

DESIGN OF A RESPIRATORY MONITOR FOR THE TEACHING LABORATORY

by

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ABSTRACT

The goal of this thesis was to develop a cost-effective, portable and user-friendly respiratory monitor that gives the opportunity to observe pressure, flow and volume waveforms as well as time based ventilation data. An electronic hardware system has been built up in order to acquire analog pressure signals from the ventilator and transmit them to a PC properly. A software system has been developed in order to process the input pressure signals; obtain flow, volume and time based ventilation data; and to display both of the measured data at the user interface window. Also a mechanical adult lung simulation system has been adapted in order to give the occasion of evaluation at standard and realistic conditions. Analog pressure signals have been acquired by a differential pressure sensor and digitalized through a standard analog to digital converter after being amplified for precise measurements. Interface between the hardware and software systems has been achieved by standard parallel port communication. Data registers of the parallel port have been used for 8-bit differential pressure signal transmission. The software system of the respiratory monitor has been created by using LabVIEW program.

Keywords: Respiratory Monitor, Ventilator, Testing Instrument, Respiration

DERS LABORATUARI İÇİN SOLUNUM MONİTÖRÜ TASARIMI

ÖZET

Bu tez çalışmasının amacı düşük maliyetli, taşınabilir ve kolay kullanılabilen; basınç, akı, hacim ve solunum zamanlarının ölçümünü yapabilen bir solunum monitörü geliştirmektir. Vantilatörden gelen analog basınç sinyallerini elde etmek ve bir bilgisayara aktarmak için bir elektronik donanım ünitesi geliştirilmiştir. Basınç sinyallerini işleyerek akı, hacim ve solunum zamanı verilerini hesaplamak ve hesaplanan verileri görüntülemek için bir yazılım programı geliştirilmiştir. Ölçüm sırasında standart ve gerçek ortam koşullarını oluşturmak amacı ile mekanik bir yetişkin ciğer simülatörü sisteme adapte edilmiştir. Bir fark basınç sensörü tarafından elde edilen basınç sinyalleri, amplifikatörler tarafından yükseltildikten sonra bir analog-dijital çevirici tarafından sayısala dönüştürülmüştür. Donanım ve yazılım sistemleri arasındaki iletişim standart paralel port haberleşmesi kullanılarak gerçekleştirilmiştir. Solunum monitörünün yazılım sistemi LabVIEW programı kullanılarak oluşturulmuştur.

Anahtar Sözcükler: Solunum Monitörü, Vantilatör, Test Cihazı, Solunum

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LIST OF SYMBOLS

K	Airway resistance constant
ΔP	Differential pressure
V	Volume
η	Gas velocity
l	Length of the tube
r	Radius of the tube

LIST OF ABBREVIATIONS

VT	Tidal volume
IRV	Inspiratory reserve volume
ERV	Expiratory reserve volume
RV	Residual volume
IC	Inspiratory capacity
FRC	Functional residual capacity
VC	Vital capacity
TLC	Total lung capacity
I:E	Inspiratory and expiratory time ratio
A/D	Analog to digital converter
VI	Virtual instrument

1. INTRODUCTION

A mechanical ventilator is simply a machine that is used to replace or supplement the natural function of breathing. Various types of ventilators have been developed over the years. Ventilators can be separated into two categories as controller ventilators and assistor ventilators. Both of these ventilators play an important role in the treatment of certain respiratory disorders.

Because ventilators are crucial in respiratory therapy, every critical care and anesthesia ventilator should be tested under true load conditions before being used for patient care and, periodically to verify that it combines with criteria established by various standards agencies and by the ventilator manufacturer [1].

In this study, a cost-effective, portable and simple to use ventilation monitoring unit has been developed for the laboratory use. A mechanical system includes the simulator for the adult pulmonary system that will be connected to the ventilator in order to establish the standardized and realistic conditions. The electronic hardware acquires the pressure signals and transmits them to a personal computer in a proper state. Software system covers the calculation and measurement of the desired parameters and the user interface part. Finally the mechanical system includes the simulation of an adult pulmonary system during the evaluation process of a ventilator in order to establish standardized and realistic conditions.

The respiratory monitor allows the user to observe the pressure, flow and the volume waveforms of the ventilator as well as the time based ventilation data.

1.1 Motivation

The motivation of this thesis is to generate a cost-effective, portable and simple to use respiratory monitor that can be used at the Biomedical Instrumentation Laboratory for the respiratory measurement experiment. The computer performs the LabVIEW program in order to process and monitor the ventilation parameters. An electronic hardware system is developed to transmit analog ventilation data to a computer and, a mechanical bellow in order to simulate realistic conditions and true load.

1.2 Objectives

The aim of this thesis study is to develop a monitoring unit that allows the user to observe respiratory pressure, flow and volume waveforms on a personal computer.

1.3 Thesis Outline

Chapter 1 introduces the thesis study. It gives a general overview about the thesis. Theoretical background about ventilators and ventilator testing units is introduced in Chapter 2. Chapter 2 consists of description and brief history of ventilators, ventilator testers and resources about ventilator testing units. Under resources section, theoretic definitions of the terms related to the ventilator testing units and formulas of the related parameters are introduced. Chapter 3 describes the setup of the respiratory monitor. Chapter 4 is describing the development of the respiratory monitor. It describes each state at the development process. The conclusion part is presented in Chapter 5. A CD that includes setup file Respiratory Monitor.exe, source code and read me file of the program is attached at the back cover.

2. OVERVIEW OF VENTILATORS

2.1 Description and Brief History of Ventilators

In general, a medical device is an instrument, apparatus, or machine used to prevent, diagnose or treat a disease. It also serves to detect, measure, restore or modify the structure or function of the body for a given health purpose [2].

A medical ventilator is a medical device designed to provide mechanical ventilation to a patient. Ventilators are chiefly used in intensive care medicine and emergency medicine and, in anesthesia [3].

Demonstrations in animals as early as 1555 by Vesalius and 1667 by Hook showed that life could be sustained by inflation of the lungs with a bellows inserted into the mouth while the chest cavity was opened. These first primitive attempts were the beginnings of ventilator therapy. In humans, ventilator therapy was initially developed to allow physicians to operate on an open chest; without proper inflation, the lungs would collapse upon opening the chest and the patient's condition would quickly deteriorate. Technological advances and greater attention to the importance of optimal ventilation led to improvements in respiratory therapy. The development of tank respirators in the 1920s and 30s was crucial in the treatment of polio patients who developed respiratory insufficiency. Today, respirators are used for a variety of reasons and are much more sophisticated than the bellows used in 1555, allowing for more specific control of a patient's oxygen delivery [4].

In its simplest form, a ventilator consists of a compressible air reservoir, air and oxygen supplies, a set of valves and tubes, and a disposable or reusable "patient tubing". The air reservoir is pneumatically compressed several times a minute to deliver an air/oxygen mixture to the patient; when overpressure is released, the patient will exhale passively due to the lungs' elasticity. The oxygen content of the inspired gas can be set from 21 percent (ambient air) to 100 percent (pure oxygen). Pressure and flow characteristics can be set mechanically or electronically [3].

Modern ventilators are electronically controlled by a small embedded system to allow exact adaptation of pressure and flow characteristics to an individual patient's needs. Fine-tuned ventilator settings also serve to make ventilation more tolerable for the patient [3]. Below in Figure 2.1 a portable ventilator is shown.



Figure 2.1 A portable ventilator

2.2 Ventilator Testers

There are many situations in which the performance of ventilators or similar respiratory care equipment must be tested. One of the most common is in the Biomedical Engineering department of a hospital, where preventive maintenance testing is routinely performed for a wide variety of medical equipment. But there are many stages in a device's life cycle at which testing is warranted. Development engineers need to test new ideas and mechanisms as new ventilator technologies are developed, and Production and Quality Control engineers must calibrate and assure the performance of the machines during manufacture.

In practice, many different types of equipment are used to test or otherwise assess the performance of ventilators and similar respiratory care equipment. These range from simple flow measuring devices, to electronically instrumented, flow meter based

“ventilator analyzers”, to fixed and variable volume test lungs, and finally all the way to instrumented, computer controlled, spontaneously breathing lung simulators. While each has their own advantages and disadvantages in terms of capabilities, convenience and cost, for most testing applications it is essential that at least the basic parameters of pulmonary compliance and resistance are realistically simulated.

A realistic compliance load, in particular, has profound affects on gas flow, the action of many ventilator components and on the performance of ventilators and similar devices in general. In fact, standards-setting organizations like ASTM and ISO mandate that ventilators be tested under conditions simulating actual use, and using realistic compliance and resistance loads in particular. Gas is a compressible substance, and even at the relatively low pressures used in Respiratory Care, volume and flow rate measurements are very sensitive to changes in pressure. This characteristic affects ventilator testing in a number of ways [5].

In order to test the accuracy and reliability of flow and volume sensors that are so frequently part of modern ventilators, pressures and changes in pressures must be realistically simulated during testing of such machines.

Many components of ventilators, and especially many components of breathing circuits used in conjunction with ventilators, are designed to work with and respond to specific flow and pressure conditions in order to function properly. Only by realistically simulating the conditions under which these components are designed to operate, they can be made to work correctly. Testing under any other conditions will cause these components to function differently, which can affect the apparent overall performance of the ventilator, and which denies the technician the ability to accurately assess the performance of the components themselves [5].

2.3 Resources

In this study a simple respiratory monitor that gives the opportunity of observation of the performance of medical ventilators to the user has been developed. The developed

respiratory monitor measures and displays differential pressure, flow, volume waveforms and time based ventilation data.

2.3.1 Respiration

Respiration refers to all the processes involved in the exchange of oxygen and carbon dioxide between the cells and the external environment. It includes ventilation, the diffusion of oxygen and carbon dioxide, and the transport of the gases in the blood. Cellular respiration deals with the cell's utilization of oxygen and release of carbon dioxide in its metabolism. The term refers to the overall process by which oxygen is abstracted from air and is transported to the cells for the oxidation of organic molecules while carbon dioxide (CO_2) and water, the products of oxidation, are returned to the environment. Mechanics of breathing is given below in Figure 2.2.

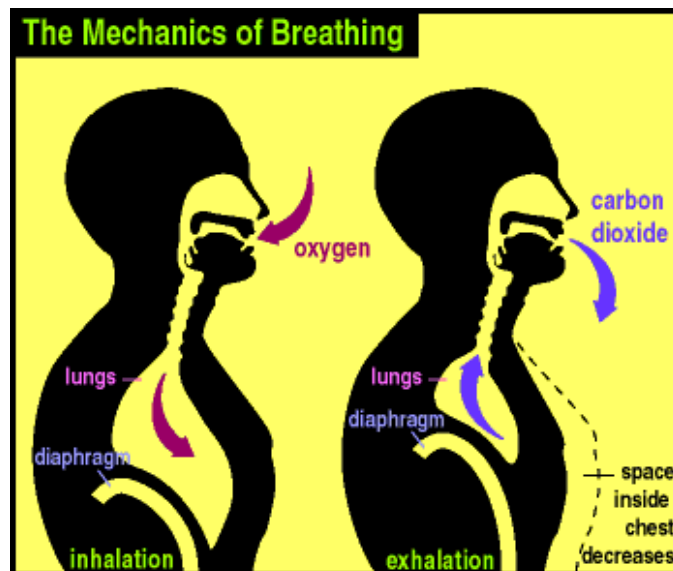


Figure 2.2 Example schema of breathing mechanics [6]

Respiration includes three separate but related functions that are ventilation (breathing), gas exchange and oxygen utilization. Oxygen utilization by the tissues is also known as internal respiration and arises in the energy-liberating reactions of cell respiration. Gas exchange function occurs between the air and blood in the lungs and between the blood and other tissues of the body.

Ventilation is the mechanical process that moves air into and out of the lungs. The inspired air passes through the nasal chambers, larynx, trachea, right and left bronchi of the lungs, and finally the lung air sacs, called the alveoli. The expired carbon dioxide returns via the same passageways to the outside environment. Since the oxygen concentration of air is higher in the lungs than in the blood, oxygen diffuses from air to blood. Carbon dioxide, conversely, moves from the blood to the air within the lungs by diffusing down its concentration gradient. In Figure 2.3 respiratory organs are given below.

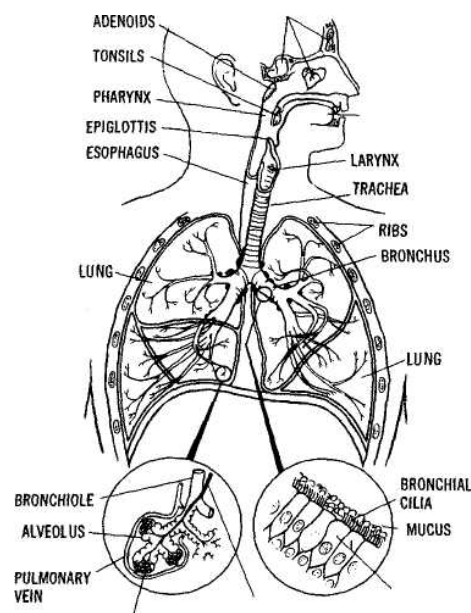


Figure 2.3 Respiratory organs [7]

2.3.2 Airway Pressure

Airway pressure is the force applied on the airway pathway by the weight of tiny particles of air molecules. Differential pressure over the airways is positive during inspiration while it is negative during expiration. Pressure can be measured directly by pressure sensors and value of it is proportionally related with flow and volume. In the Figure 2.4 below a pulmonary pressure waveform is given.

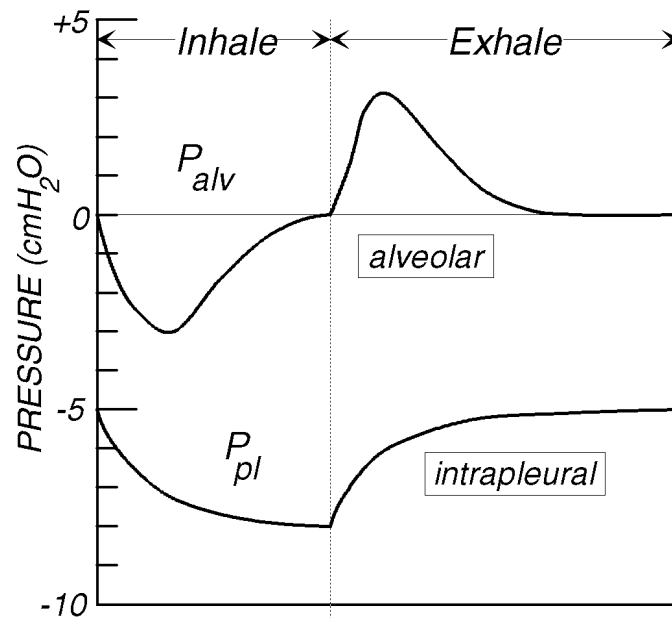


Figure 2.4 Example of pulmonary pressure waveform [8]

2.3.3 Flow

Air flow is the motion characteristics of the gases. Air flow occurs only when there is a difference between pressures. Air will flow from a region of high pressure to one of low pressure - the bigger the difference, the faster the flow. Thus air flows in during inspiration because the alveolar pressure is less than the pressure at the mouth; air flows out during expiration because alveolar pressure exceeds the pressure at the mouth such that to double the flow rate one must quadruple the driving pressure

When air flows at higher velocities, especially through an airway with irregular walls, flow is generally disorganized, even chaotic, and tends to form eddies. This is called turbulent flow, and is found mainly in the largest airways, like the trachea. [9].



Figure 2.5 Example of turbulent flow [9]

A relatively large driving pressure is required to sustain turbulent flow. Driving pressure during turbulent flow is in fact proportional to the square of the flow [9].

$$\Delta P = K \times Flow^2 \quad (\text{cmH}_2\text{O}) \quad (1)$$

Whereas, ΔP is the pressure difference across the airway resistance in cmH_2O and K is the airway resistor constant in $\text{cmH}_2\text{O} / (\text{l} / \text{s})^2$. An example waveform to the relationship between airway pressure and flow is shown in Figure 2.6.



Figure 2.6 Example waveform of the pressure and flow relationship in a volume controlled mode [10]

When flow is at a low velocity and going through narrow tubes, it tends to be more orderly and streamlined and to flow in a straight line. This type of flow is called laminar flow. Unlike turbulent flow, laminar flow is directly proportional to the driving pressure. During quiet breathing, laminar flow exists from the medium-sized bronchi down to the level of the bronchioles. During exercise, when the air flow is more rapid, laminar flow may be confined to the smallest airways.

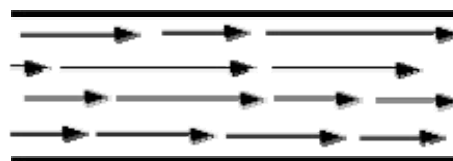


Figure 2.7 Example of laminar flow [9]

Laminar flow can be described by Poiseuille's Law [9];

$$\Delta P = Flow \times (8\eta l / \pi.r^4) \quad (\text{cmH}_2\text{O}) \quad (2)$$

Where, ΔP is the pressure difference across the airway resistance, η is gas velocity, l is length of the tube and r is radius of the tube.

In Figure 2.8 below, locations of flow patterns in lung is given. Turbulent flow is found at the airways while laminar flow is found at bronchioles. Transitional flow, which has some of the characteristics of both laminar and turbulent flow, is found between the two along the rest of the bronchial tree.

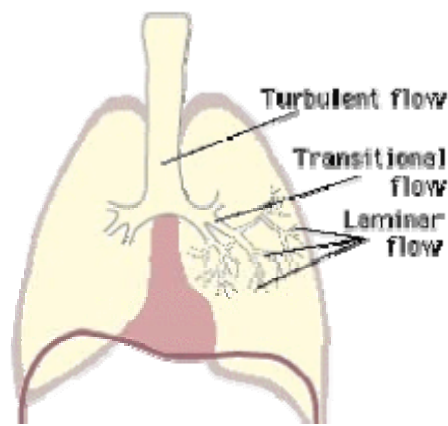


Figure 2.8 Locations of flow patterns [9]

2.3.4 Pulmonary Volumes and Capacities

Pulmonary volume is the amount of air in the lungs. In normal persons, the volume of air in the lungs depends primarily on body size and build. Additionally, volumes and capacities vary with body position. Most volumes and capacities decrease when lying down and increases when standing up. This is due to the tendency of the abdominal contents to press upward against the diaphragm when lying down and an increase in pulmonary blood volume decreasing the amount of available space for pulmonary air [11].

The pulmonary volumes can be broken down into four different lung volumes:

- Tidal Volume (VT)
- Inspiratory Reserve Volume (IRV)
- Expiratory Reserve Volume (ERV)

- Residual Volume (RV)

The pulmonary capacities can be broken down into four different lung volumes:

- Inspiratory Capacity (IC)
- Functional Residual Capacity (FRC)
- Vital Capacity (VC)
- Total Lung Capacity (TLC)

Respiratory changes in lung volume are shown in the diagram below in Figure 2.9.

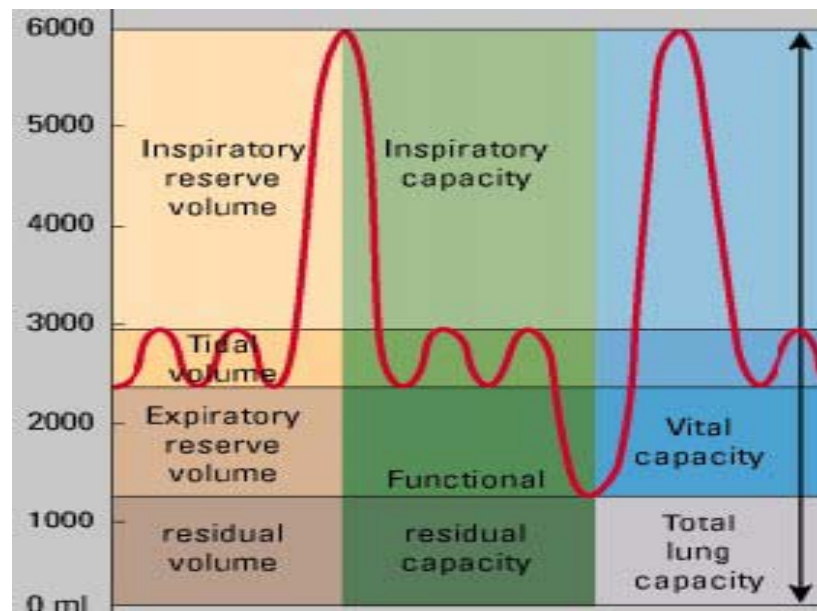


Figure 2.9 Respiratory changes in lung volume [11]

- *Tidal Volume (VT)* is the volume of air inspired or expired with each normal breath. V_t is normally 500 ml.
- *Inspiratory Reserve Volume (IRV)* is the extra volume of air that can be inspired over and beyond the normal tidal volume. IRV is approximately equal to 3000 ml.
- *Expiratory Reserve Volume (ERV)* is the extra volume of air that can be expired at the end of a normal tidal volume. ERV is approximately equal to 1100 ml.
- *Residual Volume (RV)* is the volume of air still remaining in the lungs after a forceful expiration. RV is typically 1200 ml.

- *Inspiratory capacity (IC)* equals the tidal volume plus the inspiratory reserve volume, or the amount of air that can be inspired beginning at the end of a normal expiration and distending the lungs to the maximum amount. It is normally about 3500 ml.
- *Functional Residual Capacity (FRC)* equals the expiratory reserve volume plus the residual volume. The FRC represents the amount at the end of a normal expiration, about 2300 ml.
- *Vital Capacity (VC)* equals the inspiratory reserve volume plus the tidal volume plus the expiratory reserve volume. This is the maximum amount of air that a person can expel from his lungs after first inspiring to his maximum extent, then exhaling to his maximum extent. It is normally about 4600 ml.
- *Total Lung Capacity (TLC)* is the maximum volume to which the lungs can be expanded with the greatest possible inspiratory effort, approximately 5800 ml.

Volume can be defined as the integral of flow in a mathematical form [1].

$$\Delta V = \int Flow(t).dt \quad (\text{Liter}) \quad (3)$$

2.3.5 Time Parameters of Respiration

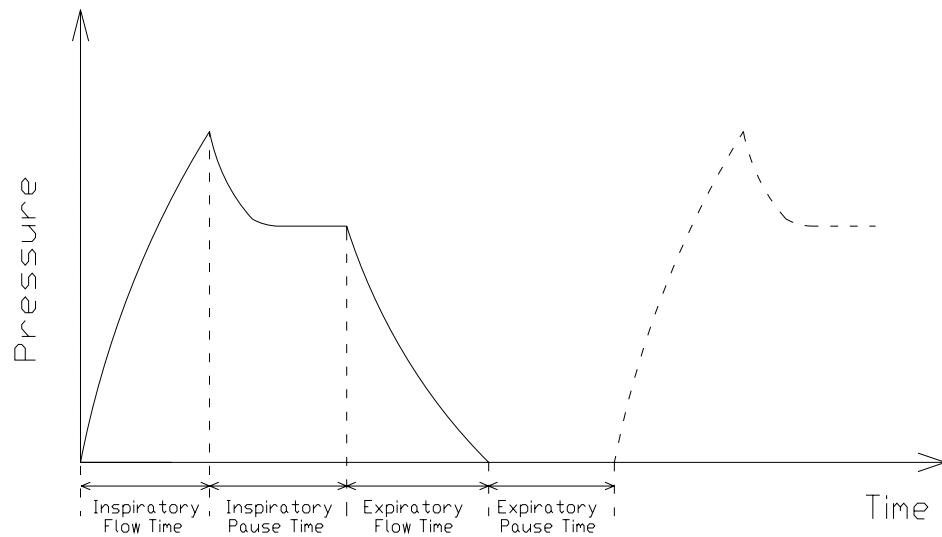


Figure 2.10 A typical Pressure-Time graph for volume controlled ventilator.

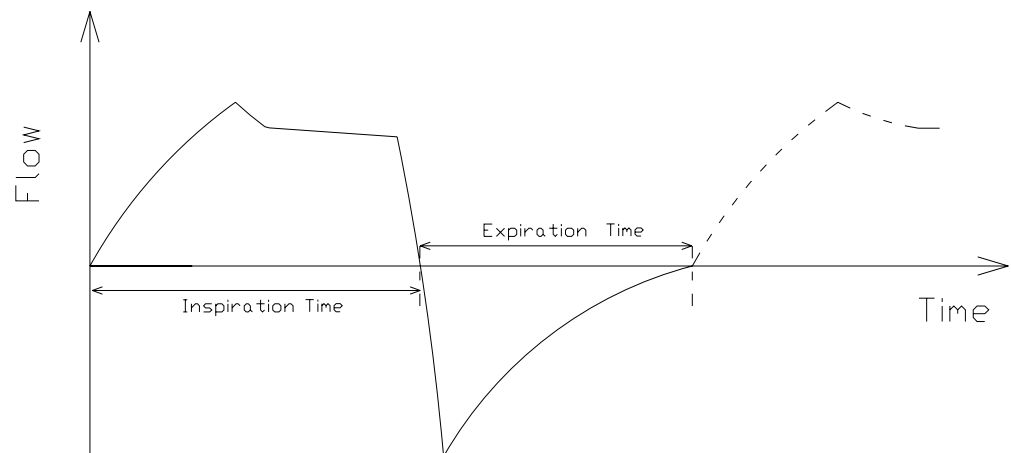


Figure 2.11 A typical Flow-Time graph for volume controlled ventilator.

- *Inspiratory Flow Time*; is the period between the beginning and end of inspiratory flow.
- *Inspiratory Pause Time*; is the period between the end of inspiratory flow and beginning of the expiratory flow.
- *Expiratory Flow Time*; is the period between the beginning and end of expiratory flow.

- *Expiratory Pause Time*; is the period between the end of expiratory flow and beginning of the inspiratory flow.
- *Inspiratory: Expiratory Phase Time Ratio (I:E Ratio)*; is the ratio of the inspiratory phase time to the expiratory phase time. I:E ratio is usually 1:2.
- *Period*; is the cycle time of ventilation. Period is equal to the total of inspiration and expiration phase time. Period can be defined as,

$$Period = Insp.FlowTime + Insp.PauseTime + Exp.FlowTime + Exp.PauseTime \quad (s)$$

- *Breath Rate*, is the number of breaths in a minute. Breath rate can be calculated by,

$$BreathRate = 60 / Period \quad (BPM) \quad (4)$$

2.3.6 Lung Simulator

The single lung simulator by Michigan Instruments is used in the thesis to realistically simulate the mechanics of the adult pulmonary system from the upper airway to the lung at the evaluation of ventilation parameters.

Michigan Instruments single lung provides an accurate simulation of the structures and mechanics of the adult pulmonary system. An elastomer bellows with a residual capacity typical of an adult is used to simulate the lung. Gas is inserted into the lung through a simulated airway. This insertion causes the vertical expansion of the lung, and corresponding rise of the top plate. The lung compliance is accurately set and adjusted using a precision steel alloy spring. Airway resistance is simulated using a turret with adjustable settings [12].

The lung simulator is shown below in Figure 3.19 and illustration of parabolic airway resistance is given in Figure 3.20.

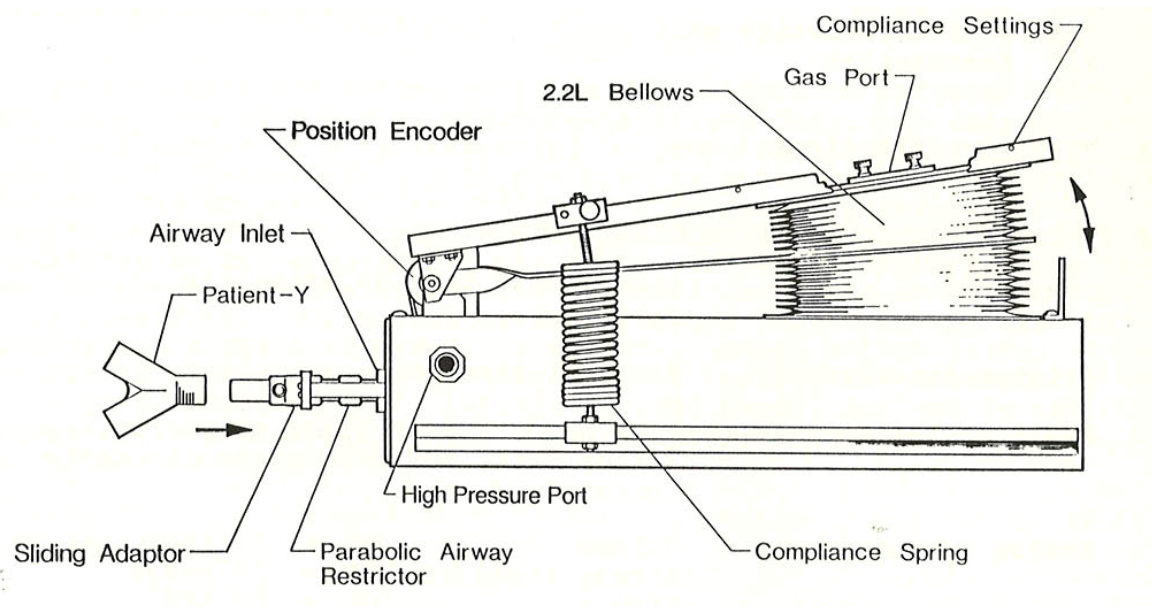


Figure 2.12 The lung simulator [12]

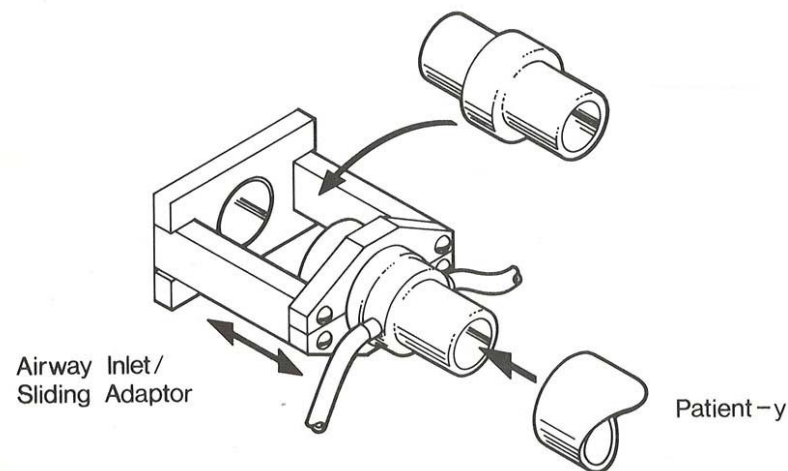


Figure 2.13 Illustration of parabolic airway resistance [12]

2.3.6.1 Airway Resistance

Airway resistance is the opposition to flow caused by the forces of friction. It is defined as the ratio of driving pressure to the rate of air flow. Resistance to flow in the airways depends on whether the flow is laminar or turbulent, on the dimensions of the airway, and on the viscosity of the gas [9].

In this study three different parabolic airway resistances by Michigan Instruments Rp5, Rp20 and Rp50 are used for the simulation of respiratory system. Parabolic airway resistors simulate airway resistance of a human patient more realistically than the linear airway resistors. The Rp5, Rp20 and Rp50 parabolic airway resistors that are used in this study are simply metal tubes with 9 mm, 7 mm and 5 mm. The airway resistance values and their K constants are given in Table 2.1 and the graphs shown below in Figure 2.14, Figure 2.15 and Figure 2.16 describe the flow rate versus pressure drop characteristic of each of the specified resistors.

Table 2.1
Airway resistor values [12]

Rp5	K = 2.7	cmH ₂ O/(l/s) ²
Rp20	K = 17.61	cmH ₂ O/(l/s) ²
Rp50	K = 108.7	cmH ₂ O/(l/s) ²

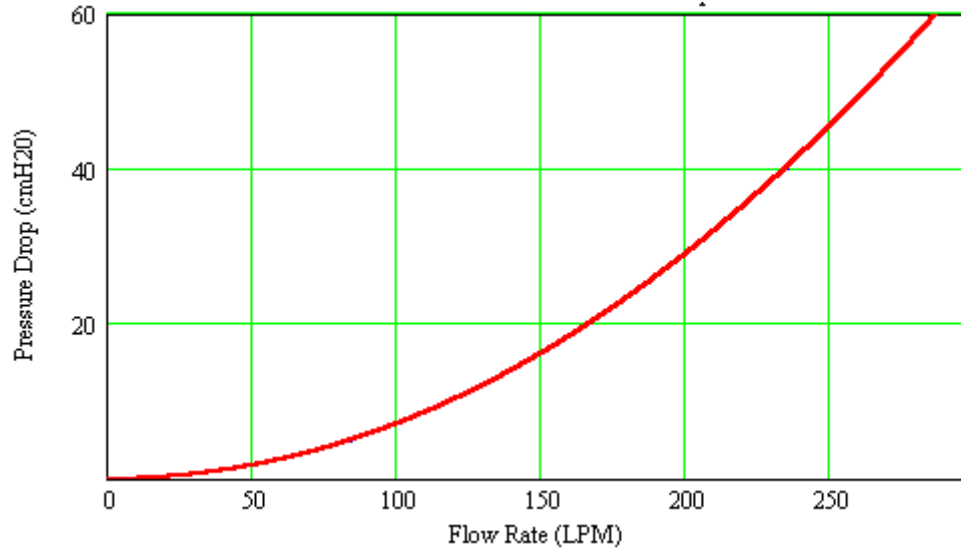


Figure 2.14 Pressure – Flow characteristics of Rp5 resistor [12]

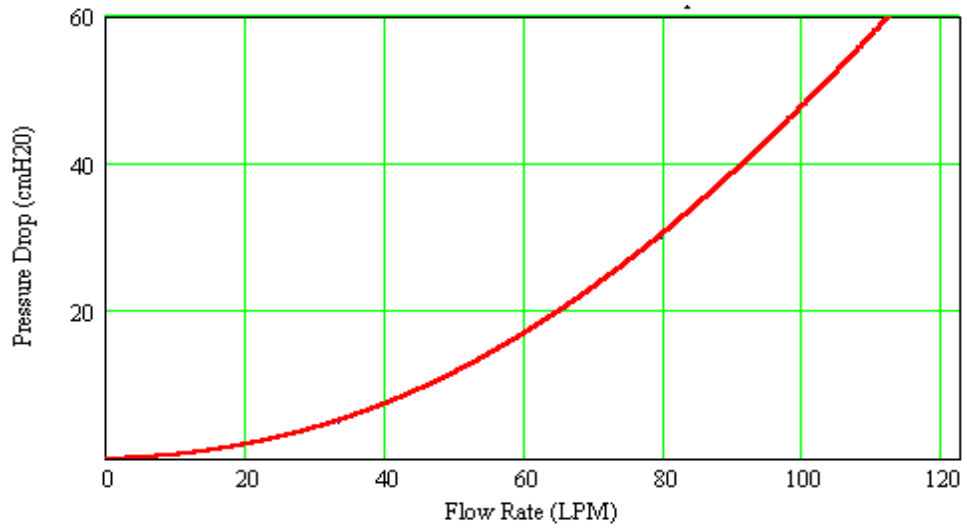


Figure 2.15 Pressure – Flow characteristics of Rp20 resistor [12]

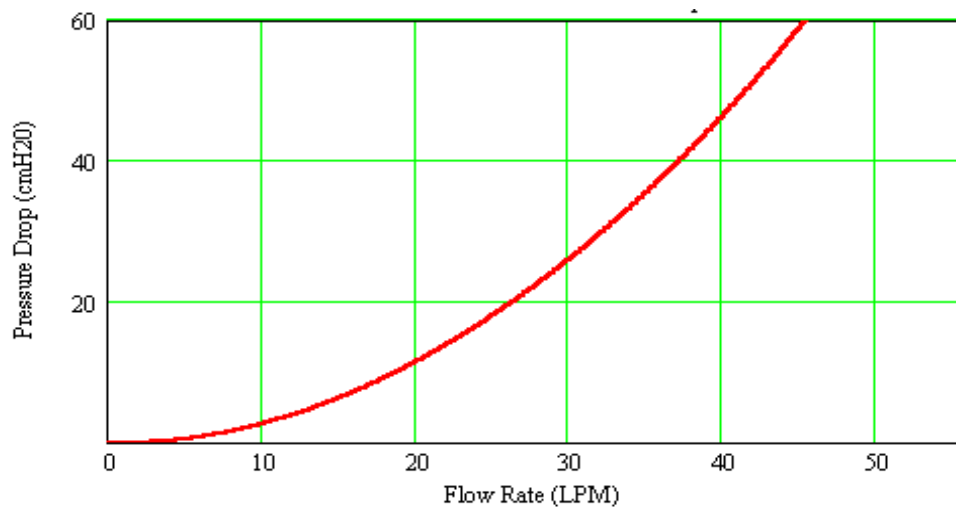


Figure 2.16 Pressure – Flow characteristics of Rp50 resistor [12]

The flow characteristics of the parabolic airway resistors can be defined as [1];

$$Flow = \frac{1}{K} \sqrt{\Delta P} \quad (\text{Liter / s}) \quad (5)$$

Where, ΔP is the pressure difference across the airway resistance in cmH₂O and K is the airway resistor constant in cmH₂O / (Liter / s)² for that specific resistor.

2.3.6.2 Lung Compliance

Lung compliance represents the expandability of the lungs and thorax. It is expressed as the volume increase in the lungs for each unit increase in the intra-alveolar pressure. Normal compliance in a healthy adult male is 100 ml / cmH₂O [9].

With the help of adult lung simulator, compliance could be varied from 0.05 to 0.15 liter / cmH₂O.

3. THE RESPIRATORY MEASUREMENT SET-UP

Monitoring of the respiratory measurements are realized by a virtual instrument which runs under LabVIEW that has aptitude to simultaneously calculate and display the desired respiratory parameters while obtaining input data via parallel port at the same time; and an electronic hardware that enables the data transmission between the pressure sensor and the personal computer.

In this study firstly a LabVIEW virtual instrument program has been designed and developed for both calculation and monitoring of the respiratory parameters. Using LabVIEW program has brought the advantage of using an extensive library during the programming phase and having a user friendly interface window. The developed software program obtains digital input signals under parallel port protocol and temporarily saves at the buffer for the selected interval by the user of the program. Then retrieves the saved data at the buffer and starts the process of calculation and projection of the respiratory parameters.

An electronic hardware has been designed and developed in order to amplify pressure signals acquired by a differential pressure sensor which is embedded in the lung simulator produced by Michigan Instruments. Operational amplifiers are used to straighten analog pressure signals and an analog to digital conversion circuitry is used to enable the transmission of the data to a computer. Also two separate power supply units are developed to feed both amplification circuitry, A/D conversion circuitry and differential pressure sensor.

Electronic hardware system and PC is connected to the lung simulator which is used for measuring respiratory parameters under standardized true load. And the lung simulator is connected to the ventilator. Air transfer is occurred among the ventilator and the lung simulator. Air pressure is converted into electrical signals by a differential pressure sensor in the lung simulator and conveyed to the personal computer through the electronic hardware system.

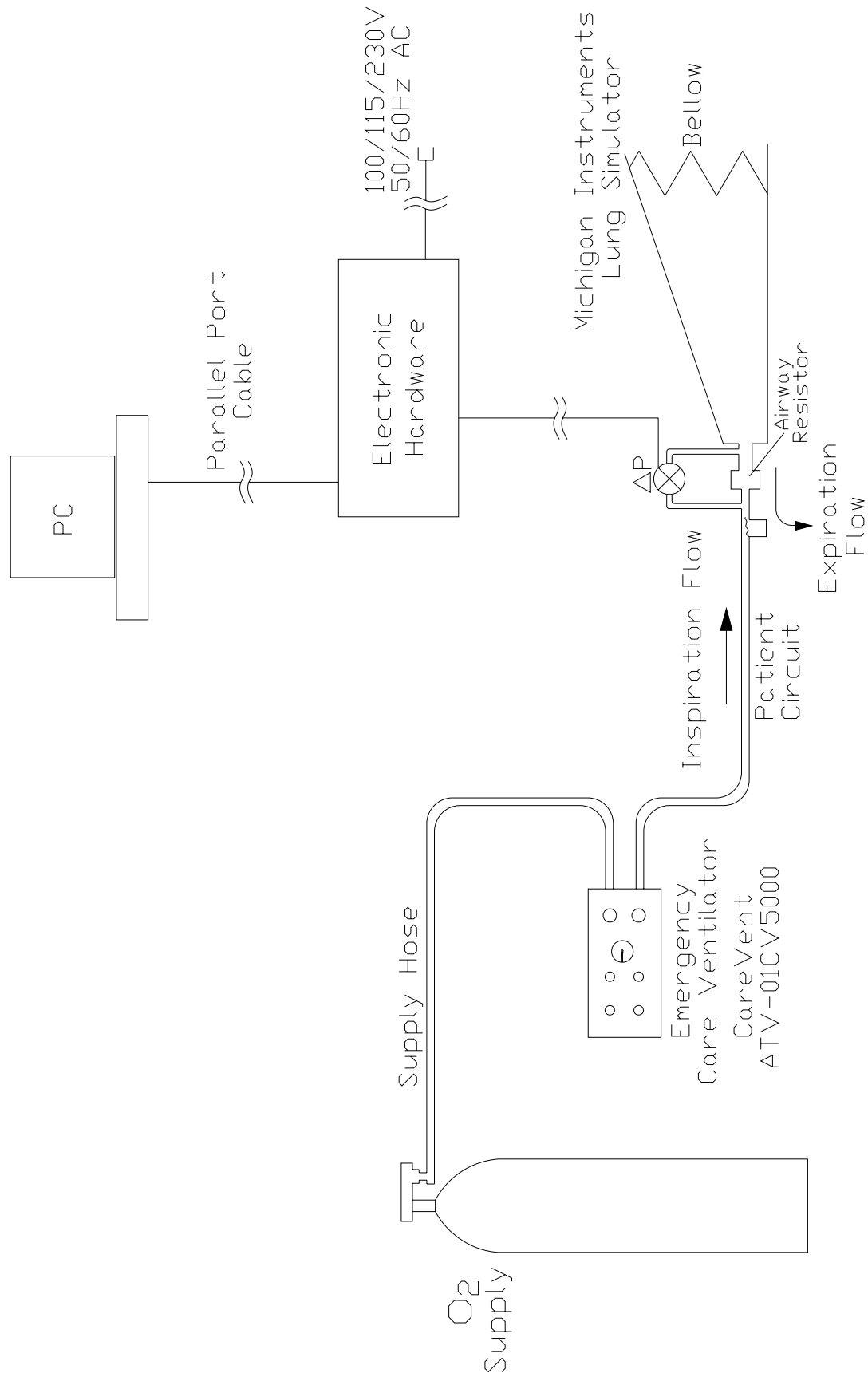


Figure 3.1 Setup diagram of the respiratory monitor

Oxygen Supply:

10 liter oxygen cylinder is used as the oxygen supply for the ventilator. Oxygen cylinder is connected to the ventilator over the supply hose of the ventilator. The oxygen supply is being connected to the ventilator, its regulator valve can be opened and set to the 50 bar pressure level.

Ventilator:

CAREvent ATV-01CV5000 model emergency care ventilator is used with the system. Specifications of CAREvent ATV-01CV5000 are given below in Table 3.1.

Table 3.1
Specifications of CAREvent ATV-01CV5000 [16]

Minute Volume Range:	2 - 14 Liters
Frequency Range:	8 - 40 BPM
I:E Ratio:	1:2
Input Pressure:	50 PSI
Oxygen Concentration:	60 - 100%
Pressure Relief Range:	20 - 60 cmH ₂ O

The patient circuit is attached to the lung simulator and desired breath rate and minute volume settings are selected on the control panel.

Parabolic Airway Resistor:

In order to match the desired flow characteristics, parabolic airway restrictor can be selected among R50, R20 and R5. Parabolic airway resistor is inserted to the sliding adaptor on the front side of the Michigan Instruments adult lung simulator.

Adult Lung Simulator:

The Michigan Instruments adult lung simulator is assembled with the gas outlet of the ventilator via patient circuit. Patient hose is connected to the airway inlet at the front side of the bellow.

Kavlico P651 differential pressure sensor in the simulator is used as the differential pressure transducer. Its input is connected to the air outlets of the simulator by flexible hose pipes and output is connected to the sensor input on the electronic hardware. The connection is established with a standard 0.3 mm² electrical cable with 3-pin socket.

Electronic Hardware:

Electronic hardware is connected to both differential sensor, computer and AC plug. Sensor input pins on the electronic hardware are connected to the differential pressure sensor via an electric cable. A personal computer is connected to the RS-232 digital output of the electronic hardware unit by a flat male parallel port cable, and power input is connected to 100/115/230 V AC plug with a power cord. The line voltage can be selected to correspond to local voltage by the line voltage select dial at the rear of the electronic hardware unit.

Personal Computer:

A PC is connected to the electronic hardware unit over the parallel port. Connection is established by a flat male parallel port cable. LabVIEW program of the respiratory monitor should be run after connection in order to start display, process and measure ventilation data.

4. DESIGN OF THE RESPIRATORY MONITOR

Block diagram of the system is shown below in the Figure 4.1. Electronic hardware part of the system is used to acquire analog pressure signals via pressure sensors and convert them into digital signals. Software part of the system realized by a computer with LabVIEW program and it is used to process the input pressure signals and calculate other desired measurements and display them. Mechanical part of the system is used to simulate adult lung and airway system in order to make the measurements at a standardized real based environment. Block diagram of the system is shown below in the Figure 4.1.

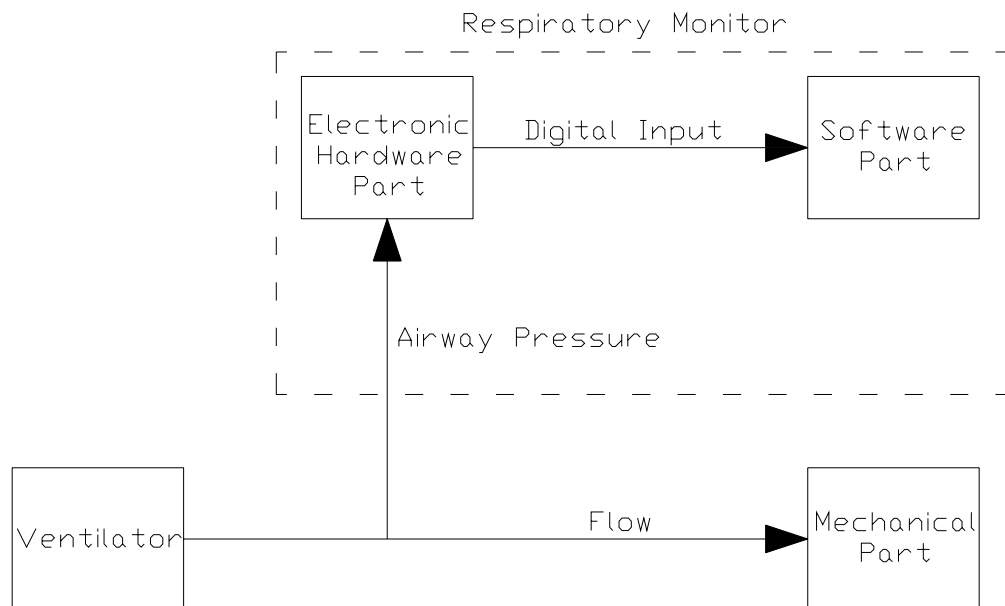


Figure 4.1.Block diagram of respiratory monitor

4.1 Electronic Hardware System

Electronic hardware system forms the most important part of the ventilator testing unit. The electronic hardware system of the ventilator testing unit consists of a differential pressure sensor, an amplification circuit, an analog to digital conversion circuit, and a power supply circuit. First, pressure is acquired by differential pressure sensor. Then obtained pressure signals are conveyed to amplification circuit and DC offset filter to bring them at a proper level, and finally precisely amplified pressure signals are converted to

digital signals to make ready their transfer to the PC. In Figure 4.2 block diagram of the electronic hardware system is given.

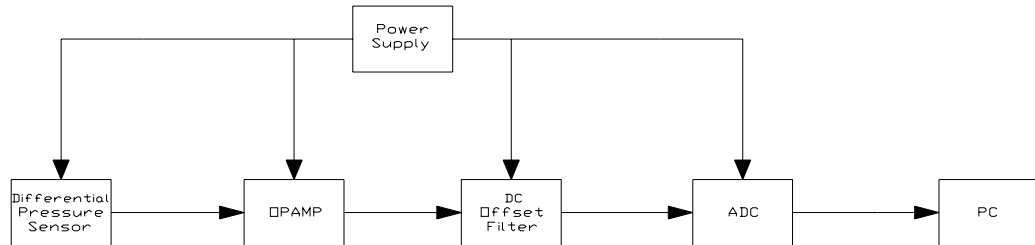


Figure 4.2 Block diagram of electronic hardware system

4.1.1 Differential Pressure Sensor

In order to acquire pressure signals from the ventilator a differential pressure sensor has been used in the hardware system of the ventilator testing unit. The differential pressure sensor is a product of Kavlico Corporation with P651 model number. Electrical characteristic of the differential pressure sensor is given below in Figure 4.3 and specifications of the sensor is given in Table 4.1.

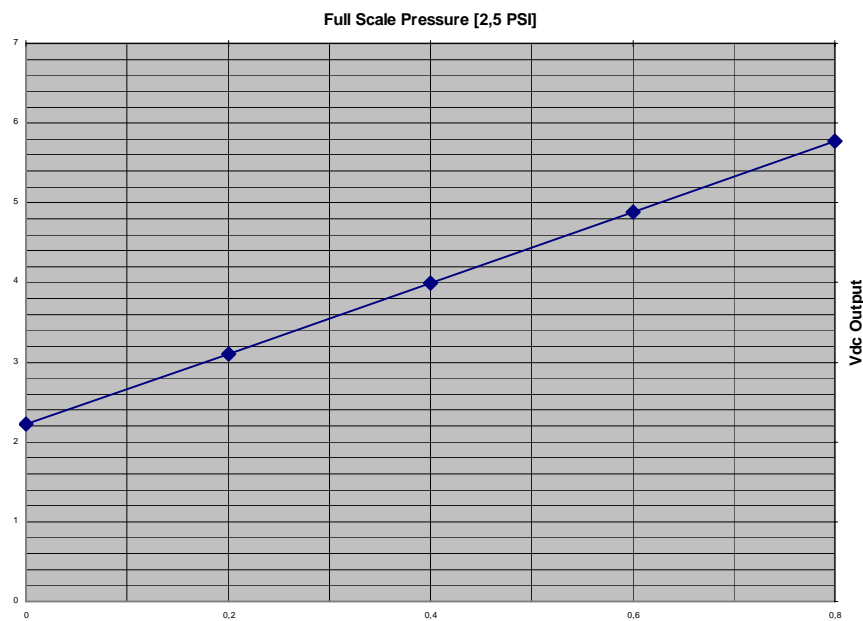


Figure 4.3 Electrical characteristics of P651 differential pressure sensor

Table 4.1
Specifications of P651 differential pressure sensor [17]

Parameter	Min	Max
	$V_s=9\text{ V}$	
Full Scale Output:	5.9 V	6.1 V
Null Offset:	3.9 V	4.1 V
Span:	1.8 V	2.2 V
Linearity:	$\pm 0.3\%$	$\pm 0.5\%$
Hysteresis:	$\pm 0.003\%$	$\pm 0.01\%$
Supply Voltage:	7.5 V	15 V
Supply Current:	8 V	15 V

Differential pressure sensor is used to measure the difference over the airway pressures before and after the airway resistor. High pressure port of the differential pressure sensor is connected between the ventilator and airway resistor, and low pressure port is connected after the airway resistor, between airway resistor and bellow. The measured differential pressure value is displayed and also used to calculate flow and volume that are supplied by the ventilator. Below in Figure 4.4 connections of the differential pressure sensor is shown.

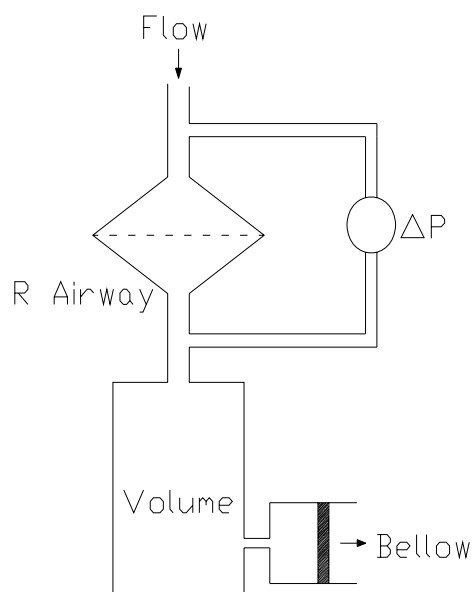


Figure 4.4 Connections of differential pressure sensor

Input of the differential pressure sensor is connected directly via the hose pipes that come from the ventilator. On the other side output of the differential pressure sensor is connected to the operational amplification circuit.

4.1.2 Amplification Circuit

Calibration of the differential pressure sensor is achieved by using a separate operational amplifier circuit. Desired output voltage can be easily obtained for calibration by changing the resistance of the potentiometers. Circuit diagram of the amplification circuit is given below in Figure 4.5.

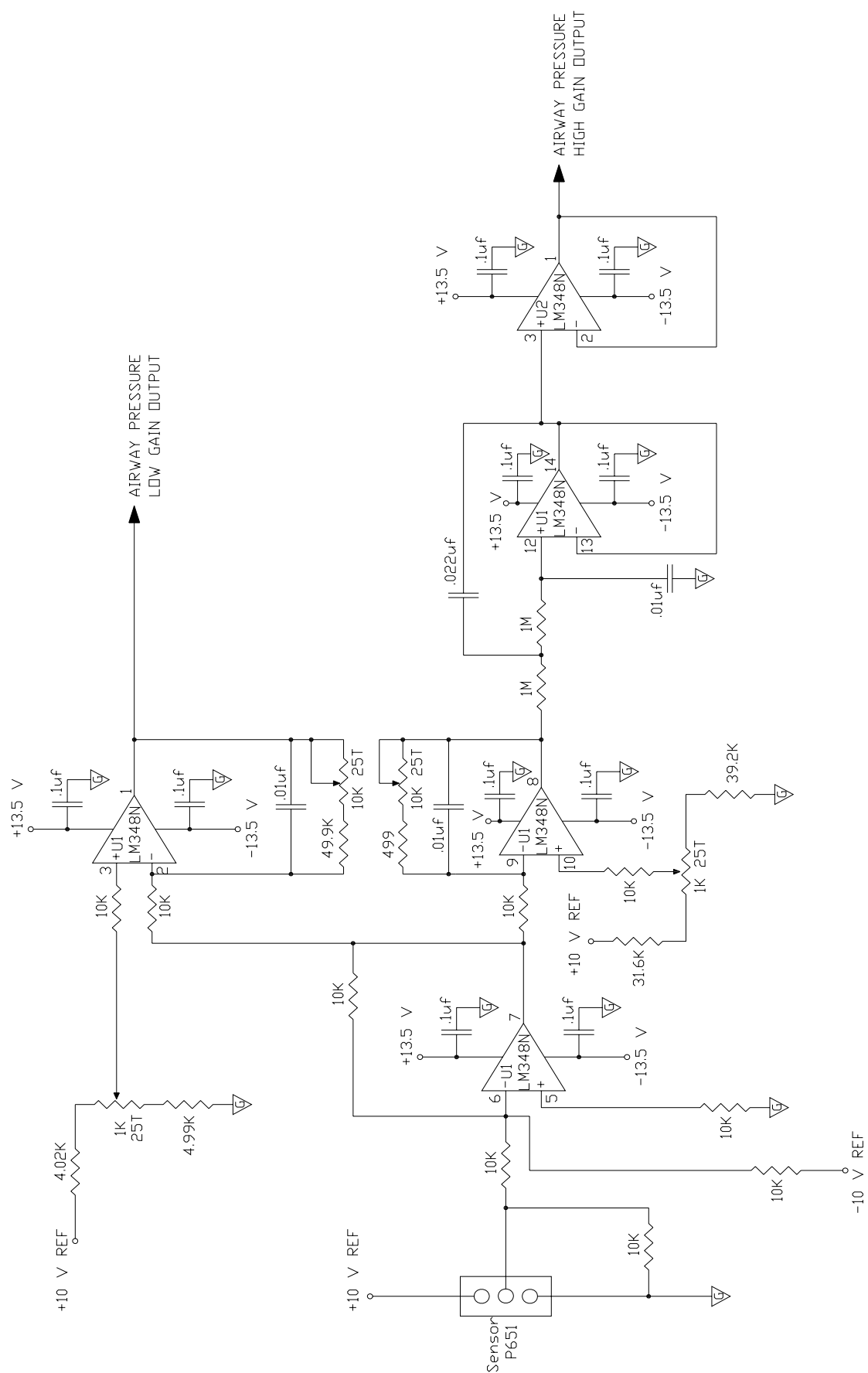


Figure 4.5 Circuit diagram of the amplification circuit

LM348N operational amplifier is chosen because of its stable characteristics, easily obtainable and low cost properties.

4.1.3 DC Offset Filter

After the amplification of the pressure signal, dc offset of the signal is diminished in order to match 0-5 V input of the analog to digital converter. DC offset filtering is achieved through a capacitor and two resistances connected between +10 V supply voltage and ground. Circuit diagram of the DC offset filter is given below in Figure 4.6.

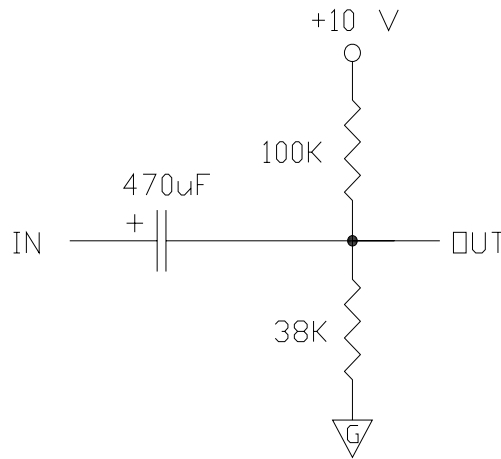


Figure 4.6 Circuit diagram of the DC offset filter

4.1.4 Analog to Digital Conversion

Analog to digital conversion constitutes the most crucial part of the hardware system of ventilator testing unit. Because the output of the pressure sensors are analog and input of the PC is digital based, analog to digital (A/D) converter plays an essential role between them in order to communicate each other.

Input of the analog to digital conversion circuits is connected to the output of the amplification circuit of the differential pressure sensor. Output of the analog to digital

converter is engaged to the parallel port connection cable. Circuit diagram of the A/D conversion is given below in Figure 4.7.

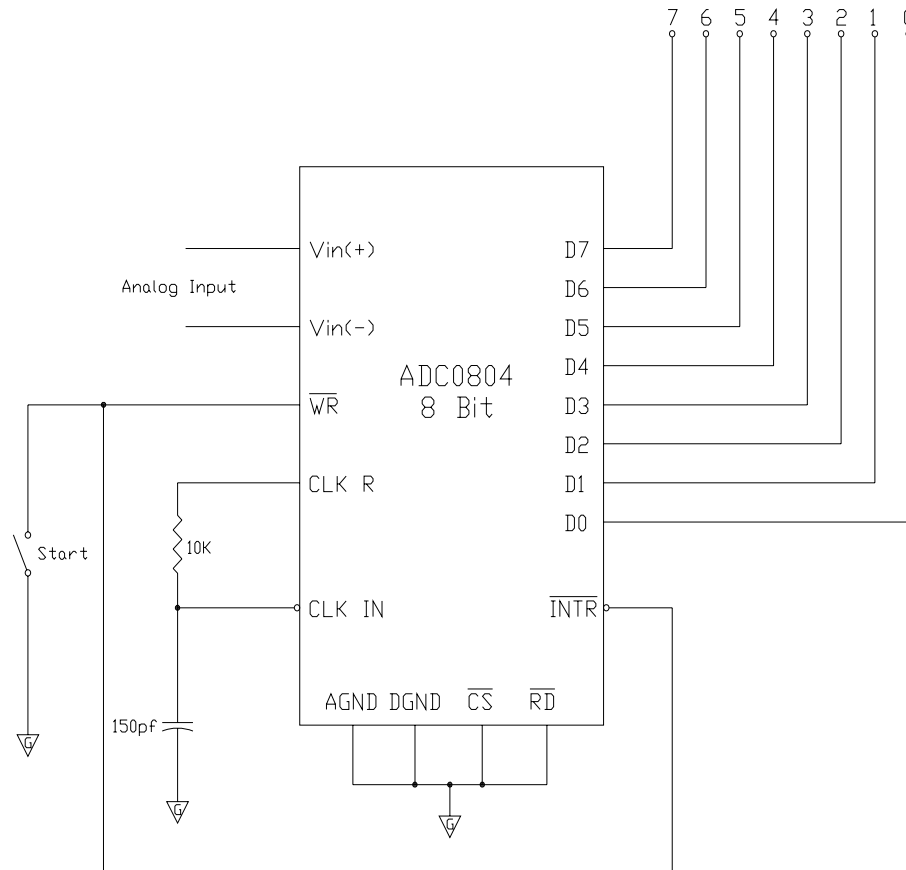


Figure 4.7 Circuit diagram of the A/D conversion circuit

ADC0804 transducer of National Semiconductor Corporation has been selected for the system. Main purposes of using ADC0804 A/D converter are its appropriate 8-bit output, low cost and easy stand alone operation.

4.1.5 Parallel Port

Parallel port is a communication terminal at personal computers date from first PC especially built to carry out the communication between PCs and printers. Parallel ports allow the input up to 9 bits or the output of 12 bits at any one given time. The port is composed of 4 control lines, 5 status lines and 8 data lines [13].

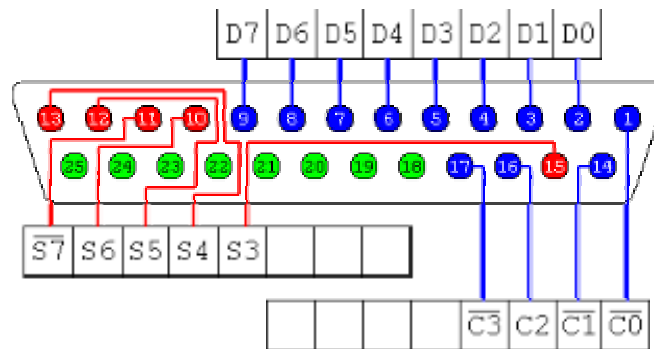


Figure 4.8 Pin out schema of parallel port [14]

In order to establish the communication between the hardware and software system of the ventilator testing unit parallel port of the PC is used as the connection terminal due to its allowance of both 8 bits and 5 bits of input at the same time.

Bi-directional data register has been used to transmit the 8 bits data. Data registers are configured as output at their default set up. In order to enable the data lines for input of the system bit 5 of the control port is pulled high by the software interface program at every start up. 8-bit differential pressure signals that are obtained by differential pressure sensor has been transmitted over data registers.

4.1.6 Power Supply

Because there has been several circuits at the hardware system, need of several different supply voltages had been occurred.

In the power supply system a transformer has been used as the main voltage supplier. The transformer used in the system has one primary winding with 220 V_{AC} primary voltage and two separate secondary windings with 18 V_{AC} and 30 V_{AC} secondary voltages respectively.

First secondary voltage with 18 V_{AC} is rectified through a bridge rectifier and filtered by two capacitors and regulated by LM7810 that is a +10 V regulator. Second secondary voltage with 30 V_{AC} is rectified through a bridge rectifier and regulated by three LM317T and two LM337T regulators after filtered by two similar capacitors in order to

acquire +5 V, +10 V, +13,5 V, -10 V and -13,5 V DC voltages respectively. At the developed hardware system, +5 V is used for A/D conversion circuit supply voltage, $\pm 13,5$ V are for operational amplifiers' supply, ± 10 V are used as input at the amplifier circuits and separate +10 V acquired by the first secondary wind is used for power supply of the differential pressure sensor. Circuit diagram of the power supply unit is shown below in Figure 4.9.

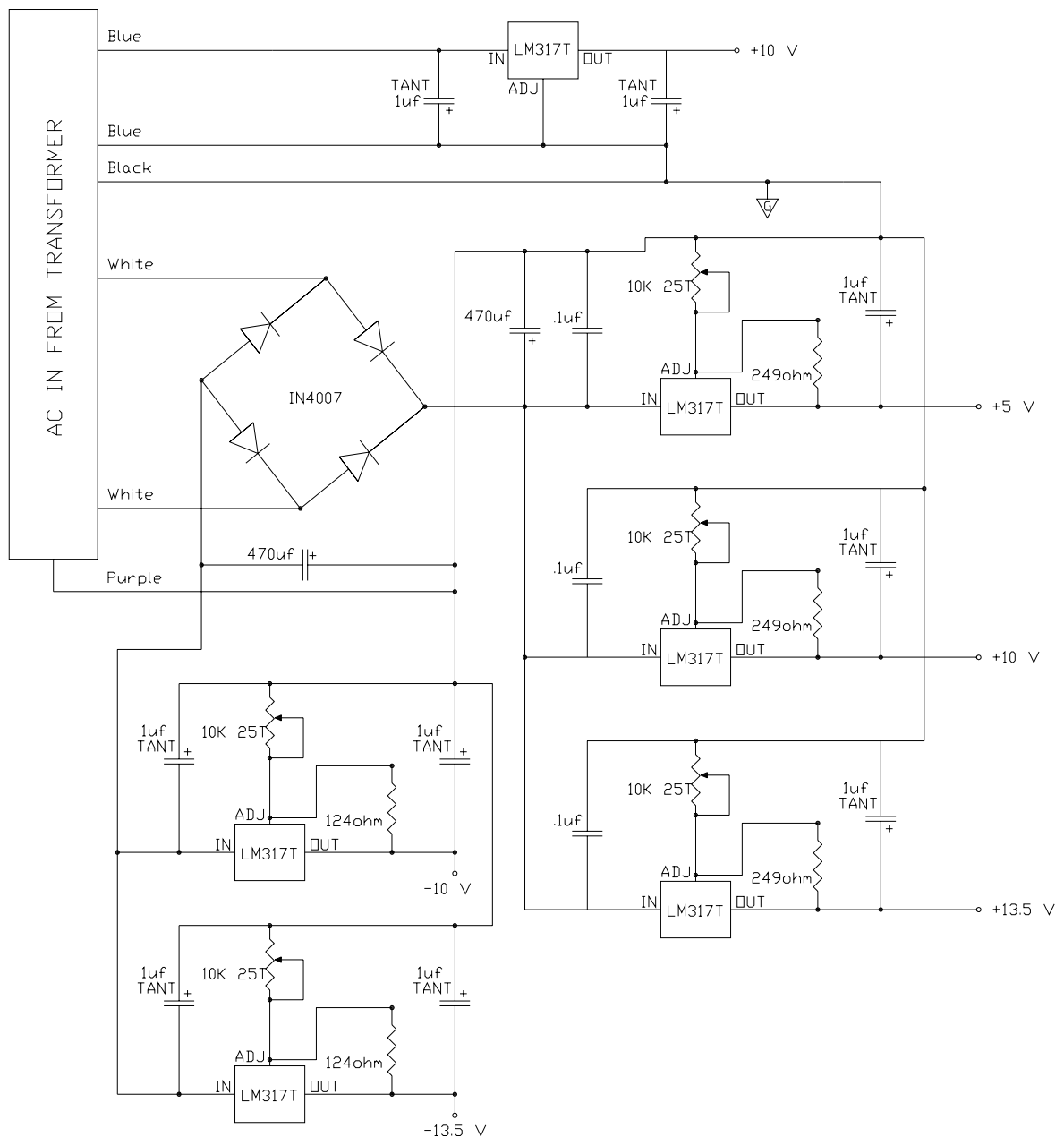


Figure 4.9 Circuit diagram of power supply unit

4.1.7 Electronic Hardware Unit

In order to establish stabilized connections and ease of use electronic hardware unit, a PCB board has been designed. Power supply, amplifier and analog to digital circuits are connected together and installed on that PCB board and enclosed in a metal box that is equipped with required outlet and inlets as AC power and parallel port sockets. Below in Figure 4.10 PCB board of the electronic hardware system is shown.

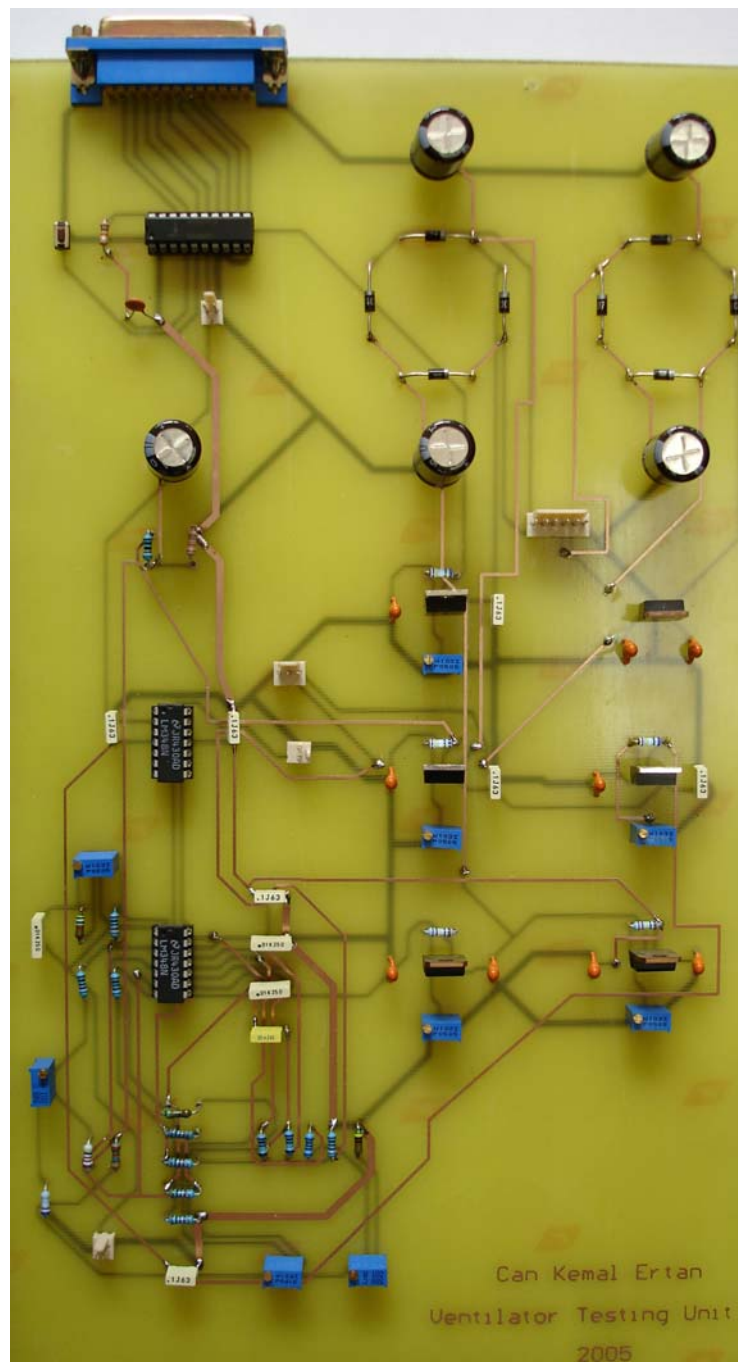


Figure 4.10 Designed PCB of the electronic hardware system

At the right hand side of the PCB power supply circuit of the system is installed and in order to enable the connection of the power supply circuit with the AC transformer that is placed in the metal enclosure a 6-pin socket is placed on the AC input side of the circuit. At the bottom left hand side of the PCB operational amplifier circuitry is installed. One 3-pin and two 2-pin sockets are placed at the amplifier circuit. 3-pin socket is used as the input of the differential pressure sensor while 2-pin sockets are used as the low and high output test pins of the circuit. These 2-pin test sockets can be used in case of necessity to display amplified analog differential pressure signals. At the top left hand side of the PCB analog to digital conversion circuit is placed. 8-bit digital output of the A/D converter is connected to 25-pin female parallel port socket which is placed at the top of the PCB. A start button is placed next to the A/D converter for giving the opportunity to start conversion manually in case of its not functioning automatically. A 2-pin socket is connected to the analog input pins of the A/D converter for testing reasons. DC offset circuitry is placed between analog to digital conversion circuit and amplifier circuit. Both resistances of the DC offset circuit is placed through small caves to give the chance to change the resistors easily.

All the ground lines on the PCB are established at 1.8 mm thick in order to diminish noise while signal lines are established at 0.8 mm thick. Each of the transistors is placed via housings to give the occasion to be changed easily. Multi turn trimming potentiometers are used at the amplifier circuitry for sensitive and precise adjustment of gain.

4.2 Software System

Digitalized pressure signals are both processed and displayed by LabVIEW program at PC. LabVIEW is a powerful and flexible instrumentation and analysis software development application. By processing of the pressure signals desired flow, volume and time measurements of the ventilator can be easily calculated and displayed at LabVIEW.

4.2.1 Design of the Interface Program

Laboratory Virtual Instrument Engineering Workbench (LabVIEW) program is used for processing the pressure signals that are detected by the pressure sensors and conveyed through the parallel port after they had been amplified and digitalized. The main reason of using LabVIEW as the signal processing program is its extensive library of virtual instruments and functions.

LabVIEW is a powerful and flexible instrumentation software and analysis development application developed by National Instruments. LabVIEW programs are called Virtual Instruments (VI) and create programs rely on graphic symbols to describe programming actions [15].

LabVIEW uses graphical programming language known as G programming language. Therefore, needed mathematical calculations and software procedure of the system is written in G language. LabVIEW program constitutes of two parts. First part is the front panel and second part is block diagram. The front panel is the control and display interface part to the users, and the block diagram is the part where G program language is being coded in order to run all the commands. The program is controlled through the front panel, and its functionality is implemented in the block diagram. The front panel is shown in Figure 4.11, and the block diagram is shown in Figure 4.12.

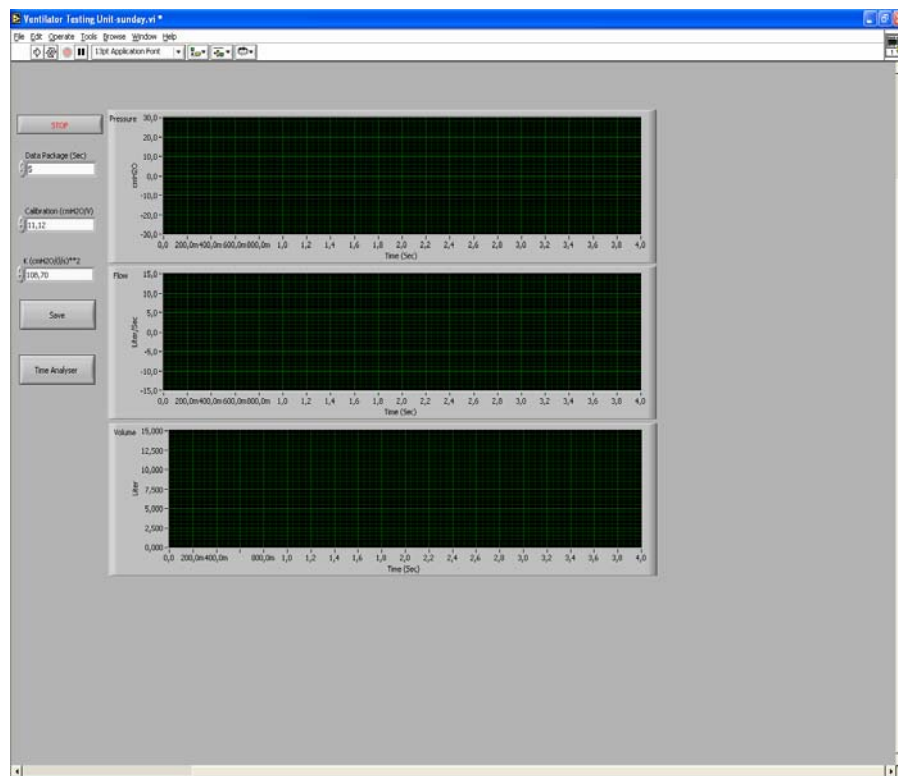


Figure 4.11 The front panel

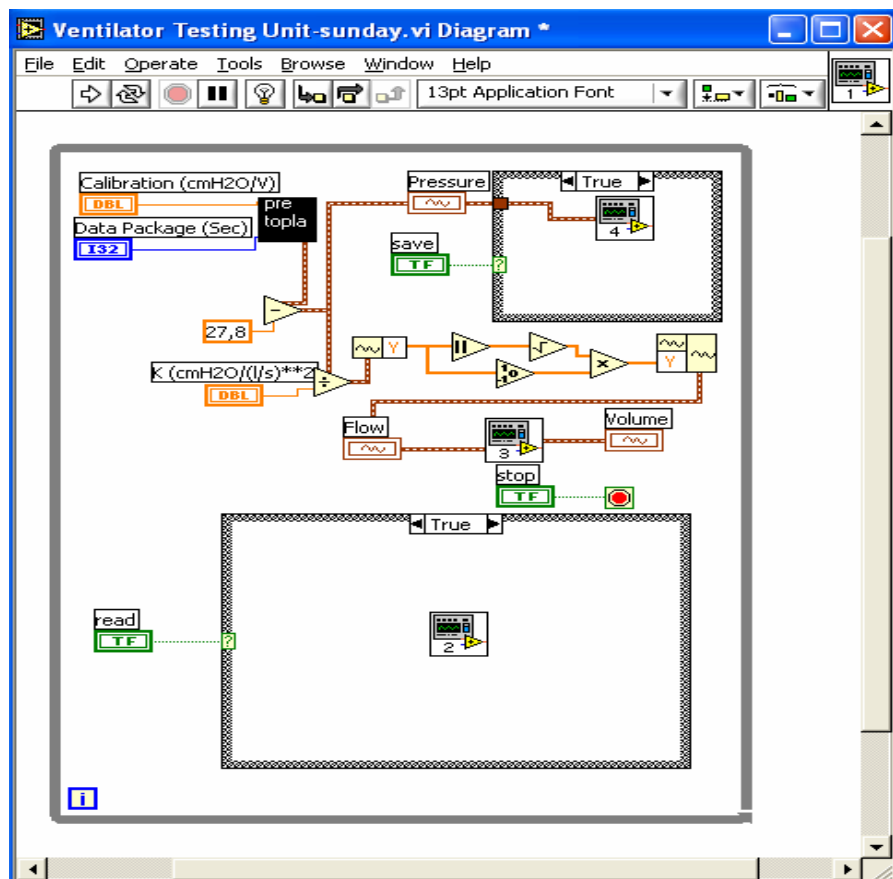


Figure 4.12 The block diagram

Designed Labview program code consists of input, output and mathematical functions under subvi programs which form the Ventilator Testing Unit program.

First subvi program is used to input pressure signals and also output them after manipulation. Firstly parallel port input has been described in the first subvi code by using the in port function. The number 888 written at in port function is the address of the data registers of the parallel port. Loading 888 to the in port function describes the address of the input data to the software program. Although there is no data output from the parallel port; an out port function is placed in the program. Data registers work as output terminals at default. In order to change them into their bi-directional characteristic and make them work as both input and output, bit 5 of the control register had to be pulled digital high. In that case an out port function is created in the program and 100000 hexadecimal number is sent to the control port whose register address is 890.

After 8 bits raw input pressure signal is entered to the program, it is first being smoothed by using a digital median filter and displayed at the front panel part of the program after multiplied by the calibration value of the pressure sensor. Manipulated pressure data is saved on the buffer for the time entered at the “Data Package” number box on the panel screen by the user. Data package value decides both the time pressure data being saved on the buffer and limits of the x-axis of both graphs. In the Figure 4.13 below the first subvi program is shown.

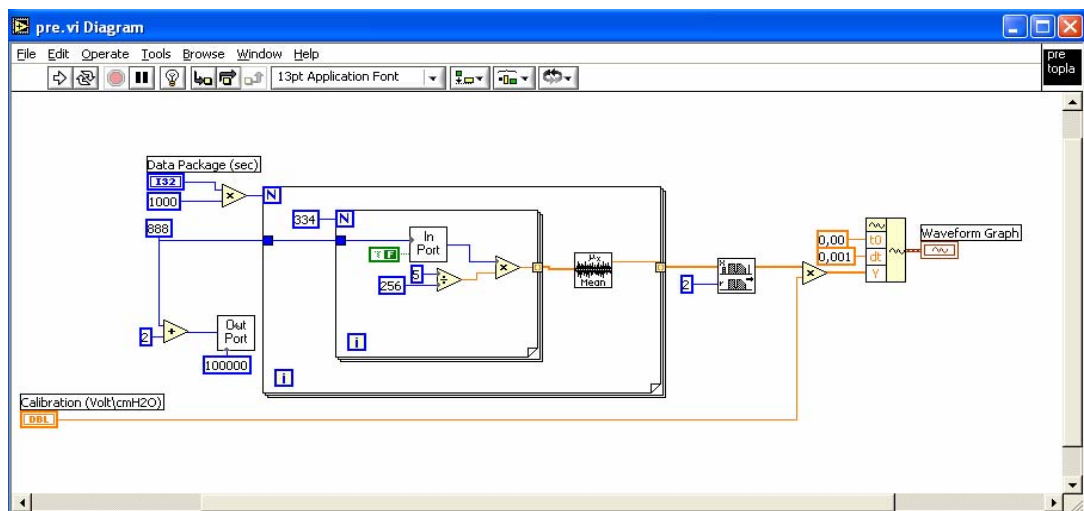


Figure 4.13 The first subvi program

Mathematical functions follow the first subvi program in order to calculate flow. Because flow is the square root of differential pressure after its division to airway resistance constant and differential pressure is both a negative and a positive number variable, absolute value function has been used before the square root function.

Under the second subvi program a time analyzer program has been developed. A virtual switch called “Read” has been developed on the program in order to open another user interface window where time measurements of the ventilation process can be easily done over the pressure waveform. Four different cursors have been adapted on the interface window. Those cursors can be moved by the virtual delta dial and placed to the desired position on the pressure graph. By subtracting x-axis values of the two consecutive cursors, inspiration and expiration times can be calculated. Afterwards, period, breath rate and I:E ratio are calculated and displayed. In Figure 4.14 front panel of the “Time Analyzer” program and, in Figure 4.15 diagram of the “Time Analyzer Program” is shown.

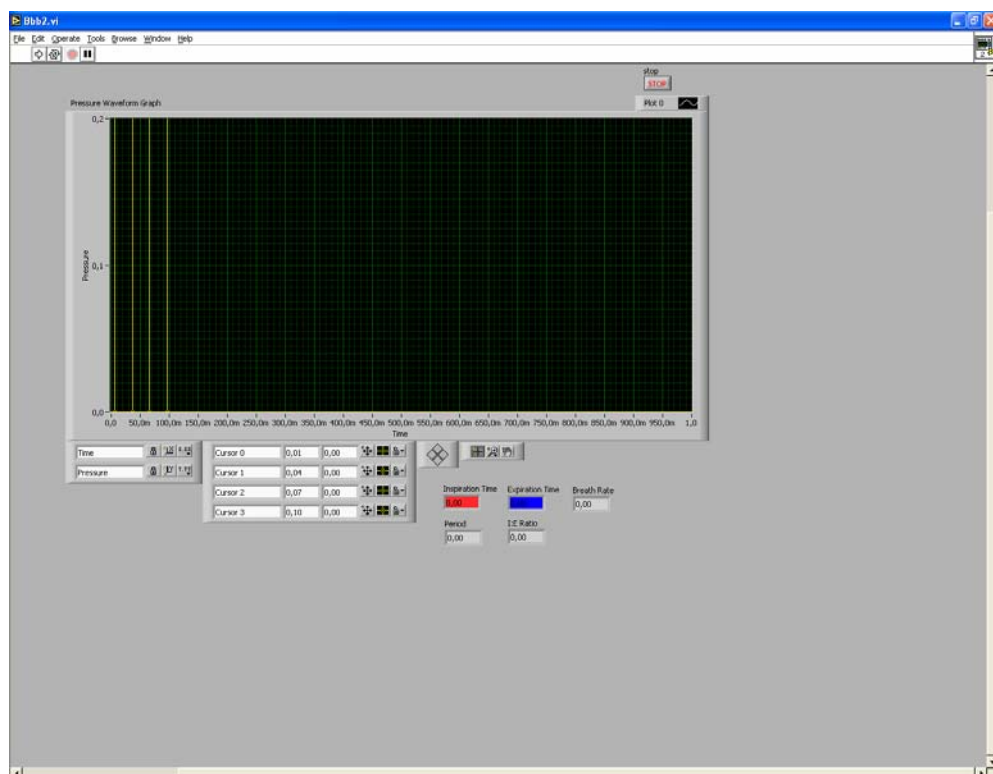


Figure 4.14 The time analyzer panel

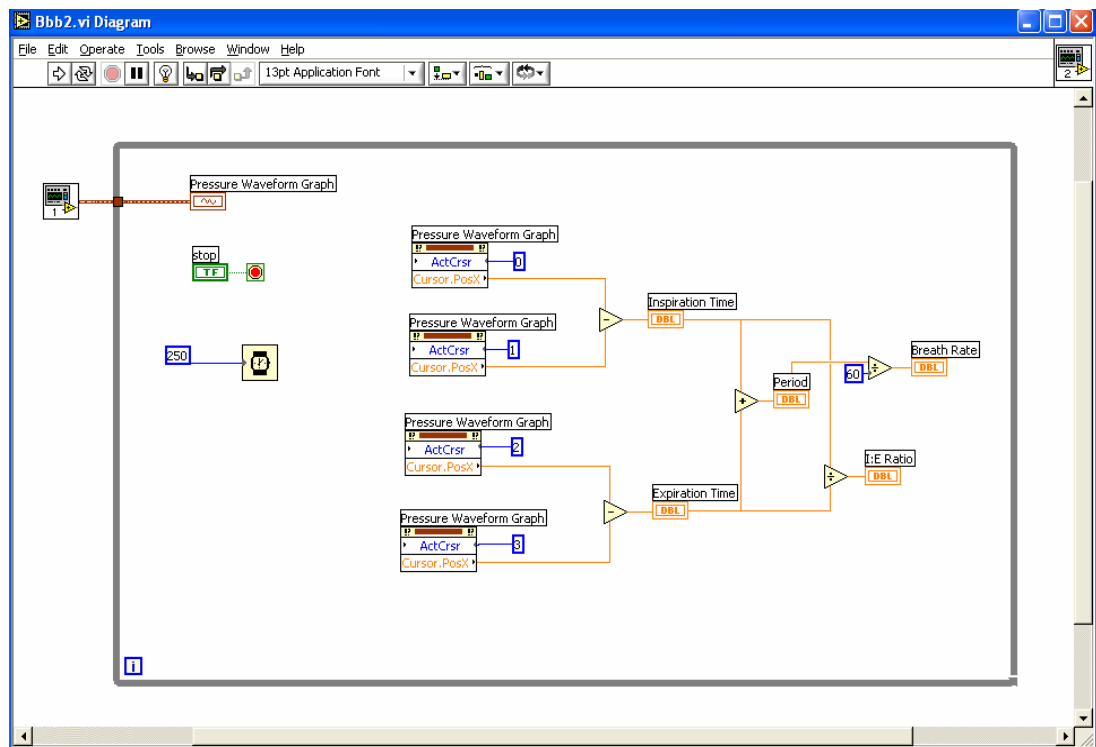


Figure 4.15 The block diagram of the time analyzer program

Under the third subvi program volume calculation and output has been done. Because volume is the integral of flow function, it is calculated by the total area under the flow wave form. After volume is calculated it is displayed on the panel screen as a graph. In the Figure 4.16 below the block diagram of the third subvi program is shown.

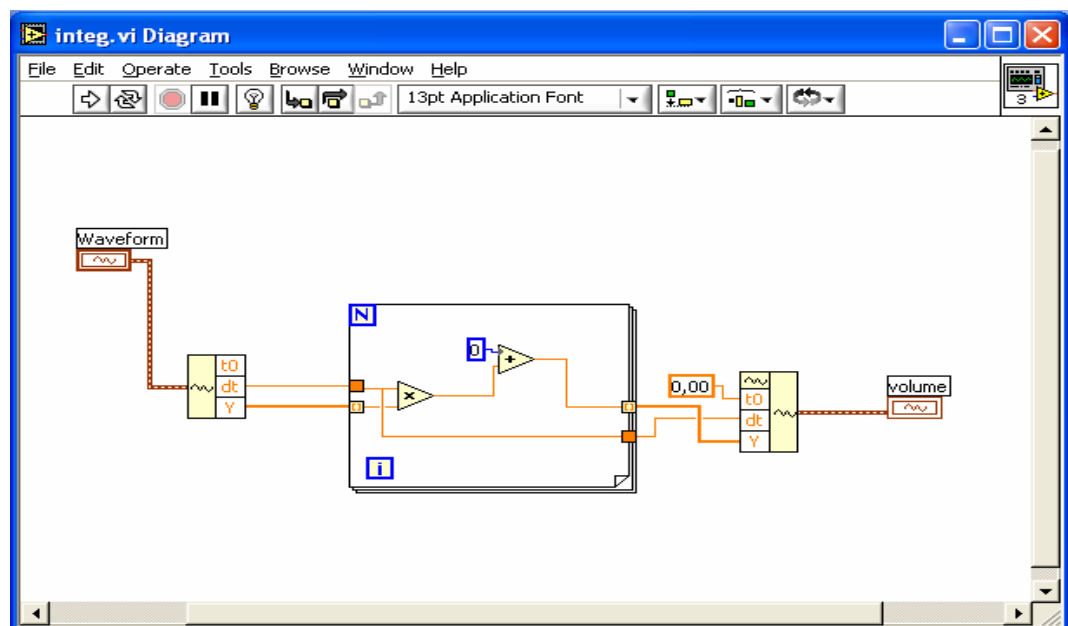


Figure 4.16 The third subvi program

At the fourth subvi program, save function of the pressure output is being coded. Saving the pressure output is necessary to open the pressure wave form data at Time Analyzer program in order to calculate and display time based ventilation data. Saving the pressure wave form can be simply achieved by pressing the “Save” button on the front panel screen of the main program and opened at the Time Analyzer program by pressing the “Time Analyzer” button. In the Figure 4.17 the fourth subvi program is shown.

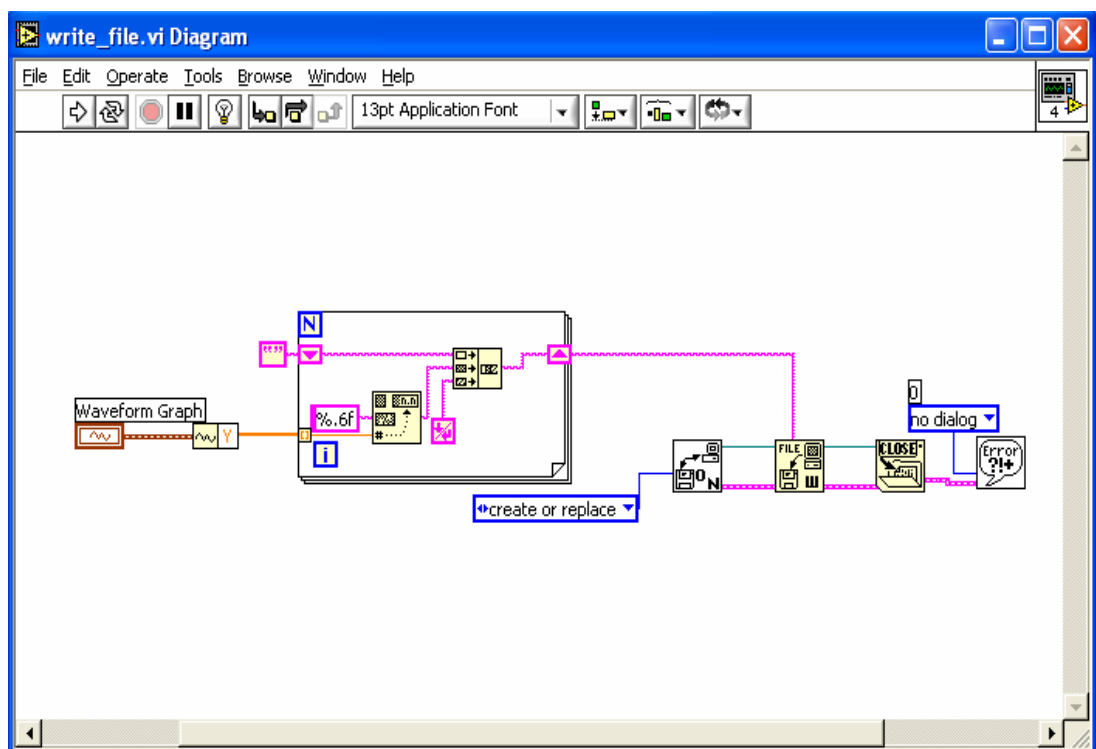


Figure 4.17 The fourth subvi program

4.2.2 User Interface of the Software Program

One of the major advantages of LabVIEW program is its user friendly interface feature. LabVIEW uses the front panel part for the user interface window. Three different graphs are displayed at the front panel. Differential pressure signal, flow and volume measurements are displayed in graphs simultaneously. Time axis values of the graphs, K constant value and pressure sensor calibration value can be selected at the front panel by the help of virtual number boxes. The virtual “Save” switch triggers the saving of the next pressure wave form package and the virtual “Time Analyzer” switch on the front panel

opens the analyze window where such time based respiration parameters are measured and displayed;

- Inspiration Flow Time
- Inspiration Pause Time
- Expiration Flow Time
- Expiration Pause Time
- Period
- Breath Rate
- I:E Ratio

The main program can easily be started just by pressing the run button. After the program is being started, both differential pressure, flow and volume graphs are displayed on the screen. While monitoring the graphs, the duration of the data being displayed can be selected from the number box called “Data Package”. K constant value which is related to the parabolic airway restrictor tube being used can also be selected from the number box called “K” and, calibration value of differential pressure sensor that is the equivalent pressure value to electrical output of the sensor in Volts can be selected by the number box. Interface window of the main program is given in Figure 4.18.

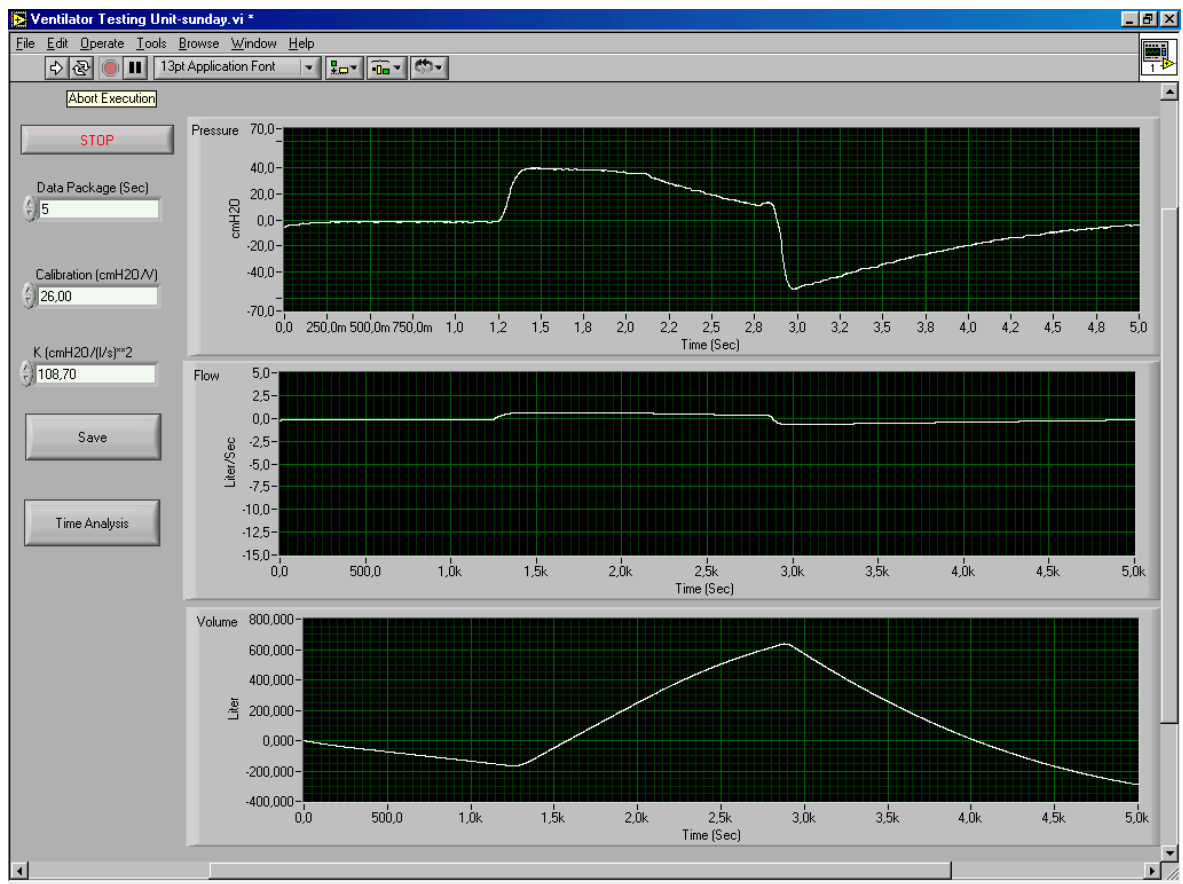


Figure 4.18 Interface window of the main program

By the help of the save button, differential pressure graph can be saved for the selected time interval at the Data Package number box in order to reopen the file at the Time Analyze program. Saved pressure data can easily retrieved by pressing the Time Analyzer button. When Time Analyzer button is pressed once, the second program that is used to measure and monitor the time based ventilation data will be opened. In order to measure the time intervals of the pressure graph, each of the four cursors can be moved to desired positions by using the cursor dial. Interface window of the Time Analyzer program is shown in Figure 4.19.

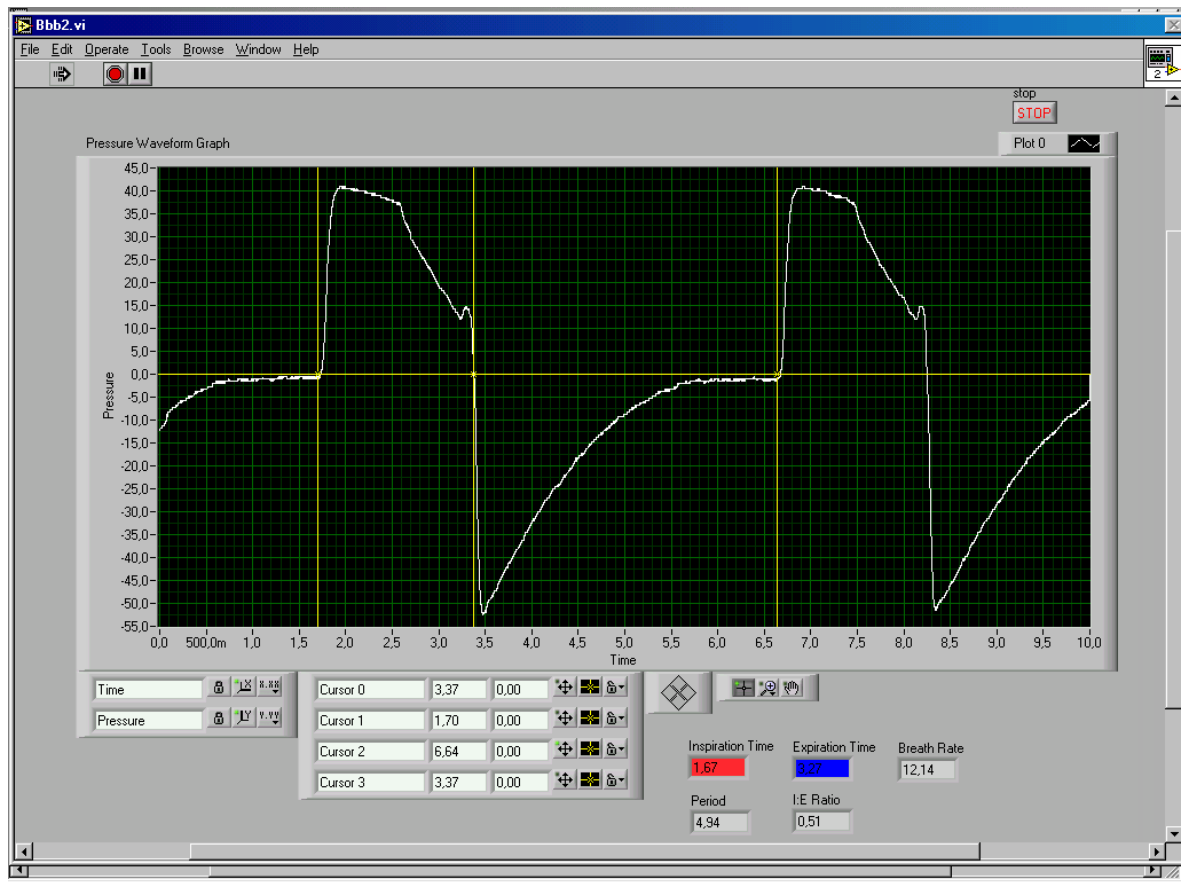


Figure 4.19 Interface window of the time analyzer program

The chart palette of the software, given below in Figure 4.20, allows users to pan and zoom in and out.



Figure 4.20 The chart palette buttons

5. CONCLUSION

In this thesis, a cost effective, portable and user-friendly respiratory monitor was developed in order to observe typical ventilation parameters such as pressure, flow and volume waveforms besides time based ventilation data.

An electronic hardware unit composed of operational amplifiers and an analog to digital converter was established to strengthen analog pressure signals and convert them into digital. A differential pressure sensor was adapted to the developed electronic hardware unit in order to acquire pressure signals.

The software system of the respiratory monitor was created by using LabVIEW program, also known as virtual instrument and G programming language because of its being user-friendly, easy to program and extended source of library. Developed software system was used to process input pressure signals; obtain flow, volume and time based ventilation data and monitor both of the measured data at the user interface window.

A lung simulator by Michigan Instruments was combined to the developed respiratory monitor in order to realistically simulate the mechanics of the adult pulmonary system from the upper airway to the lung with different characteristics.

Standard parallel port communication system was used to assure transmission between the electronic hardware and software systems. 8-bit differential signal transmission was achieved by using data registers of the parallel port.

For the future work, resolution of the data that is transferred between the electronic hardware system and the computer can be developed for more sensitive measurements. Also, an absolute pressure sensor can be integrated to the system for additional measurements.

APPENDIX

Enclosed inside back cover the followings can be found in CD-ROM.

- Source code of LabVIEW Virtual Instrument
- Respiration Monitor.exe file
- Read Me file

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