17.21	44.60
23.10	82.50	18.26	49.40
24.13	89.90	19.09	53.50
25.28	98.30	20.40	61.40
26.24	105.20	21.30	66.10
27.14	112.00	22.00	70.00
28.16	120.60	23.14	76.80
29.19	128.80	24.33	85.00
30.19	137.30	25.04	89.30
		26.08	96.10
		27.08	103.00
		28.25	111.70
		29.10	118.00
		30.05	125.70

Table A.4
Results of the pressure measurement

Reference (mmHg)	Output (mmHg)
-200	-200.94
-195	-196.66
-190	-190.01
-185	-186.24
-180	-180.58
-175	-175.68
-170	-170.78
-165	-165.26
-160	-160.98
-155	-156.40
-150	-150.05
-145	-146.65
-140	-141.00
-135	-136.47

Table A.4 continued

Reference (mmHg)	Output (mmHg)
-130	-130.44
-125	-127.80
-120	-120.26
-115	-116.49
-110	-110.46
-105	-105.56
-100	-101.04
-95	-96.89
-90	-91.23
-85	-85.96
-80	-81.06
-75	-76.13
-70	-70.50
-65	-65.22
-60	-60.32
-55	-56.17
-50	-50.52
-45	-45.99
-40	-40.72
-35	-35.81
-30	-30.54
-25	-25.01
-20	-21.11
-15	-15.46
-10	-11.02
-5	-6.03
0	0.00
5	4.90
10	10.18
15	15.08
20	20.36
25	24.88
30	29.78
35	34.68
40	39.58
45	44.86
50	50.14
55	54.66
60	59.17
65	64.71
70	69.37
75	75.02
80	79.17
85	84.82
90	89.73

Table A.4 continued

Reference (mmHg)	Output (mmHg)
95	95.38
105	104.81
110	109.71
115	114.61
120	119.13
125	125.16
130	129.31
135	134.59
140	138.74
145	144.39
150	149.29
155	154.57
160	159.47
165	165.13
170	169.27
175	174.17
180	179.07
185	184.35
190	189.25
195	194.53
200	199.06

Table A.5
Results of the flow measurement

Oxygen Flow		Air flow	
Reference (lpm)	Differantial Output(mV)	Reference (lpm)	Differential Output(mV)
0.00	0.00	0.00	0.00
1.18	1.24	1.48	2.71
2.06	1.24	2.63	2.71
3.15	5.06	3.71	3.52
4.08	5.68	4.62	4.69
5.11	6.25	5.58	5.58
6.00	7.03	6.24	6.34
7.03	7.96	7.23	7.62
8.10	8.39	8.25	8.69
9.07	9.57	9.04	9.16
10.08	10.30	10.15	10.29
11.08	11.46	11.11	11.29
12.13	12.23	12.12	12.20
13.19	13.37	13.36	13.37
14.40	14.43	14.48	14.59
15.23	15.30	15.39	15.43

Table A.5 continued

Oxygen Flow		Air flow	
Reference (lpm)	Differantial Output(mV)	Reference (lpm)	Differential Output(mV)
16.10	16.12	16.50	16.49
17.13	17.13	17.47	17.60
18.03	18.39	18.26	18.30
19.00	19.19	19.30	19.31
20.18	20.14	20.16	20.16
22.14	22.27	22.60	22.60
24.30	24.53	24.08	24.03
26.05	26.39	26.10	26.72
28.22	28.13	28.13	28.18
30.11	30.15	30.00	30.10
32.28	32.41	32.21	32.29
34.11	34.36	34.20	34.46
36.26	36.20	36.33	36.33
38.10	38.06	38.19	38.10
40.30	40.03	40.68	39.96
42.70	42.90	42.12	41.87
44.22	44.18	44.30	43.70
46.16	45.26	46.70	45.77
48.45	47.94	48.58	47.32
50.34	48.74	50.50	49.07
51.50	49.56	52.30	49.86

APPENDIX B. DESIGN DETAILS

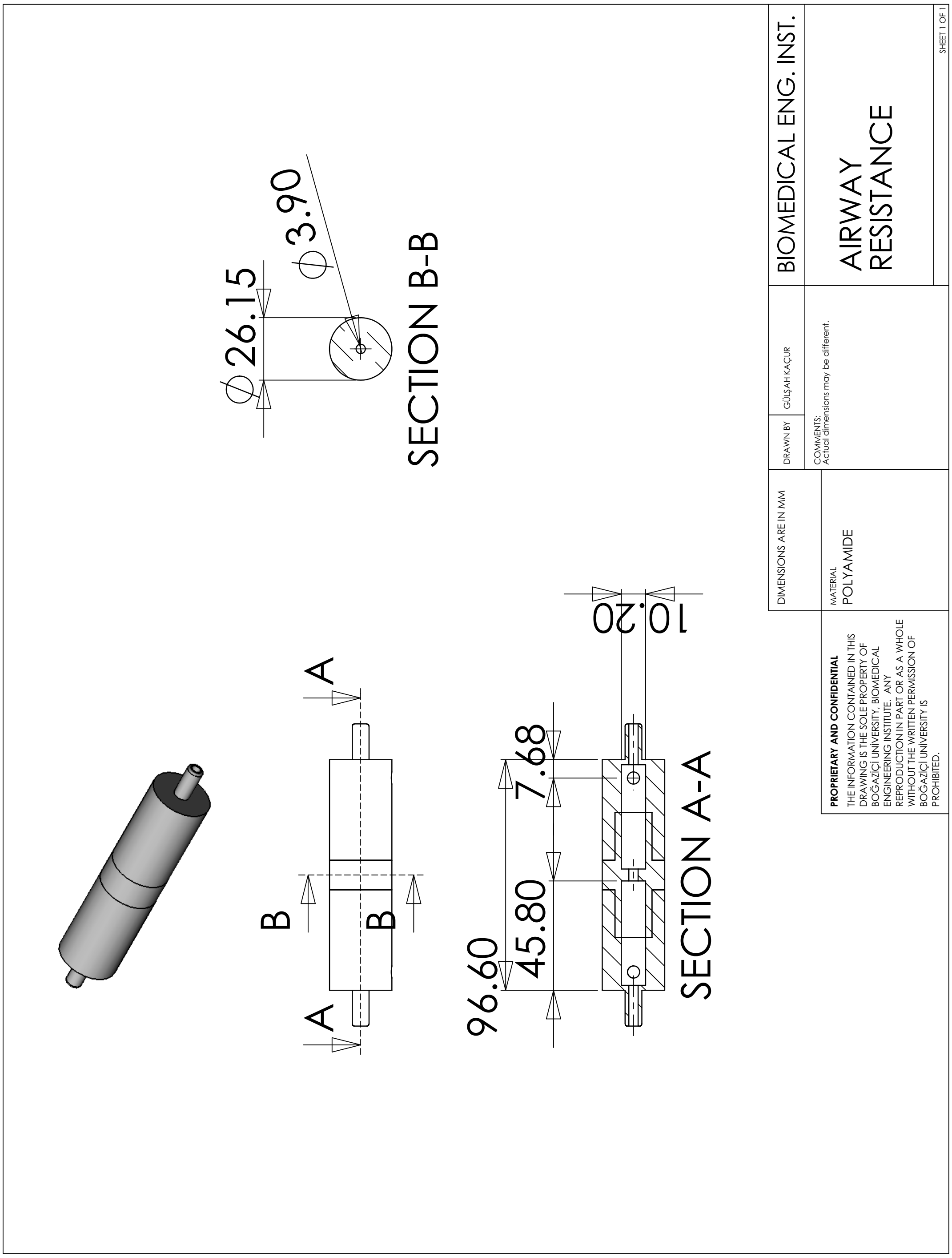


Figure B.1: Technical drawing of the airway resistor

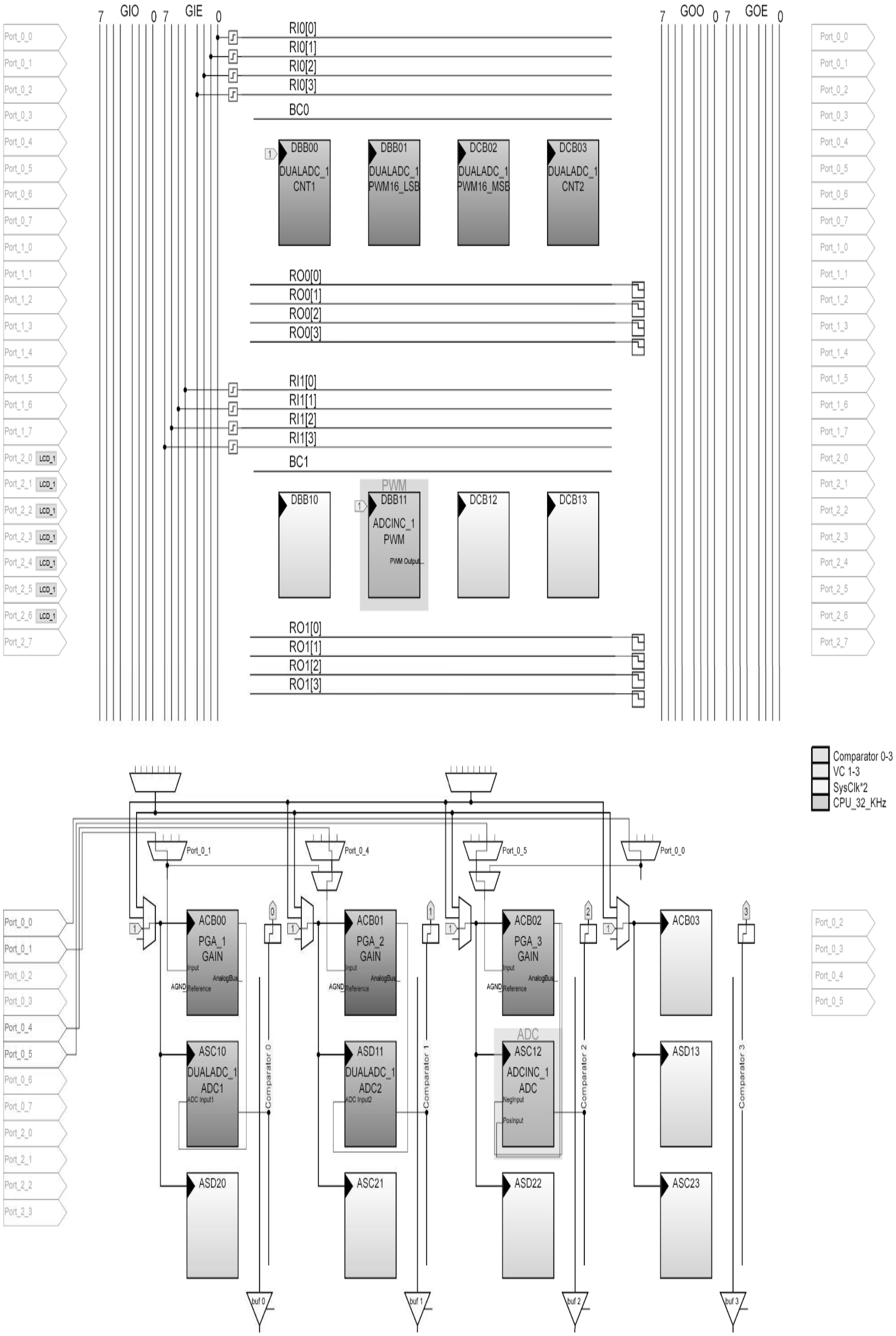


Figure B.2 Digital and analog row interconnects

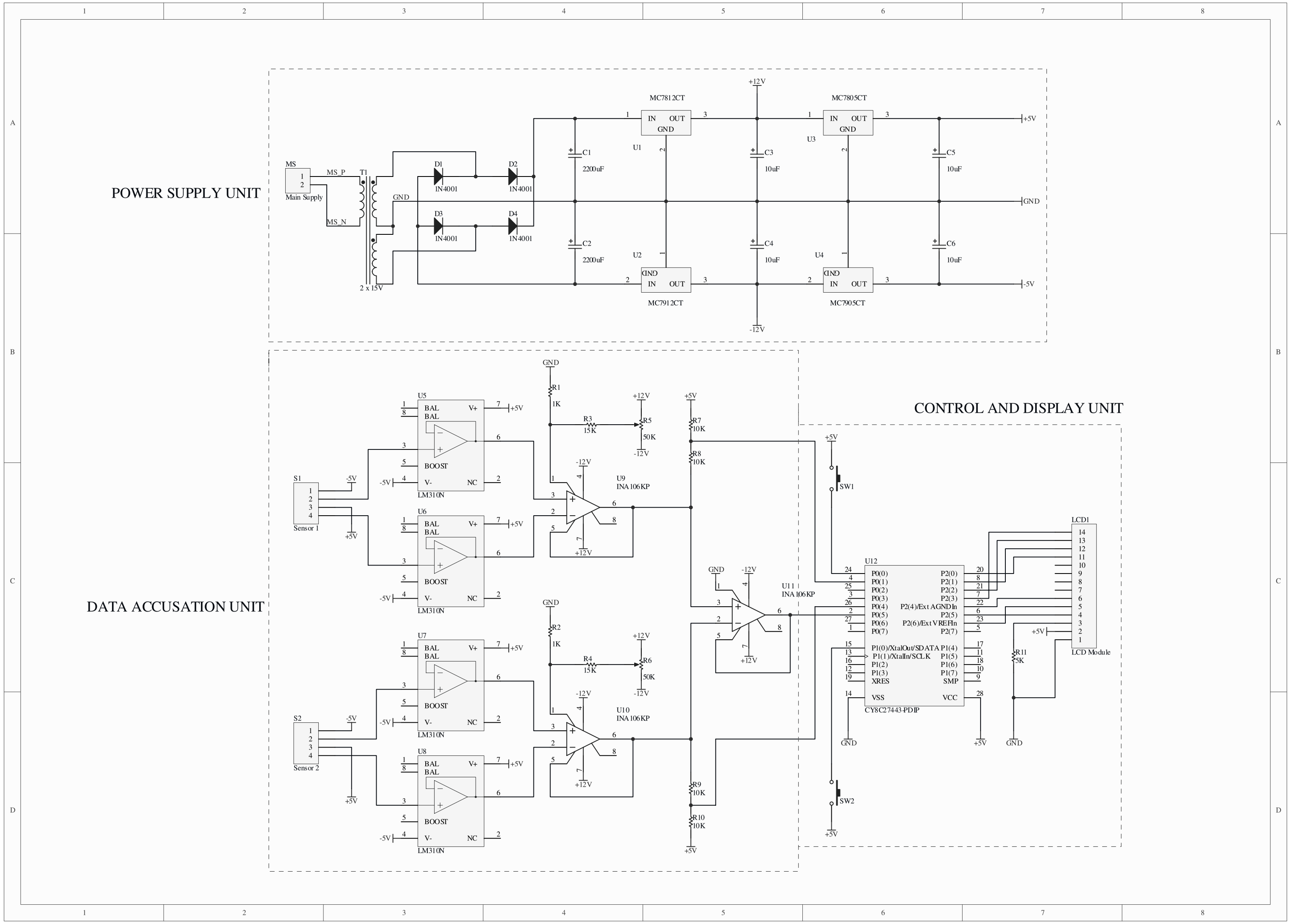


Figure B.3 Schematic of the Pressure and Flow Analyzer

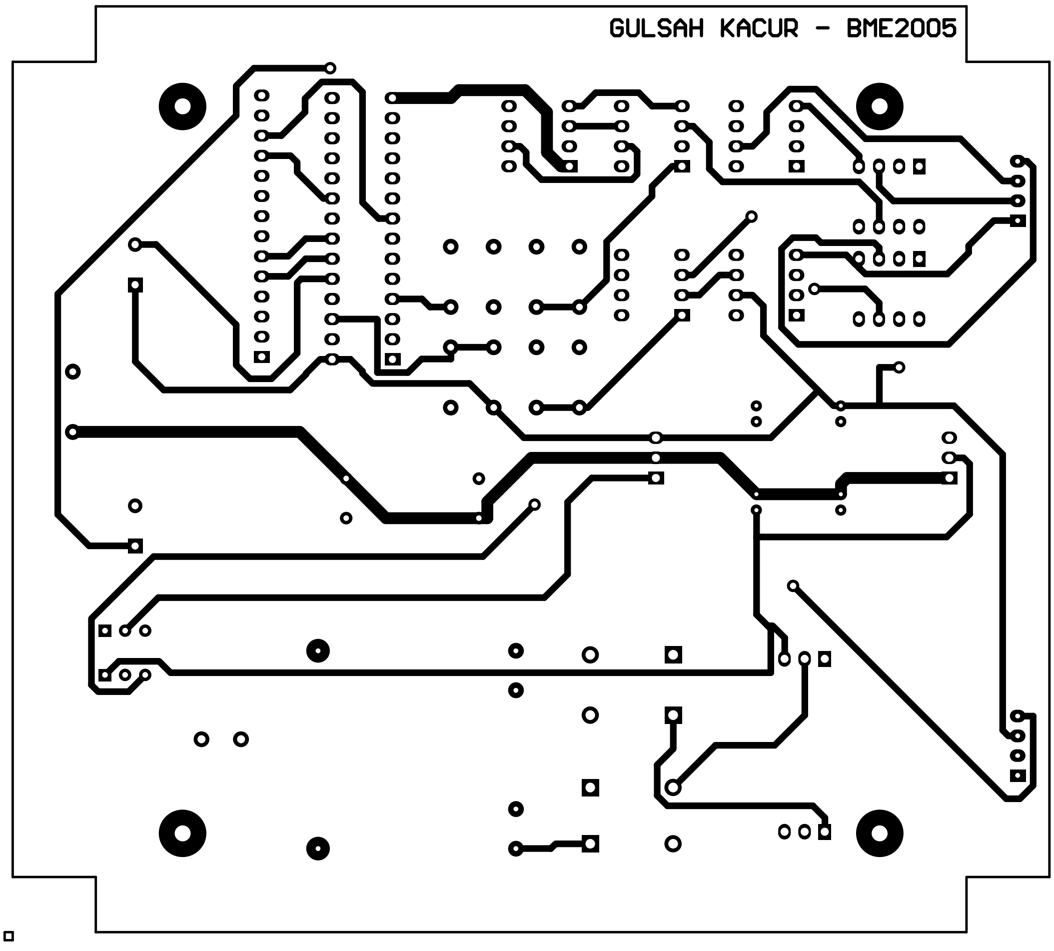


Figure B.4 PCB top layer

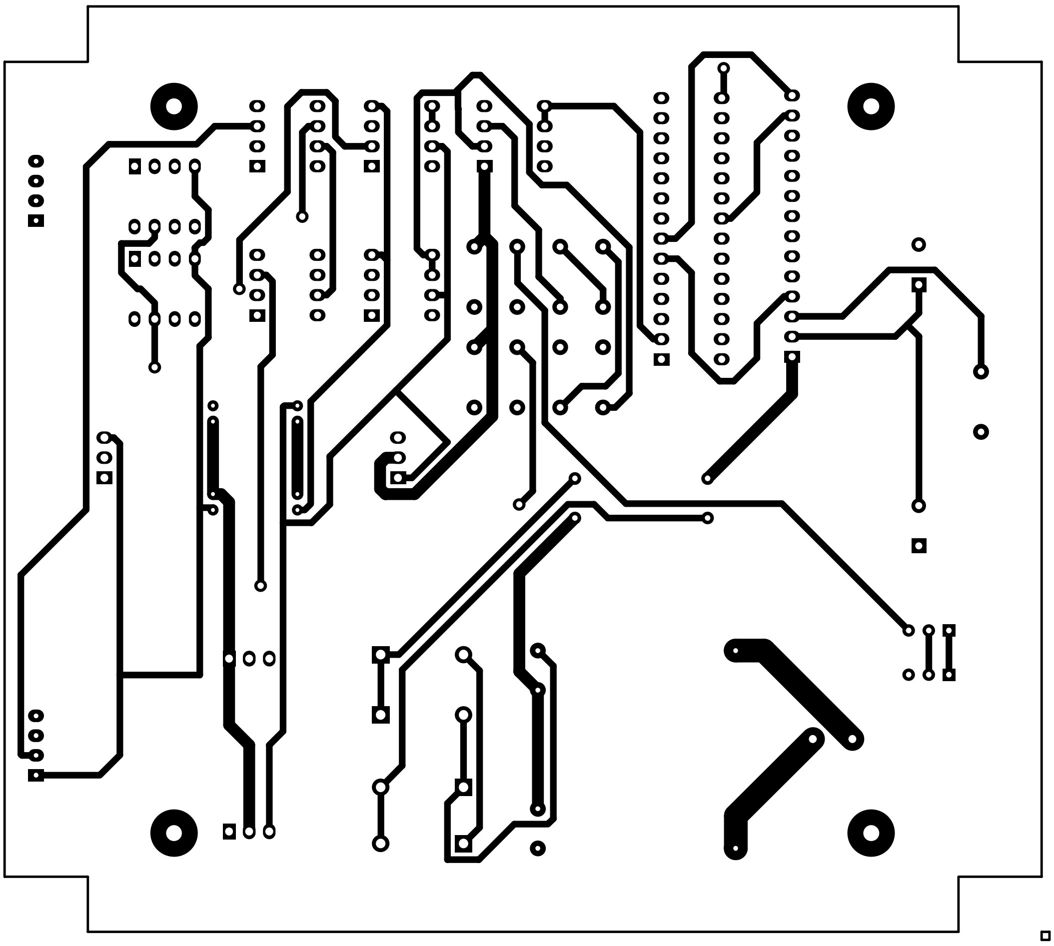


Figure B.5 PCB bottom layer (mirrored)

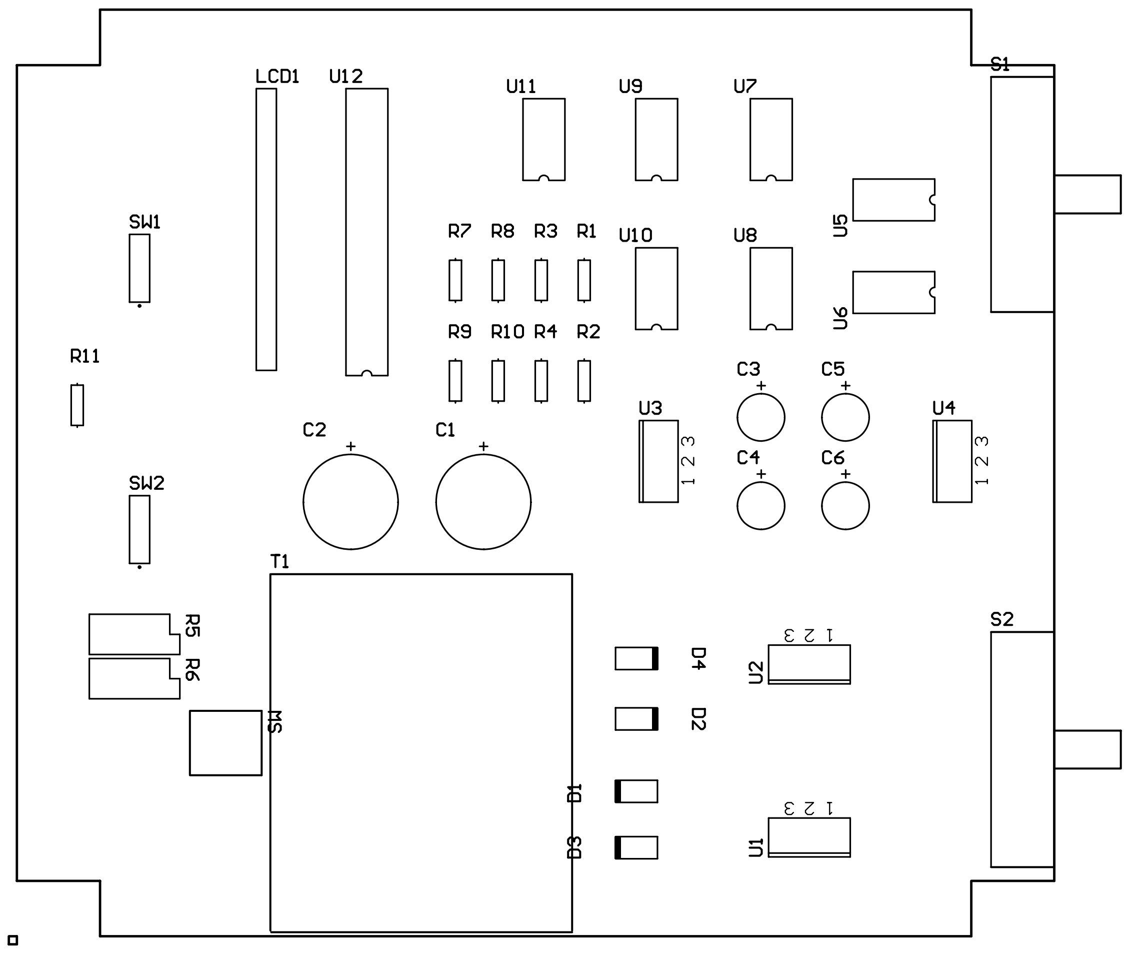


Figure B.6 PCB components placement

APPENDIX C. LISTING: MICROCONTROLLER CODE

```
/*
This program was coded as a part of the graduate study titled "DEVELOPMENT OF A PRESSURE AND FLOW
ANALYZER AS A TEST EQUIPMENT" by Gulsah Kacur in Bogazici University, Biomedical Engineering
Institute.

June 2005.
*/

#include <m8c.h>        /* Part specific constants and macros */
#include "PSoCAPI.h"    /* PSoC API definitions for all User Modules */
#include <math.h>
#include <stdlib.h>

/* Custom definitions */
#define TSTBUTTON_MODE (PRT1DR)
#define TSTBUTTON_GAS (PRT0DR)
#define AIR    1
#define OXYGEN 2

/* Prototype decleration */
char *gm_ftoa (float x, char *str);

/* Look-up Tables including bitwise corrections related to the ADC*/
const float air_LUT[602] = {
0, 2.715, 3.5241, 4.153, 4.686, 5.157, 5.5836, 5.9764, 6.3423, 6.6862, 7.0117, 7.3214, 7.6175,
7.9015, 8.1749, 8.4387, 8.694, 8.9414, 9.1817, 9.4154, 9.6431, 9.8652, 10.082, 10.294, 10.502,
10.705, 10.904, 11.099, 11.291, 11.48, 11.665, 11.847, 12.026, 12.203, 12.377, 12.548, 12.717,
12.884, 13.048, 13.21, 13.37, 13.529, 13.685, 13.839, 13.992, 14.143, 14.292, 14.44, 14.586,
14.73, 14.873, 15.015, 15.155, 15.294, 15.432, 15.568, 15.703, 15.837, 15.97, 16.102, 16.232,
16.362, 16.49, 16.618, 16.744, 16.869, 16.994, 17.117, 17.24, 17.362, 17.482, 17.602, 17.721,
17.84, 17.957, 18.074, 18.19, 18.305, 18.419, 18.533, 18.646, 18.758, 18.87, 18.981, 19.091,
19.2, 19.309, 19.418, 19.525, 19.632, 19.739, 19.845, 19.95, 20.055, 20.159, 20.262, 20.365,
20.468, 20.57, 20.671, 20.772, 20.873, 20.973, 21.072, 21.171, 21.27, 21.368, 21.466, 21.563,
21.659, 21.756, 21.852, 21.947, 22.042, 22.136, 22.231, 22.324, 22.418, 22.511, 22.603, 22.695,
22.787, 22.878, 22.969, 23.06, 23.15, 23.24, 23.33, 23.419, 23.508, 23.596, 23.685, 23.772, 23.86,
23.947, 24.034, 24.12, 24.207, 24.293, 24.378, 24.463, 24.548, 24.633, 24.718, 24.802, 24.885,
24.969, 25.052, 25.135, 25.218, 25.3, 25.382, 25.464, 25.546, 25.627, 25.708, 25.789, 25.869,
25.95, 26.03, 26.109, 26.189, 26.268, 26.347, 26.426, 26.505, 26.583, 26.661, 26.739, 26.816,
26.894, 26.971, 27.048, 27.125, 27.201, 27.277, 27.353, 27.429, 27.505, 27.58, 27.655, 27.73,
27.805, 27.88, 27.954, 28.028, 28.102, 28.176, 28.25, 28.323, 28.396, 28.469, 28.542, 28.615,
28.687, 28.759, 28.831, 28.903, 28.975, 29.046, 29.118, 29.189, 29.26, 29.331, 29.401, 29.472,
29.542, 29.612, 29.682, 29.752, 29.821, 29.891, 29.96, 30.029, 30.098, 30.167, 30.236, 30.304,
30.372, 30.441, 30.509, 30.576, 30.644, 30.712, 30.779, 30.846, 30.913, 30.98, 31.047, 31.114,
31.18, 31.247, 31.313, 31.379, 31.445, 31.511, 31.576, 31.642, 31.707, 31.772, 31.837, 31.902,
31.967, 32.032, 32.097, 32.161, 32.225, 32.289, 32.353, 32.417, 32.481, 32.545, 32.608, 32.672,
32.735, 32.798, 32.861, 32.924, 32.987, 33.05, 33.112, 33.175, 33.237, 33.299, 33.361, 33.423,
33.485, 33.547, 33.608, 33.67, 33.731, 33.792, 33.853, 33.915, 33.975, 34.036, 34.097, 34.158,
34.218, 34.278, 34.339, 34.399, 34.459, 34.519, 34.579, 34.638, 34.698, 34.758, 34.817, 34.876,
34.935, 34.995, 35.054, 35.113, 35.171, 35.23, 35.289, 35.347, 35.406, 35.464, 35.522, 35.58,
35.638, 35.696, 35.754, 35.812, 35.869, 35.927, 35.984, 36.042, 36.099, 36.156, 36.213, 36.27,
36.327, 36.384, 36.441, 36.497, 36.554, 36.61, 36.667, 36.723, 36.779, 36.835, 36.891, 36.947,
37.003, 37.059, 37.115, 37.17, 37.226, 37.281, 37.336, 37.392, 37.447, 37.502, 37.557, 37.612,
37.667, 37.721, 37.776, 37.831, 37.885, 37.94, 37.994, 38.048, 38.103, 38.157, 38.211, 38.265,
38.319, 38.372, 38.426, 38.48, 38.533, 38.587, 38.64, 38.694, 38.747, 38.8, 38.853, 38.906,
38.959, 39.012, 39.065, 39.118, 39.171, 39.223, 39.276, 39.328, 39.381, 39.433, 39.485, 39.538,
39.59, 39.642, 39.694, 39.746, 39.798, 39.849, 39.901, 39.953, 40.004, 40.056, 40.107, 40.159,
40.21, 40.261, 40.312, 40.364, 40.415, 40.466, 40.517, 40.567, 40.618, 40.669, 40.72, 40.77,
40.821, 40.871, 40.922, 40.972, 41.022, 41.072, 41.123, 41.173, 41.223, 41.273, 41.323, 41.372,
41.422, 41.472, 41.522, 41.571, 41.621, 41.67, 41.72, 41.769, 41.818, 41.868, 41.917, 41.966,
42.015, 42.064, 42.113, 42.162, 42.211, 42.26, 42.308, 42.357, 42.406, 42.454, 42.503, 42.551,
42.6, 42.648, 42.696, 42.744, 42.793, 42.841, 42.889, 42.937, 42.985, 43.033, 43.08, 43.128,
```

```
43.176, 43.224, 43.271, 43.319, 43.366, 43.414, 43.461, 43.509, 43.556, 43.603, 43.65, 43.698,
43.745, 43.792, 43.839, 43.886, 43.933, 43.979, 44.026, 44.073, 44.12, 44.166, 44.213, 44.26,
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44.86, 44.906, 44.952, 44.998, 45.043, 45.089, 45.135, 45.18, 45.226, 45.271, 45.317, 45.362,
45.408, 45.453, 45.498, 45.543, 45.589, 45.634, 45.679, 45.724, 45.769, 45.814, 45.859, 45.903,
45.948, 45.993, 46.038, 46.082, 46.127, 46.172, 46.216, 46.261, 46.305, 46.35, 46.394, 46.438,
46.483, 46.527, 46.571, 46.615, 46.659, 46.704, 46.748, 46.792, 46.835, 46.879, 46.923, 46.967,
47.011, 47.055, 47.098, 47.142, 47.186, 47.229, 47.273, 47.316, 47.36, 47.403, 47.447, 47.49,
47.533, 47.576, 47.62, 47.663, 47.706, 47.749, 47.792, 47.835, 47.878, 47.921, 47.964, 48.007,
48.05, 48.092, 48.135, 48.178, 48.221, 48.263, 48.306, 48.348, 48.391, 48.433, 48.476, 48.518,
48.561, 48.603, 48.645, 48.688, 48.73, 48.772, 48.814, 48.856, 48.898, 48.94, 48.982, 49.024,
49.066, 49.108, 49.15, 49.192, 49.234, 49.275, 49.317, 49.359, 49.4, 49.442, 49.484, 49.525,
49.567, 49.608, 49.649, 49.691, 49.732, 49.774, 49.815, 49.856, 49.897, 49.938, 49.98, 50.021
};

const float o2_LUT[662] = {
0, 1.2417, 2.0024, 2.6044, 3.1186, 3.5748, 3.9892, 4.3715, 4.7282, 5.0638, 5.3817, 5.6844,
5.9739, 6.2517, 6.5193, 6.7777, 7.0276, 7.27, 7.5055, 7.7346, 7.9578, 8.1755, 8.3882, 8.5962,
8.7997, 8.9991, 9.1945, 9.3863, 9.5746, 9.7595, 9.9414, 10.12, 10.296, 10.47, 10.64, 10.808,
10.974, 11.138, 11.299, 11.459, 11.616, 11.771, 11.925, 12.076, 12.226, 12.375, 12.521, 12.666,
12.81, 12.952, 13.092, 13.232, 13.369, 13.506, 13.641, 13.775, 13.908, 14.04, 14.17, 14.3, 14.428,
14.555, 14.681, 14.806, 14.931, 15.054, 15.176, 15.298, 15.418, 15.538, 15.656, 15.774, 15.891,
16.008, 16.123, 16.238, 16.352, 16.465, 16.577, 16.689, 16.8, 16.91, 17.02, 17.129, 17.238,
17.345, 17.452, 17.559, 17.665, 17.77, 17.875, 17.979, 18.082, 18.185, 18.288, 18.389, 18.491,
18.592, 18.692, 18.792, 18.891, 18.99, 19.088, 19.186, 19.283, 19.38, 19.476, 19.572, 19.668,
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23.903, 23.982, 24.061, 24.139, 24.217, 24.295, 24.373, 24.451, 24.528, 24.605, 24.682, 24.758,
24.835, 24.911, 24.987, 25.062, 25.138, 25.213, 25.288, 25.363, 25.437, 25.512, 25.586, 25.66,
25.734, 25.807, 25.88, 25.954, 26.027, 26.099, 26.172, 26.244, 26.316, 26.388, 26.46, 26.532,
26.603, 26.674, 26.745, 26.816, 26.887, 26.958, 27.028, 27.098, 27.168, 27.238, 27.307, 27.377,
27.446, 27.515, 27.584, 27.653, 27.722, 27.79, 27.858, 27.927, 27.995, 28.062, 28.13, 28.198,
28.265, 28.332, 28.399, 28.466, 28.533, 28.599, 28.666, 28.732, 28.798, 28.864, 28.93, 28.996,
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30.594, 30.656, 30.718, 30.78, 30.842, 30.904, 30.966, 31.027, 31.088, 31.15, 31.211, 31.272,
31.333, 31.394, 31.454, 31.515, 31.576, 31.636, 31.696, 31.756, 31.816, 31.876, 31.936, 31.996,
32.055, 32.115, 32.174, 32.234, 32.293, 32.352, 32.411, 32.47, 32.528, 32.587, 32.646, 32.704,
32.762, 32.821, 32.879, 32.937, 32.995, 33.053, 33.11, 33.168, 33.226, 33.283, 33.34, 33.398,
33.455, 33.512, 33.569, 33.626, 33.683, 33.739, 33.796, 33.853, 33.909, 33.965, 34.022, 34.078,
34.134, 34.19, 34.246, 34.301, 34.357, 34.413, 34.468, 34.524, 34.579, 34.634, 34.69, 34.745,
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37.959, 38.009, 38.06, 38.11, 38.16, 38.211, 38.261, 38.311, 38.361, 38.411, 38.461, 38.511,
38.56, 38.61, 38.66, 38.709, 38.759, 38.808, 38.858, 38.907, 38.956, 39.006, 39.055, 39.104,
39.153, 39.202, 39.251, 39.3, 39.349, 39.397, 39.446, 39.495, 39.543, 39.592, 39.64, 39.689,
39.737, 39.785, 39.833, 39.882, 39.93, 39.978, 40.026, 40.074, 40.122, 40.169, 40.217, 40.265,
40.313, 40.36, 40.408, 40.455, 40.503, 40.55, 40.597, 40.645, 40.692, 40.739, 40.786, 40.833,
40.88, 40.927, 40.974, 41.021, 41.068, 41.115, 41.161, 41.208, 41.255, 41.301, 41.348, 41.394,
41.44, 41.487, 41.533, 41.579, 41.625, 41.672, 41.718, 41.764, 41.81, 41.856, 41.902, 41.947,
41.993, 42.039, 42.085, 42.13, 42.176, 42.221, 42.267, 42.312, 42.358, 42.403, 42.449, 42.494,
42.539, 42.584, 42.629, 42.674, 42.719, 42.764, 42.809, 42.854, 42.899, 42.944, 42.989, 43.033,
43.078, 43.123, 43.167, 43.212, 43.256, 43.301, 43.345, 43.389, 43.434, 43.478, 43.522, 43.566,
43.611, 43.655, 43.699, 43.743, 43.787, 43.831, 43.874, 43.918, 43.962, 44.006, 44.05, 44.093,
44.137, 44.18, 44.224, 44.267, 44.311, 44.354, 44.398, 44.441, 44.484, 44.527, 44.571, 44.614,
44.657, 44.7, 44.743, 44.786, 44.829, 44.872, 44.915, 44.958, 45, 45.043, 45.086, 45.129, 45.171,
45.214, 45.256, 45.299, 45.341, 45.384, 45.426, 45.469, 45.511, 45.553, 45.595, 45.638, 45.68,
45.722, 45.764, 45.806, 45.848, 45.89, 45.932, 45.974, 46.016, 46.058, 46.1, 46.141, 46.183,
46.225, 46.266, 46.308, 46.35, 46.391, 46.433, 46.474, 46.515, 46.557, 46.598, 46.64, 46.681,
46.722, 46.763, 46.804, 46.846, 46.887, 46.928, 46.969, 47.01, 47.051, 47.092, 47.133, 47.174,
```

```
47.214, 47.255, 47.296, 47.337, 47.377, 47.418, 47.459, 47.499, 47.54, 47.58, 47.621, 47.661,
47.702, 47.742, 47.782, 47.823, 47.863, 47.903, 47.943, 47.984, 48.024, 48.064, 48.104, 48.144,
48.184, 48.224, 48.264, 48.304, 48.344, 48.384, 48.424, 48.463, 48.503, 48.543, 48.583, 48.622,
48.662, 48.701, 48.741, 48.781, 48.82, 48.86, 48.899, 48.938, 48.978, 49.017, 49.056, 49.096,
49.135, 49.174, 49.213, 49.253, 49.292, 49.331, 49.37, 49.409, 49.448, 49.487, 49.526, 49.565,
49.604, 49.643, 49.681, 49.72, 49.759, 49.798, 49.837, 49.875, 49.914, 49.953, 49.991, 50.03
};

void main()
{
        int P1, P2;
        float S1_mmHg, S2_mmHg;
        float S1_cmH2O, S2_cmH2O;
        float Diff_volt;
        float Flow_LPM, Diff_cmH2O, temp, cal;
        int i, Pr1[3], Pr2[3], index, Diff_bit;
        long int ii;
        char mode = 0, GasSelect = 1, ModeReadyFlag = 1, GasReadyFlag = 1;
        char t_1[10], t_2[10];

        /* Enable global interrupts */
        M8C_EnableGInt;

        /* Initializes LCD to use the multi-line */
        LCD_1_Init();

        /* Performs required initialization for PGA_1, PGA_2 and PGA_3 with set power level */
        PGA_1_Start(PGA_1_MEDPOWER);
        PGA_2_Start(PGA_2_MEDPOWER);
        PGA_3_Start(PGA_3_MEDPOWER);

        /* Performs required initialization for AD Converters and sets the power levels */
        DUALADC_1_Start(DUALADC_1_HIGHPOWER);
        DUALADC_1_SetResolution(12);
        ADCINC_1_Start(ADCINC_1_HIGHPOWER);


        /* Starts ADC's to read continuously */
        DUALADC_1_GetSamples(0);
        ADCINC_1_GetSamples(0);

        /* Displays Info */
        LCD_1_Start();
        LCD_1_Position (0,0);
        LCD_1_PrCString("P & F ANALYZER");
        LCD_1_Position (1,0);
        LCD_1_PrCString("B1-Gas, B2-Mode");

/* Waits until the gas selected */
        while(TSTBUTTON_GAS != 1);
        for(i = 1 ; i < 10000; i++);

        for(;;){

                /* Mode Selection */
                if(ModeReadyFlag){
                        if (mode == 4)
                                mode = 1;
                        else
                                mode = mode + 1;

                        ModeReadyFlag = 0;
                };

                /* Gas Selection */
```

```
        if(GasReadyFlag){
                if (GasSelect == 2){
                        GasSelect = 1;
                        LCD_1_Start();
                        LCD_1_Position(0,0);
                        LCD_1_PrCString("AIR Selected");}
                else {
                        GasSelect = GasSelect + 1;
                        LCD_1_Start();
                        LCD_1_Position(0,0);
                        LCD_1_PrCString("OXYGEN Selected");
                }
                GasReadyFlag = 0;
        };

        /* prevents LCD from blinking and checks if the buttons are pressed */
        for(ii = 1 ; ii < 40000; ii++){
                if(TSTBUTTON_MODE == 1)
                        ModeReadyFlag = 1;
                if(TSTBUTTON_GAS == 1)
                        GasReadyFlag = 1;
        }

        switch (mode) {
                /* Zeroing Mode */
                case 1:
                        /* Waits for data to be ready */
                        while(DUALADC_1_fIsDataAvailable() == 0);
                        /* Gets Data from ADC Output2 and clear data ready flag */
                        P1 = DUALADC_1_iGetData1() - 2039;
                        P2 = DUALADC_1_iGetData2ClearFlag() - 2047;
                        itoa (t_1, P1, 10);
                        itoa (t_2, P2, 10);

                        LCD_1_Start();
                        LCD_1_Position (0,0);
                        LCD_1_PrCString("ZEROING MODE");
                        LCD_1_Position (1,0);
                        LCD_1_PrCString("1:");
                        LCD_1_Position (1,2);
                        LCD_1_PrString(t_1);
                        LCD_1_Position (1,9);
                        LCD_1_PrCString("2:");
                        LCD_1_Position (1,11);
                        LCD_1_PrString(t_2);
                        break;

                /* Pressure Measurement */
                case 2:
                        /* Waits for data to be ready */
                        while(DUALADC_1_fIsDataAvailable() == 0);
                        /* Gets Data from ADC Output1 and clear data ready flag*/
                        P1 = DUALADC_1_iGetData1ClearFlag();

                        /* ADC Response Corrections */
                        if (P1 <= 1772)
                                P1 = (1.0658 * P1) + 113.5;
                        else if ((P1 > 1772) && (P1 < 2025))
                                P1 = 2001 + (P1 - 1772) / 9.4;
                        else if ((P1 >= 2025) && (P1 <= 2263))
                                P1 = (1.0132 * P1) - 27.388;
                        else if ((P1 > 2263) && (P1 < 2527))
                                P1 = 2266 + (P1 - 2263)/ 9.8;
                        else /* if (P1 >= 2527)*/
                                P1 = (1.0246 * P1) - 294.8;
```

```
                /* Pressure Calculations */
                S1_mmHg = -768.7030 + (P1 * 0.377);
                S1_cmH2O = S1_mmHg * 1.3595;

                LCD_1_Start();
                LCD_1_Position (0,0);
                LCD_1_PrCString("S");
                LCD_1_Position (0,2);
                LCD_1_PrString(gm_ftoa(S1_mmHg,t_1));
                LCD_1_Position(0,11);
                LCD_1_PrCString("mmHg");
                LCD_1_Position(1,0);
                LCD_1_PrCString("1");
                LCD_1_Position(1,2);
                LCD_1_PrString(gm_ftoa(S1_cmH2O,t_2));
                LCD_1_Position(1,11);
                LCD_1_PrCString("cmH2O");
                break;

        /* Differential Pressure Measurement in cmH2O */
        case 3:
                /* Loop until value ready */
                while(ADCINC_1_fIsDataAvailable() == 0);

                /* Clear ADC flag and get data */
                Diff_bit = ADCINC_1_wClearFlagGetData();

                /* Differential pressure calculations */
                Diff_volt = (Diff_bit * 5.00 / 1020.00) / 10.00;
                Diff_cmH2O = (Diff_volt / 0.0065) * 1.3595;

                if(GasSelect == AIR){

                        LCD_1_Start();
                        LCD_1_Position(0,0);
                        LCD_1_PrCString("Diff");
                        LCD_1_Position (0,9);
                        LCD_1_PrString(gm_ftoa(Diff_cmH2O,t_1));
                        LCD_1_Position(1,0);
                        LCD_1_PrCString("AIR");
                        LCD_1_Position(1,11);
                        LCD_1_PrCString("cmH2O");
                        break;
                }
                else if (GasSelect == OXYGEN){
                        LCD_1_Start();
                        LCD_1_Position(0,0);
                        LCD_1_PrCString("Diff");
                        LCD_1_Position (0,9);
                        LCD_1_PrString(gm_ftoa(Diff_cmH2O,t_1));
                        LCD_1_Position(1,0);
                        LCD_1_PrCString("O2");
                        LCD_1_Position(1,11);
                        LCD_1_PrCString("cmH2O");
                        break;
                }

                /* Flow Measurement in LPM */
                case 4:
                        if(GasSelect == AIR){

                        /* Loop until value ready */
                        while(ADCINC_1_fIsDataAvailable() == 0);

```

```
                        /* Clear ADC flag and get data */
                        index = ADCINC_1_wClearFlagGetData();

                        if(index >= 602){
                                LCD_1_Start();
                                LCD_1_Position(0,0);
                                LCD_1_PrCString("WARNING");
                                LCD_1_Position(1,0);
                                LCD_1_PrCString("Out of Range");
                                break;
                        }

                        LCD_1_Start();
                    LCD_1_Position(0,0);
                        LCD_1_PrCString("Flow");
                        LCD_1_Position (0,9);
                        LCD_1_PrString(gm_ftoa(air_LUT[index],t_1));
                        LCD_1_Position(1,0);
                        LCD_1_PrCString("AIR");
                        LCD_1_Position(1,11);
                        LCD_1_PrCString("lpm");
                        break;
                        }/*End of AIR flow*/

                        else if (GasSelect == OXYGEN){

                                /* Loop until value ready */
                                while(ADCINC_1_fIsDataAvailable() == 0);

                                /* Clear ADC flag and get data */
                                index = ADCINC_1_wClearFlagGetData();

                                if(index >= 662){
                                        LCD_1_Start();
                                        LCD_1_Position(0,0);
                                        LCD_1_PrCString("WARNING");
                                        LCD_1_Position(1,0);
                                        LCD_1_PrCString("Out of Range");
                                        break;
                                }

                                LCD_1_Start();
                            LCD_1_Position(0,0);
                                LCD_1_PrCString("Flow");
                                LCD_1_Position (0,9);
                                LCD_1_PrString(gm_ftoa(o2_LUT[index],t_1));
                                LCD_1_Position(1,0);
                                LCD_1_PrCString("O2");
                                LCD_1_Position(1,11);
                                LCD_1_PrCString("lpm");
                                break;
                        }/* End of OXYGEN flow */
                }/* End of SWITCH */
        }/* End of continuous 'for_loop' */
}/* End of MAIN */

char *gm_ftoa (float x, char *str)
{

        int ie, i, k, ndig;
        int prec = 2; /* decimal places*/
        double y;
        char *start;

        start = str;
```

```
/* based on percission sets number digits */
ndig = prec + 1;

/* initially sets exp. power to zero, e.g. XXX.XE+00 */
ie = 0;

/* if x negative, write minus and reverse */
if ( x < 0)
{
  *str++ = '-';
  x = -x;
}

/* if x < 0.0 then multiply by 10 until it becomes a number between 1.0 and 10.0 */
if (x != 0.0)
{
        while (x < 1.0)
        {
          x = x * 10.0;
          ie--;
        }
}

/* if x > 10 then shift it down */
while (x >= 10.0)
{
        x = x * (1.0 / 10.0);
        ie++;
}

ndig = ndig + ie;

/* - rounding - */
y = 1;
/* finds lest significant digit */
for (i = 1; i < ndig; i++){
  /* multiply by 1/10 is faster than dividing */
  y = y * (1.0/10.0);
}

/*adds rounding*/
x = x + y * (1.0/2.0);

/* repairs rounding disasters */
if (x >= 10.0)
{
        x = 1.0;
        ie++;
        ndig++;
}

/* checks if the number is less than 1.0 */
if (ie < 0)
{
        *str++ = '0';
        *str++ = '.';
        if (ndig < 0)
                ie = ie - ndig; /* limit zeros if underflow */
        for (i = -1; i > ie; i--)
                *str++ = '0';
}

/* for each digit */
for (i = 0; i < ndig; i++)
```

```
        {
                float b;
                k = x;                          /* k = most significant digit */
                *str++ = k + '0';  /* output the char representation */
                if (i == ie)
                        *str++ = '.';  /* output a decimal point */
                b = (float)k;

                /* multiply by 10 before subtraction to remove errors from limited number of bits
                   in float. */
                b = b * 10.0;
                x = x * 10.0;

                /* subtracts k from x */
                x = x - b;
        }

        *str++ = '\0';

        /* returns the starting address */
        return start;
}/*End of gm_ftoa */
```

APPENDIX D. INSTRUCTIONS FOR USERS

When the pressure and flow analyzer is switched on, the screen in Figure D.2 (a) appears. If the button 1 (B1) is pressed, zeroing mode starts [in Figure D.2 (c)]. Two potentiometer is used for zero adjustment, top potentiometer is for Sensor 1, bottom potentiometer is for Sensor 2 (see the front panel view in Figure D).

Button 1 is used for gas selection in the flow measurement. If this button is pressed, the operator can select the gas (oxygen or air) flowing through the system [see Figure D.2(b)]. Button 2 is used for mode selection. The operator chooses the measurement s/he wants to perform by pressing this button. The instrument can display the pressure applying to Sensor 1 [Figure D.2(d)], or the differential pressure between two sensors [Figure D.2(e)], or gas flow rate passing through the airway resistor [see Figure D.2(g) for oxygen flow, see Figure D.2(f) for oxygen]. For flow measurement above 50 lpm, a message appears on the screen indicating that flow rate is out of the limits [Figure D.2(h)].

Airway resistor connection to the analyzer is also seen on Figure D.3.

D.1 Calibration Task

The most vulnerable components of the instrument are the pressure sensors. They must be never exposed to pressure outside their operating range ($\pm$15 psi) and must also be protected from objects such as dust, grease, and etc.

Although all precautions are taken, the instrument may measure different from actual values as time passes. Disagreement between the readings might be observed when the instrument is tested against a calibrated reference test equipment. In such case, enough data should be collected and a new relationship should be established to deduce a calibration factor. This calibration factor, for the pressure measurement,

Figure D.1 Front Panel

(a)

(b)

(c)

(d)

(e)

(f)

(g)

(h)

Figure D.2 Display views

Figure D.3 Connection of the airway resistor to the analyzer

may be added to the code immediately after line 260 to recalculate corrected pressure values. For the flow measurements, this calibration factor may be multiplied by the values returned from the look-up tables (line 338 and line 367 for air and oxygen respectively) before passing them to the gm_ftoa function as an input argument.

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