Addis Ababa University

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Center of Biomedical Engineering

Customizing the Pressure-Based Spirometer for Improving Its Accuracy for Chronic Obstructive Pulmonary Disease Tests

By:

Ashebir Worku (ID: GSR 0982/08)

Advisor:

Dr. Masreshaw Demelash

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Dr. Masreshaw Demelash	____________
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Declaration

I, the undersigned, declare that this thesis is my original work, has not been presented for a degree in any other university, and all sources of materials used for the thesis have been fully acknowledged.

Ashebir Worku ________________
Name Signature

Place: **Addis Ababa University, Ethiopia**

Date of Submission: ________________________

This thesis has been submitted for examination with my approval as a University advisor.

Dr. Masreshaw Demelash ________________
Advisor Signature

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I would like to express my gratitude to the Almighty God, who has enabled me to work hard mentally and spiritually for the completion of this thesis work.

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Abstract

Breathing is a fundamental and essential process for the proper functioning of human being. It is a process of supplying oxygen to the body and removing of waste carbon dioxide from the body. This process takes place through the respiratory system and organs involving the nose, pharynx, trachea, bronchi, and lungs. There are several diseases and conditions that can affect the respiratory system. Chronic obstructive pulmonary disease (COPD) is the most common one that arises from airway obstructions. A differential pressure-based spirometer is the most dominant device used to perform pulmonary function test in order to diagnose and monitor COPD. This pressure-based spirometer acquires a signal through a pressure difference between two points of the patient breathing tube (PBT) created by the patient against the atmospheric pressure. This means as the COPD patient finds it hard to breath, the differential pressure created by the patient across PBT reduces. The detection sensitivity of the PBT and altitude correction are the main important parameters which need to be considered while designing a differential pressure-based spirometry. So the aim of this thesis is to customize a differential pressure-based spirometer device with respect to altitude in order to improve accuracy of COPD tests. In this thesis, LabVIEW software has been used to generate and analyze the measured signal, and customize the spirometer with respect to altitude. Moreover, Bernoulli's equation has been used to calculate air flowrate and forced vital capacity of the lung which are used to grade COPD test result. Finally, the designed PBT is simulated using Ansys 19.2 software and a prototype is created using 3D printer. The flowrate vs time, and volume vs time graphs were obtained and displayed using LabVIEW. The result shows that the accuracy of COPD test can be improved by increasing the sensitivity of PBT and customizing a pressure-based spirometer using the altitude correction factor during designing and operating the pressure-based spirometer.

Keywords: Bernoulli's Equation, COPD, LabVIEW, PBT, Spirometer, Lung

Contents

List of Figures

List of Tables

Abbreviation

CAM	Computer Aided Manufacturing
COPD	Chronic Obstructive Pulmonary Disease
DP	Differential Pressure
FEV1	Forced Expiratory Volume in One Second
FRC	Functional Residual Capacity
FVC	Forced Vital Capacity
GOLD	Global Initiative for Chronic Obstructive Lung Disease
GUI	Graphical User Interface
LabVIEW	Laboratory Virtual Instrumentation Engineering Workbench
LED	Light Emitting Diode
LDR	Light Dependent Resistor
WHO	World Health Organization
VC	Vital Capacity
VI	Virtual Instrumentation
RV	Residual Volume
PBT	Patient Breathing Tube
PEF	Peak Expiratory Flow
PFT	Pulmonary Function Test

CHAPTER ONE

INTRODUCTION

1.1. Background

Breathing is a fundamental and essential process for the proper functioning of every living organism. It is a process of supplying oxygen to the body and removing of waste carbon dioxide from the body. These processes take place through involvement of the respiratory system. The respiratory system consists of all organs involved in the breathing. These are the nose, pharynx, trachea, bronchi and lungs. The lungs are the main organ of the breathing system that are responsible for gas exchange, namely the transfer of oxygen into the body and the transfer of carbon dioxide out of the body.

A healthy respiratory system will cause a flow of air into the lungs through the bronchial tubes of the airways and remove air to the environment. When a person inhales air through the nose or mouth, it passes to the trachea and bronchial tubes, and is forced into smaller and thinner bronchioles, eventually arriving at a vast amount of tiny round sacs of air called alveoli. The exchange of oxygen from the lung into tiny blood vessels, capillaries is by diffusion of gas done through the walls of these alveoli. The blood inside the capillaries circulates across these walls alveoli, exchanging oxygen from the alveoli and into the blood, and vice versa for the carbon dioxide [1].

There are several diseases and conditions that can affect the respiratory system. Chronic obstructive pulmonary disease (COPD) is the most common respiratory disease that arises from airway obstructions. The world health organization (WHO) reported that COPD is the most cause of mortality in the Sub-Saharan countries including Ethiopia due to an increase in demographic environmental changes [2][3]. COPD is a term that is used to include chronic bronchitis, emphysema and Asthma. Chronic bronchitis is a condition of increased swelling and mucus production in the air ways tubes. Due to the swelling and extra mucus, the airway obstruction can occur that causes the breathing trachea tube and bronchitis to be smaller than the normal. Emphysema is a condition that involves damage to the walls of the air sacs (alveoli) of the lung which is the functional units of lungs that absorbs only oxygen gas from the other gas that breathes all gases into the lungs [4].

Asthma is also one of the inflammatory COPD of the airways that is associated with an exaggerated airway narrowing and obstruction response by coughing, wheezing and etc.

As these diseases do not progress quickly, it can be difficult to detect early warning signs through symptoms alone [5]. COPD test is the test to diagnosis, monitor, and screen COPD patients in the health facilities. Therefore to diagnosis and monitor COPD, pulmonary function test (PFT) is performed. In PFT, abnormality of the respiratory system is determined through analyzing a patient's breaths. This test has proven to be the most effective way to screen, diagnose, and monitor COPDs [6][7]. Spirometer is an essential device to perform PFT procedure.

There are different types of spirometer designs, and differential pressure-based spirometer is the most dominant method for PFT [8]. The differential pressure is used for flow meter and volume measurement while the patient breathes to the venture tube called patient breathing tube. The sensitivity of the patient breathing tube that is used to create the differential pressure with altitude correction are the main important parameters used to improve the result of a differential pressure-based spirometry.

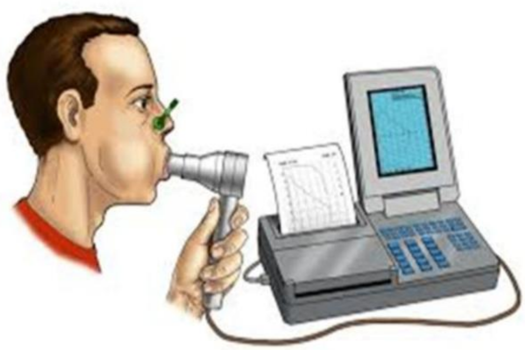


Figure 1: Pulmonary Function Test using Spirometer [9].

1.2. Problem Statement

In a differential pressure-based spirometer, the signal is acquired through pressure difference between two points of the patient breathing tube (PBT) that is created by the patient by breathing outward forcefully against the atmospheric pressure. If the patient finds it hard to breathe, then the sensitivity of the differential pressure across the PBT created by the patient reduces.

The design of the PBT provides the maximum pressure difference when the patient breaths across it in order to maximize the sensitivity. The detection sensitivity of the PBT and altitude correction are the main important parameters which need to be considered while designing a differential pressure-based spirometry. Based on literature review and clinical visits, it is seen that most of the existing pressure-based spirometers do not consider the sensitivity of the PBT and altitude correction. So, the current thesis aims to design and simulate a simple customized pressure-based spirometer by increasing the sensitivity of the PBT and taking the altitude correction factor into account.

1. 3. Objective

1. 3.1. General Objective

The main objective of this thesis is to customize a differential pressure-based spirometer device with respect to altitude correction factor in order to improve the accuracy of COPD tests.

1.3.2. Specific Objectives

The specific objectives of the thesis are: -

- ✓ Review the existing literature and conduct clinical visits at selected Hospitals
- ✓ Design and simulate the patient breathing tube
- ✓ Redesign the differential pressure-based spirometer that considers sensitivity PBT and taking the altitude correction factor into account.
- ✓ Simulate the air flow from patient by software during breathing

1.4. Significance of the Study

This thesis work intended to design PBT and customize a simple differential pressure-based spirometer for use in COPD testing (diagnosis and monitoring of COPD) with increasing the sensitivity of the PBT and taking the altitude correction factor into account. The differential pressure-based spirometer was designed to measure the amount of lung capacity volume and air flowrate graph with numerical values during breathing by increasing the sensitivity of the PBT that helps a pulmonologist to detect the low signal of the patient with hardness in breathing out ward with pressure based spirometer. This helps the diagnosis and monitoring of the COPD patient and the progress of the patient to measure the lung capacity to check how the patent is breathing.

1.5. Delimitation

The differential pressure-based spirometer that has been redesigned in this thesis does not differentiate the exact type COPD, like emphysema, bronchitis or asthma.It is used at the primary level just to identify whether a patient is affected by obstruction of the airflow or not in the emergence and primary health care during inhalation and exhalation.

In this thesis, a PBT that is used to create enough differential pressure is designed. Finally patient breathing and signal outputs of differential pressure based spirometer have been simulated on a computer.

1.6. Organization of the Thesis

This rest of the thesis has been organized into six chapters. Chapter two gives an overview of anatomy of the human respiratory system, and breathing mechanisms. This chapter provides a basic understanding of the mechanics of forced expiration which will be discussed along with the mechanics of airflow that is relevant to the current thesis work and the relation of pressure to volume in the human breathing system. The chapter also provides the history of pulmonary function test measurements mainly on spirometer equipment, defines and discuss its parameters, as well as the graph of the spirometry testing. Chapter three provides review of published researches on the design of different types of spirometers recently and gaps behind the design. Chapter four provides detailed presentation of the materials and methods used in the research. Chapter five provides the necessarily design and simulation. Chapter six presents and discusses the results obtained in the study and Chapter seven gives a summary and the conclusions arrived as well as suggestions for further research into this topic.

CHAPTER TWO

HUMAN RESPIRATORY SYSTEM AND SPIROMETER

2.1. Introduction

All cells inside any living organism continuously use oxygen for their metabolic activities in order to get energy from nutrients. During the metabolism, carbon dioxide is produced as waste, however if not removed from the body can cause acidity that can be toxic to cells [9]. Therefore**,** the main function of the respiratory system is gas exchange between the human body and environment in order to release energy from nutrients. Oxygen which is needed for cells from the external environment is transferred into our bloodstream while carbon dioxide as a waste product is expelled into the outside air [9].

The respiratory system performs several other functions, including the regulation of acid-base balance in the blood, enabling vocalization, participating in defense against pathogens and foreign particles in the airways, providing a route for water and heat losses i.e. by the expiration of air that was moistened and warmed during inspiration.

2.2. Anatomy of the Human Respiratory System

The main parts of the human respiratory system are the oral, nasal cavities, the airways, the lungs, and the chest structure. The airways are a path of tubes leading to the two lungs. The chest structure provides the function of moving air in and out of the lungs during breathing. While breathing, air passes through the nose and mouth that enters to the respiratory system that passes to the pharynx, which is a passage for both air and food. The pharynx is split up into two tubes to enter the two lungs, the esophagus, used for food passage. The larynx is connected to the trachea, which air is pulled down into trachea. The trachea branches into two bronchi which carry oxygen further into each lung [10].

The bronchi is divided into a branches of small tubes the bronchioles. The bronchioles, are surrounded by very smooth muscles, carry oxygen deep into the lungs.

The alveoli are tiny air sacs situated at the end of each bronchiole. The alveoli lie in clusters called alveolar sacs which have extremely thin walls through which oxygen is diffusing into the blood and carbon dioxide is taken out of the blood as shown in figure 2 below.

The surface of the alveolus is surrounded by the network of small blood capillaries where the passage of blood carrying the gases that are exchanged by diffusion [11].

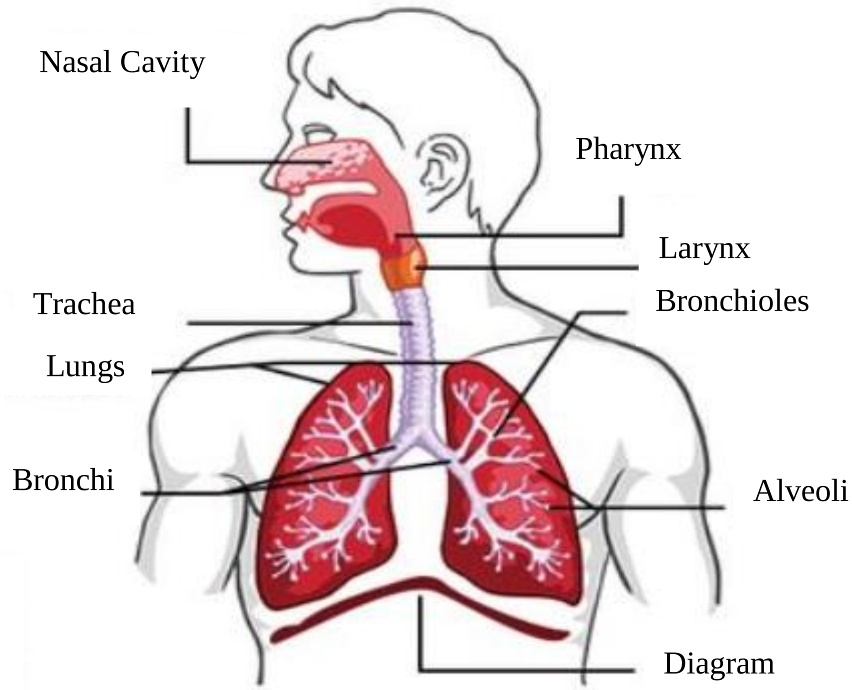


Figure 2: Overview of Human Respiratory System Anatomy Diagram[11].

The diaphragm is also the main part of the respiratory system. It is a skeletal muscle sheet attached to the lower part of the chest thorax. When the diaphragm expands lungs are inflated oxygen is pulled into the lungs. When this muscle contracts the lungs deflate carbon dioxide is pumped out of the lungs. A respiratory cycle includes inspiration, the movement of air from the external environment to the alveoli, and expiration, movement in the opposite way [9] [10].

To accomplish this function three processes, collectively called respiration, must occur. These are:-

1. Pulmonary ventilation: is called breathing that the flow of air moved into the lungs during inspiration and out of the lungs during expiration. Therefore, gases are changed between the lungs and environments and refreshed frequently.

2. Internal respiration: is gases exchange between blood in capillaries and tissue cells. The blood gains oxygen and loses carbon dioxide.

3. External respiration: gases exchange between the alveoli and the pulmonary capillaries. In this process, pulmonary capillary blood gains oxygen and loses carbon dioxide [9] [11].

2.3. Mechanics of Breathing

Airflow into and out of the lungs is driven by pressure differences between the atmospheric pressure and air pressure in the lungs. When atmospheric pressure exceeds intrapulmonary pressure (alveoli pressure) air flows into the lungs, and when intrapulmonary pressure exceeds atmospheric pressure, air flows out of the lungs, exhalation. Thus increasing the volume of the lungs will decrease intrapulmonary pressure, whereas decreasing lung volume will increase intrapulmonary pressure as shown below in figure (3). Therefore, the respiration pulmonary ventilation process commonly called breathing that consists of two phases [9]. These are inspiration and expiration.

2.3.1 Inspiration

Inspiration is the active part (need muscle contraction of diagram) of the breathing process, which is initiated by the respiratory control center in the medulla oblongata (Brain stem). When medulla oblongata activate a contraction of the diaphragm is happen and intercostal muscles leading to an expansion of the thoracic cavity and a decrease in the pleural space pressure [9].

The diaphragm is a dome-shaped structure that separates the thoracic and abdominal cavities. When the contraction has occurred, the diaphragm moves downward and because it is attached to the lower ribs and also rotates the ribs toward the horizontal plane that expands the chest cavity. When they contract the ribs are pulled upward and forward causing a further increase in the volume of the thoracic cavity. During this happen fresh air flows along the branching airways into the alveoli until the alveolar pressure equals the pressure at the airway opening [9].

2.3.2 Expiration

Expiration does not require muscle contraction during breathing, so it is a passive process. Because at the end of inspiration, the chest walls and lungs are expanded by muscle contraction. This is achieved by simply relaxing to regain their original shape, these muscles occur when the brain stem is no activated. As consequence, the elastic chest walls and lungs recoil to their original positions i.e. resting. As the chest wall and lungs recoil, the volume of the lungs decreases, causing alveolar pressure to increase to a value greater than atmospheric pressure which is used to diffuse the air from alveolar to the atmosphere. Air flows out due to the pressure gradient until the volume in the lungs equals the functional residual capacity (FRC) [10].

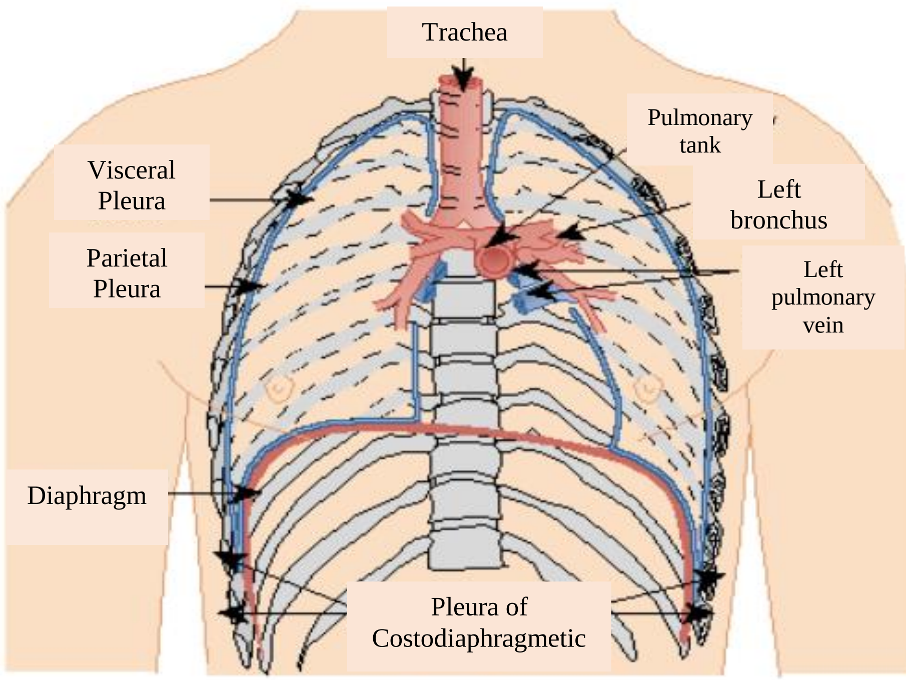


Figure 3:Human Breathing Mechanism [10]

2.4. Pressure Changes in the Thoracic Cavity

The difference of atmospheric pressures and the pressure of the respiratory system are expressed as relative pressures to the atmospheric pressure. When alveolar pressure is zero, it means that alveolar pressure equal to atmospheric pressure at this time the breathing is not happening at all [11].

The values of respiratory pressures are described relative to atmospheric pressure $Patm$, which is the pressure exerted by the mixture of atmospheric gases surrounding the body. A negative respiratory or inhalation pressure in the respiratory area indicates that the pressure in that region is lower than atmospheric pressure so the air diffuses to the lung from higher pressure of the atmosphere.

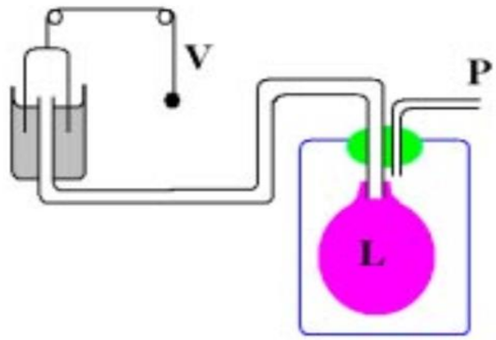


Figure 4:Pressure Volume Relationship [11]

Boyle's law states that the pressure of a given quantity of gas is inversely proportional to its volume at constant temperature.

$$P_1V_1 = P_2V_2 \tag{1}$$

Volume changes lead to pressure changes, and pressure changes lead to the flow of gasses to equalize the pressure. This follows from the above equation where P is the pressure of the gas, V is its volume, and subscripts 1 and 2 represent the initial and resulting conditions respectively.

At sea level, atmospheric pressure is 760 mm Hg. This pressure can also be expressed in different units: $Patm$ = 760 mm Hg ≈ 1 atm ≈ 101.325 kPa. When the altitudes is higher than sea level, atmospheric pressure decreases [12].

2.4.1 Primary Pressures Associated with Respiratory Ventilation

The diffusion of gases brings the partial pressures of oxygen and carbon dioxide in the blood and alveolar gas to equilibrium at the pulmonary blood gas barrier. In order to make their function properly the lung has to deal with three pressures which are related to the respiratory system [13][10].

1. **Intrapulmonary pressure: -** this pressure is called alveoli pressure is the pressure of air within the alveoli. The difference between intra-alveolar pressure and atmospheric pressure is the pressure gradient that drives ventilation gas in and out. At rest, it is equal to atmospheric pressure [10].

2. **Intrapleural pressure (Pip): -** is the negative pressure inside the pleural space. It is always negative during normal breathing because opposing forces exerted by the chest wall and the lungs tend to pull the parietal pleura and visceral pleura apart. Intrapleural pressure can be positive during forced expiration [10].
3. **Transpulmonary pressure (Ptp): -** is the difference between the intrapulmonary and intrapleural pressures. Transpulmonary pressure keeps the lungs from collapsing. The size of the transpulmonary pressure determines the size of the lungs at any time[10].

2.5. Lung Volume

One of the fundamental processes in the respiratory system is the ventilation, which takes place in the lungs. So the main function of the lungs is to transfer oxygen from the atmosphere to the bloodstream and to release carbon dioxide in the reverse direction.

Lung capacity is the volume of air in the lungs. The total lung capacity is the volume of gas in the lungs after maximal inspiration. The total lung capacity of an adult human male is about 6 liters (L). The volume of air left in the lungs after maximal expiration is called the residual volume (RV), whereas the volume of air left in the lungs after a normal expiration is called functional residual capacity (FRC). The residual volume is about 1.2L/1.1L for an adult human male/female respectively.

The difference in volume between the total lung capacity and the residual volume is called the vital capacity (VC). Figure(5) shows below a time-volume diagram with a graph presenting the lung volumes to each other [14][15].

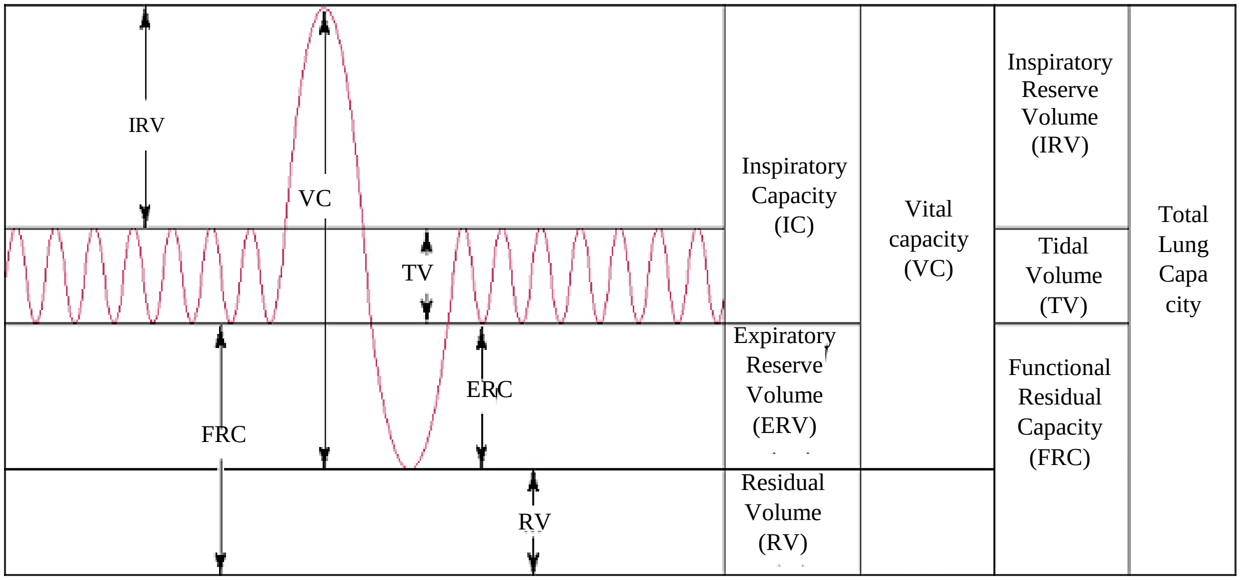


Figure 5: Lung Compliance Volume [14]

2.6. Diseases Related to the Respiration System

Chronic respiratory diseases are major contributors to mortality rate, disability, and medical cost. This disease refers to many disorders affecting the different respiratory system organs including lungs that range from infections such as influenza to life-threatening ones like lung cancer. Some of these can lead to respiratory failure. Every day a human breathes thousands of times. During this breathing, the lung takes in oxygen from the air and delivers it to the bloodstream and expels carbon dioxide from the body. A problem in any part of it could lead to respiratory disease that makes a person difficulty in breathing. Different factors could lead to respiratory diseases such as smoking, environmental pollution, genetics, and infections, which are considered the leading causes of lung diseases. Among the most common lung diseases, chronic obstructive pulmonary disease (COPD) is the most cause of death in the world. So this thesis mainly focuses on improving the diagnosis of CPOD through the designing of optimum PBT and reduce the effect of altitude in the pressure-based spirometer.

2.6.1 Chronic Obstructive Pulmonary Disease

The term COPD is used to describe chronic respiratory diseases characterized by airway obstruction, and a long time blockage of respiration airway [14]. COPD commonly the combination of emphysema, chronic bronchitis, and Asthma as shown in the Figure (6).

COPD can arise from genetic or environmental reasons such as exposures to air pollutions and smoking. COPD is also progressive and an irreversible decline in lung functions.

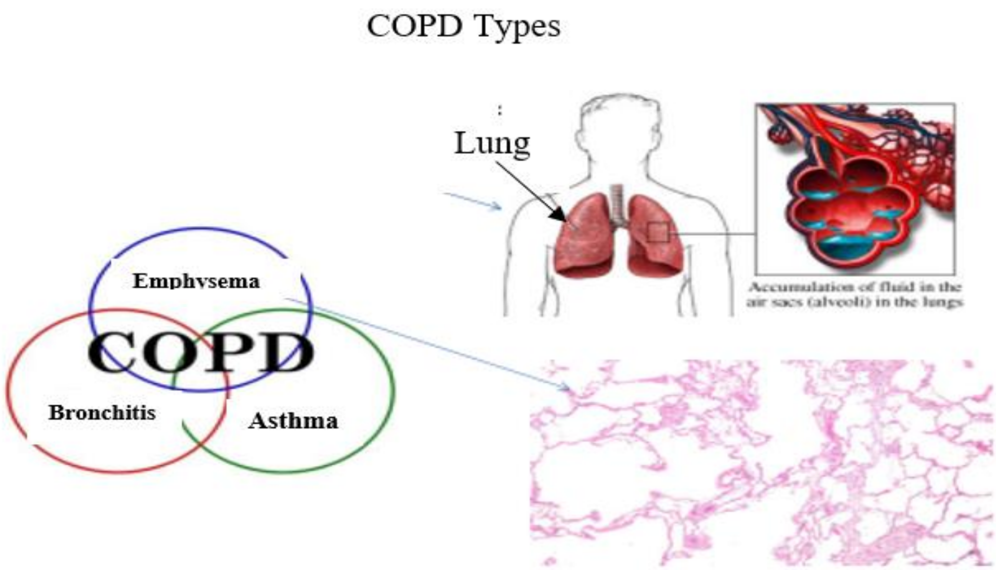


Figure 6:Chronic Obstructive Pulmonary Disease [14]

1. **Emphysema**

A condition that damages the walls of the alveoli of the lungs. This may result in loss of the stretchiness of the alveoli. So, this disease is make the COPD patient difficult to push the air out of the lung that contains the air more than the normal content.

2. **Bronchitis**

This condition of increasing the production of swelling and mucus in the breathing tube. So the airway obstruction occurs in chronic bronchitis because the swelling and extra mucus cause inside the breathing tube to be smaller than normal [14].

3. **Asthma**

Asthma is a disease that affects the respiratory system of the main organ lung and trachea. It is a chronic disease of airway inflammation that manifests clinically with coughing, breathlessness, and wheezing and chest tightness.

2.7. Pulmonary Function Tests

Pulmonary function tests (PFT) are very useful tests to diagnose several lung diseases. There are different techniques of performing PFT.

Diffusion capacity: Estimates the transfer of oxygen from the alveolar air to the blood cell for metabolic activities.

Maximal Voluntary Ventilation (MVV): The subject breaths as hard and fast as possible for 10-15 sec, and then adjust it to 1 min.

Maximal Inspiratory Pressure (Pi max) & Maximal Expiratory Pressure (Pe max): to measure the strength of the respiratory muscle by the amount of pressure can generate.

Walking Oximetry: To detect the hidden diffusion defect of oxygen through oxygen saturation at rest, 4 and 6 minutes' walk technique [16].

Spirometer: Simplest and most common technique of performing PFT for COPD tests.

2.7.1. Spirometry and its Measurements

Spirometer is used to analyze how well the lungs receive, hold, and utilize an air. It provides quantitative and qualitative data with graph display important to make the decision on diagnostics of COPD.

This device is used to measures lung volumes and air flow, during diagnosis the patient was ordered to exhale forcefully as quickly as possible. It produces a graphical representation (spirogram) of the forced expiratory maneuver that plots volume expired over time and flow rate over time [18] [19]. These parameters are calculated based on an individual's age, height, sex, and race since the diagnostic thresholds for obstructive lung disease differ by body size as well as are affected by altitude. The following standard measurements will be used in the spirometry parameters:

1. Forced Vital Capacity (FVC): The maximum volume of air exhaled forcefully after a maximal inspiration.

This result may differ based on the age and the condition of the patient and is also affected by the altitude. For adults, this forced exhalation should last at least be 6 seconds; however, persons with COPD may take considerably longer time to exhale air.

2. Forced Expiratory Volume in One Second (FEV1): The volume of air exhaled during the first second. Since the normal patient exhales almost 80% from the total forced vital capacity.

3. Peak Expiratory Flow (PEF): The highest airflow rate measured during the FVC maneuver. PEF is measured in liters per second.

4. FEV1/FVC: The ratio of FEV1 to FVC is expressed in a percentage. Currently, the spirometry primary measurement used to assess obstructive lung disorders will be the ratio of forced expiratory volume in first second to the forced vital capacity. The use of FEV1 is a useful diagnostic method for the early detection of chronic obstructive pulmonary disease.

This information can help the physicians in the management and treatment of COPD. If FEV1/FVC is less than 0.7, then the subject is considered as a potential COPD subject suffering from airflow obstruction. Normalization of FEV1 according to expected value based on age, height, sex is called FEV1% predicted of that specific patient [17][18]. COPD can be predicted based on the ranges of the table below:-

No	**COPD Stages**	**Spirometry**
1	Normal	FEV1≥80%
2	Moderate	50%≤FEV1<80%
3	Severe	30%≤FEV1<50%

Table 1: Ranges of COPD prediction

Measurements of exhaled volume (in liters), time (in seconds), and airflow rates (in liters per sec) are determined and displayed on the spirograms. Two types of graphical representation, spirograms will be used in the spirometry component:

- **Volume-Time**: The basic volume vs. time curve contains points corresponding to the FEV1 and FVC.

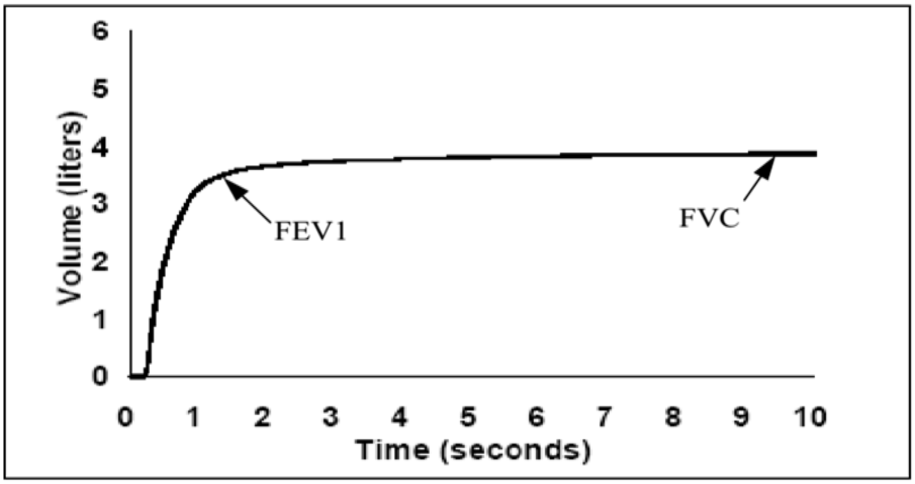


Figure 7:Volume Vs Time graph [19]

- **Flow-Volume:** The expiratory flow vs volume curve displays instantaneous airflow rates as a function of volume exhaled. This curve contains points corresponding to the PEF and FVC.

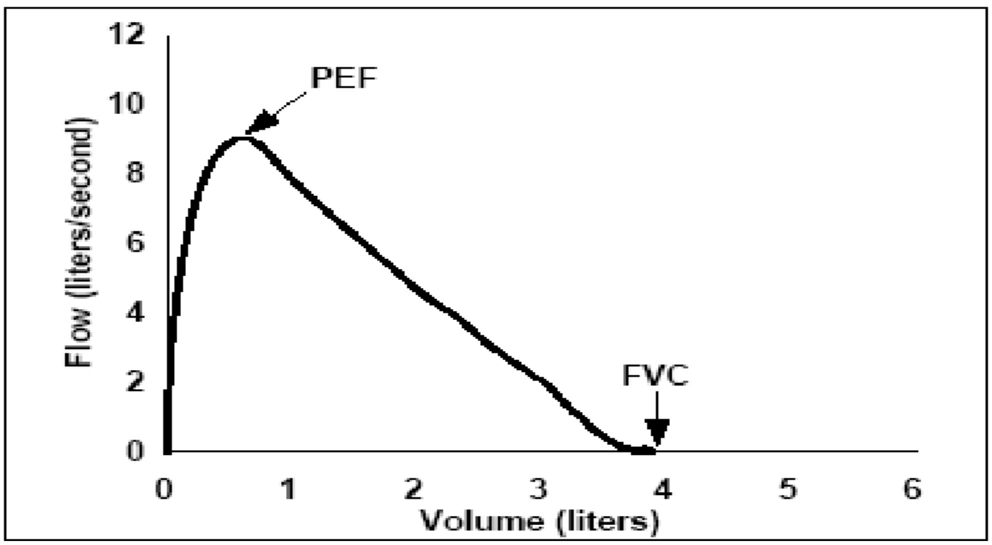


Figure 8: Flow Vs Time graph [19]

Since the development of COPD is slow and symptoms tend to be noted by patients only after there has been a significant loss of lung function, often to 50-60% of the predicted value. So the spirometer is most predominant PFT to detect and screen out the stage of COPD at the primary care health care [20].

CHAPTER THREE

LITERATURE REVIEW

There are two methods of spirometer design, these are either volume-based spirometer or flow-based spirometer design.

3.1. Volume-Based Method

3.1.1. Water Sealed Spirometer

Water sealed spirometers work on the principle of measurement of air volume through the displacement of water while the patient is breath outward into the water sealed jar. When the patient exhales the water changes its position and measures the volume of air that can exhale [8].

This design is the simplest and oldest design, it uses a hollow cylinder that is inverted and lowered into a jar of water containing a tube due to the movement of gas. The bell rises and lowers as gas moves into or out of the gas space between the bell and the water when the patient is exhaling. A pen is attached to the tip of the bell, which is used to draw the volume-time graph on paper [8]. The basic principle of operation is that the height of the bell is related to its volume by the formula for the volume of a cylinder as shown from the figure 8 below:

$$V = \pi r^2 h \tag{2}$$

However, this device is quite large in comparison to the other designs and less accuracy so we can't get the market right now.

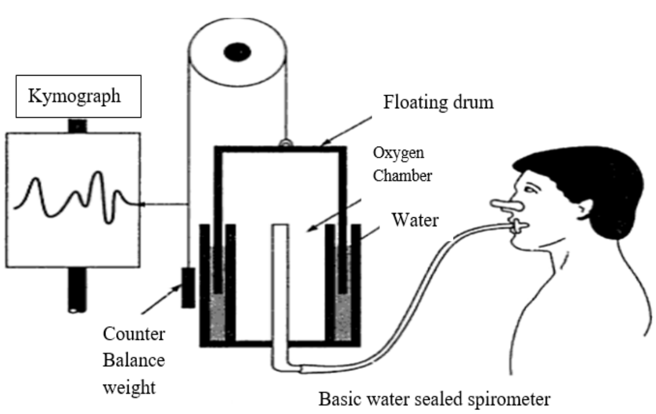


Figure 9:Basic Water Sealed Spirometer [8]

3.1. 2. Dry Seal Spirometer

A dry seal spirometer design is a thin layer of flexible material that used to seal the bell to its base. When the patient exhales into the latex the gas enters the bell, the latex keeps the gas and forces the bell to move up due to the pressure created during expiration. As water-sealed spirometer dry-seal spirometers also less accurate and reliable [8]. The figure below show that the dry seal spirometer.

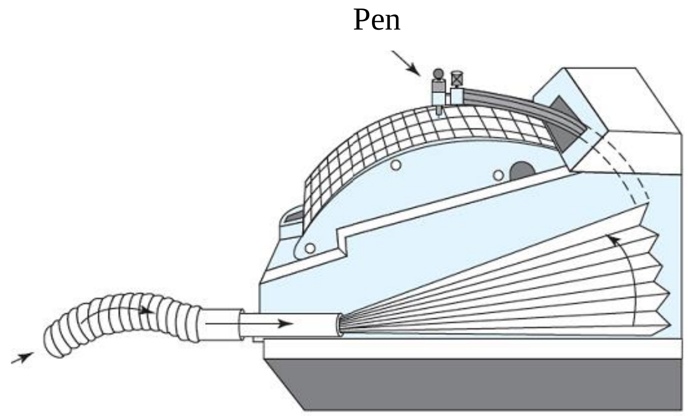


Figure 10: Dry Seal Spirometer [8]

3.2. Flow-based Spirometer

Also called electronic sensor, flow-based spirometer is the latest technology, which acquires the signal while the patient is breathing. Currently most of the researches to develop spirometers focuses on their electronic sensor. There are many kinds of electronic spirometer designs but the thermal sensor, turbine flow sensor, and differential pressure sensors are the most dominant ones for electronic spirometers design [8].

3.2.1. Thermal Flow Sensor

This type of pulmonary function test device is a thermistor-based circuit used to monitor respiration. The thermistor sensor is placed in the PBT at the output airflow by sensing differences in the temperature of the warmer expired air and cooler inhaled ambient air from the environment during air breathing. The flow electrical signal generated is related directly proportional to the sensor temperature. Under the constant current, a thermistor will change its resistance as a function of the temperature as shown in figure (11).

This value is converted as a voltage and amplified for further analysis. Finally, the voltage gain is filtered and analyzed to get the necessarily graph output with numerical spirometric parameters. However this types of spirometer lack accuracy due to poor sensitives of the thermistor since the difference of temperature between exhalation and inhalation is almost the same [8][20].

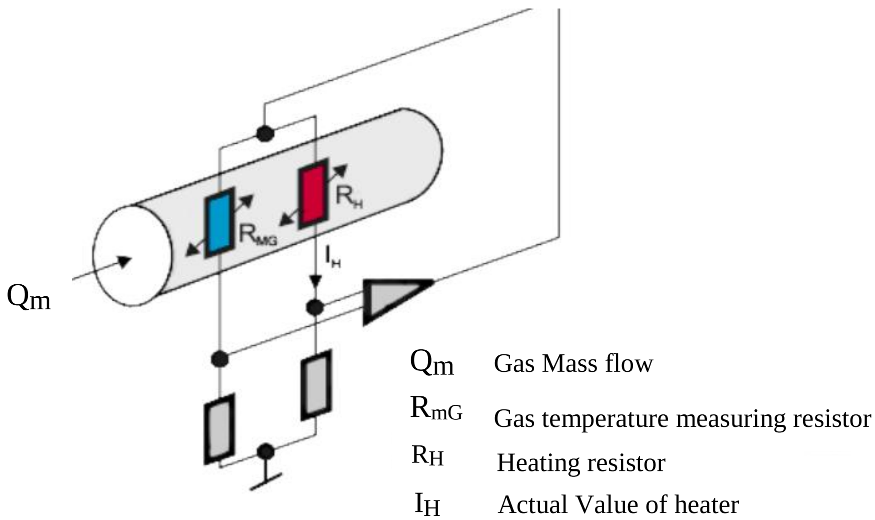


Figure 11: Thermal Flow Diagram [20]

3.2.2 Turbine Flow Sensor

The turbine flowmeter requires well designed and placed aerodynamic blades that are suitable for the air flow condition. The bearings are both smooth and durable to survive the sustained high-speed rotation of the turbine as shown below in the figure 12.

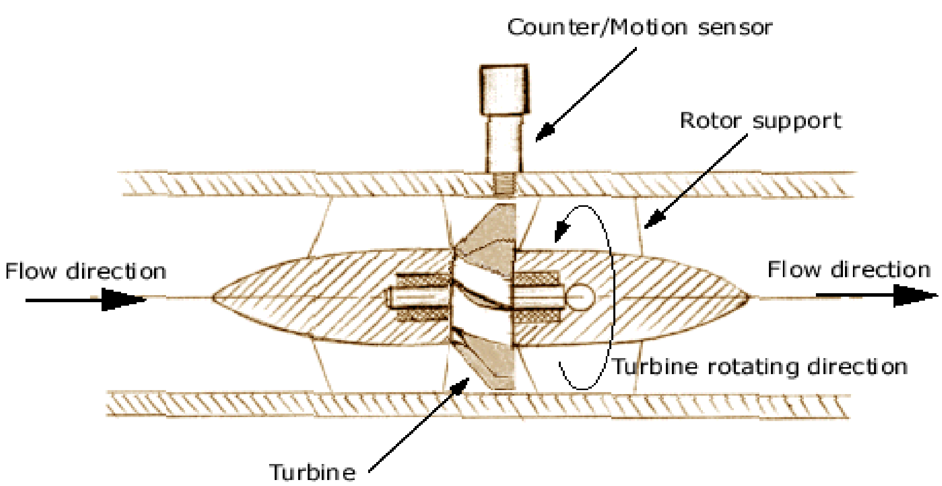


Figure 12:Turbine Flow Diagram [21]

To measure the rotation speed this type of sensor measures the angular velocity or rotation speed using visible Light Emitting Diode (LED) and Light Dependent Resistor (LDR). While the patient is exhale the turbine rotates its makes the LDR exposed to the LED light on /off through it measures rotation speed finally it calculates the volume and flow rate of the patient exhale [22].

LDR is a light-dependent resistor that provide change in resistance when exposed to LED light. Based on the light intensity the resistance of LDR will decreases or increases. The spectral response of LDR depends on the wavelength of visible spectrum. LDR gives high response having wavelength ranges 530 nm to 560 nm. Green visible light has a wavelength range of 495nm to 570 nm which cover the highest spectral response range of LDR hence it would be better to use green visible light LED [22].

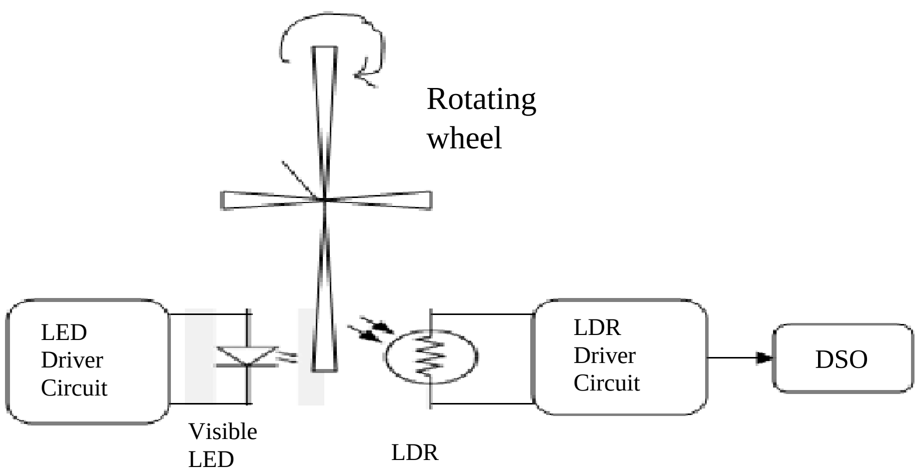


Figure 13:Turbine Flow Sensor Diagram [22]

The following are the drawbacks turbine flow meters sensor:

- ✓ Installation must be done carefully to avoid errors
- ✓ Accuracy is affected by bearing degradation due to age
- ✓ Errors caused by viscosity changes
- ✓ Require frequent calibration

3.2.3. Flow Sensor using Differential Pressure Sensor

Differential pressure-based spirometer design mainly deals with gas continuity measurement theory. During breathing the airflow rate is measured by a differential pressure sensor.

Differential pressure sensor measures the pressure drop across the two opening of PBT. From the measured pressure, the velocity of the air moving through the PBT is calculated using the Bernoulli's equation as shown below.

$$P_1 + \frac{1}{2}\rho V_1^2 + \rho g y_1 = P_2 + \frac{1}{2}\rho V_2^2 + \rho g y_2 \qquad (3)$$

From the dynamic pressure

$$P = \frac{1}{2}\rho v^2 \qquad (4)$$

Where: ρ = Density of Air at 300^O K V = Velocity of the flow of fluid

Once the dynamic pressure is extracted from the sensor, the velocity of flow can be determined using the above equation. From the flow continuity when a gas flow through a pipe of varying diameter the same volume flow is done across cross-sectional.

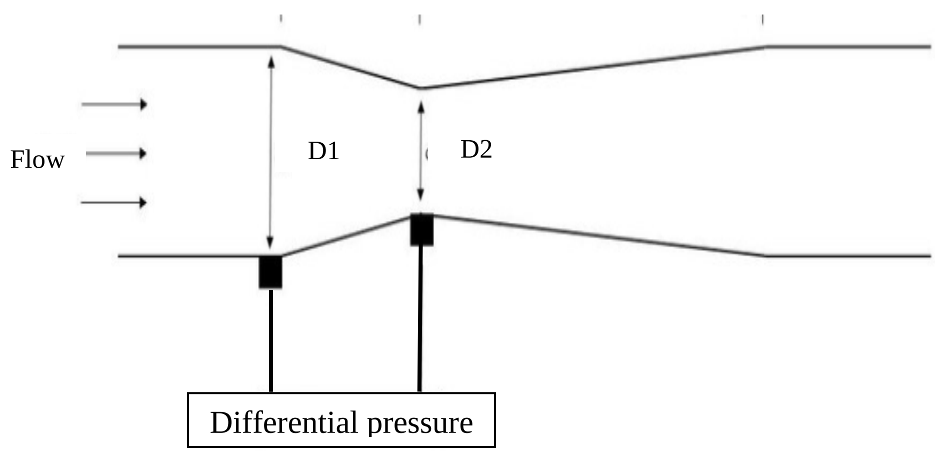
$$Q = \frac{V}{t} \qquad (5)$$

Where Q= volumetric flow rate
V= volume and t= time

But the volume is the product of cross-sectional area and length V= A*L

$$Q = \frac{A*L}{t} \qquad (6)$$

But the L/t is equal to the velocity



Therefore, Flow-rate: = A * Velocity Where: A = Area of cross-section of tube

So the volume can be calculated from the integration of the flow rate

$$V = \int Q.\, dt \qquad (7)$$

After that, we can get all the parameters related to the spirometry measurement stated above that help for the diagnostic pulmonary disease. So far, many kind research have been conducted on the differential pressure-based spirometer. Some of the selected literature conducted on this thesis is reviewed below.

In September 2012, a group of researchers proposed “Using a Differential Pressure Sensor as Spirometer” to design the differential based spirometer that used diagnosis the COPD using Bernoulli’s equation to get the spirometry parameters. The methodology is acquiring a signal using differential pressure by PBT and analyzed using Matlab software. However, this paper does not mention what type of venture tube was used to create the DP across the PBT and the effect of altitude correction on the pressure measurement [24].

On August 1, 2012, Jorge González also proposed a “Spirometer Demo with Freescale microcontroller” using a differential pressure sensor and microcontroller to get the pressure difference in the computer for analysis of the spirometer. In this paper the venturi effect is discussed and presented in the diagram method. Finally, the flow diagram of how to implement is also mentioned in this paper. However, the same problem, they are not mention how the data is acquired and effect of altitude correction accordingly is not mentioned [23].

The work of Alejos-Palomares *et al* on “Digital Spirometer with LabVIEW Interface” published in 2008 at IEE computer society also uses differential pressure-based spirometer design using the LabVIEW software for signal acquisition and Bernoulli’s equation for further analysis. However, this paper doesn’t mention how the air is flowing through the breathing tube while the patient is breathing and lacks customization of pressure according to altitude[24].

From the literature review, we have found out that the previous reserches conducted in the field did not consider the effect of altitude and its impact in affecting the accuracy of pressure-based spirometer measururements [25][26][27].

Moreover, I visited the clinical setup of two hospitals found in Addis Ababa where the pressure-based spirometer is found and interviewed the physician. These hospitals are Black Lion Specialized hospital and Teklehaimanot general hospital.

During my visit, I spoke with the doctors about the sensitivity of the PBT test for COPD patients who are finding it difficult to breathe and the altitude correction factor for improving COPD diagnosis. However, I has been unable to find a physician who considers the impact of PBT and altitude correction as a potential factor influencing PFT results.

To conclude, the reviewed paper and the clinical visits showed that the pressure-based spirometer did not incorporate the sensitivity of PBT on the pulmonary function test for COPD patient. Therefore, during spirometer design the PBT diameter and altitude correction are important parameters for pressure-based spirometer.

CHAPTER FOUR

METHODS AND MATERIALS

4.1. Methods

The method was utilized to improve COPD diagnosis and adapt a simple pressure-based spirometer by creating the PBT and altitude correction. The pressure-based spirometer procedure has three unique processes after the PBT is designed. The first step is using LabVIEW software to create a simulated raw signal. While the patient is exhaling forcefully into the PBT, this signal is similar to the obtained signal from the differential pressure sensor. The PBT is currently being developed in order to improve the sensitivity of the DP sensor's differential pressure measurement.

The second step is the recorded signal is then analyzed and filtered, which is the signal processing. Following signal processing, the differential pressure output is modified with altitude adjustment to increase the spirometric parameter, which may lead to COPD patient diagnosis..

Finally, the general description of the theory of operation and governing physics were elaborated which will provide full information about the differential pressure-based spirometer parameters. At the end, the system is optimized and calibrated using leading standards in the field of spirometry in the signal processing. The general method to approach from the signal acquisition through patient breathing to the final output (spirometer graph) is shown below in the figure (14).

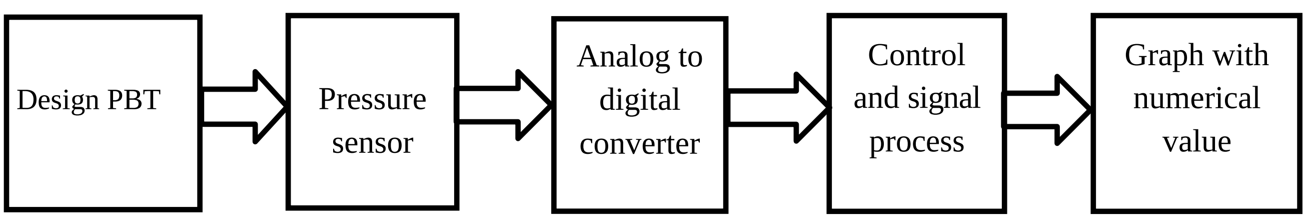


Figure 14: Flow Diagram

4.2. The Signal Acquisition Materials

The standard spirometers measure the flow of air breathing that passes through the sensor to measure air flowrate and volume of the lung with the graph.

To get spirometer parameters and graph the patient must exhale through the venture narrow tube called the patient breathing tube and make the deformability of the pressure in terms of electrical parameters. So that to acquire the raw signal from the patient and analyze different materials were used. The materials needed to conduct the design consists the following:-

- ✓ Differential Pressure transducer
- ✓ 3D printer
- ✓ Filament
- ✓ Ansys 19.2 software and
- ✓ LabVIEW

4.2.1 Differential Pressure Transducer

The difference between two pressure reading measurements at one tube is called differential pressure. The differential pressure transducer is a sensor that measures the changes of pressure across the venture tube while the patient is breathing. This sensor produces in terms of analog electrical signals.

To do this the piezo resistive pressure sensor type is used since it's available easily and cheaply. This sensor contains a silicon chip with an integral sensing diaphragm and four piezo-resistors; when pressure is applied on the diaphragm causes it to change the resistance with an output voltage proportional to pressure. A pressure transducer-based instrument allows continuous monitoring using a recorder or input to electronic storage or control equipment [24]. The figure (15) show that how the differential pressure sensor is connected to the PBT and direction of airflow.

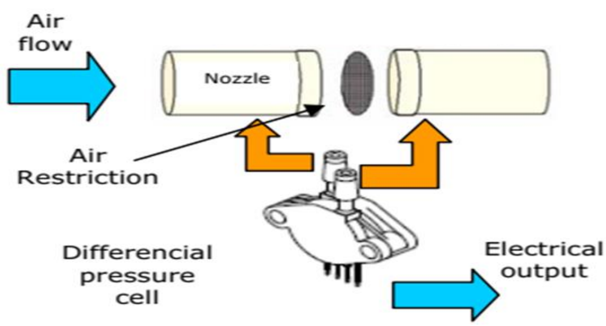


Figure 15:-Differential Pressure Sensor Working Principles [24]

The main implementing of pressure-based spirometer is a converting pressure into the airflow and volume that cross the venture tube which processed by software for analyzing. To achieve this, an integrated silicon pressure sensor is used which is a free scale, MPX5010 differential pressure sensor. It is a differential pressure sensor that delivers an analog output voltage proportional to the applied differential pressure on the sensor.

The key features of this differential pressure sensor used are listed below:

- ✓ Pressure range from 0 to 10 kilo pascal
- ✓ The output is from 0.2 to 4.7 V.
- ✓ 5.0 % Maximum error over 0° to 85°C.
- ✓ Temperature-compensated over -40° to +125°C.

The sensors contain a pair of pipes in its shell to allow a different pressure on each one of them and being capable of detecting the direction of the airflow as shown the figure (16).

Figure 16: Differential Pressure Sensor

Since the differential pressure sensor not find in the market, the out output of this sensor is replaced by generating the signal by the LabVIEW software.

4.2.2. Ansys 19.2 Software

Ansys or analysis system software is used to simulate computer models of structures, electronic, strength, toughness, elasticity, temperature distribution, electromagnetism, and fluid flow. There are different version Ansys modules. From that Ansys fluent software is used in this thesis because it contains the broad physical (geometrical) modeling, mesh and gas flow setup capabilities needed to model flow.

This software is used for modeling the geometrical measurement of PBT and simulate the differential pressure across the PBT that used test the minimum differential pressure measurement of PBT.

4.2.3. 3D Printer and Filament

A 3D printer is a device that uses computer-aided manufacturing (CAM) software to generate three-dimensional items. It takes digital data from a computer as input, which is then transformed into a finished product using CAM software. A 3D printer, on the other hand, creates a three-dimensional model out of a specific material rather than printing the product on paper. 3D printers manufacture tangible objects layer by layer until the model is complete, using a technique known as additive manufacturing. To accomplish this, a material known as filament is utilized to create the physical thing.

Filaments used in 3D printing are thermoplastics, which are plastics that melt rather than burn when heated, molded, and solidify when cooled for the desired shaped of the parts. The filament is fed into a heating chamber in the printer's extruder assembly. This a place where it is heated to its melting point and then extruded through a metal nozzle as the extruder assembly moves, tracing a path programmed into a 3D object file to create, layer by layer, the printed object. This device used to print the PBT prototype which is modeled and simulated in this thesis.

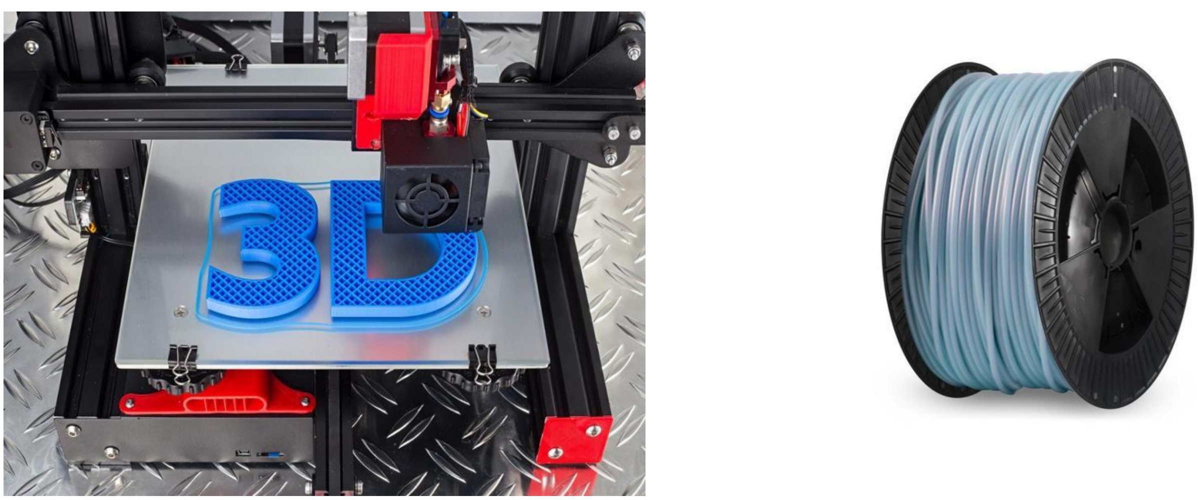


Figure 17:3D Printer and Filament

4.2.4. LabVIEW

LabVIEW, short for Laboratory Virtual Instrument Engineering Workbench, is a programming environment in which you create programs using a graphical notation (connecting functional nodes via wires through which data flows) instead of writing the text. LabVIEW ties the creation of user interfaces called front panels. LabVIEW programs are called virtual instruments (VIs). Each VI has three components:

a. Front panel

The front panel is what allows the operator or end-user to control and monitor the process. It includes software controls and indicators that mimic physical controls such as buttons, sliders, LEDs, and charts.

b. Block Diagram

The block diagram is a graphical representation of the underlying software program. It consists of icons that represent typical programming elements such as constants, variables, subroutines, and loops. Terminals, Sub VIs, functions, constants, structures, and wires are just a few of the block diagram components that transport data. A block diagram is a form of flowchart that is specialized and high-level. It's a helpful tool for both creating new processes and refining old ones. The block diagram provides a rapid, high-level picture of the work in both scenarios.

c. Terminals

This a wiring tool to connect the different objects and to create the logical flow between the object on the Block Diagram, A connector shows available terminals for data transfer and executes the program.

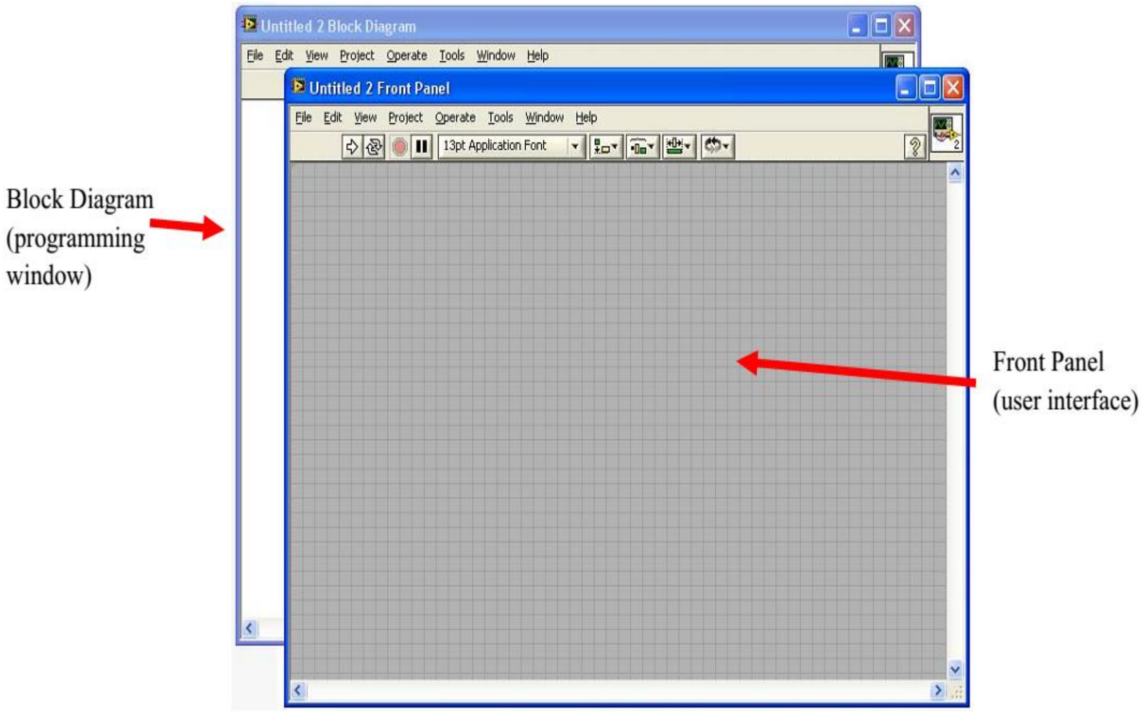


Figure 18: LabVIEW Block Diagram and Front Panel

Benefits of LabVIEW

- ✓ It provides extensive support for accessing instrumentation hardware.
- ✓ Real-time visual debugging features.
- ✓ Built-in drivers and function libraries for the serial, parallel, and network computer ports.
- ✓ Plug and play interface devices for external equipment.
- ✓ Built-in interactive graphic control and display.
- ✓ A graphical user interface and compiled language for fast execution.

The programming language used in LabVIEW also referred to as G, is a dataflow programming language.

Execution is determined by the structure of a graphical block diagram on which the programmer connects different function nodes by drawing wires. These wires propagate variables and node can execute all its input data. I used this LabVIEW software for the following main.

a. Signal generating

Since the differential pressure sensor isn't available on the market, we'll use the LabVIEW program to generate a signal that's equivalent to the sensor's original signal. This includes all of the settings (frequency, amplitude, period and noise).

b. Signal processing

This software include all the modules that used to filtered the original signal from noise using specific cutoff frequency and processed for further analyzing.

c. Display

I used this software to display the signal or wave form on the front panel of the LabVIEW in order to visualized and interpret easily. Therefore, output of the pressure-based spirometer will display using this software.

CHAPTER FIVE

DESIGN AND SIMULATION

The design of the optimal PBT to obtain the largest differential pressure measurement while a COPD patient breathes is discussed in this chapter. It also comprises using Ansys software to simulate differential pressure, analyzing differential pressure in the pipe, and converting spirometric values using Bernoulli's equation while accounting for altitude correction.

5.1. Differential Pressure Measurement

Differential pressure measurement devices can be separated into restrictive and non-restrictive devices by changing their cross-sectional area within one tube which is used to create the pressure difference across PBT. The pressure difference within the patient breathing tube helps to measure low velocity at a high cross-sectional area, high-pressure upstream section, and the high velocity at low cross-sectional area, low-pressure downstream section as shown in the figure (19). When breathing into tube, air flows through a restrictive tube, this leads to a change of differential pressure across the tube. This change of pressure difference is directly related to the airflow rate and lung volume.

Therefore, differential pressure flow measurement is one of the most common technologies for measuring flow rate and volume gases while the COPD patient is subjected to forcefully exhale.

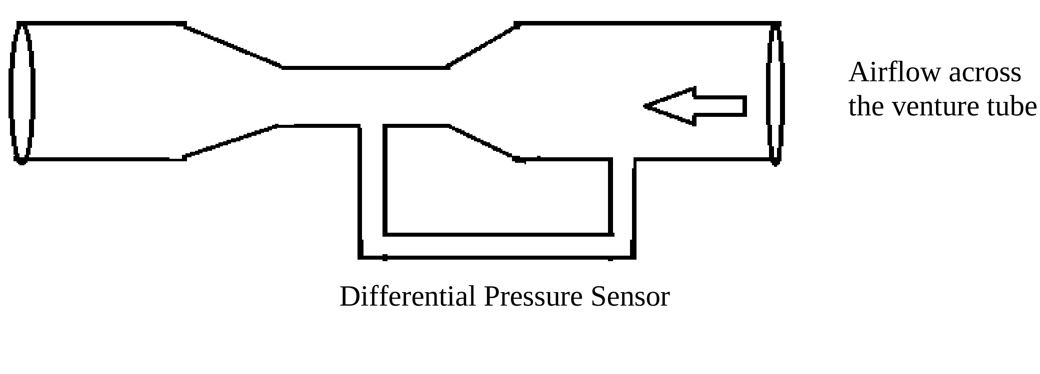


Figure 19: Direction of Air Flow across the Restrictive Tube

5.1.2. Designing Patient Breathing Tube

In the pressure-based spirometer, the patient breathing circuit consists of the PBT that connected with the differential pressure sensor to create the pressure difference across the tube. The main criteria to the design of a PBT is measure the maximum differential pressure present in the PBT and ability to detect the low differential pressure output created by the patient during exhaling. Because COPD patient may induce very low differential pressure which is difficult to detect this low signal. To increase the sensitivity of PBT, the two diameters of PBT are the main parameters that can affect the differential pressure output during breathing. To do this the Ansys 19.2 software were used to design the appropriate PTB and simulate the minimum air velocity that can induce by the COPD patient. So that the design of PBT and analysis of the airflow passes through the following steps.

1. **Geometric Modeling**

The initial step in determining the ideal diameter of PBT is geometric modeling design in order to obtain the largest differential pressure measurement across the PBT. Two parameters were taken into account throughout the modeling process. The first is the bigger diameter D1 at the PBT intake, which is used to breathe in the patient. D1 depends on how wide the patient can open the mouth. The second one is Reynolds Number (Re) which is a convenient parameter that helps in predicting whether a air flow condition will be laminar or turbulent due to the surrounding air. Based on the above information, the lowest air velocity 1m/s as an initial velocity from the surrounding air with the maximum number of Reynolds number equation for laminar flow[28]. Based on the equation (8), the Reynold equation is used to determine the types of flow.

$$R_e = \frac{\rho V D}{\mu} \tag{8}$$

Where ρ density of air 1.225 kg/m^3

V Initial velocity 1m/s

D Larger diameter at the inlet 6cm

μ Viscosity of air 1.81 × 10-5 kg/ (m·s)

R_e Reynolds number i.e. 2000 for laminar flow

As a result, the larger diameter is defined by the Reynold number, which is 2000, because laminar flow occurs when the Reynold number is 2000 and below as shown below:-

$R_e \leq 2000$ Laminar flow

$2000 \leq R_e \leq 4000$ Transitional flow

$R_e \geq 4000$ Turbulent flow

Based on the above parameters when we are calculated, the larger diameter is 3 cm that used to measure the maximum pressure during exhaling forcefully and more convenient for the across the PBT. However to get the maximum differential pressure created across the PBT, the smaller diameter (D2) is the main designed part in this paper. To decide the smaller diameter of the PBT, the minimum and maximum air velocity as initial velocity which is from 2.2m/s to 9.9m/s were simulated with minimum pressure created in the mouth (0.2 to 0.5 kilo pascal) [29][30]. Most of the time to get the maximum pressure difference and smooth flow across the PBT the ratio of the smaller diameter to larger diameter is between 0.25 and 0.75. Therefore to get the smaller diameter of the PBT the minimum air inlet velocity is 2.2m/s and minimum pressure created in the mouth 0.2kilo pascal were simulated, similarly the patient during exhaling. The pressure measured at the point of smaller diameter is positive or converge to zero which makes the patient easy to breathe without any difficulties and blockage.

Finally the maximum differential pressure is created at the 1.2cm diameter which obtained in the figure below (22).This help to detect the low inlet pressure at the larger diameter D1 which is maximum pressure and smaller diameter D2 is minimum pressure that the difference is directly proportional to the differential pressure out of the PBT [31]. The length of PBT is 20cm which more suitable and more appropriate for handling the device as shown in the figure below (20).

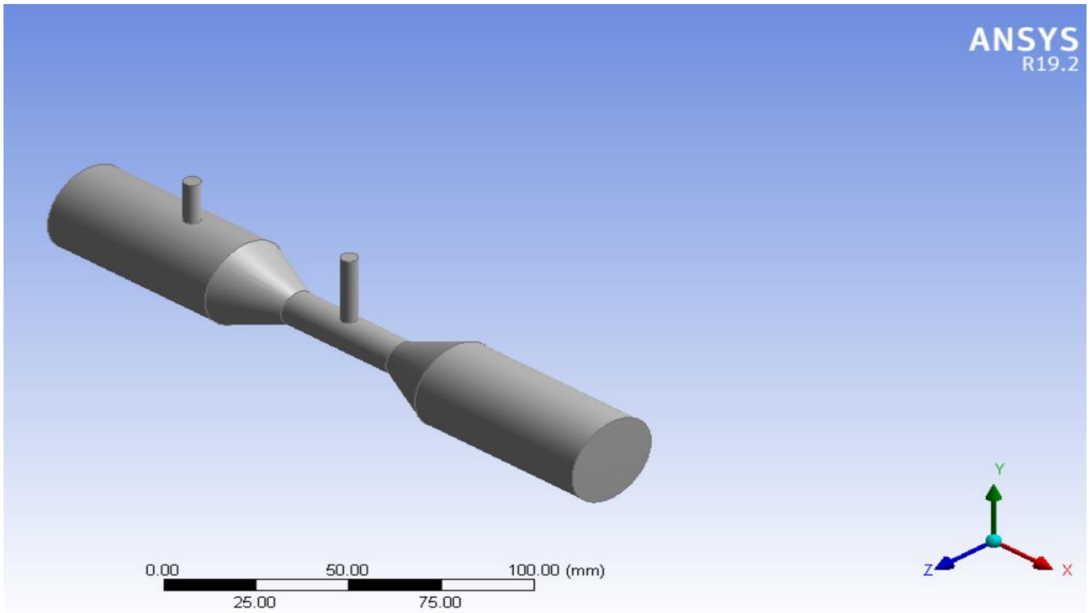


Figure 20: Patient Breathing Tube Design by Ansys 19.2 Software

From the above design parameters, the PBT prototype were produced using 3D printer which is connected to the differential pressure sensor, as shown below in figure (21).

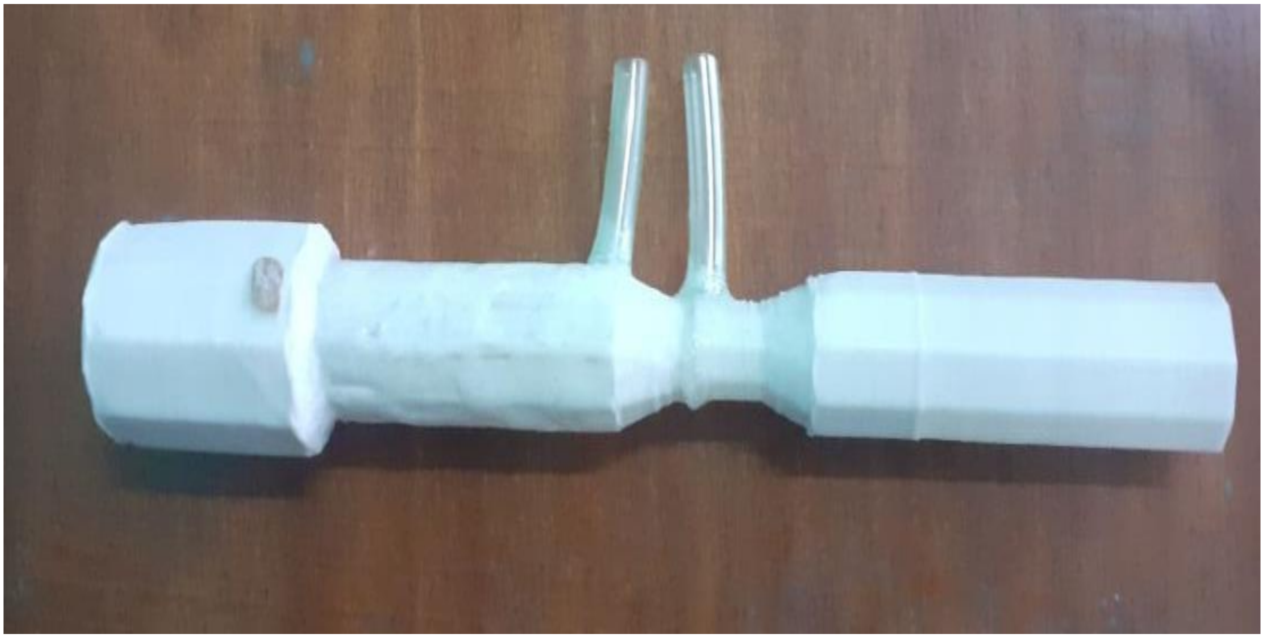


Figure 21: Patient Breathing Tube prototype

2. **Mesh**

Meshing is a step in the engineering simulation process that involves breaking down complex geometries into simple parts that can be used as discrete local approximations to the wider domain. The mesh can affect the simulation's accuracy, convergence, and speed if the geometry is not adequately assessed. By replicating the initial low inlet velocity and high inlet velocity, this component is utilized to assess the influence of the inlet pressure and to visualize where the pressure and velocity are high and low in the PBT. To do this, two main tasks were done here:

Sizing: - used to localize where we are going to visualize the effect of pressure and velocity at specific sizing area

Facing: - used to label the inlet, wall pipe and outlet of the pressure where the effect of pressure and velocity could be measure [32].

3. **Setup**

This comes after fully completing the meshing of your design geometrical modeling. Under this, the following parameters is discussed listed below:-

a. **Model:** used to determine what type of flow is undergoing. In this case, laminar air flow is preferred which is almost similar to air moving with the same direction towards the outlet and wall pipe without the crossover.
b. **Material:** used to select the materials that are either air, fluid or solid. In this case, air is obviously selected.
c. **Boundary condition:** this boundary condition needs a pair of inlet and outlet zone. The boundary conditions includes the following points:

- Identifying the location of the boundaries where we are going to measure the effect of the pressure and velocity (inlets, wall-pipe, outlet)
- Supplying information at the boundaries (pressure, velocity and roughness surface of materials)

The data required at a boundary depends upon the boundary condition type and the physical models employed. Following section gives details about the setup for inlet and outlet in the pair.

1. **Conditions at Inlet**

It is the side where the patient exhales towards the PBT to measure pressure difference and the air velocity which is similar to while the patient is exhaling.

So that at the initial condition the minimum initial air velocity of 2.2m/s and, the minimum pressure created at the mouth during exhalation is 0.2kilopascal were taken. Because the purpose of this thesis is to detect the low signal that difficult to get enough differential pressure in the mouth i.e 0.2kpa. ,

2. **Conditions at Outlet**

On the outlet part, only two parameters were measured which deal with this thesis. These are,

a. **Velocity:** We can specify total mass flow rate through supply section or extraction section.

The velocity at the large diameter (D1) is very low and at the small diameter (D2) high as shown below in the figure (22).

D2 (cm)	**Velocity(m/s) at D1**	**Velocity(m/s) at D2**
0.9	2.2	21.4
1.2	2.2	23.13
1.5	2.2	19.7
1.8	2.2	19.2
2.1	2.2	18.7

Table 2:-Velocity output at different diameter

So the maximum velocity is obtained at the smaller diameter of 1.2cm that is 23.13m/s with simulation as shown in the figure (22) below.

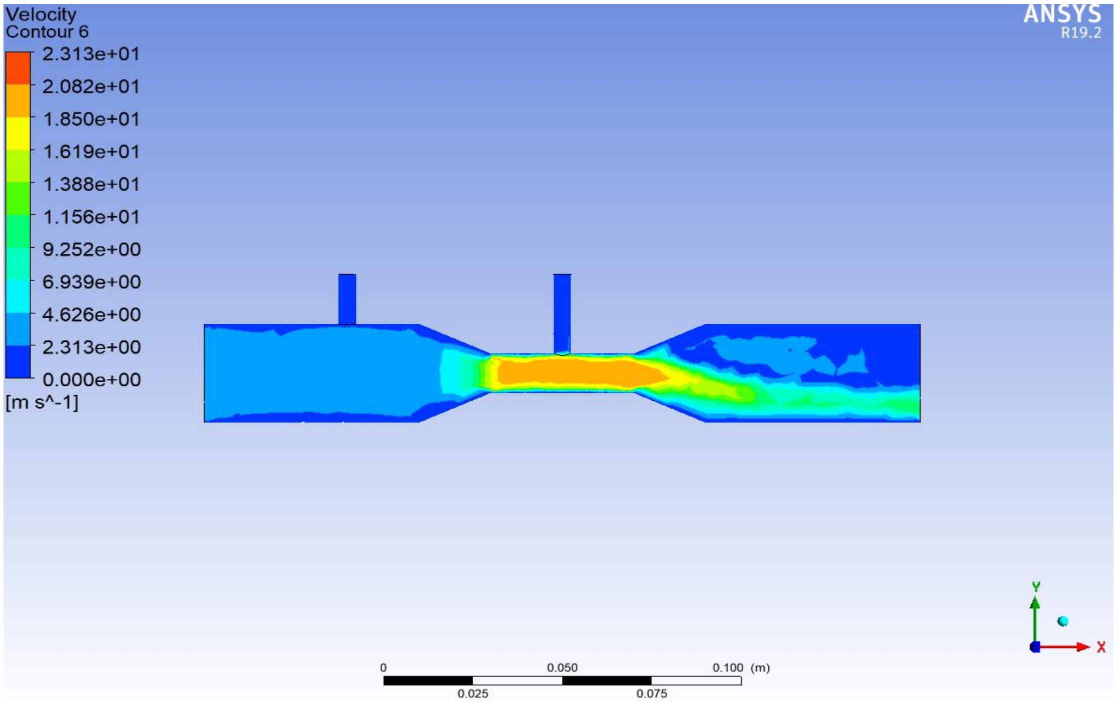


Figure 22:-Velocity Simulation across the PBT

5.2. Signal Conditioning

Signal conditioning includes generating the original raw signal acquired from the DP sensors and filter it for further analysis until the desired output is obtain.

5.2.1 Generating Original Signal Using LabVIEW

Since the differential pressure sensor didn't get from the market, LabVIEW software is used to generate the raw signal which is equivalent to DP sensor output [33]. This signal has the following parameters as shown in the figure (21).

a. Amplitude

Amplitude is the maximum value of voltage. It indicates the maximum potential that the patient can induced while breathing forcefully after inhalation. Therefore, since the thesis is dealing with maximum differential pressure output the voltage is equal to the maximum signal amplitude induced during signal generating.

b. Frequency

The signal's frequency determines how rapidly the differential pressure sensor can convert changes in pressure into electrical impulses. There are two ways to describe this frequency response, which is based on how quickly the DP sensor responds to pressure changes. The terms reaction time and flat response are used to describe these concepts.

Response frequency is sensor time constant which is required the sensor to changes from 0 to 63.2% of the full scale under the pressure changes. This time constant frequency response is used for slow signal sensing. In this thesis to generate the signal constant frequency response is used which provide the undamped frequency within constant time.

c. Noise

Since the signal we acquired from the DP sensor is full of noise. In this paper white noise signal is available in the signal because white noise is a random signal having equal intensity at different frequencies. The below figure (23) is signal generating using the LabVIEW software.

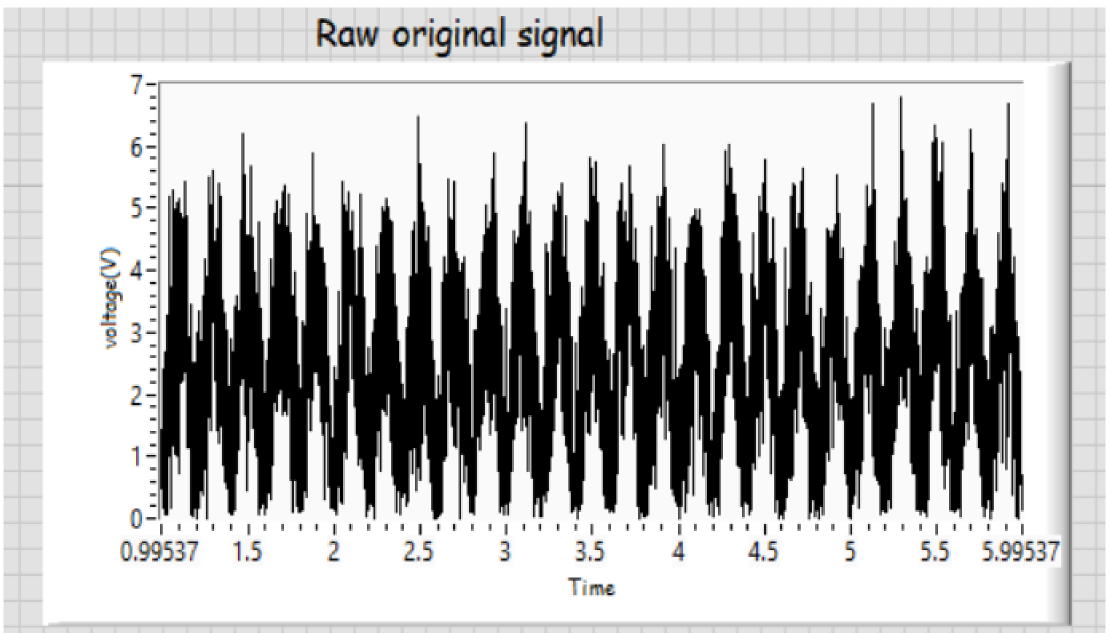


Figure 23: Original Raw Signal with Noise

5.2.1. Filtering

Because the information we receive from the pressure sensor is intermingled with noise, filtering the signal is required after acquisition to acquire the required signal without losing critical parameters. As a result, one of the key phases of signal processing utilizing LabVIEW software for the next analysis is filtering.

LabVIEW offers a wide range of filters, such as Butterworth filters, Chebyshev filters, and elliptic filters, and so on. In this thesis, to filter without losing the original signal we received from the differential pressure sensor, a low pass filter at cutoff frequency of 10.1 Hz is applied. The cutoff frequency is a frequency at which the output voltage equals to above 70.7% of the input voltage. This cutoff frequency helps us to get the best filtered signal without losing the necessarily signal. The LABVIEW filter module is used as shown in the figure below (24).

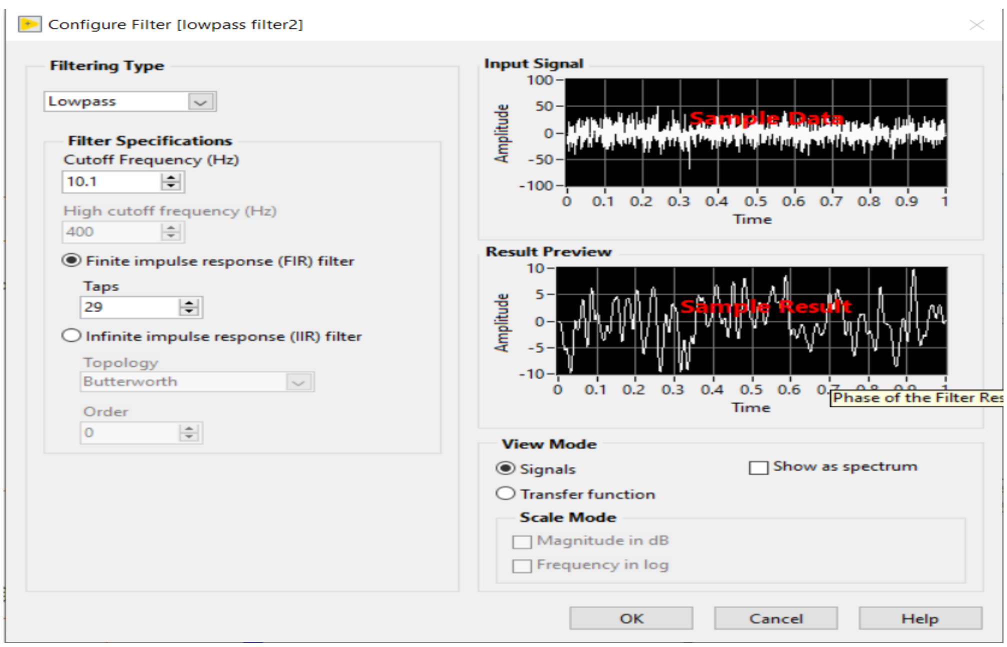


Figure 24: Signal Filter Module

Then the filtered signal is used to convert the maximum analog voltage into equivalent pressure units. Converting a voltage output into the pressure unit, kilopascal (Pa) is needed, to customize the differential pressure by considering the effect of altitude correction factor and for further analysis. Thus scaling method is a simple to scale any sensor.

Using this scaling formula below:

Y=MX+B (9)

Y is the output

M is the slope or the scale factor

X is the input (millivolts, volts,) and

B is the offset

Therefore, from the datasheet of differential pressure sensors, the range of output is from 0 to 10 kilopascal which is indicated in volts from 0.2 to 4.7 V. Using the equation 9, the M variable is calculated from the following table which we get from DP sensor datasheet as shown from the figure shown below. .

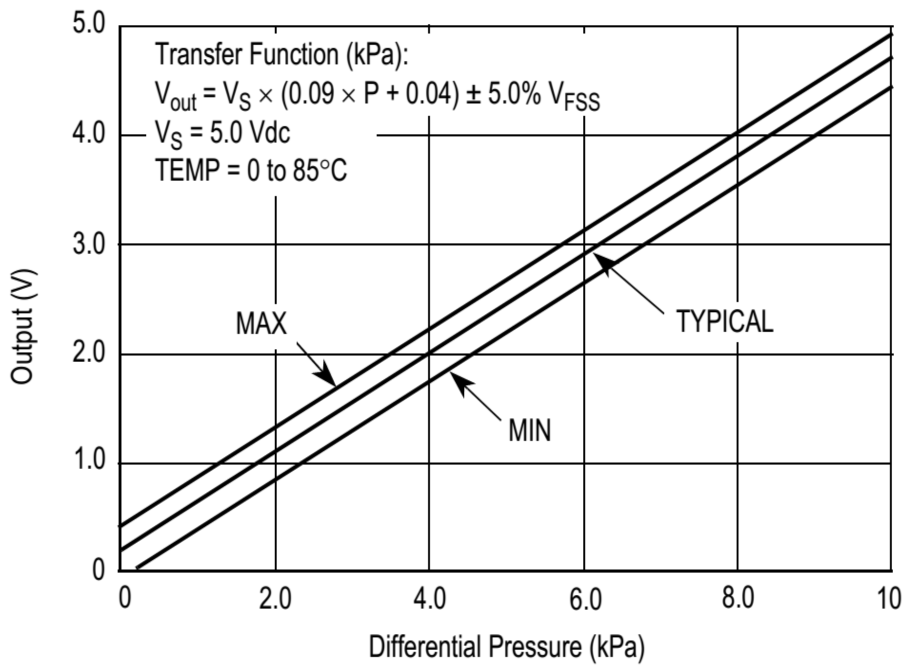


Figure 25: Differential pressure vs voltage output

$$M = \frac{10-0}{4.7-0.2} = 2.2$$

Since the reading of the sensor started from 0.2, it is offset value. So, B is 0.2.

Y=2.2X+0.2

$$Y = 2.2X + 0.2, \tag{10}$$

Where Y is the output of pressure in kilo pascal and X is the electrical equivalent in millivolts

The next step is to develop and optimize of post-processing methods to convert these differential pressure output into meaningful spirometry parameters such as PEF, FEV1 and FVC.

The lung is a respiratory system organ that interfaces with the existing environment. Exposure to high altitude plays a crucial role in breathing since the distribution of environmental air pressure decreases as altitude increases. This reduction of environmental pressure and oxygen concentration reduces the air density as altitude varies. Therefore before proceeding to the spirometer parameters the differential pressure output is calibrated with altitude correction based on the following table and equation.

So, based on the DP sensor datasheet information, the calibration is done for altitude, since the device is operated at a different range of altitudes, which in turn is affected by pressure as shown in table 3.

Altitudes (meters)	**Atmospheric Pressure (kpa)**	**Correction factor (Pcal/Pamb)**
0	101.3	0.95
250	98.4	0.98
425	96.6	1.00
500	95.8	1.01
750	92.5	1.04
1000	84.2	1.15
2250	76.6	1.26
3000	69.7	1.38

Table 3: Altitude vs Atmospheric pressure correction of sensor [29]

Equation of the altitude using the excel

$$\mathbf{DPeff = DPsensor * Pcal/Pamb} \quad (11)$$

DPeff = effective differential pressure

DPsensor =differential pressure indicated by sensor

Pamb = actual atmospheric pressure

Pcal = absolute pressure at calibration

Therefore the final Differential pressure (ΔP) the product of the correction factor and pressure acquiring from the differential pressure sensor.

$$\Delta P = (2.2 * X + 0.2) \text{ correction factor} \quad (12)$$

Using the above table, according to the area the spirometer is being used, we can insert the altitude in meters and the program automatically executes, optimizes the altitude and compensates the pressure [34][35].

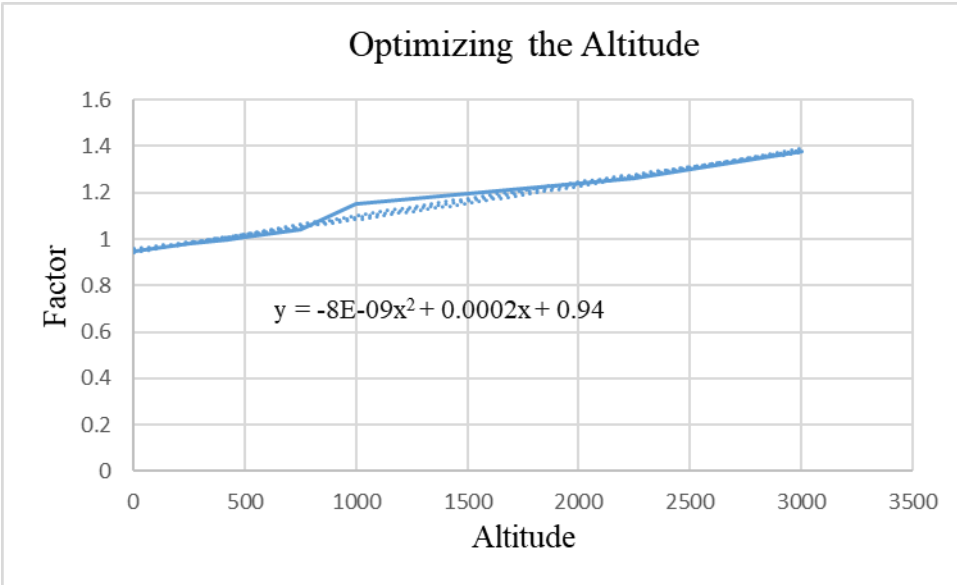


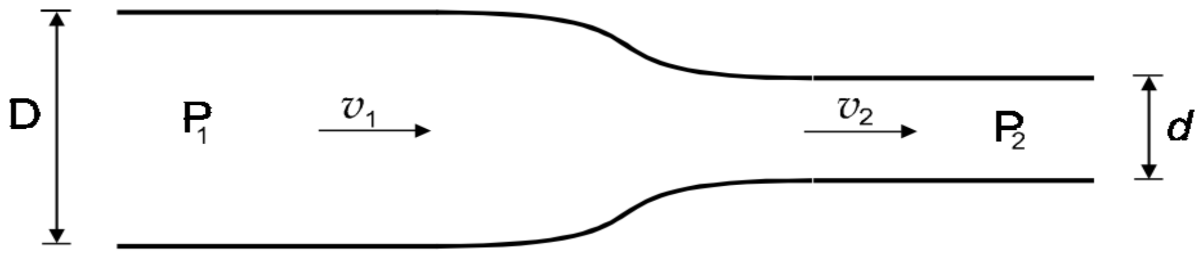
Figure 28: Altitude optimizing equation graph

5.4. Flow Measurement

As mentioned in the above, DP flow meters operates by restricting the cross-sectional area of a flowing fluid gases while the patient exhales through the designed BPT. To relate the pressure difference with the flow rate and volume, the governing equations are Bernoulli's theorem and the continuity equation of fluid flow. By combining these two equations, the relationship between the flow rate and differential pressure drop were analyzed.

5.4.1 Bernoulli's Equation and Derivation

The below aperture tube is similar to the patient breathing tube that designed in the above which is connected with the differential pressure sensor. Using the Bernoulli's equation the breathing air flow rate and the lung volume from differential pressure are calculated.



At the wider aperture, there is P1 with larger diameter D1. At this position, there is a force due to pressure one P1 is: -

$$F_1 = P_1 A_1 \quad (13)$$

However we have V is Volume, X is the length of the tube and A is the cross sectional area

$$\Delta V_1 = \Delta X_1 A_1 \quad (14)$$

With the same fashion, we have a force at the position two due to Pressure two P2.

$F_2 = P_2 A_2$ However, we have $\Delta V_2 = \Delta X_2 A_2$

But from basic mechanic's theorem, the work done due to these forces are: -

$$\Delta W_1 = F_1 \Delta X_1 \quad (15)$$

$\Delta W_1 = P_1 A_1 \Delta X_1$ From the volume calculation V=AX at point one

$\Delta W_1 = P_1 \Delta V_1$ So with the same formula

$\Delta W_2 = P_2 \Delta V_2$

For incompressible gases the volume at two points are equal.

$$\Delta V1 = \Delta V2 = \Delta V$$

So the net work done due to this forces is: -

$\Delta W = \Delta W_1 + \Delta W_2$

$$\Delta W = P_1 \Delta V - P_2 \Delta V$$
$$\Delta W = (P_1 - P_2) \Delta V \quad (16)$$

Kinetic Energy

The network done due to the change of kinetic energy also happens with the parcel.

$$\Delta W = K_2 - K_1 \quad (17)$$

Where kinetic energy

However $k_i = \frac{1}{2}\Delta m v_i^2$ (18

but from the density equation $m = \rho\Delta V$

$$\Delta W = K2 - K1 \quad \text{Therefore} \quad \Delta W = \frac{1}{2}\Delta m v_2^2 - \frac{1}{2}\Delta m v_1^2$$

$$\Delta W = (P_1 - P_2)\Delta V = \frac{1}{2}\rho\Delta V v_2^2 - \frac{1}{2}\rho\Delta V v_1^2$$

$$(P_1 - P_2)\Delta V = (\frac{1}{2}\rho v_2^2 - \frac{1}{2}\rho v_1^2)\Delta V$$

$$P_1 - P_2 = \frac{1}{2}\rho v_2^2 - \frac{1}{2}\rho v_1^2 \tag{19}$$

Therefore, the above equation can be summarized as: -

$$P_1 + \frac{1}{2}\rho v^2 = constant \tag{20}$$

With same principles, the potential energy due to the height of Y can be calculated in the same fashion.

Potential energy

$$(P_1 - P_2)\Delta V = mgY \quad \text{But the mass is } m = \rho\Delta V \tag{21}$$

$$(P_1 - P_2)\Delta V = \rho\Delta V gY$$

So

$$P + \rho gY = constant \tag{22}$$

Therefore, by combining the above basic equation for calculating the pressure drop through an orifice with the Bernoulli equation, if the flow is at the elevation h, there is the potential energy that is combined to give this equation.

$$P + \rho gY + \frac{1}{2}\rho v^2 = constant \tag{23}$$

I can rewrite the above equation by using the subscripts as if subscript “1” refers to the high pressure to be measured and the subscript "2" to the low pressure at the point of pressure measurement which makes pressure difference at the two-point, then assuming that change in temperature is negligible and no change in elevation, i.e. flow in a horizontal of patient breathing pipe. Thus, Y becomes zero, and the Bernoulli equation simplifies to,

$$P_1 + \frac{1}{2}\rho_1 V_1{}^2 = P_2 + \frac{1}{2}\rho_2 V_2{}^2 \tag{24}$$

$$P_1 - P_2 = \frac{1}{2}\rho_2 {V_2}^2 - \frac{1}{2}\rho_1 {V_1}^2$$

$$\Delta P = \frac{1}{2}\rho_2 {V_2}^2 - \frac{1}{2}\rho_1 {V_1}^2$$

From gas flow rate Q is the volumetric flow rate of the gases which is equal across the tube,

Q = $A_1 V_1 = A_2 V_2$ where A1 and A2 are the cross-sectional areas at points "1" and "2" where the pressures are measured. Thus we can write:

$$\Delta P = \frac{1}{2}\rho_1 \left(\frac{Q}{A_1}\right)^2 - \frac{1}{2}\rho_1 \left(\frac{Q}{A_2}\right)^2$$

Therefore, solving for Q the volumetric flow rate,

$$Q = A2\sqrt{\frac{2\Delta P/\rho}{1-(A2/A1)^2}} \qquad (25)$$

Hence, the nature of the gas is a compressible fluid, density at the two positions is different. However, the change of temperature while the patient while is exhaling is very small. This makes the breathing air density to be almost the same. So that the density for all the patients is the same, which is the density at Standard Temperature and Pressure, STP.

Standard temperature and pressure, STP, indicates the nominal conditions in the atmosphere at sea level. Standard temperature is defined as zero degrees Celsius (0 ^{0}C), which translates to 32 degrees 32 ^{0}F or 273.15 degrees kelvin. This is essentially the freezing point of pure water at sea level, in the air at standard pressure[36].

Standard pressure supports 760 millimeters in a mercurial barometer (760 mmHg). The density of air at STP is approximately 1.29 kilograms per meter cubed (1.29 kg/m^3).

5.4.2 Volume Calculation from Bernoulli's Equation

Flow is the time-derivative of volume, as shown below:

But the volume is the product of cross-sectional area (A) and length (L)

$$V= A*L \qquad (26)$$

$$Q = \frac{A * L}{t}$$ But the L/t is equal to the velocity

Therefore, Flow-rate: = A * Velocity Where: A = Area of cross-section of tube

$$V = \int Q dt \qquad (27)$$

Thus the mathematical procedure to get the spirometric parameters will conclude as shown in the figure (26).

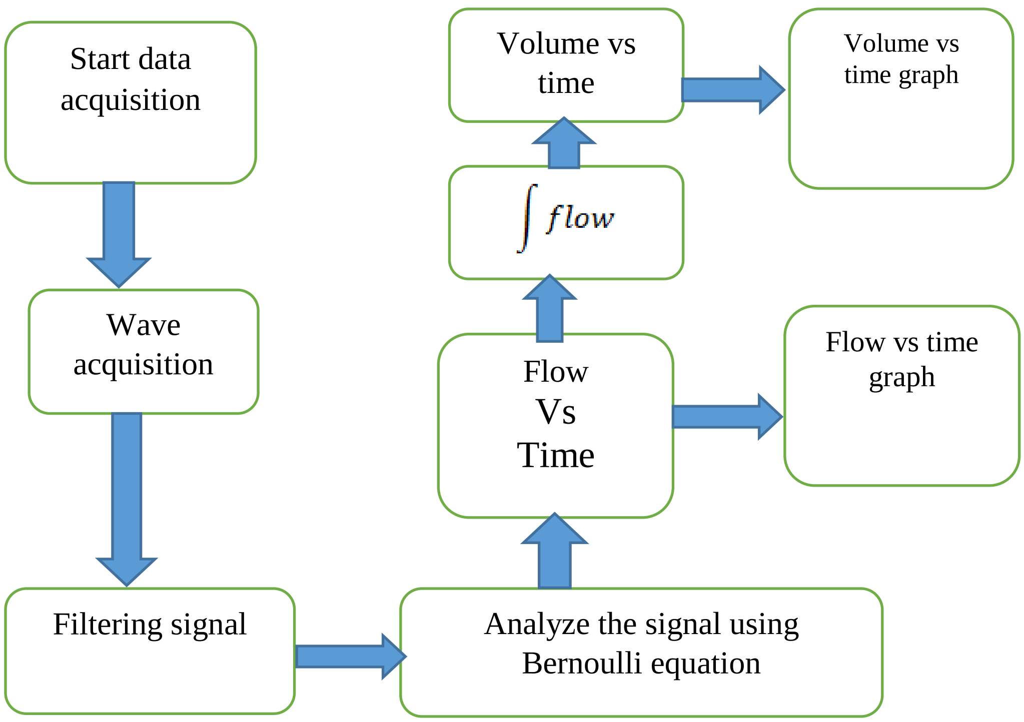


Figure 25: Volume and Flow calculation flow diagram

Finally, compare the numerical value of spirometer with the gender and age of the patient and classify the stage of the COPD as intermittent, mild persistent, moderate persistent and severe persistent.

5.5. Calibration

During calibration, a scaling procedure will be approved that converts millivolts to pressure units. To execute this calibration, a flow producing device with a known flow source, such as a sphygmomanometer with differential pressure sensor, must be linked to the differential pressure sensor and circuitry. The output of the pressure transducer, which is utilized to determine how the conversion from pressure difference to flow rate differs from the theoretical equations, is calculated using this method.

The first steps in calibrating the differential pressure-based spirometer are optimized and calibrated using leading spirometry standards. Finally, the technique is used to simulate the validation as well as the critical parameters of the graph using numerical data as a validation reference from a commercial spirometer.

If this effect is found to be significant, the device will need to be adjusted to get it closer to optimal behavior, i.e. scaling. The tube position of pressure detecting locations, as well as the pressure transducer sensor port and PBT design, are all variables that can be adjusted.

5.6. Age and Height Comparison

The spirometer measurement parameters are compared with the standard equation that the patient can give expected value based on age, height, and sex. The below equation is derived from different tests and worldwide accepted equation to analyze the spirometry parameters, conducted at the Japanese institute [37].

Males

- FVC(L)=5.76*height(m)-0.026*age(yr.)-4.34
- FEV1(L)=4.30*height(m)-0.029*age(yr.)-2.49

Female

- FVC(L)=4.43*height(m)-0.026*age(yr.)-2.89
- FEV1(L)=3.95*height(m)-0.025*age(yr.)-2.60

The conditioned signal is acquired and analyzed by our LabVIEW Virtual Instrumentation VI. Once a measurement is complete, the analysis portion of the VI integrates flow for the entire data set to obtain volume vs. time and flow vs. volume. It also calculates the parameters PEF, FVC, FEV1 and FEV1/ FVC. The VI front panel displays these parameters as well as graphs of the volume data.

CHAPTER SIX

RESULT AND DISCUSSION

This thesis is carried out with the design of PBT and breathing air simulation using Ansys and LabVIEW software respectively. This can be done by generating the original signal and designing the patient breathing tube which is used to create differential pressure and increase the sensitivity of PBT detection and simulate the effect as well as customize pressure by altitude correction.

6.1. Differential Pressure Simulation

The effect of differential pressure measurement at the restricted (D2) and non-restricted (D1) was measured after developing the venturi tube (PBT) shown in figure (20). As indicated in table 4, the differential pressure of breathing air is simulated using an inlet air velocity of 2.2m/s and an intake pressure of 0.2kpa at various tiny diameters.

D2(cm)	P1(Pa) at D1	P2 (Pa) at D2
0.9	197.8	-10.0
1.2	196.2	-9.0
1.5	189.6	-54
1.8	162.9	-147.8
2.1	121.3	-182.5

Table 4:-Differential pressure output at different diameter

The differential pressure is formed based on the boundary conditions when the patient breathes forth into the PBT. The highest pressure differential is achieved at the smaller diameter of 1.2cm, as seen in the table above. Because of the head loss of inlet pressure from high to low, the inlet pressure drops. The negative pressure indicates that the air flow is obstructed by the PBT restriction diameter rather than flowing outward,that making it more difficult for the patient to breathe.

To get the effective pressure difference across the PBT the pressure at the smaller point to be positive or zero. So that the optimum smaller diameter of PBT is 1.2 cm which is the ratio of the larger diameter to smaller 0.4 as shown in the above figure (26) below.

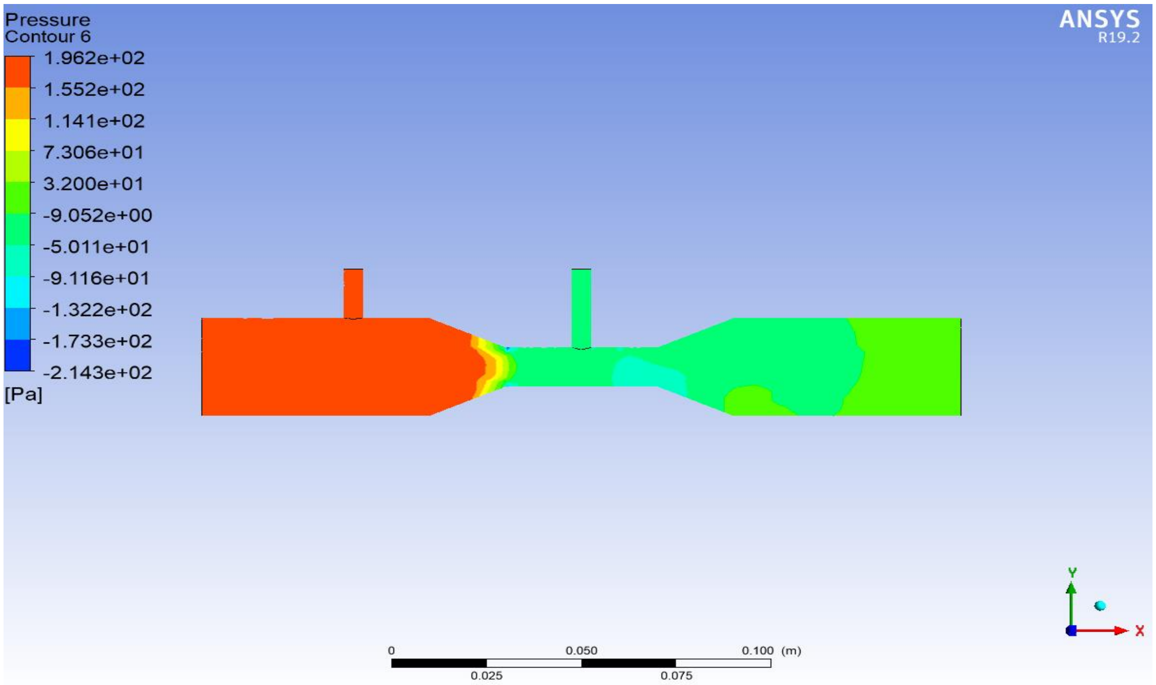


Figure 26: Differential pressure simulation

From the figure above, the pressure produce at the restricted (middle) is low and pressure at a larger diameter is high. This relationship is vice versa with velocity across PBT. The differential pressure is the difference between the restrictive and non-restrictive pressure is used to calculate the spirometric parameters using Bernoulli's and gas continuity equation.

6.2. Voltage vs Time

The data acquisition take place by generating the original signal by LabVIEW software which is equivalent to the signal acquired by the DP sensor. Since the acquired from the DP sensor is with the noise, and filtered out at cutoff frequency 10.1Hz. Figure 27 shows the voltage signal induced during exhaling through the patient breathing tube.

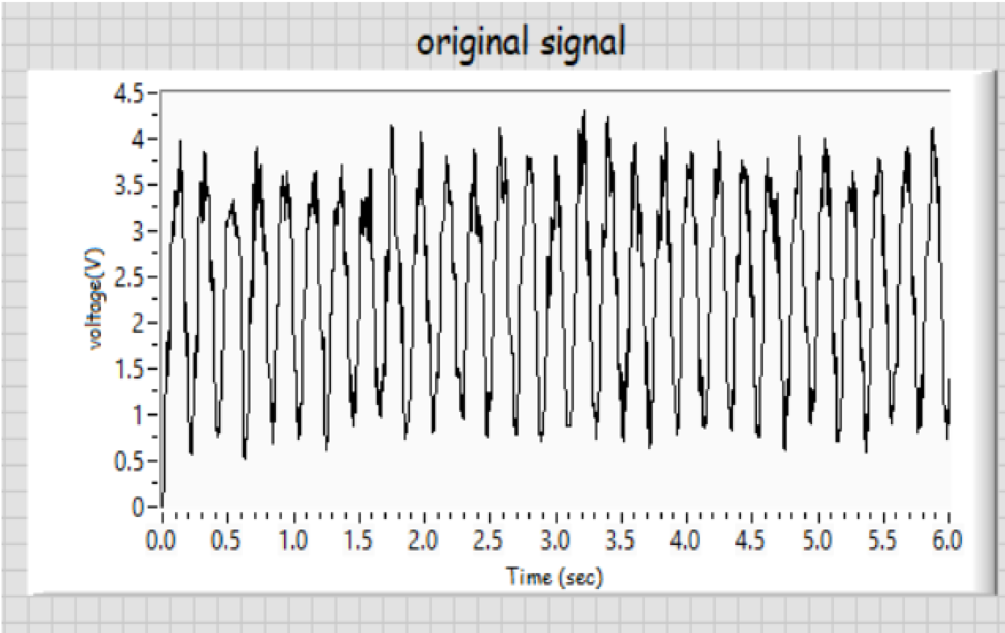


Figure 27: Voltage vs time graph

This induced voltage has a 2.5 Hz frequency response which indicates how quickly the differential pressure sensor can respond to changes in measured differential pressure.

The amplitude is the greatest voltage that the patient can create by forcibly inhaling. It varies depending on age, gender, height, and the PBT's sensitivity. The maximum amplitude is used to convert the voltage to equivalent pressure units for analysis and to adjust the pressure to the altitude. The recorded differential pressure is multiplied by a standard coefficient of altitude correction during customizing with an altitude correction factor.

6.3. Spirometer Parameters

From the above result, after the voltage is converted into equivalent differential pressure output, the spirometric parameters are displayed in the front panel of LabVIEW.

6.3.1. Front Panel

After connecting the differential pressure sensor with patient breathing tube, the signal data can be analyzed further and displayed. The end users can easily interact with this front panel by inserting the following parameters as shown in the front panel.

These are:-

- ✓ Height
- ✓ Age
- ✓ Sex

These parameters are used to compare with the standard and to what extent does the patient exhale in order to give the clinical decision for the patient, as shown in the figure (31). This figure indicates that the end user can fill and interpret the output of the result on the front panel using LabVIEW.

Figure 28: Front Panel using LabVIEW

Other factors influences the PFT reading of the numerical output and curve structure of FEV1 and FVC. These are gender, height, and age. Therefore the interpretation of lung function relies on the comparison to reference values derived from a healthy population based on the value inserted. There are two different graph were plotted in the frontline of the LabVIEW.

6.3.2. Air Flow vs Time

This graph is the flowrate of breathing air calculated from the differential pressure. It shows how fast the patient breathe outwards. At higher altitude, lower density of air reduces respiratory resistances and increases airflow rate.

The maximum point of the airflow vs time graph is called peak expiratory flow (PEF). The graph of flowrate is the same as the graph of flowrate vs volume graph which is used take the maximum PEF. The PEF will calculate from this graph i.e. 2.45m/s for the normal person. The output of the flowrate vs time graph is not difference from the commercial spirometer used in the COPD test as shown in the figure (32). The flowrate time also varies according to the age. For adults, the patient is expected to breath outward maximum for the six seconds while for children, it's approximately three to four second.

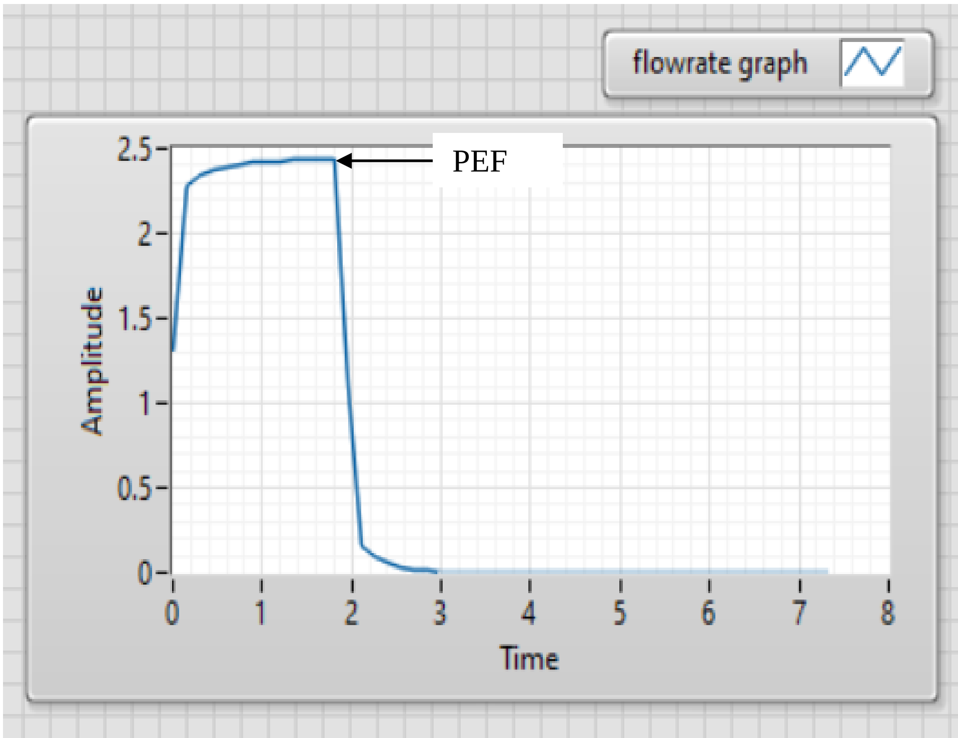


Figure 29: Flowrate Vs Time graph

6.3.3. Volume vs Time

With the integration of flowrate vs time result, we will get the volume in liters that is the total lung capacity the patient breathes after a full inhaling. The results of a spirometry testing maneuver can be displayed as a volume-time tracing as shown in Figure 33 below.

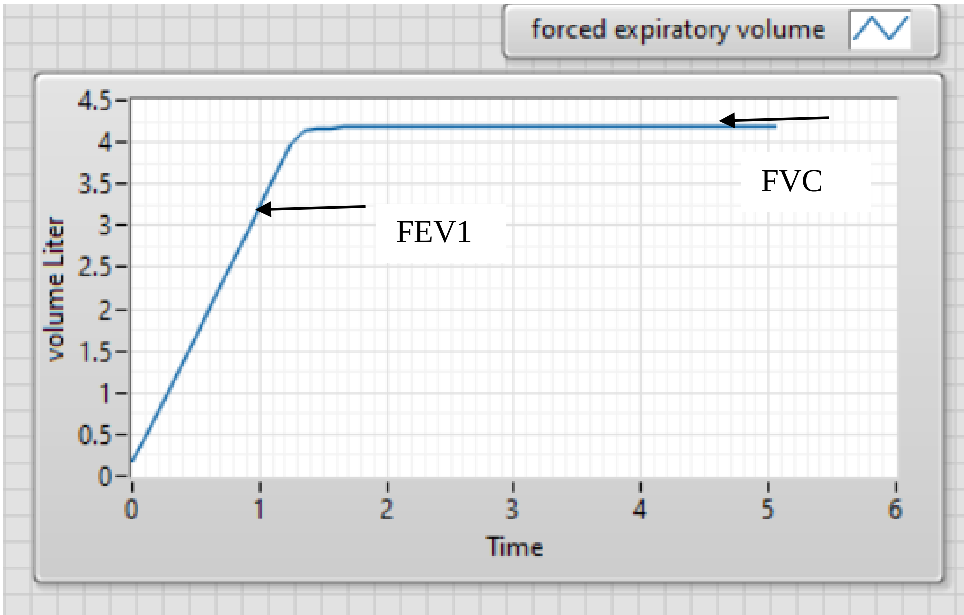


Figure 30: Volume Vs Time graph

This result is helpful to give a decision on diagnosis and identification of COPD at healthcare. In the spirometric measurement, this result is called vital capacity volume. From this graph, the forced expiratory volume at first second, FEV1 and FVC can be obtained. From volume vs. time, it calculates the parameters PEF, FVC, FEV1 and FEV1/ FVC as shown above front panel.

From this graph FEV1=3.33L and FVC= 4.2L

- **FEV1/FVC Ratio**

FEV1% = FEV1/FVC *100

FEV1% =3.33/4.2*100=79.3% which is moderate

Note that the ratio of FEV1/FVC from Volume vs Time curve is used to determine the COPD conditions.

The sensitivity of each of the above parameters affects the spirometer parameters measured here. This means that the characteristics of spirometers used to diagnose and screen COPD patients are affected by altitude, gender, and age. Finally, by increasing the sensitivity of the PBT and altitude correction while the patient is undergoing a pulmonary function test, customizing a pressure-based spirometer for improving diagnosis of chronic obstructive lung disease is the solution.

CHAPTER SEVEN

CONCLUSION AND RECOMMENDATION

7.1. Conclusion

In this thesis, customize a simple pressure-based spirometer were proposed to improve its accuracy for COPD device with respect to altitude in order to improve accuracy of COPD tests. The PBT were designed, simulate breathing air across the PBT using Ansys 19.2 software to find an optimum design of PBT which gives better sensitivity, and a prototype is created using 3D printer. LabVIEW software is used to generate and analyze the measured signal by the differential sensor, and customize the spirometer with respect to altitude. Moreover, Bernoulli's equation has been used to calculate air flowrate and forced vital capacity of the lung which are used to grade COPD test result. The flowrate vs time, and volume vs time graphs were obtained and displayed using LabVIEW. The result shows that the accuracy of COPD test can be improved by increasing the sensitivity of PBT and customizing a pressure-based spirometer using the altitude correction factor during designing and operating the pressure-based spirometer.

7.2. Recommendation and Future Work

Here are the recommendations to be considered while designing a differential pressure-based spirometer:

- Improving the sensitivity if the spirometer lays on design of the PBT and the corresponding locations on the PBT where we measure the differential pressure
- Altitude correction helps helps improve accuracy of differential pressured-based spirometer

Customizing the overall mechanical design and getting ethical clearance for conducting clinical tests and comparing the results with commercially available spirometery will be among the major works to be done in the future.

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