



ADDIS ABABA SCIENCE AND TECHNOLOGY UNIVERSITY

**FUZZY BASED CONTROL OF VOLUME AND
PRESSURE ASSIST MEDICAL VENTILATOR**

By

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to

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ENGINEERING**

October, 2021


Declaration

I hereby declare that this thesis entitled “**Fuzzy Based Control of Volume and Pressure Assist Medical Ventilator**” was prepared by me, with the guidance of my advisor. The work contained herein is my own except where explicitly stated otherwise in the text, and that this work has not been submitted, in whole or in part, for any other degree or professional qualification. Parts of this work have been published in [state previous publication].

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Abstract

Positive pressure ventilation (PPV) mode is the direct application of air to the patient's lungs and is the most commonly used ventilation mode for passive patients. Mechanical ventilators are life-saving medical equipment if a person is unable to breathe on their own. Due to the scarcity and unavailability of the machines, patients are not getting assistance. The proposed thesis aims to design a low-cost volume and pressure-controlled mechanical ventilator for pandemic emergencies and scarcity. A Fuzzy logic controller is used to control both the pressure and volume parameters. The designed system can be used for different patients that need different tidal volumes and pressure. The clinician can set the tidal volume and airway pressure depending on the lung requirements. A piston cylinder mechanism is used to deliver the required volume and pressure of gas to the pneumatic circuit. Armature controlled DC motor is used to drive the ventilator mechanism, and its model is used to determine the maximum displacement, velocity, and delivery capacity of the mechanism. The testing is done by interfacing the sensors and actuators and deploying the Simulink model on an Arduino mega controller. A medical linear test lung simulator is used for the test.

Keywords:— Mechanical ventilator, Positive pressure ventilation, Fuzzy controller, Pneumatic circuit, Airway pressure, LTL, Tidal volume.

To my brother: ZEBENE BESHAWURED GEBREMARIAM

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List of Abbreviation and Acronyms

ARDS	Acute Respiratory Distress Syndrome
CPAP	Continuous Positive Airway Pressure
DC	Direct Current
FL	Fuzzy Logic
FIO ₂	Fraction of Inspired Oxygen
FLC	Fuzzy Logic Controller
HFV	High Frequency Ventilation
HFJV	High Frequency Jet Ventilation
HFOV	High Frequency Oscillatory Ventilation
ICU	Intensive Care Unit
IPPV	Invasive Positive Pressure Ventilation
KVL	Kirchhoff Voltage Law
LTL	Linear Test Lung
LMICs	Low Middle Income Countries
MVM	Mandatory Ventilation Mode
NPV	Negative Pressure Ventilation
NIPPV	Non Invasive Positive Pressure Ventilation
PSO	Particle Swarm Optimization
PIP	Peak Inspiratory Pressure
PPV	Positive Pressure Ventilation
PEEP	Positive End Expiratory Pressure
PSV	Pressure Support Ventilation
PCV	Pressure Controlled Ventilation
PID	Proportional Integral Derivative
PSOL	Proportional Solenoid Valve
RPM	Revolution Per Minute
SLPM	Standard Liters Per Minute
SIMV	Synchronized Intermittent Mandatory Ventilation
VT	Tidal Volume
VALI	Ventilator Associated Lung Injury
VILI	Ventilator Induced Lung Injury
VSV	Volume Support Ventilation

Chapter1

INTRODUCTION

1.1 Lung Anatomy

The respiratory system has two zones, the conduction zone or physical structure and organs that bring air to the lung from the atmosphere, and the respiratory zone in which gas exchange takes place. The lungs are a pair of spongy or balloons, air-filled primary or-

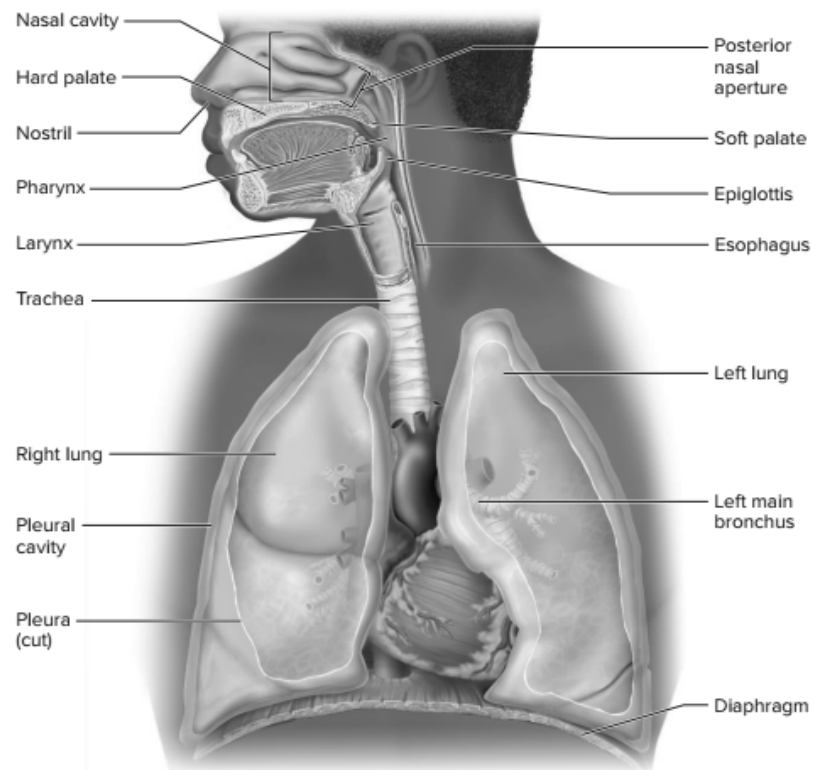


Figure 1.1: Structure of respiratory system [1].

gans. Two lungs the right lung and left lung as shown in Figure 1.1, are located around the backbone on both side of the chest. Their purpose in the breathing system is to trans-

port oxygen from the surrounding environment and supply it into the bloodstream, and to release carbon dioxide from the bloodstream into the atmosphere, in the process of gas transport. Respiration is driven by different muscular systems to support and foster breathing. The diaphragm is the primary muscle of respiration in humans, and it is responsible for breathing. Bronchioles are the end terminals in the conducting zone. Due to the elastic properties of the respiratory system, gas exchange between atmosphere and lung is driven by differential pressure, volume, pressure and velocity of inspiration, and expiration can be affected by the respiratory impedance. The process of breathing in healthy person is

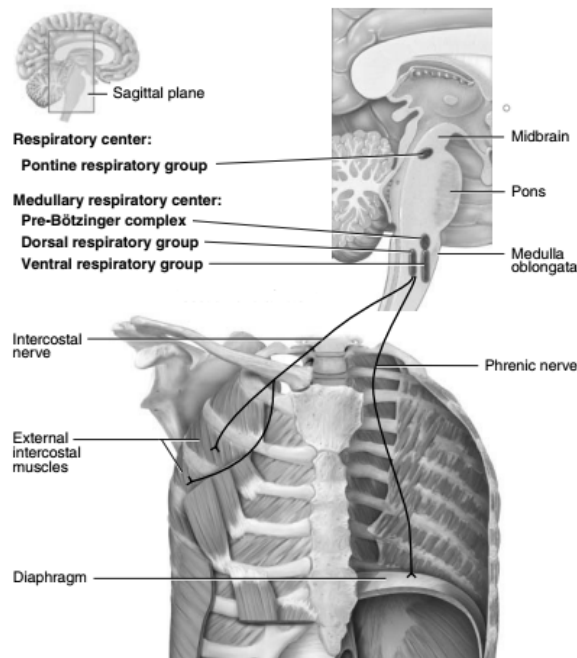


Figure 1.2: Control of respiration [2].

controlled by detectors in the brain called central chemo-receptors as shown in Figure 1.2. Central chemo-receptor, are sensitive to the concentration of CO_2 and PH levels in blood vessels, send a signal to the brain to remake breathing. The message sent from the brain alters the rate of contraction and relaxation of the intercostal muscles. During inspiration intercostal muscles contract, the thoracic cavity gets larger, the diaphragm contracts, the lung expands and gets filled with air [6]. when the reverse happens expiration will take place.

1.2 Motivation

A variety of respiratory illnesses, such as pneumonia and lung cancer, can damage the tissue of the lungs. When the tissue of the lung is affected by such diseases, the muscle of respiration cannot perform the activity of gas exchange. For a person having a problem with respiration, a breathing machine is required. A medical ventilator/breathing machine/mechanical ventilator is a machine that helps the lungs work. It is used for a person with a variety of conditions having breathing problems. A medical ventilator can be life-saving when a person cannot breathe properly or when they cannot breathe on their own at all. Mechanical ventilators are commonly employed in critical care units to provide appropriate blood oxygenation and to keep patients' metabolisms stable. Humans require oxygen because it is a source of energy. The transfer of oxygen and carbon dioxide is known as respiration. Respiration has two stages: inspiration or inhalation and expiration or exhalation. Inspiration is the act of taking air into the lungs, whereas expiration is the process of out-breathing. Mechanical ventilators are used to artificially carry out the breathing process in patients. A mechanical ventilator is a mechanism that uses mechanical means to transfer breathing air into and out of the lungs. Respiratory assistance can be provided in PPV, NPV, and high-frequency ventilation (HFV).

1. Positive pressure ventilation (PPV)

Respiratory support is provided by applying effective pressure to the patient's airway opening. The ventilator system is linked to the patient's breathing system. Positive pressure is delivered to the patient's airway on a regular basis. Gas is forced into the lungs when the airway pressure is slightly higher than the alveolar pressure, a process known as inspiration. The gas is evacuated from the lungs when alveolar pressure is greater than airway pressure, a process known as expiration. In PPV gas can be delivered in ether forms:

- (a) Non-invasive positive pressure ventilation (NIPPV): A customized face mask with a tight seal is used to supply air (air travels through anatomical airways)
- (b) Invasive positive pressure ventilation (IPPV): includes the use of an endotracheal tube to apply positive pressure to the lungs (or any other device that delivers gas by passing parts of the anatomical ducts).

2. Negative pressure ventilation (NPV)

In NPV respiratory support is given by applying vacuum pressure to the chest wall [7]. Ventilated patient's nose and mouth remain fully open to the atmosphere. Whenever the alveolar pressure varies compared to atmospheric air pressure gas can flow

in and out. A pressure is exerted to the surface of the chest wall during inspiration, this force will temporarily decrease alveolar pressure, fresh air will flow into the respiratory organ. Respiratory elastic recoil force creates a temporary positive alveolar pressure that can force the gas out of the lung. Because the tidal volume and respiratory rate are comparable to physiological values, both IPPV and INPV are considered as conventional mechanical ventilation.

3. High-frequency ventilation (HFV) The respiratory rate in HFV is significantly greater (150 b/min or more) than in the other kinds. The tidal volume is significantly lower than the physiological range, and in some cases, it is even smaller than dead space. Gas transport and exchange processes in HFV differ significantly from those in PPV and NPV, and they are rarely discussed. HFV has developed into five types due to variations in implemented technology, breathing rate, pressure ranges, and base pressure [8]. The most often utilized versions are HFJV and HFOV.

A Mechanical ventilator is used in different modes of ventilation. The mode of ventilation relates to how it provides inspiratory assistance. Inspiration is delivered by a unique mix of breathing patterns and control factors. The Common mode of ventilatory assistance includes [9][10]:

- Pressure support ventilation: instead of a set V_t as in volume-targeted modes, volume in pressure support ventilation is variable. The mandatory ventilation mode (MVM) delivers pressure to assist the patient in breathing in PSV mode, while the patient regulates the respiration rate. This mode is not recommended for patients who are unable to take their own breaths.
- Volume support ventilation: V_t is the targeted parameter in this mode of ventilation, and a fixed V_t is supplied with each breath. The most often utilized modes are volume-focused modes.
- Volume and pressure targeted or dual (hybrid) ventilation: in an attempt to prevent both the high peak airway pressures of volume ventilation and the variable tidal volumes that can occur with pressure ventilation, newer ventilators provide hybrid modes that combine elements of volume-targeted and pressure-targeted ventilation.
- Pressure controlled ventilation: PCV is a time-cycled breathing method in which the inspiratory pressure, PEEP, the period of the inspiratory stage of the breathing period, and the respiratory rate are all controlled by the medical practitioner. The resultant patient tidal volume fluctuates based on patient effort, lung elasticity and resistance, and inspiratory pressure since the volume and flow are not monitored.

- Other Modes: Synchronized Intermittent Mandatory Ventilation (SIMV) Plus Pressure Support. There are two modalities in use: SIMV and pressure assist mode. Both the volume of obligatory breaths is controlled and the pressure of spontaneous breaths is controlled. The patient is required to take a certain number of volume-targeted obligatory breaths at a specific V_t .

1.3 Problem Statement

There are different conditions that need a medical ventilator for a person whose breathing system is affected because of different issues. A medical ventilator is an indispensable machine after the occurrence of covid-19. While recovering from an illness, babies, children, and adults may require a medical ventilator for a brief period of time. It is also required during surgery, recovering from surgery, when breathing on their own is very difficult because of having lung diseases such as pneumonia, covid-19, lung cancer, lung infection, or another condition that makes breathing difficult. Some condition that may also need the use of medial ventilator includes brain injury, coma or loss of consciousness and drug overdose. For a person having such a problem of breathing a controlled artificial breathing machine (medical ventilator) should be installed to help the patient. A patient is usually vented for a duration lasting from a few hours to several days, however, this might be considerably longer. Mechanical ventilation is an extremely costly treatment. It determines the length of stay in the critical care unit to a large extent (ICU). According to large-scale research, the average added cost of mechanical breathing for individuals in United States was \$1522 per day [11] [8]. In principle, if every ventilated patient's therapy could be cut by only one hour, huge savings may be realized. Medical equipment to combat the COVID-19 pandemic is now in short supply worldwide, particularly in poor and medium-income countries. This is especially true for ventilators, which are required for COVID-19 therapy both in the acute phase, when intrusive completely regulated breathing is required, and in the subacute phase ventilation is needed for a long duration. Due to cost and lack of availability of the machine patients could not get treatment [12].

1.4 Objective

General Objective

The general objective of this research is to design a fuzzy based control of volume and pressure assist medical ventilator.

Specific Objective

The specific objective of the research has been pointed as follows:

- Develop the system mathematical model.
- Collect data.
- Design a fuzzy based control algorithm.
- Simulate the system and analyze the result.
- Develop the plant prototype and evaluate simulation and prototype results.

1.5 Scope of the Study

The scope of the study will cover developing a system mathematical model, collecting data for different lung capacities, designing a controller, controlling the valve, system simulation, and develop a prototype.

1.6 Significance of the Study

The proposed research has a significant role for a person who has several respiratory diseases such as pneumonia, covid-19, lung cancer, lung infection. The proposed medical machine may increase the availability, reduce the total cost of the machine and the cost required for importing the medical machine. It provides an easily understandable system and controls action for the clinician, mass productions of machines for pandemic emergencies. The fuzzy controller is expected to provide the required volume and pressure of air by controlling the angular speed, airway and expiratory valves. This will provide safe breathing for the patient because of its volume and pressure controlled mode [13].

1.7 Thesis Outline

This thesis is structured as follows::

Chapter 1: has established the basic introduction about the anatomy of lung (how the lungs work), motivation on different mode of mechanical ventilation, problem statement, objective, scope of the study and its significance.

Chapter 2: covers related studies in previous work on lung mechanics, ventilator design

and control strategies.

Chapter 3: verifies over all methodology followed in this thesis. Overview of mechanics of gas exchange, system mathematical modeling, prototype mechanical structure, pneumatic circuit of the system and mechanism response is included in this portion.

Chapter 4: covers fuzzy logic controller design, input fuzzification, fuzzy rule, output defuzzification and Simulink model of the system.

Chapter 5: shows the implementation of the designed system, detail prototype accessories, prototype Simulink model, system circuit design and its operation principle.

Chapter 6: includes open loop response and closed loop response of system simulation model, and prototype real time simulation result with discussion.

Chapter 7: conclude and summarize the work, and give recommendation for future work.

Chapter2

LITERATURE REVIEW

The control of mechanical ventilators has been the subject of numerous articles in the literature. Nowadays, world researchers are pointing towards the design and control of medical ventilators to provide safe and controlled breathing for lung patients. Different researches have been done to develop medical ventilators for lung patients. This device has different mechanical structures and control strategies to perform their breathing task. Since the main focus of this research is on volume and pressure control strategies for the designed actuation mechanism. Many literature are reviewed to understand the current control method. Researchers initially devised an open-loop control approach for ventilators. However, due to the fact that this control increases clinician burden, researchers concentrated on closed-loop control to minimize clinician workload. Both patients and physicians benefit greatly from this method. Many attempts have been made to regulate respiratory variables in order to aid clinicians.

Zar et al. [14] explore that in poor and middle-income nations, pneumonia remains the top and challenging cause of infant mortality and the most prevalent reason for adult hospitalization despite advancements in prevention and control techniques. Adequate medications and supportive care, including oxygen, remain the cornerstones of successful therapy for infant pneumonia. The use of oxygen devices can lower mortality in individuals with hypoxic pneumonia by around 20%.

Inglis et al. [12] points that in low-middle-income countries (LMICs), the treatment of patients with acute respiratory failure should seek to avoid intubation if practicable, optimizing the quality of mechanical ventilation, and completing weaning as quickly as possible. The implementation of advanced and situational respiratory assistance technologies should be a priority in the future.

Fan et al. [15] addressed the use of respiratory support in adult patients with ARDS in this publication, which includes scientific proof clinical guidelines. Medical ventilation with reduced tidal volumes (4–8 ml/kg estimated body weight) and reduced inspiratory pressures (plateau pressure, 30 cmH₂O) is strongly recommended for patients with ARDS. The advice is strong against regular use of HFOV in patients with serious ARDS and is dependent on greater PEEP and recruitment procedures. In certain patients with acute respiratory distress syndrome, greater Positive end-expiratory pressure may enhance alveolar activation, decrease lung stress and strain, avoid atelectrauma, and reduce alveolar damage and excess destination.

Bugedo et al. [16] describes that the increases in chest wall elastance are the major cause of changes in transpulmonary driving pressure and trachea (airway) working pressure. Driving pressure is airway pressure minus PEEP. They recommend modifying medical ventilatory assistance using standard protective settings such as V_t 6 to 8 mL/ kg body weight index and moderate positive end expiratory pressure levels (PEEP ≈ 10 cmH₂O), as well as driving pressure, which should preferably be less than 15 cmH₂O, and plateau pressures of < 30 cmH₂O.

Tonetti et al. [17] identified a number of variables that are potential causes of lung damage while breathing in breathing machine (VILI). The first is pressure followed by tidal volume and the lung units cyclic expanding and contracting. However both theoretical approaches and practical studies indicate a substantial role in the development of VILI, less emphasis has been given to breathing rate and flow rate. Mechanical power may initially cause injury to the lung parenchyma. Mechanical power component includes breathing rate, flow, and PEEP. Higher tidal volume and relatively low PEEP statistics indicate that driving pressure, rather than tidal volume, is the greatest predictor of VILI. Higher pressure causes observable fractures of the lung parenchyma, which is the major cause of VILI. Even at higher pressure, low tidal volume does not result in a rupture in the lung, but higher tidal volume will. Clinical experimental studies show that a higher tidal volume of 12 ml/kg BW is associated with 10% higher mortality compared to a lower tidal volume of 6 ml/kg BW.

Bellani et al. [18] verify how the compliance of respiratory system and driving pressure are independently linked with mortality in pressure assist ventilation. In patients with ARDS high driving pressure is linked to a higher death rate. Following respiratory support in patients with acute respiratory distress syndrome, driving pressure, plateau pressure, and

respiratory compliance may all be monitored, and both higher driving pressure and lesser compliance are linked to a higher death rate. Peak pressure is always less than plateau (Ppl) pressure in controlled ventilation mode.

Müller-Redetzky et al. [19] investigate the effect of high I/E ratio. Lung-protective ventilation decrease mortality in patients with ARDS. Tidal volume must be restricted, high plateau pressures are eliminated, and PEEP is used to reduce ventilator-induced lung damage (VILI). Increasing inspiratory duration and therefore the inspiratory/expiratory ratio (I:E ratio) can help with oxygenation, but it can also be hazardous since relative stress and strain rise with time. As a result, they postulated that VILI would be worse while increasing inspiratory duration and the I: E ratio.

De Prost and Dreyfuss [20] investigated years before the publication of the ARDS Network study, which showed a 22 percent higher survival rate in patients receiving lower 6 mL/kg than those receiving larger 12 mL/kg tidal volumes, clinicians began using lower tidal volumes and improved the prognosis of ARDS patients based on experimental studies supporting the deleterious effects of ventilation with high peak inspiratory pressures. The study also suggested that ventilator could not overload the lungs through excessive pressure and volume application, and excessive removal of air from the end respiratory system. The discouragement of ventilator induced lung injury (VILI) has to be the major goal of future research related to medical ventilators.

Chen et al. [13] suggested good pressure stability and pressure down grading is an important issue during exhalation for better efficiency and comfort of breathing. The study's goal was to compare the pressure stability of the prototype CPAP ventilator to that of commercially available CPAP machines. They Developed continuous positive airway pressure (CPAP) ventilator using PID controller to control pressure. The electrified blower draws air from the atmosphere through the air filter before being compressed. Then compressed air is directed through flexible tube to patients lung. The speed of brush-less DC motor is regulated by microprocessor by varying the motor driver input voltage which in turn control airway pressure. The paper recommends more accurate control strategy is required to control the pressure and volume of exhaled air. The study followed only pressure control ventilation.

Yuan et al. [21] investigated the effect of respiratory impedance on velocity and volume of respiration. Managing the output phase, triggering time, flow-velocity, pressure,

levels of oxygen, reaction to respiration, and air humidity and temperature are all critical for regulating its air flow output in a medical ventilator. It also explore assist control ventilation, SIMV, PSV and VSV are medical ventilator control mode accepted by patients. The main rationale for evaluating the therapeutic impact of a mechanical ventilator is whether the ventilator's airflow can satisfy the breathing physiology requirements, and oxygen exchange between the organism and the environment is improved. The goal of this study was to enhance and maximize the ventilator's performance by evaluating the whole system design and development process for a modern ventilator. For design, a central control unit (MCU) is used to process the information from the pressure sensor and flow sensor, and to operate the motor and valve. The outgoing airflow is in the process of being regulated. In this investigation, the continuous positive airway pressure (CPAP) control mode was accomplished and to simulate the patient, a continuously breathing simulator, the Series 1101 was used.

Yimchunger et al. [5] improve the dynamic behaviour of a volume controlled artificial ventilation system. Particle swarm optimization (PSO) algorithm based PID controller was designed for the purpose. POS is considered to tune optimization gain the value of the designed controller parameters. First order model is used as a breathing mechanics. Piston drive mechanism acts as an inspiration driving mechanism and PEEP valve act as out breathing actuator. This paper concentrated on volume dynamics control of artificial medical ventilator and does not consider pressure control strategies.

Zhu and Moller [22] developed nonlinear adaptive fuzzy-neural network controller. A fuzzy controller is used to regulate flow pressure and a neural network is used to identify the nonlinear respiratory system. Ventilation flow control is used to make the ventilated patient get the required alveolus pressure set by the clinician. This research also focuses on pressurized air control strategies, however, it does not specified the actuator controlling mechanism and studied only volume controlling technique.

SAYIN and ERDAL [7] developed a standard biomedical device compatible mechanical ventilator prototype for use in traveling critical care babies between medical facilities. If necessary, the ventilator prototype may function as a CPAP (Continuous Positive Airway Pressure) device for infant survival. The prototype was done while keeping in mind the safety measures and sensitivity parameters stated in the applicable newborn ventilator standards. Pressure or volume control may be possible with the ventilator. Proportional Flow Control Valve was used to control the PIP pressure or tidal volume of the breathing gas and

Piezoelectric Pressure Regulating Valve was used to determine PEEP value at the end of expiration. Medical gas compatible pressure and flow sensors were used for the prototype. The control strategy employed was a real-time PID control technique. For volume and pressure regulated ventilation, two different models have been proposed. In this research, a separate model was developed to control the volume or pressure separately. Dual controlling technique is not used in this study.

Muraveva et al. [23] explain the challenge of synthesizing mathematical models and control algorithms with insufficient knowledge about the control process, valid signals, and interferences impacting. Over time, it has become obvious that traditional methods for managing such systems using automated control theory are insufficient, requiring the development of new methodologies and approaches. Fuzzy sets and fuzzy logic are one of these techniques. This method was initially used to create expert systems and demonstrating its efficacy. Later, it launched an application for the creation of expert control systems and for the synthesis of controller and process control application. This article will look at how to use a fuzzy controller for a new mechanism and air pump combination for pressure and volume assist ventilators.

Chapter3

MATHEMATICAL MODELING

3.1 Overview of mechanics of gas exchange

The mechanics of gas exchange in breathing includes lung and chest wall elastic properties, and air flow resistance. The lung is an elastic balloon which deflates to let air in and out of it. Deflation force applied by lung is called elastic recoil pressure which increases with increase in lung volume and maintains inflation. This pressure is considered as positive and is always applied by the lung during expiration. Air flows into the lungs when atmospheric pressure is greater than pressure inside the lungs and when pressure in the surroundings is less than pressure inside the lung, breathing out will take place [2]. The basic elements (settings) in mechanical ventilators are:-

- Pressure(P)
- Pressure gradient (ΔP)
- Volume (V)
- Gas flow (\dot{V})
- Resistance (R)
- Compliance(C) and
- Time constant (RC)

Pressure

Is the force applied per unit area. Pressure has an inverse relation to volume and flow-rate. As the pressure increases, the volume will decrease in the closed system and when the flow rate increases, the pressure will drop. Pressure in breathing has two parts: resistive pressure and elastic pressure. Resistive pressure is encountered when air is pushed through the

airway and elastic pressure is the pressure which inflates the lung and chest wall. Peak pressure is the maximum pressure in the airway at the end of inspiration, related to the resistive pressure, depends on resistance in the airway and compliance of the lung. plateau pressure is end-inspiratory alveoli pressure and related to the elastic pressure. High peak pressure causes airway issues and high plateau pressure causes compliance issue [24]. Except when the volume capacity has been dropped to zero, the pressure inside an alveolus will always be higher than the pressure in the neighboring interstitial tissue. Lung elastic recoil pressure (P_L) or Trans-pulmonary pressure (P_{TP}) is the difference in pressure between alveolar lumen pressure (P_A) and pressure in pleura space (P_{PL}).

$$P_{TP} = P_A - P_{PL}$$

Intrapleural pressure = -Lung elastic recoil pressure

The difference between alveolar lumen pressure and chest wall outer surface or body surface pressure gives the trans-thoracic pressure. Trans-thoracic pressure is also the sum of lung elastic recoil and chest wall pressure, and active inspiratory (expiratory) pressure. The total pressure in respiratory system (P_{RS}) is equal to the sum of lung elastic recoil pressure (P_L) and chest wall, and active inspiratory or expiratory effort (pressure in thorax) (P_{CW}).

$$P_{RS} = P_L + P_{CW}$$

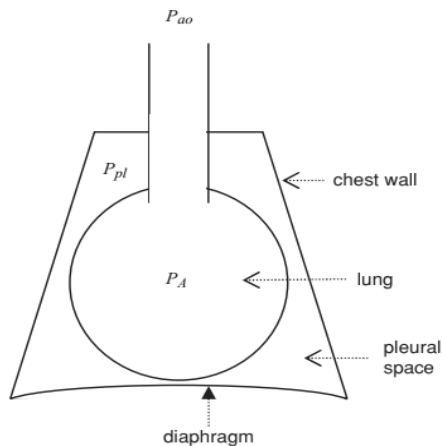


Figure 3.1: Respiratory pressure [3].

Pressure gradient

Pressure gradient (usually of air, but more broadly of any fluid) is the difference in pressure at different area in the system. It is a physical quantity that represents the direction and rate at which pressure rises the fastest around a given point and is expressed in units of pressure per meter.

Volume

Volume is the space occupied by gas molecule at a specific pressure. Lung volume changes when the internal pressure level changes.

Capacity of the lung

- Tidal volume (V_T):- the volume of air inhaled or exhaled in normal breathing $\approx 500\text{mL}$ in an adult.
- Residual volume (RV):- Volume of air remain in the lungs past maximum exhalation ≈ 25 to 35% of (TLC) $\approx 1200\text{mL}$ in males and $\approx 700\text{mL}$ in females.

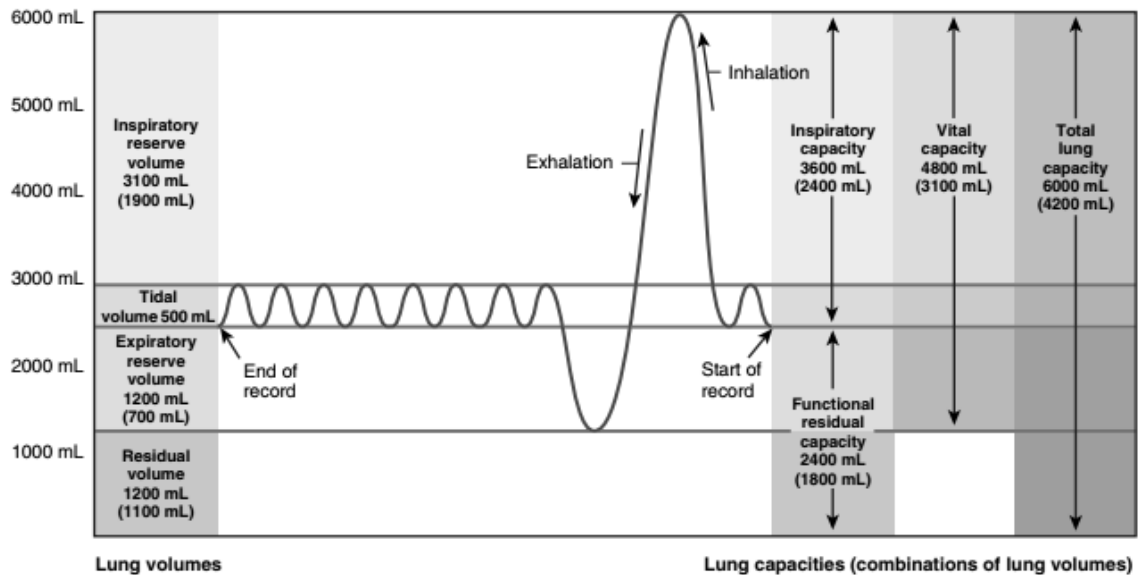


Figure 3.2: Lung capacity [2].

- Functional residual capacity (FRC):- volume of air remain in lungs past tidal exhalation $=ERV + RV \approx 2400$ in males and $\approx 1800\text{mL}$ in females.
- Inspiratory capacity (IC):- maximum volume of air inhaled after $V_T = V_T + I_{RV} \approx 3600\text{mL}$ in males and $\approx 2400\text{mL}$ in females.
- Expiratory reserve volume (ERV):- volume of air exhaled after V_T exhalation ≈ 1200 in adult males and $\approx 1100\text{mL}$ in females.
- Inspiratory reserve volume (IRV):- volume of air inhaled after normal tidal inspiration $\approx 3100\text{mL}$ in male and $\approx 1900\text{mL}$ in female.
- Vital capacity (VC):- total volume of air a person breath $= TV + ERV + IRV \approx 4800$ mL in male and ≈ 3100 mL in adult female.

- Total lung capacity (*TLC*):- total volume of air in lungs = $TV + IRV + ERV + RV = VC + RV \approx 6000$ mL in male and ≈ 4200 mL in female.

Gas flow or flow rate

The movement of gas volume over time. At low gas flow rate gas molecule moves smoothly and the resulting flow pattern is laminar flow. Laminar flow exists from the medium-sized bronchi down to the level of the bronchioles during calm breathing. Laminar flow may be limited to the smaller airways during exercise, when the air flow is more fast. At a higher gas flow rate gas molecules moves unevenly faster and the resulting flow pattern is turbulent flow. The biggest airways, such as the trachea, have turbulent flow. As the pressure gradient rises, so does the flow. The amount of air flow to the lung is proportional to the change in volume of lung per time.

$$Q = \frac{dV}{dt} \quad (3.1)$$

The change in volume and pressure difference $\Delta P = P_{rs}$ for laminar air flow can be described by Hagen Poiseville law [21].

$$Q = \frac{dV}{dt} = \frac{\Delta P}{R} \quad (3.2)$$

Where Q is air flow, R is flow resistance in respiratory system and t is time.

Resistance of respiratory system

Respiratory resistance is the force that opposes the movement or flow of gas and is flow dependent [25].

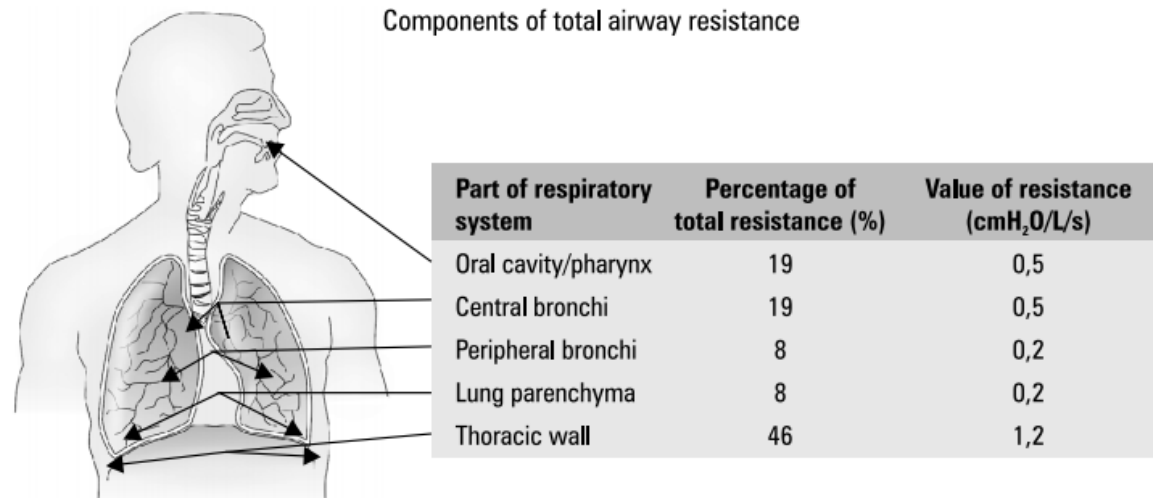


Figure 3.3: Resistance of the respiratory system [4].

The resistance level in gas flow can be determined by the following three factors. The first is flow rate as the flow rate increase the resistance will increase and vice versa. The second is physical properties, internal diameter, curvature, inner surface and length of the tube. The third is gas properties including viscosity and density. Flow resistance R depends on fluid or air viscosity (η) length (L) and tube radius (r) [26] [27].

$$R = \frac{8}{\pi} \eta \frac{L}{r^4} \quad (3.3)$$

Figure 3.3 shows that the total airway resistance is determined by the contribution of airway resistance from the mouth to the level of peripheral bronchi, lung parenchyma resistance, and chest wall resistance. Resistance of the nasal cavity accounts more than half of total airway resistance during tidal breathing through the nose.

Compliance

Respiratory compliance is a measure of volume and pressure relation ship in the system. Elastic properties of respiratory system is measured by compliance (C). Compliance is the ability of pressure to change the volume of lung in the respiratory system [21].

$$C = \frac{dV}{dP} \quad (3.4)$$

Where dV is change of lung volume and P is gauge pressure in lung.

Compliance of the chest wall (thorax)

$$C_{CW} = \frac{dV}{dP_{CW}} \quad (3.5)$$

Compliance of the lung

$$C_L = \frac{dV}{dP_L} \quad (3.6)$$

Compliance of the respiratory system (lung-thorax)

$$C_{RS} = \frac{dV}{dP_{RS}} \quad (3.7)$$

Total respiratory system compliance is calculated by adding the reciprocal of lung and chest wall compliance:

$$\frac{1}{C_{RS}} = \frac{1}{C_L} + \frac{1}{C_{CW}} \quad (3.8)$$

Lung and chest wall average compliance in adult over a normal tidal volume are approximately $200 \text{ ml/cmH}_2\text{O}$. Therefore the average C_{RS} in adult is $100 \text{ ml/cmH}_2\text{O}$.

Time constant

The time constant describes how quickly the respiratory system responds to a pressure change. A rapid reaction correlates to a small time constant, whereas a sluggish response corresponds to a long time constant. Time constant τ is theoretically equivalent to the product of resistance and compliance. This indicates that the slower the Volmax, the higher the compliance and/or resistance. When applying continuous pressure to the respiratory system under muscle relaxation (passive patient), a time constant (τ) shows the time necessary to accomplish 63 % percent of the maximal volume variation [7].

$$\tau = R \times C = \frac{R}{E} \quad (3.9)$$

The mechanics of respiratory system

To deliver tidal volume of air medical ventilator has to supply specific driving ventilator pressure to overcome the opposing force. Opposing force includes elastic pressure (which inflate lung and chest wall) and resistive pressure (air way pressure). Breathing is controlled by single variable, pressure or flow rate depending on modes of ventilator [28]. Respiratory system mechanics can be used to model, control and monitor mechanical ventilator for delivering specific volume of air for the patient [3, 24, 29–31].

$$P_{rs} = P_{aw} + p_{mus} = E_{rs}\Delta V + R_{rs}\dot{V} + PEEP \quad (3.10)$$

Eq.3.10 is the respiratory mechanics model obtained from equation of motion by dynamic

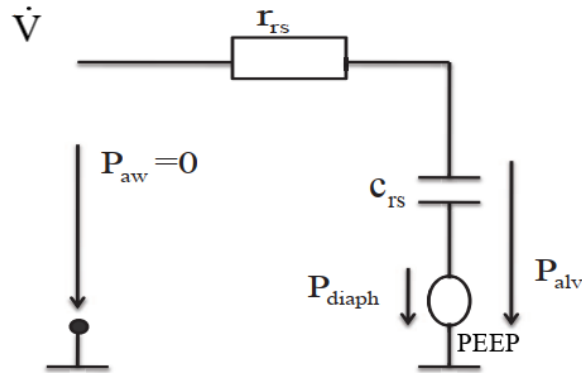


Figure 3.4: Breathing mechanics [5].

and static analysis. where P_{rs} is respiratory pressure, $P_{aw}=P_{ao}$ is air way pressure at the opening or mouth (ventilator air driving pressure), p_{mus} is respiratory muscles pressure=0

for passive breathing patients, E_{rs} respiratory elastance (inverse of compliance), ΔV is change in lung volume, R_{rs} respiratory resistance, $Q = \dot{V}$ is flow rate, P_{pl} is pleural pressure, P_A is alveolar pressure and $PEEP$ is positive end expiratory pressure (alveolar end expiratory pressure). The term $E_{rs}\Delta V$ in *eq.3.10* is lung and chest wall elastic pressure, and $R_{rs}\dot{V}$ is resistive pressure P_{res} or pressure at the air way and endotracheal tube. The elastic and resistive pressure of respiratory system can be expressed by the following equation [31]:

$$E_{rs}\Delta V = \Delta P \Leftrightarrow E_{rs} = \frac{P_{plat} - PEEP}{\Delta V} \quad (3.11)$$

$$R_{rs}\dot{V} = P_{peak} - P_{plat} \Leftrightarrow R_{rs} = \frac{P_{peak} - P_{plat}}{\dot{V}} \quad (3.12)$$

where P_{peak} is peak pressure and P_{plat} is plateau pressure.

3.2 System Mechanical Structure

Mechanism is used to deliver uncontrolled air to the lung simulator through the pipe and the operation is the same as piston and cylinder. This part includes

- Housing that cover the mechanical structure of the system, protect the entire system from atmospheric contact, and other mechanism are mounted on it.
- DC motor as system actuator that rotate at speed of 8 RPM. The system is designed for an adult that breath at the rate of 8 breath per minute. The Motor rotation will cause the one end fixed bar to rotate on its fixed axis. One end of the bar is attached to the vertical sliding bar this will make the vertical bar to move up and down or vertical linear translation.
- Piston cylinder operating mechanism: One big and one small plastic container are used for this purpose. Bigger container is full of pure water and smaller one is inverted. The bottom end of vertically sliding bar is attached to the inverted (smaller) container. An alternative external diaphragm can be used for the purpose instead of a water filled container, but the cost of the system might get higher.

When the motor starts rotating, one end fixed bar starts to oscillate (rotate clockwise and counter-clockwise). This rotational motion will cause the slider with attached container to move up and down and deliver air through the airway tube to the lung simulator. System mechanical design is shown in Figure 3.5.

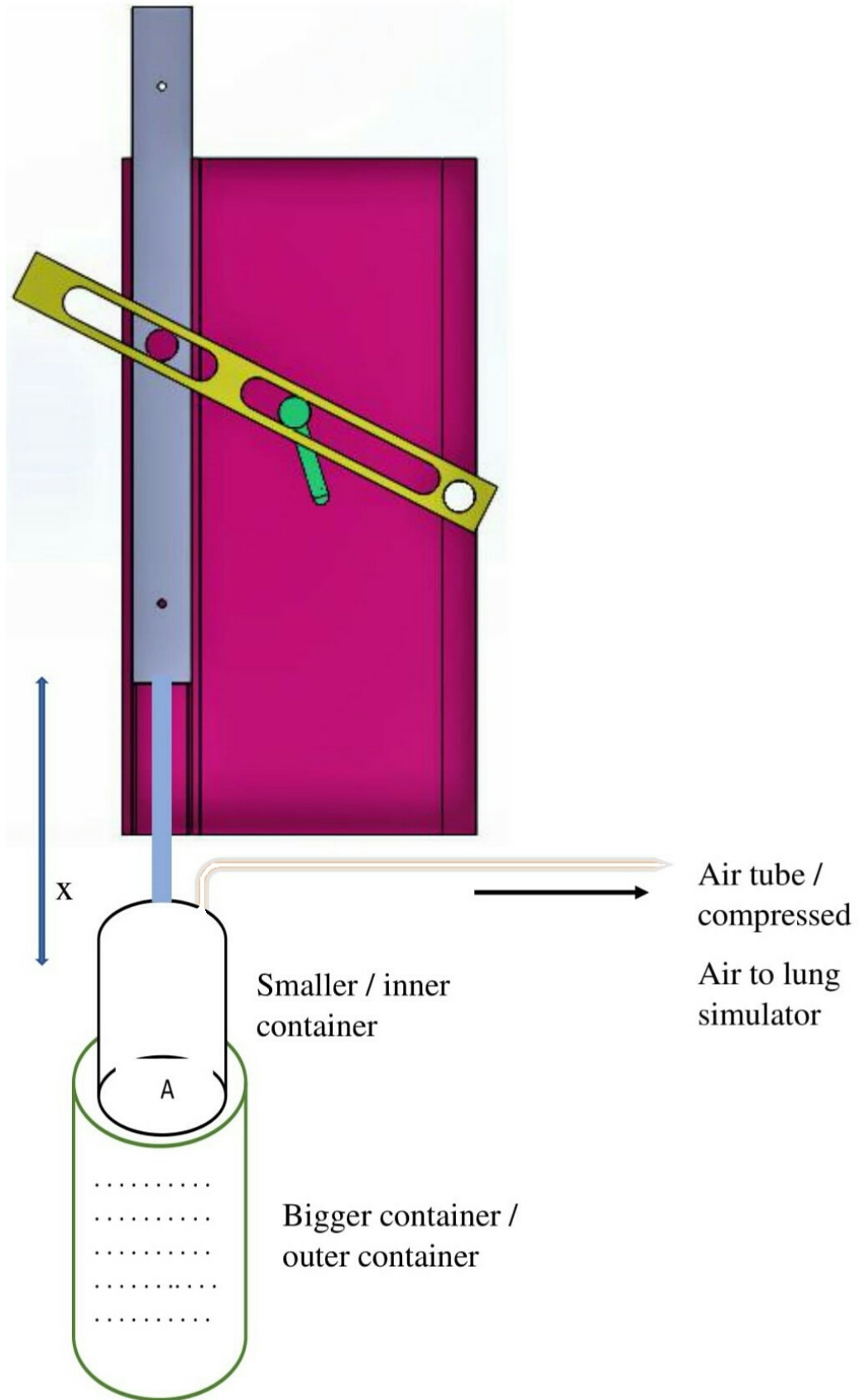


Figure 3.5: System mechanism.

Armature controlled DC motor Dynamic Equation

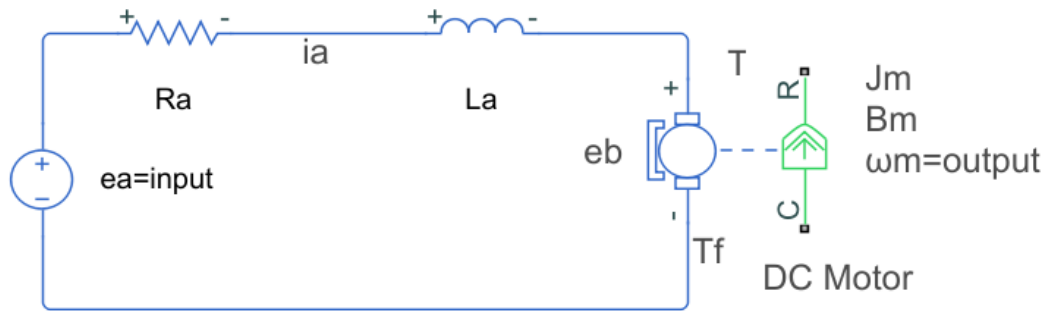


Figure 3.6: Armature controlled dc motor circuit diagram.

i. Steady state motor equation

Developed torque (motor torque)

$$T = K_t \times i_a - T_f \quad (3.13)$$

Armature circuit

$$e_a = i_a \times R_a + e_b \quad (3.14)$$

Back emf

$$e_b = K_e \times \omega \quad (3.15)$$

ii. Dynamic equation of the motor

Electrical dynamics using KVL for electrical circuit

$$e_a(t) = R_a \times i_a(t) + L_a \frac{d}{dt} i_a + e_b(t) \quad (3.16)$$

Mechanical dynamics

$$T(t) = J \times \frac{d}{dt} \omega(t) + B \times \omega(t) \quad (3.17)$$

Electro mechanical motor equation

$$e_b(t) = K_e \times \omega(t) \quad (3.18)$$

$$T(t) = K_t \times i_a(t) \quad (3.19)$$

ω is an output and $e_a(t)$ is an input variable

Laplace transforms of electrical equation

where

e_a	armature terminal voltage	e_b	back emf
i_a	armature current	T	motor torque
T_f	static frictional torque	ω	motor speed
R_a	armature resistance	B	Viscous friction
L_a	armature inductance	J	rotational inertia
θ	angular displacement		

$$\begin{aligned} E_a(s) &= L_a \times s \times I_a(s) + R_a \times I_a(s) + E_b(s) \\ &= (L_a \times s + R_a)I_a(s) + E_b(s) \end{aligned}$$

$$I_a(s) = \frac{E_a(s) - E_b(s)}{L_a \times s + R_a} = \left[\frac{1}{L_a \times s + R_a} \right] [E_a(s) - E_b(s)] \quad (3.20)$$

Laplace transforms of Mechanical equation

$$\begin{aligned} T(s) &= (J \times s + B)\omega(s) \\ \omega(s) &= \left[\frac{1}{J \times s + B} \right] T(s) \end{aligned} \quad (3.21)$$

Laplace transforms of electro-mechanical equation

$$E_b(s) = K_e \times \omega(s) \quad (3.22)$$

$$T(s) = K_t \times I_a(s) \quad (3.23)$$

Angular velocity transfer function

$$\frac{\omega(s)}{E_a(s)} = \frac{G(s)}{1 + H(s)G(s)} \quad (3.24)$$

$$G(s) = K_t \times \left[\frac{1}{L_a \times s + R_a} \right] \left[\frac{1}{J \times s + B} \right] \quad (3.25)$$

$$H(s) = K_e \quad (3.26)$$

$$\frac{\omega(s)}{E_a(s)} = \frac{K_t}{L_a \times s(J \times s + B) + K_t \times K_e}$$

$$= \frac{K_t}{L_a \times J \times s^2 + (R_a \times J + B \times L_a)s + (K_t \times K_e + R_a \times B)} \quad (3.27)$$

Position transfer function

$$\theta(t) = \int \omega(t) dt$$

$$= \left[\frac{1}{s} \right] \omega(s) \quad (3.28)$$

$$\frac{\theta(s)}{E_a(s)} = \frac{1}{s} \left[\frac{K_t}{L_a \times J \times s^2 + (R_a \times J + B \times L_a)s + (K_t \times K_e + R_a \times B)} \right]$$

$$= \frac{K_t}{L_a \times J \times s^3 + (R_a \times J + B \times L_a)s^2 + (K_t \times K_e + R_a \times B)s} \quad (3.29)$$

Graphical analysis

By taking eq. 3.29 and finding the angle β using trigonometric relation.

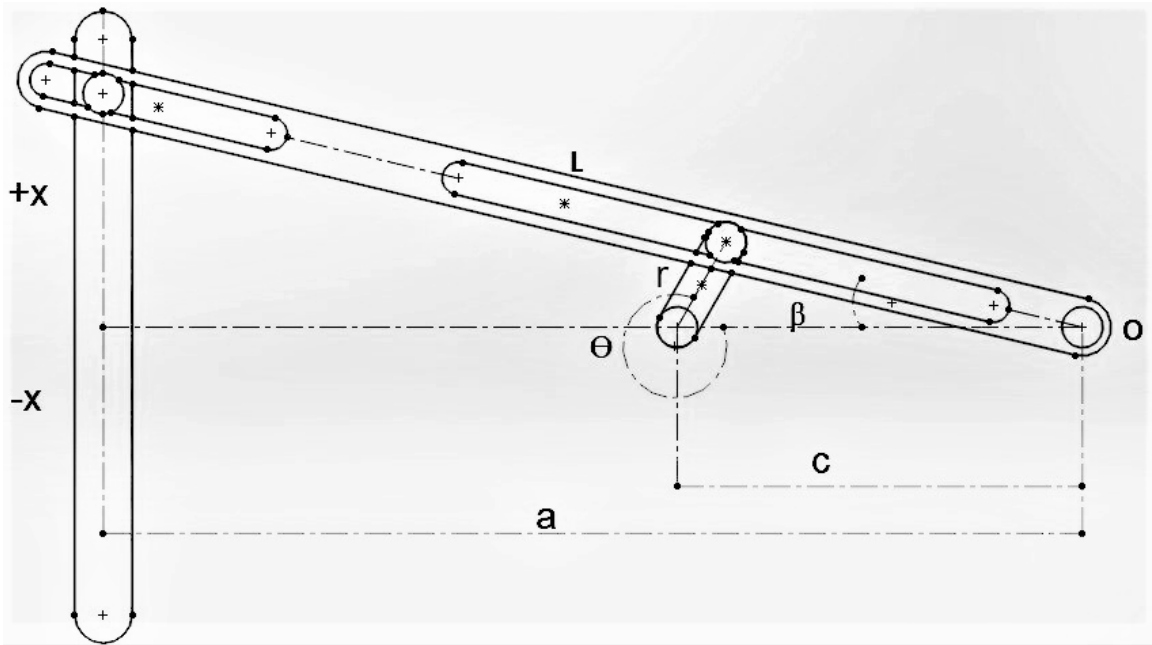


Figure 3.7: 2D model of system mechanism.

$$a = L \times \cos(\beta)$$

$$x = L \times \sin(\beta)$$

$$\frac{x}{a} = \frac{L \times \sin(\beta)}{L \times \cos(\beta)} = \frac{-r \sin(\theta)}{c - r \times \cos(\theta)}$$

$$x = \frac{-a \times r \times \sin(\theta)}{c - r \times \cos(\theta)} \quad (3.30)$$

From fig.3.7 β is rotational angle from fixed point and θ is angular position measured clockwise. $r = 0.0442$ m, $a = 0.149043$ m, $c = 0.0657$ m and $L = 0.26$ m and maximum displacement of x is $+x_{max} = 13.55$ cm and $-x_{max} = -13.55$ cm this is shown in simulation result in Figure 3.10. Volume of gas displaced

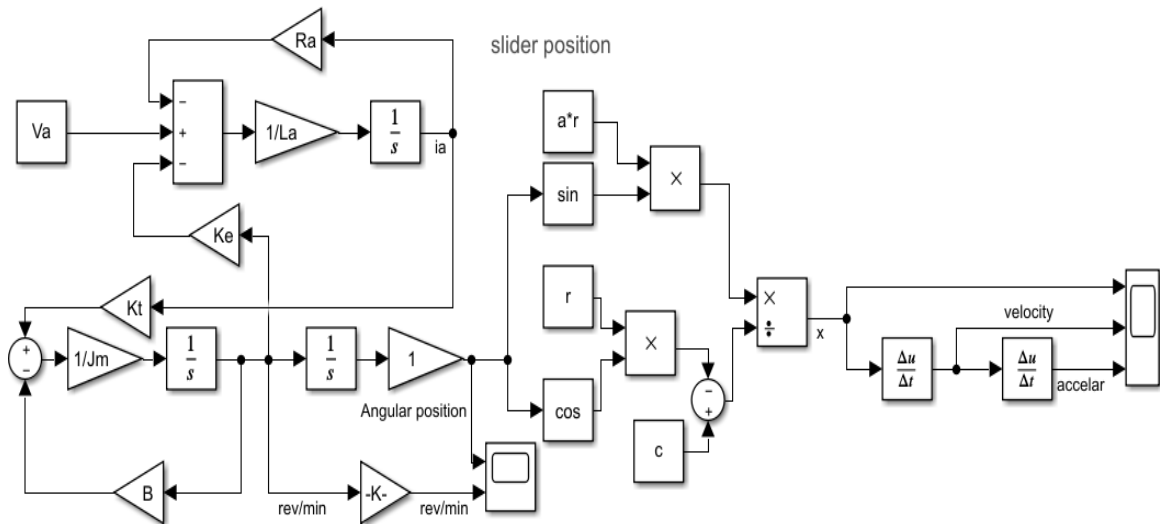


Figure 3.8: Simulink model.

$$V = A \times x$$

$$dV = A \times dx$$

Differentiating dV with respect to time

$$\frac{dV}{dt} = A \times \frac{d}{dt}x \quad (3.31)$$

$$Q = A * v \quad (3.32)$$

Where Q is flow rate, V is volume, A is area of inner cylinder, x displacement of inner cylinder and v is velocity.

The downward speed of the mechanism is faster than the upward speed, as shown on the velocity graph in Figure 3.10. Downward motion is taken as positive and upward is taken as negative. The time required for the mechanism to complete the positive cycle is 1/3 of

the full period and the negative cycle will be 2/3 of a single revolution. The mechanism was designed to be able to have approximately 1/3 of an inspiration period from a single breathing cycle (inspiration plus expiration). The positive cycle is taken as inspiration and the negative cycle as the expiration of the single respiration period. Position, velocity and acceleration response of the mechanism using Simulink model is shown in graph in Figure 3.10. The designed mechanism can deliver a volume up to 600 milliliter of air to the lung simulator.



Figure 3.9: Angular displacement and angular speed of the driving motor.

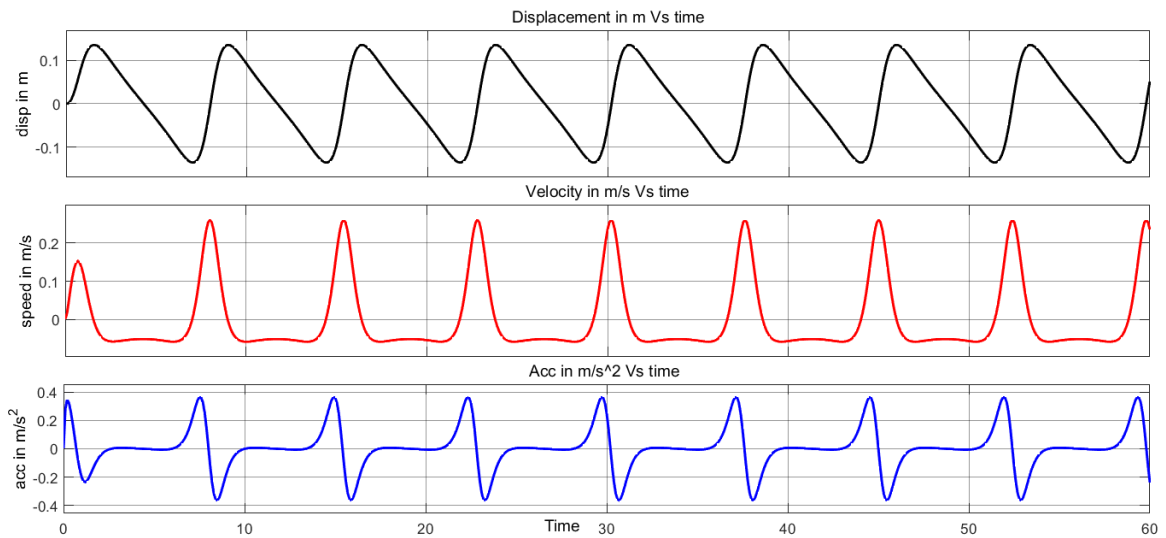


Figure 3.10: Displacement, velocity and acceleration response of the mechanism.

3.3 Pneumatic Unit

The designed pneumatic has two independently functioning parts; the filling part and the gas mixture supplying the pneumatic circuit to the patient. The filling part has air and oxygen input pneumatic lines. Mould, pollen, dust mites, and other impurities are removed from the air using air purifiers. For those with serious allergies, a better air filtration system may make a significant impact because pollutants are removed through filter systems, which improves air quality. The average atmospheric humidity range from 30% to 55%. Depending on the percentage of air moisture atmospheric air could be humidified or dehumidified. Mold and fungal spores may spread if the humidity is too high. Dehumidifiers remove excess moisture from indoor environment and inhibit mold and bacterial development. Humidifiers is recommended in dry environment. Filtered and humidified air is filled into one input line. Oxygen is supplied into the second input line. A mixture of oxygen and filtered air will be available for supplying the lungs (lung simulator). This will be done in the expiration period of breathing and compressed into the targeted patient during the cycle of inspiration. The gas reservoir (plastic container or external diaphragm) could have a capacity of air sufficient for a patient who needs a high tidal volume of air. As indicated in the literature review, an adult may consume as much as 6 L of air per minute. If a person has a breathing rate of 8 breaths per minute, the reservoir could contain 750 ml to deliver a sufficient tidal volume of air and oxygen mixture. Fraction of inspired oxygen (FIO_2) concentration in the inspired gas can be determined by the clinician. The percentage of O_2 in the gas mixture varies from 21 % (atmospheric air) to 100%.

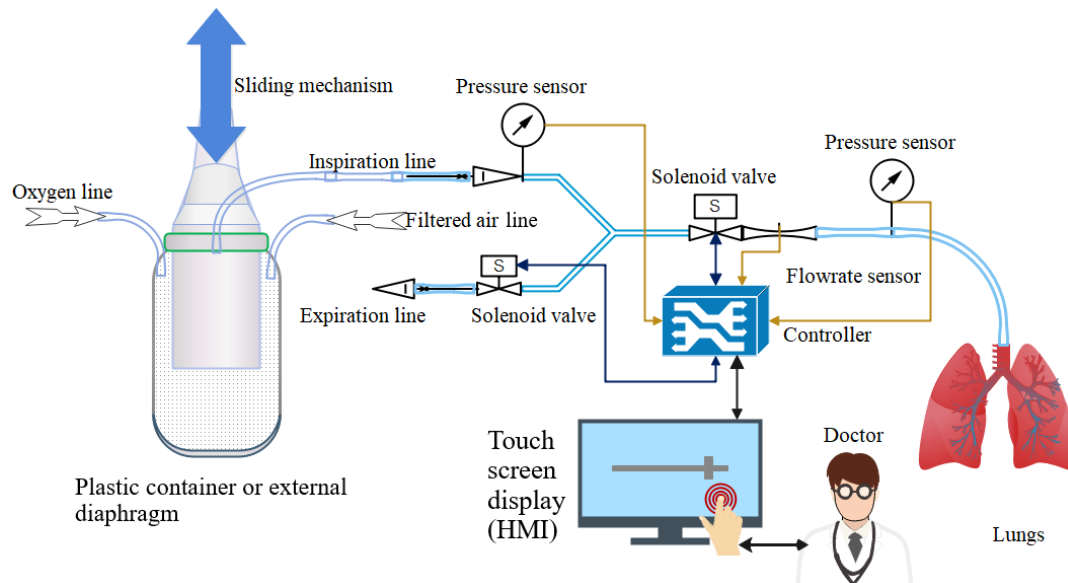


Figure 3.11: Inspiration and expiration air line.

Chapter4

CONTROLLER DESIGN

4.1 Fuzzy Logic Controller Design

The notion of fuzzy sets is used to translate crisp inputs into crisp outputs in a fuzzy logic system. It is made up of three stages: input, processing and output. Sensor or other inputs, like switches and so on, are mapped to the proper membership functions and truth values at the input stage. The processing step calls each applicable rule and generates a result for each, before combining the results. Finally, at the output step, the combined result is converted back into a specified control output value. An inference engine in a fuzzy logic system operates with fuzzy rules. Rather of attempting to represent a system analytically, Fuzzy Logic uses a basic, rule-based IF X AND Y THEN Z method to solve control issues. In computers, fuzzy logic offers an option to representing linguistic and subjective characteristics of the actual world. FL requires various numerical parameters to function, such as considerable error and significant rate-of-change-of-error, although the precise values of these numbers are rarely important unless particularly responsive performance is desired, in which case empirical tuning would be used to find them. Instead of a system model, the FLC method is built on experiences and linguistic meanings.[32][33]

Error (e) is the difference between reference value (r) and reading or measured value of the sensor (y).

$$e(t) = r(t) - y(t) \quad (4.1)$$

The following stages are often employed in numerous applications:

- Assemble the system's inputs.
- Make the inputs fuzzier (make them appropriate fuzzy values).

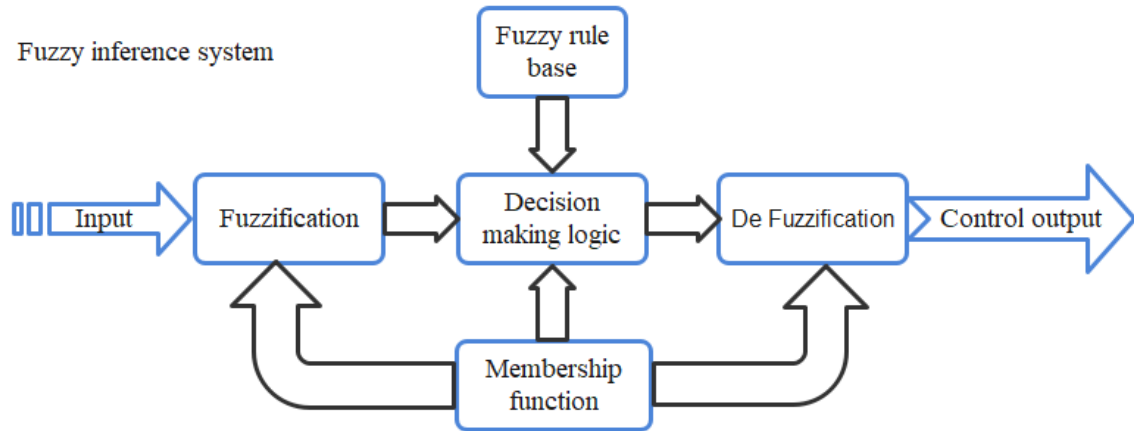


Figure 4.1: A fuzzy controller’s process blocks for the system.

- Apply the inputs to all of the system’s rules, one by one, then conduct a global accumulation of the outputs.
- Remove the fuzziness from the outputs (create crisp numbers from the output fuzzy values).
- Apply the system’s precise outputs.
- Repeat all of the procedures until the system is under control.

The fuzzy logic controller design have four input variables (pressure error, volume error, flow rate and change in pressure) and two output variables (airway valve and expiration valve).

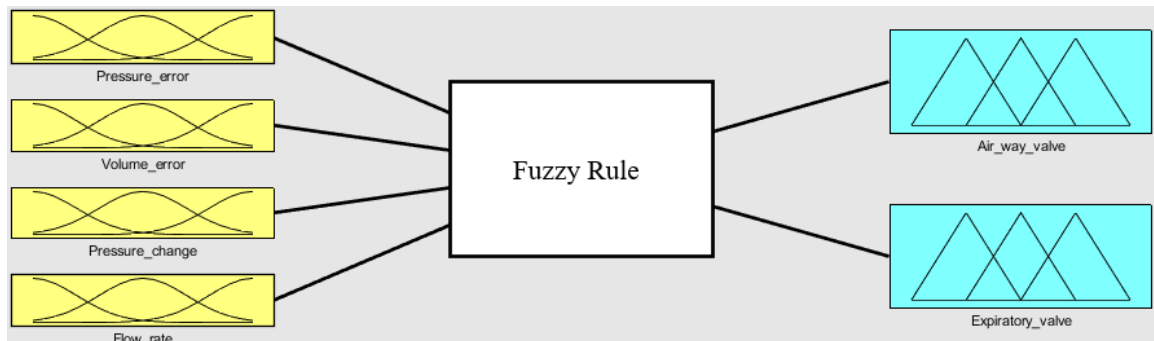


Figure 4.2: Input output fuzzy logic controller design.

4.1.1 Fuzzification

The controller’s initial step is fuzzification, which uses a search in one or more membership functions to transform each piece of input data into degrees of membership. The fuzzifi-

cation block next compares the input data to the rules' conditions to see how well each rule's condition fits that specific input instance. Each linguistic word that relates to that input variable has a degree of membership. Fuzzy membership function for the pressure error, volume error, flow rate and the change in pressure are shown in the figure 4.3,4.5,4.6 and 4.7 respectively. The first input variable is the difference between set pressure and measured airway pressure (pressure error signal). Reference pressure is set to 20 cmH_2O and airway pressure is measured from the system or from analog pin A0 of Arduino mega 2560. Membership function of pressure error signal have two values. When the difference between reference and reading signal is positive it will have positive (PV) value. If the difference is negative it will have (NV) value.

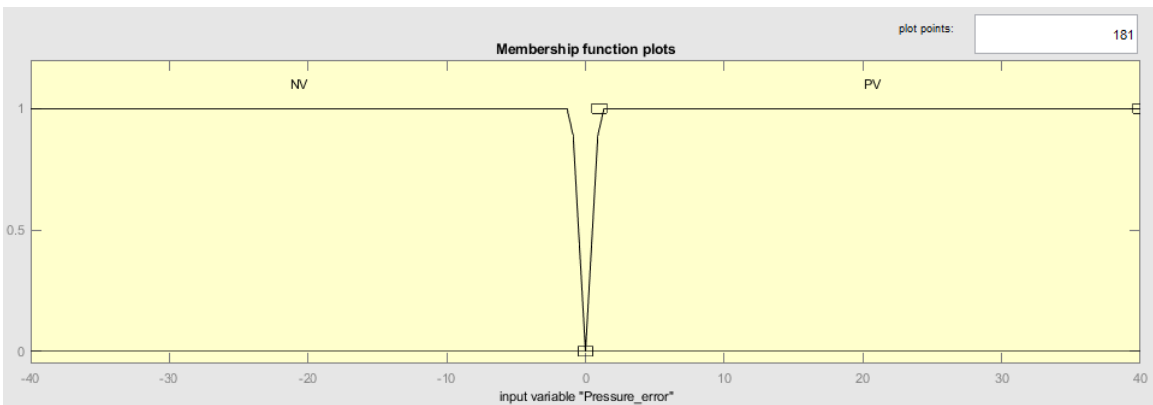


Figure 4.3: Pressure error input variable.

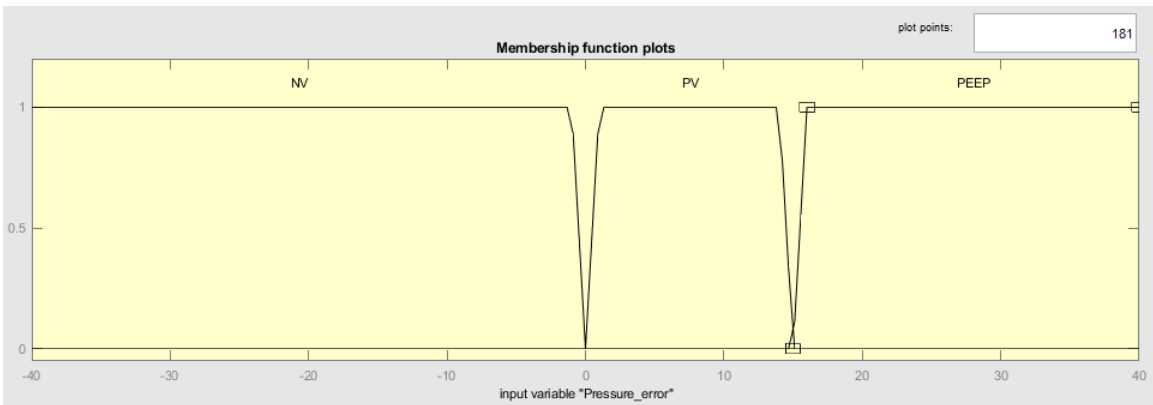


Figure 4.4: Pressure error input variable (membership) function for PEEP control.

The second input variable is the difference between set volume and measured value (Volume error). The amount of volume required for the patient depends on weight of the person (8 ml/kg). Considering the mass flow rate the set volume can be varied from 160 to 800 ml for the weight of a person from 20-100 kg respectively. The volume error membership

function have two values as the pressure error membership function. Changing the pressure error membership function is required to control the airway valve at the end of expiration to monitor the PEEP level.

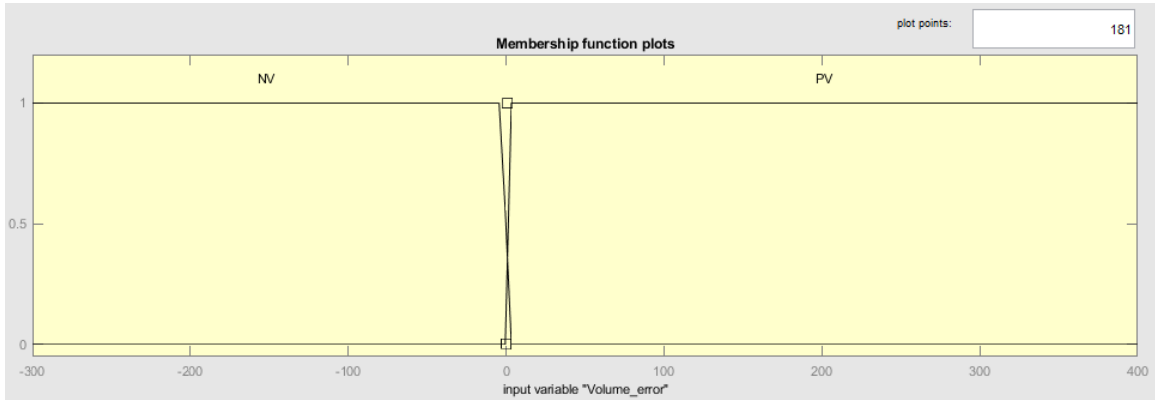


Figure 4.5: Volume error input variable.

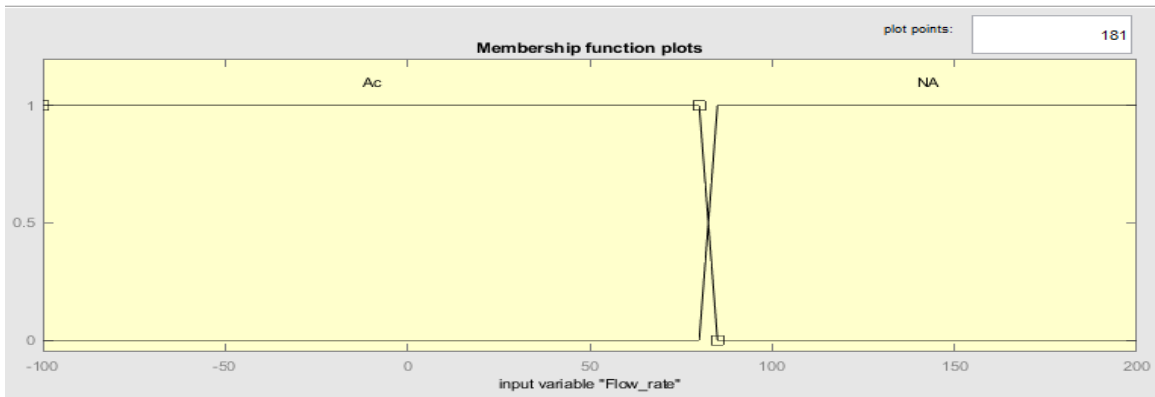


Figure 4.6: Flow rate input variable.

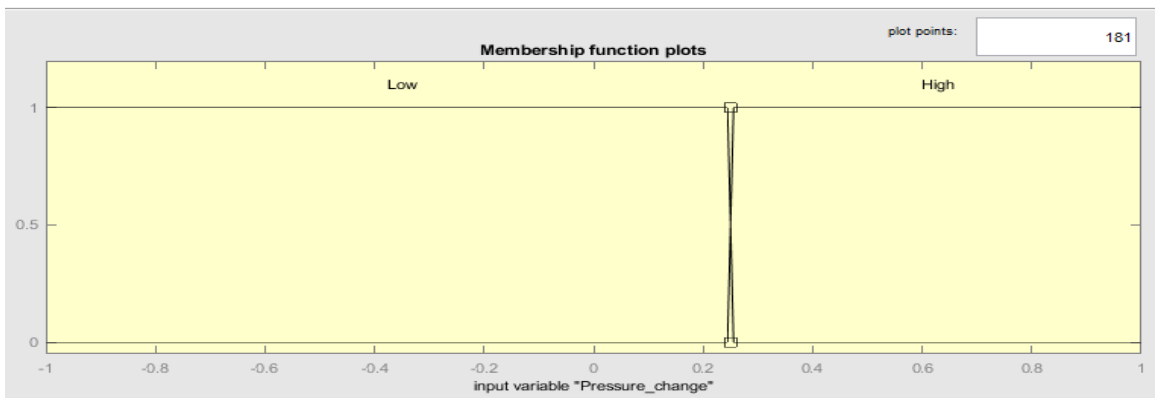


Figure 4.7: The change in pressure input variable.

4.1.2 Fuzzy rules

A rule basis refers to a set of rules. The rules are written in a “If Then” style, with the If side referred to as the conditions and the Then side as the conclusion. Based on the measured inputs, the computer may execute the rules and calculate a control signal. The control strategy is contained in a rule-based controller in natural language. In "fuzzy logic," fuzzy rules have been proposed as a significant technique for articulating bits of knowledge. There is no such thing as a single form of fuzzy rule, nor is there a single sort of "fuzzy logic". Fuzzy logic systems employ fuzzy rules to derive an outcome from input variables. The form of a fuzzy logic controller rule is:

The premise is that x equals A.

IF x equals A, then y equals B.

As a result, y equals B.

The premise x is A can only be true or false in crisp logic. In a fuzzy rule, however, the premise x is A and the consequent y is B might be somewhat true rather than fully true or fully false. This is accomplished by utilizing fuzzy sets to represent the linguistic variables A and B. Figure 4.10 shows fuzzy logic controller input and output relation rule viewer.

No	Rule
1	If (Pressure error is NV) and (Volume error is NV) and (Pressure change is Low) then (Air way valve is OFF)(Expiratory valve is ON)
2	If (Pressure error is NV) and (Volume error is NV) and (Pressure change is High) then (Air way valve is OFF)(Expiratory valve is ON)
3	If (Pressure error is NV) and (Volume error is PV) and (Pressure change is Low) then (Air way valve is OFF)(Expiratory valve is ON)
4	If (Pressure error is NV) and (Volume error is PV) and (Pressure change is High) then (Air way valve is OFF)(Expiratory valve is ON)
5	If (Pressure error is PV) and (Volume error is NV) and (Pressure change is High) then (Air way valve is OFF)(Expiratory valve is ON)
6	If (Pressure error is PV) and (Volume error is PV) and (Pressure change is Low) then (Air way valve is ON)(Expiratory valve is ON)
7	If (Pressure error is PV) and (Volume error is PV) and (Pressure change is High) then (Air way valve is ON)(Expiratory valve is OFF)
8	If (Pressure error is PV) and (Volume error is NV) and (Pressure change is Low) then (Air way valve is ON)(Expiratory valve is ON)
9	If (Flow rate is NAC) then (Air way valve is OFF)(Expiratory valve is OFF)

Fuzzy rules for PEEP control

No	Rule
1	If (Pressure error is NV) and (Volume error is NV) and (Pressure change is Low) then (Air way valve is OFF)(Expiratory valve is ON)
2	If (Pressure error is NV) and (Volume error is NV) and (Pressure change is High) then (Air way valve is OFF)(Expiratory valve is ON)
3	If (Pressure error is NV) and (Volume error is PV) and (Pressure change is Low) then (Air way valve is OFF)(Expiratory valve is ON)
4	If (Pressure error is NV) and (Volume error is PV) and (Pressure change is High) then (Air way valve is OFF)(Expiratory valve is ON)
5	If (Pressure error is PV) and (Volume error is NV) and (Pressure change is High) then (Air way valve is OFF)(Expiratory valve is ON)
6	If (Pressure error is PV) and (Volume error is PV) and (Pressure change is Low) then (Air way valve is ON)(Expiratory valve is ON)
7	If (Pressure error is PV) and (Volume error is PV) and (Pressure change is High) then (Air way valve is ON)(Expiratory valve is OFF)
8	If (Pressure error is PV) and (Volume error is NV) and (Pressure change is Low) then (Air way valve is ON)(Expiratory valve is ON)
9	If (Flow rate is NAC) then (Air way valve is OFF)(Expiratory valve is OFF)
10	If (Pressure error is PEEP) and (Volume error is PV) and (Pressure change is Low) then (Air way valve is OFF)(Expiratory valve is ON)
11	If (Pressure error is PEEP) and (Volume error is PV) and (Pressure change is High) then (Air way valve is OFF)(Expiratory valve is ON)

4.1.3 Output defuzzification

Defuzzification is the final stage of a fuzzy logic system and is the polar opposite of fuzzification, which creates crisp output y from the aggregated output of the fuzzy logic system. Given fuzzy sets and accompanying membership degrees, defuzzification is the process of creating a measurable outcome in crisp logic. It's the procedure for converting a fuzzy set to a crisp set. It's most commonly used in fuzzy control systems. These systems will have a set of rules for transforming a collection of variables into a fuzzy outcome, which is defined in terms of fuzzy set membership. Defuzzified output of Patient air way valve and expiration valve for the simulation and experiment is shown in figure 4.8 and 4.9 respectively. Same two controlled output variables with different fuzzy rules are used to control the two valves (airway and expiration). Valves to be controlled are connected on two PWM pin of Arduino as shown in Simulink model figure 5.13 and Fritzing circuit design in figure 5.14. Figure 4.11 and 4.12 shows input output relation surface view of the controller.

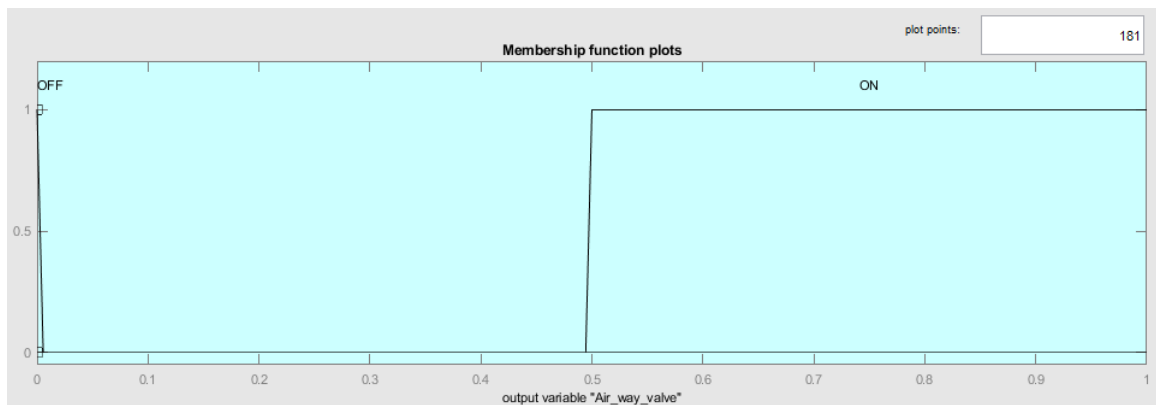


Figure 4.8: Patient air way and expiratory valve output variable.

Prototype controller output command for the valves.

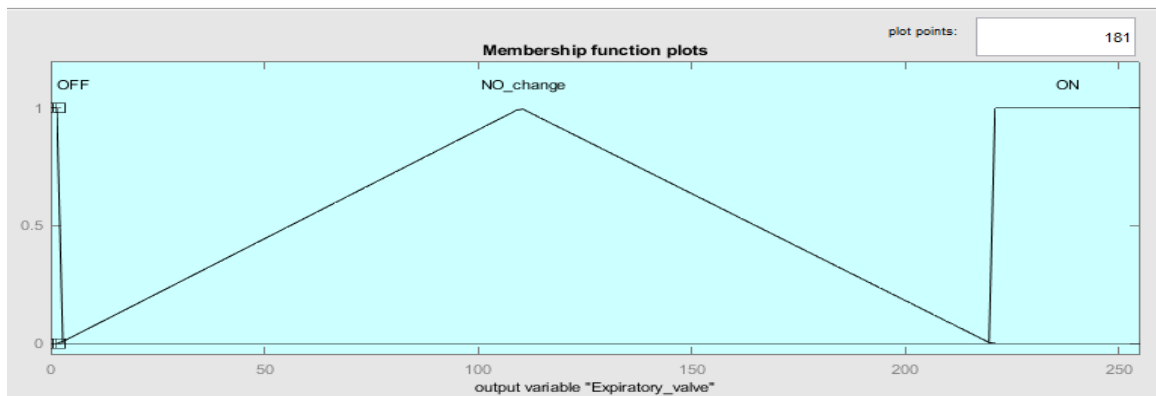


Figure 4.9: Inspiratory and expiration valve output variable for prototype.

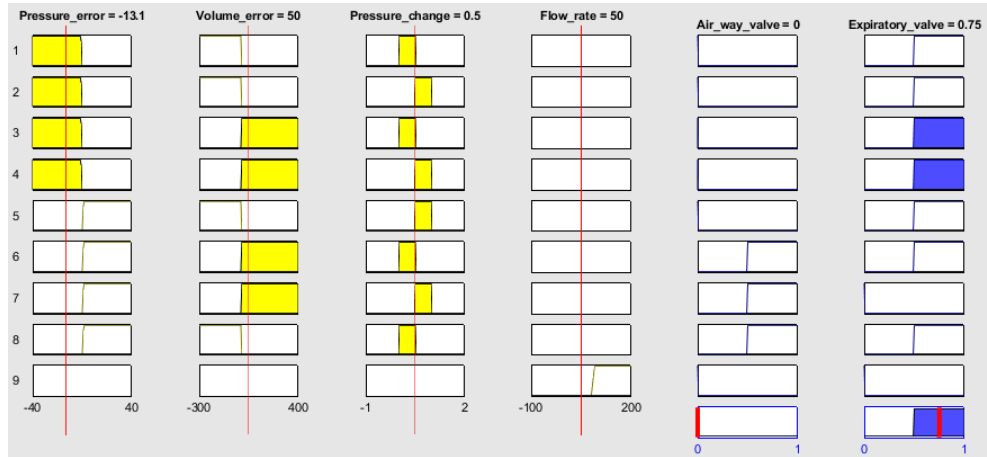


Figure 4.10: Fuzzy rule viewer.

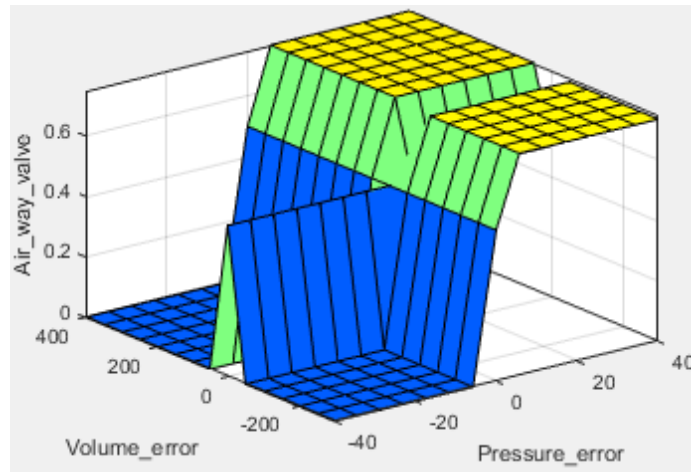


Figure 4.11: Pressure, volume and airway valve relation surface viewer.

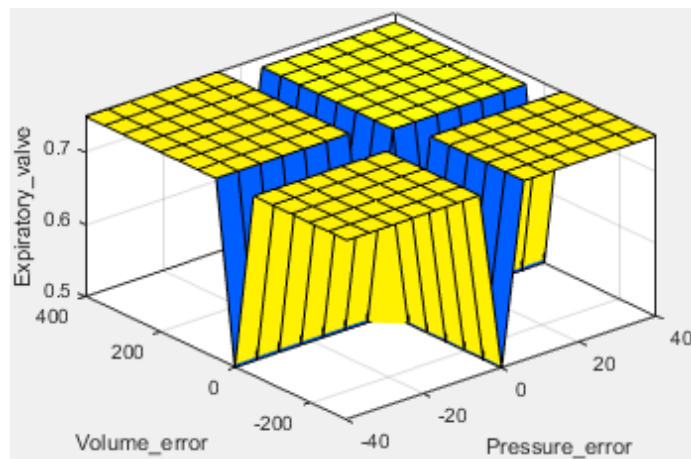


Figure 4.12: Pressure, volume and expiratory valve relation surface viewer.

4.2 Simulink Model

The open loop (with no controller) response of the mechanical ventilator simulation result for the system is shown in the figure 4.13. This modal contains the delivering mechanism (system mechanism and slider position subsystem/ constant volumetric flow rate source), inspiration line (tube and nasal mask), sensors and lungs subsystem. Open loop model is used to show the total delivering capacity of the system. Open loop system model Simulink simulation result is shown in Figure 6.1.

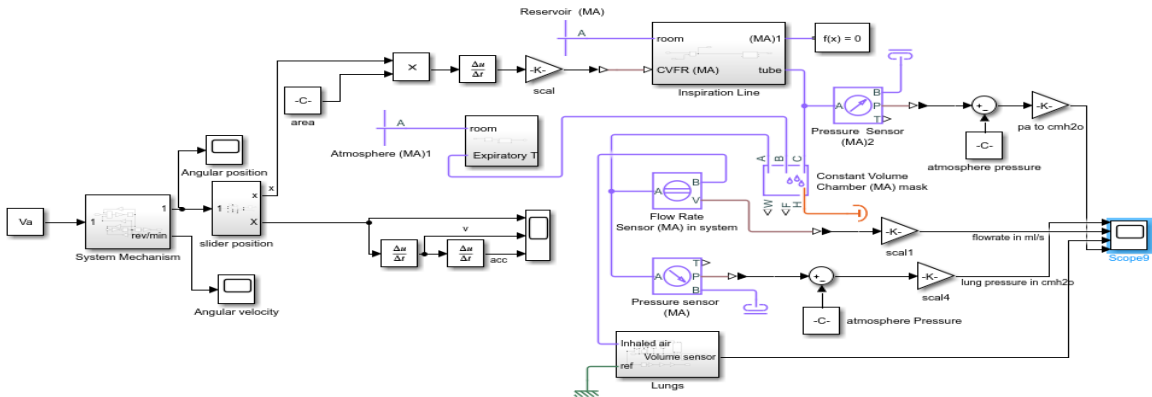


Figure 4.13: Simulink model with out controller (open loop).

The simulation result for the closed loop (with controller) of the mechanical ventilator system is shown in the figure 4.14. This model have an additional fuzzy logic controller and controlling valve (inspiratory and expiratory) model system on the open loop model. Closed loop system response is shown in Figure 6.2.

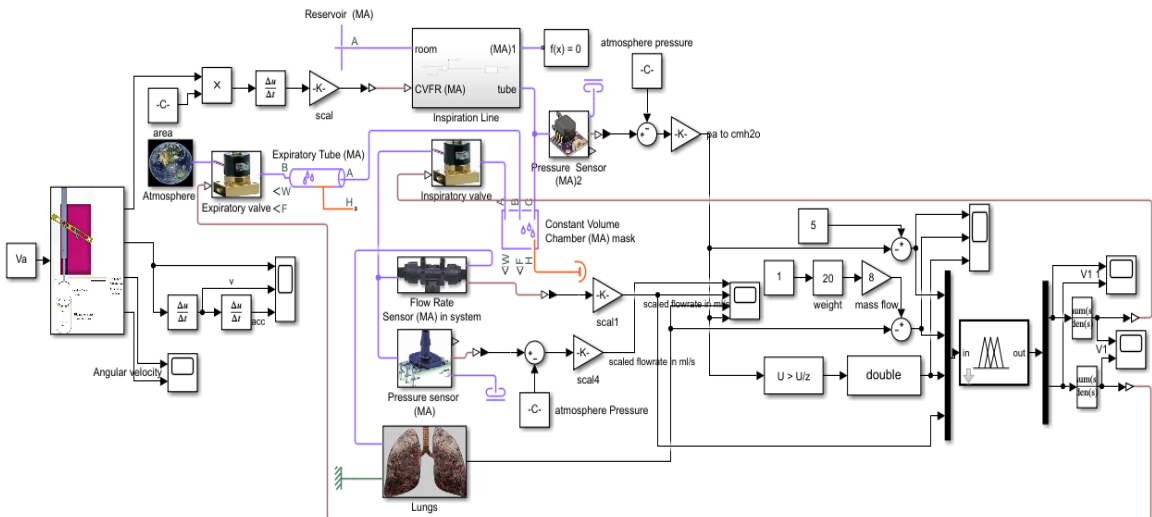


Figure 4.14: Simulink model with controller (closed loop).

Chapter5

PROTOTYPE IMPLEMENTATION

5.1 Prototype Accessories

MPXV7002DP differential pressure sensor

Transfer function for MPXV7002DP differential pressure sensor

$$V_{out} = V_s \times (0.2 \times p(kpa) + 0.5) \pm 6.25\%V_{FSS} \quad (5.1)$$

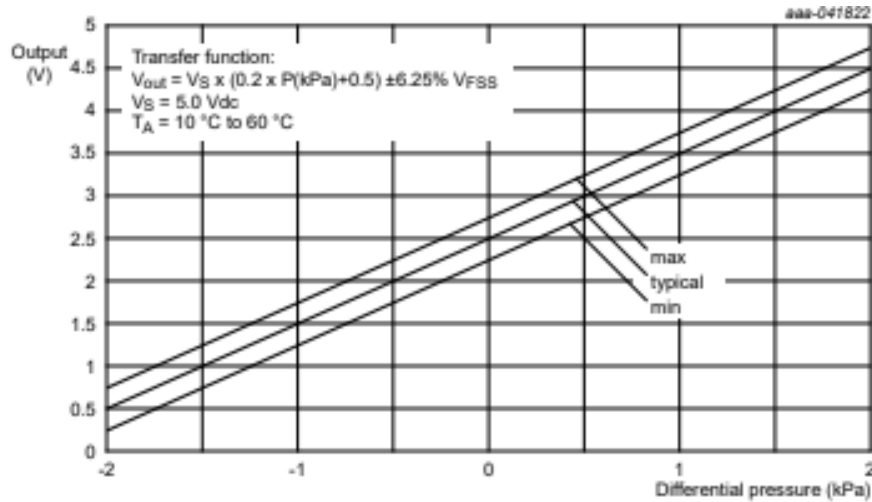
$$\begin{aligned} p &= 1000 \times \left(\frac{V_{out}}{V_s} - 0.5 \right) \div 0.2 \text{ Pa} \\ &= 10.197162 \times \left(\frac{V_{out}}{V_s} - 0.5 \right) \div 0.2 \text{ cmH}_2\text{O} \end{aligned} \quad (5.2)$$

Table 5.1: MPXV7002DP specification (operating characteristics at $V_s=5$ VDC, 25°C).

Characteristics	Minimum	Type	Maximum
Supply voltage VDC	4.75	5	5.25
Supply current mADC	-	-	10
Pressure range (kPa)	-2	-	2
Response time (msec)	-	1	-
Temperature range ($^\circ\text{C}$)	10	25	60
Pressure offset (Vdc) at $V_s=5$ V 10-60 $^\circ\text{C}$	2.25	2.5	2.75
Full scale output (Vdc) at $V_s=5$ V 10-60 $^\circ\text{C}$	4.25	4.5	4.75
Full scale span (Vdc) at $V_s=5$ V 10-60 $^\circ\text{C}$	3.5	4	4.5
Accuracy ($\%V_{FSS}$) 10-60 $^\circ\text{C}$	-	± 2.5	± 6.25
Sensitivity(V/kPa)	-	1	-
Warm up time (msec)	-	20	-



(a)



(b)

Figure 5.1: (a) MPXV7002DP differential pressure sensor and (b) Output vs differential pressure relation with its transfer function .

Measuring Gas Flow in a Mechanical Ventilator Using a Differential Pressure Sensor

By using conservation of energy of Bernoulli's principle the amount of energy that get in is equal to the energy leave out in a closed system.

$$E_{in} = E_{out} \quad (5.3)$$

Total energy in and out of the system is the sum of work done, potential energy and kinetic energy.

$$W_1 + PE_1 + KE_1 = W_2 + PE_2 + KE_2$$

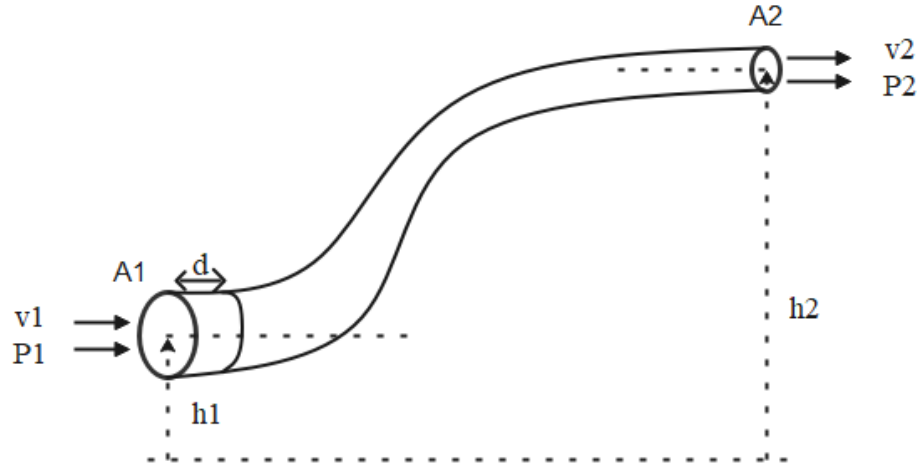


Figure 5.2: Variable pipe cross-section.

Work done:

$$W = Fd = \frac{F}{A}Avt = \rho Avt = \frac{Pm}{\rho}$$

Potential energy and Kinetic energy respectively:

$$PE = mgh$$

$$KE = \frac{1}{2}mv^2$$

$$\frac{P_1 m_1}{\rho_1} + m_1 g h_1 + \frac{1}{2} m_1 v_1^2 = \frac{P_2 m_2}{\rho_2} + m_2 g h_2 + \frac{1}{2} m_2 v_2^2$$

$$P_1 + \frac{1}{2} \rho v_1^2 + \rho g h_1 = P_2 + \frac{1}{2} \rho v_2^2 + \rho g h_2$$

For the case of $h_1 \approx h_2$, the above Bernoulli's equation can be simplified to:

$$P_1 + \frac{1}{2} \rho v_1^2 = P_2 + \frac{1}{2} \rho v_2^2$$

Differential pressure ΔP between the inlet and outlet point becomes:

$$\begin{aligned} \Delta P = P_1 - P_2 &= \frac{1}{2} \rho v_2^2 - \frac{1}{2} \rho v_1^2 \\ &= \frac{1}{2} \rho (v_2^2 - v_1^2) \end{aligned} \quad (5.4)$$

Where $v, h, A, t, d, g, P, m, \rho$ the velocity, height, area, time, displacement, gravity, pressure, mass and density and the subscript 1, 2 indicates inlet and outlet point respectively.

Continuity equation

When gas velocity is sufficiently less than speed of sound (343 m/s) the change in density of the gas is negligible and the gas is considered as in-compressible.

The equation of continuity for in-compressible fluid is:

$$A_1v_1 = A_2v_2 \quad (5.5)$$

Volumetric flow rate of the fluid can be calculated by combining continuity and Bernoulli's equation.

$$Q = \frac{\text{Volume}}{\text{time}} = \frac{A_1 \times \Delta x_1}{\Delta t} = A_1v_1$$

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

$$\Delta P = \frac{1}{2}\rho v_2^2 \left(1 - \left(\frac{A_2}{A_1}\right)^2\right)$$

$$Q = A_1v_1 = A_2v_2 = A_2 \sqrt{\frac{\Delta P \times 2}{\rho \left(1 - \left(\frac{A_2}{A_1}\right)^2\right)}} \quad (5.6)$$

ABPMANN005PGAA3 Gauge pressure sensor: High accuracy, compensated/Amplified basic board mount piezo-resistive silicon pressure sensors is used. The sensor offer a proportional analog output for sensing pressure over the specified temperature range and full scale pressure span. Output is proportional to the difference between applied pressure and atmospheric pressure. The sensor can operate in temperature range of -40 to 85 °C and Compensated temperature range from 0 to 50 °C. Calibrated output values are updated at 1 ms or 1 kHz . The part number ABPMANN005PGAA3 indicates the specification of the sensor.

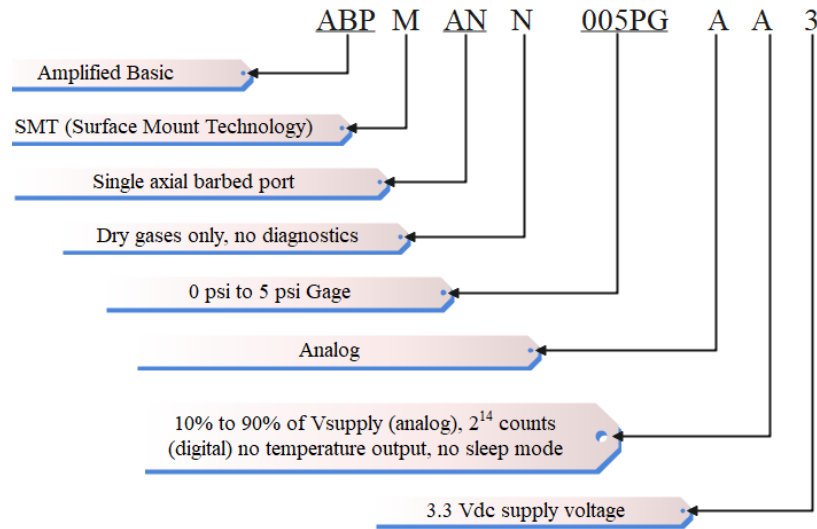


Figure 5.3: ABPMANN005PGAA3 sensor description.

Table 5.2: ABPMANN005PGAA3 detail specification.

Characteristics	Minimum	Type	Maximum
Excitation VDC (Vsupply)	3	3.3	3.6
Long term stability (1000 hr, 25°C)	-	-	±0.25
Response time (msec)	-	1	-
Accuracy	-	-	±0.25
Operating pressure (psi)	0	-	5
Operating temperature range (°C)	-40	-	85
Compensated temperature range (°C)	0	-	50

Relation between applied pressure and output voltage in compensated temperature range is shown in Fig.5.4.

The sensor transfer function that describes the relation ship between applied pressure and

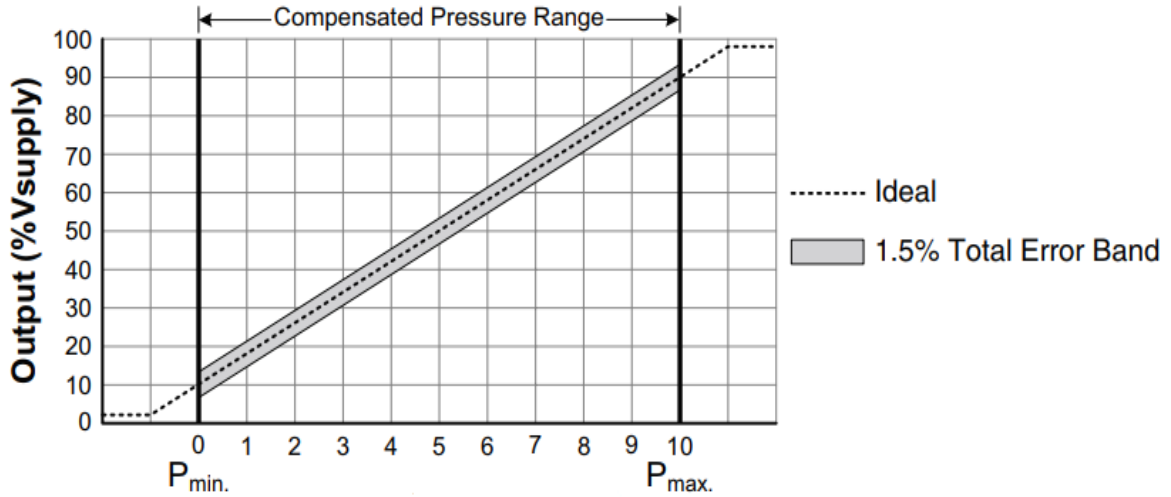


Figure 5.4: Analog version ABS output signal and pressure relation.

output voltage is shown in equ.5.7.

$$V_{out} = \frac{0.8 \times V_{supply}}{P_{max} - P_{min}} \times (P_{applied} - P_{min}) + 0.10 \times V_{supply} \quad (5.7)$$

Rearranging equ.5.7 the applied pressure will be:

$$P_{applied} = \frac{(V_{out} - 0.10 \times V_{supply})(P_{max} - P_{min})}{0.8 \times V_{supply}} + P_{min} \quad (5.8)$$

SMT AN: Single axial barbed port

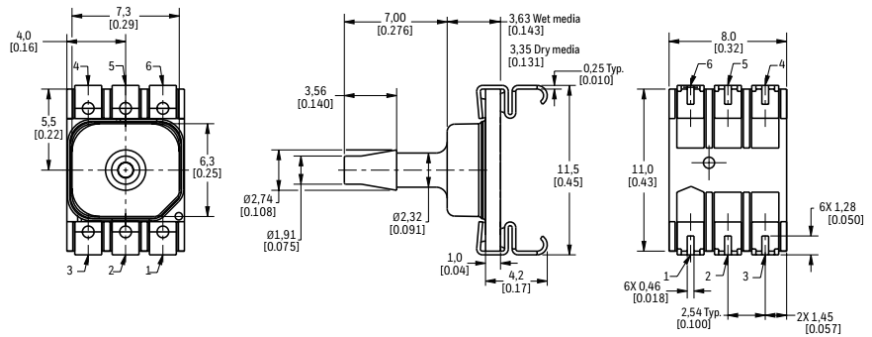


Figure 5.5: ABPMANN005PGAA3 dimensional drawing and pin number.

Table 5.3: ABPMANN005PGAA3 pin specification

Output type	Pin 1	Pin 2	Pin 3	Pin 4	Pin 5	Pin 6
Analog	GND	NC	V_{out}	NC	NC	V_{DD}

AWM5102VN Amplified Airflow Sensors AWM5102VN is 10 SLPM (Standard Liters Per Minute), N_2 calibration venturi type housing flow Sensors in AWM5000 Series Micro-bridge Mass Airflow Sensors. A 1 to 5 VDC linear output with 10 VDC supply over given flow range for a specific calibration gas. Output at laser trim point is 5 VDC at Full Scale Flow.

Table 5.4: AWM5102VN specification (performance characteristics at 10.0 ± 0.01 VDC, 25 °C).

Characteristics	Minimum	Type 2	Maximum
Excitation VDC	8	10 ± 0.01	15
Power consumption	-	-	100
Response time (msec)	-	-	60
Null output VDC	0.95	1	1.05
Null output shift -20 to 70 °C	-	± 0.05 VDC	± 0.200 VDC
Common mode pressure (psi)	-	-	50
Temperature range (°C)	-20	25	70

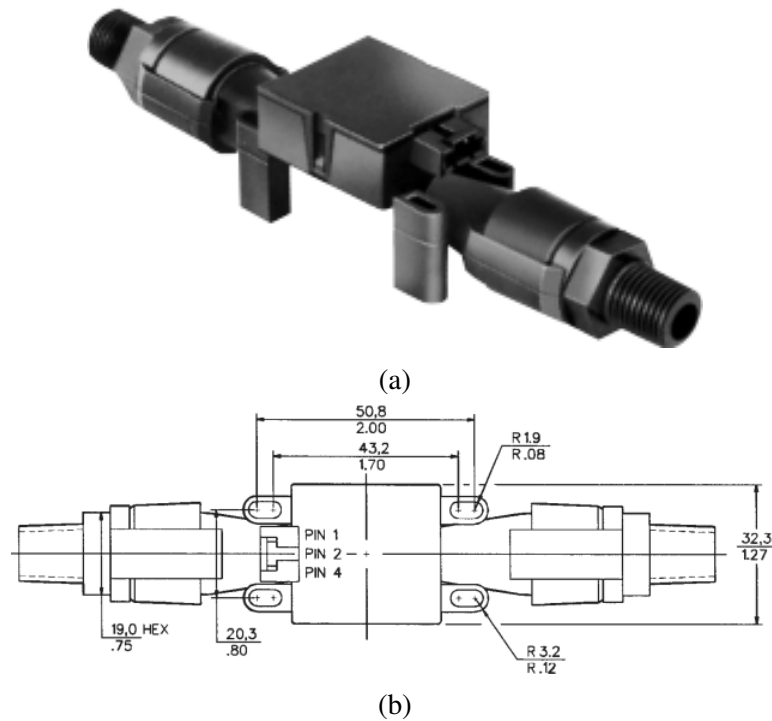


Figure 5.6: (a) 3D structure of AWM5102VN sensor and (b) detail structure of AWM5102VN mass air flow sensor from data-sheet.

The output connections of the sensor on Figure.5.6b are:

- Pin 1 + supply voltage
- Pin 2 ground connection
- Pin 4 output voltage connection

Flow direction is indicated by arrow on bottom of housing.

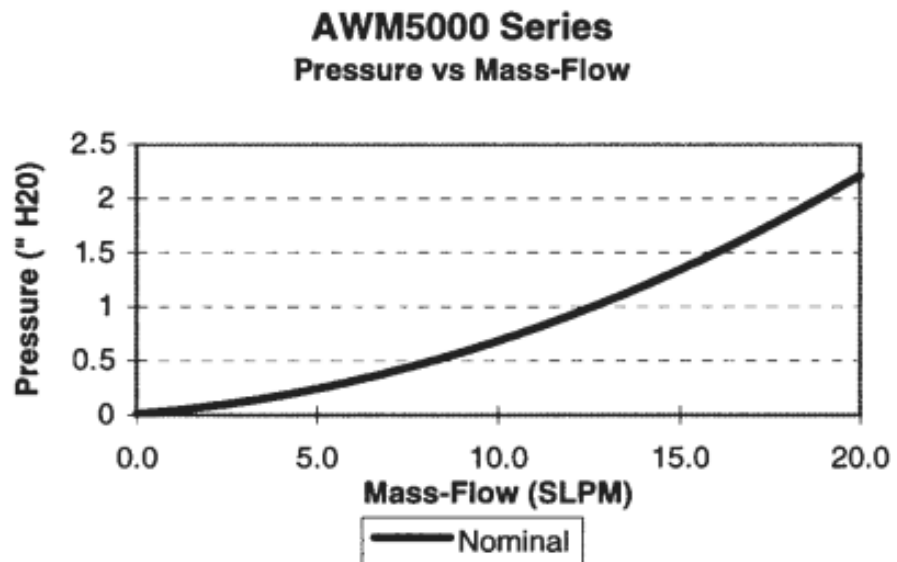
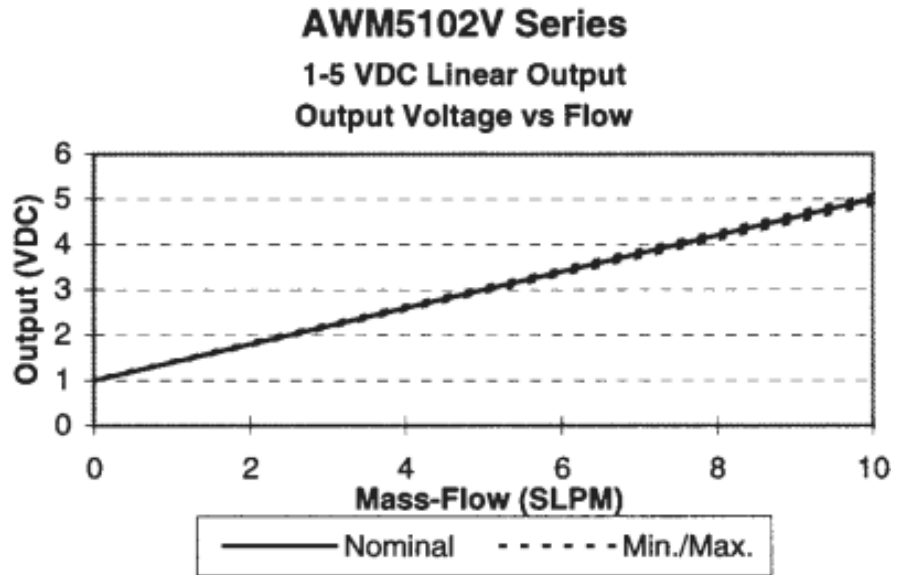


Figure 5.7: AWM5102VN (a) output (b) pressure curves (performance characteristics at 10.0 ± 0.01 VDC, 25°C).

L298N H-Bridge motor driver Board Specification

- Double H-Bridge Drive
- Operating voltage (logical Voltage): 5V (All logic pins)
- Drive Voltage: 5V-35V
- Logical Current: 0mA-36mA
- Drive Current: 2A (Max Single Bridge)
- Max Power: 25W

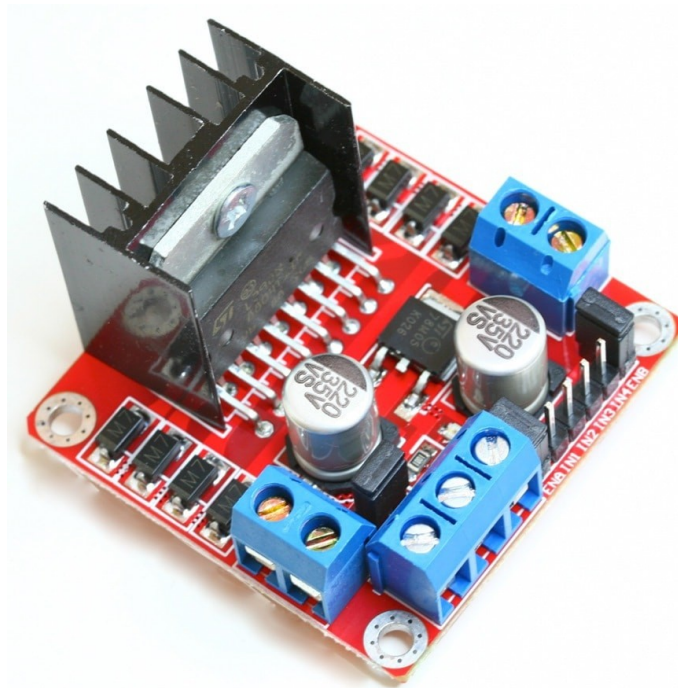


Figure 5.8: L298N H-Bridge.

Arduino Mega2560

- Operating voltage: 5V
- Input voltage (Recommended): 7-12V
- Input voltage (Limit): 6-20V
- Digital I/O PINS: 54 (of which 15 provide PWM output)

- Analog input PINS: 16
- DC current per I/O PIN: 20 mA
- DC current for 3.3V PIN: 50 mA

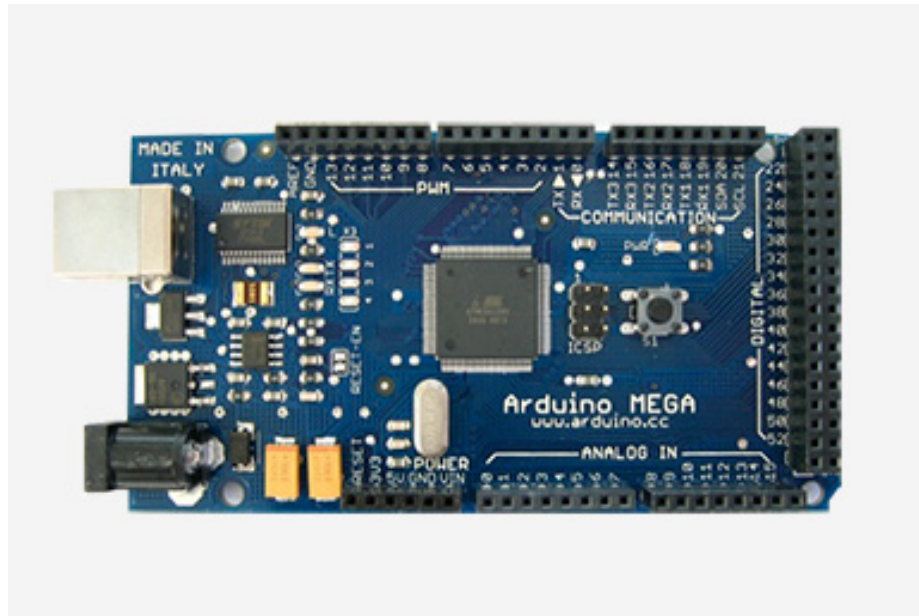


Figure 5.9: Arduino Mega controller board.

Relay

- One coil end used to trigger(On/Off) the relay, normally one end is connected to 5V and the other end to ground.
- Common (C): common is connected to one end of the Load that is to be controlled.
- Normally Close (NC): the other end of the load is either connected to NO or NC. If connected to normally closed the load remains connected before trigger.
- Normally Open (NO): the other end of the load is either connected to NO or NC. If connected to normally open the load remains disconnected before trigger.
- Maximum AC load current of 10A at 250/125V AC.
- Maximum DC load current of 10A at 30/28V DC.
- 5-pin compact configuration with plastic molding.

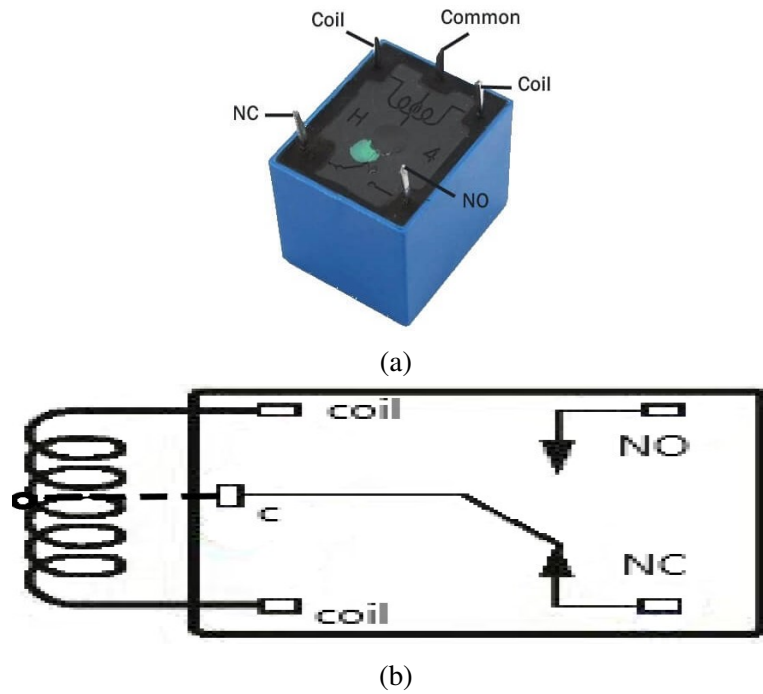


Figure 5.10: (a) Single Pole Double Throw SPDT relay and (b) relay circuit connection.

Solenoid valve

Electric Solenoid Valve magnetic DC NC (normally closed) air (water) inlet flow switch

- Nominal operating voltage: 12V
- Material: Metal + Plastic
- Rated Power: 5W
- Pressure: 0- 0.01Mpa
- Interface Modes: 1/2" Hose

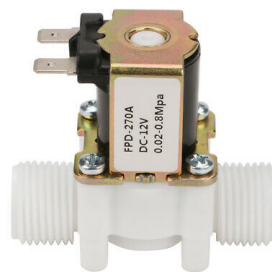


Figure 5.11: Solenoid valve.

5.2 Prototype Model

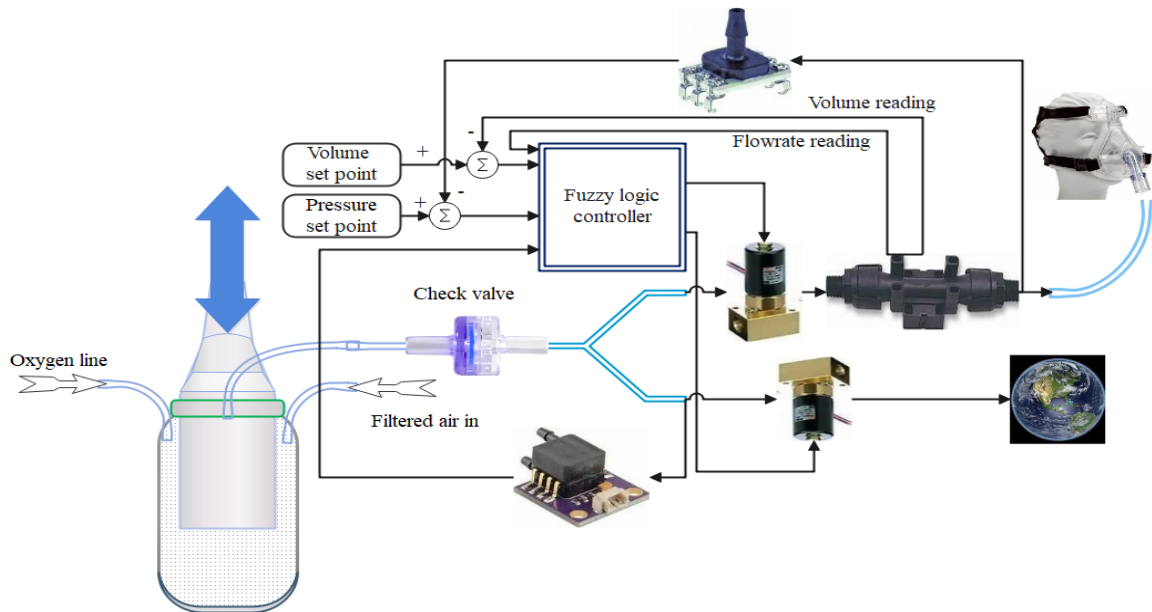


Figure 5.12: Practical prototype design with controller.

Steps to design and setup real time Arduino mega Simulink model for the prototype.

- Install Arduino Simulink Support Package
- Set up the Simulink support package for Arduino.
- Build the Simulink model for controlling ventilator using Arduino as in Figure 5.13.
- Interface physical parameters (sensors and actuators)
- Generate, download and run code on the Arduino.

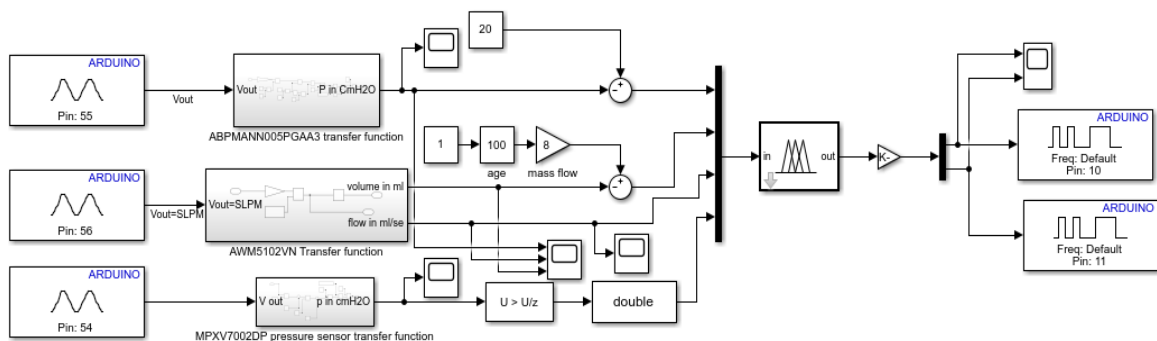


Figure 5.13: System Simulink model.

Detail Circuit Design The H-bridge is powered by a 12V Arduino power charger. High power-consuming devices like two valves (airway valve and expiratory valve) are directly supplied from a 12V 2A Arduino power supply cable and GND pin of all sensors and actuators are connected to ground pin pin of Arduino. Low power-consuming devices like sensors (AWM5102VN Amplified Airflow Sensors and MPXV7002DP differential pressure sensor) are supplied by the H-bridge 5V pin. The ABPMANN005PGAA3 airway pressure sensor is supplied by a 3.3V Arduino mega pin. One pin of the each valve actuator relay air way valve and expiratory valve is connected to OUT3 and OUT4 of the H-bridge respectively. Arduino Pin 10 and 11 are connected to IN3 and IN4 pins of the H-bridge. The enable pin of the H-bridge is connected to 5V to enable OUT3 and OUT4. The signal output pins of differential, gauge and flow sensors are connected to pins A0, A1 and A2 of the controller separately as shown in system circuit design figure 5.14. Controller board can be powered either from pc or from ac source using Arduino power charger.

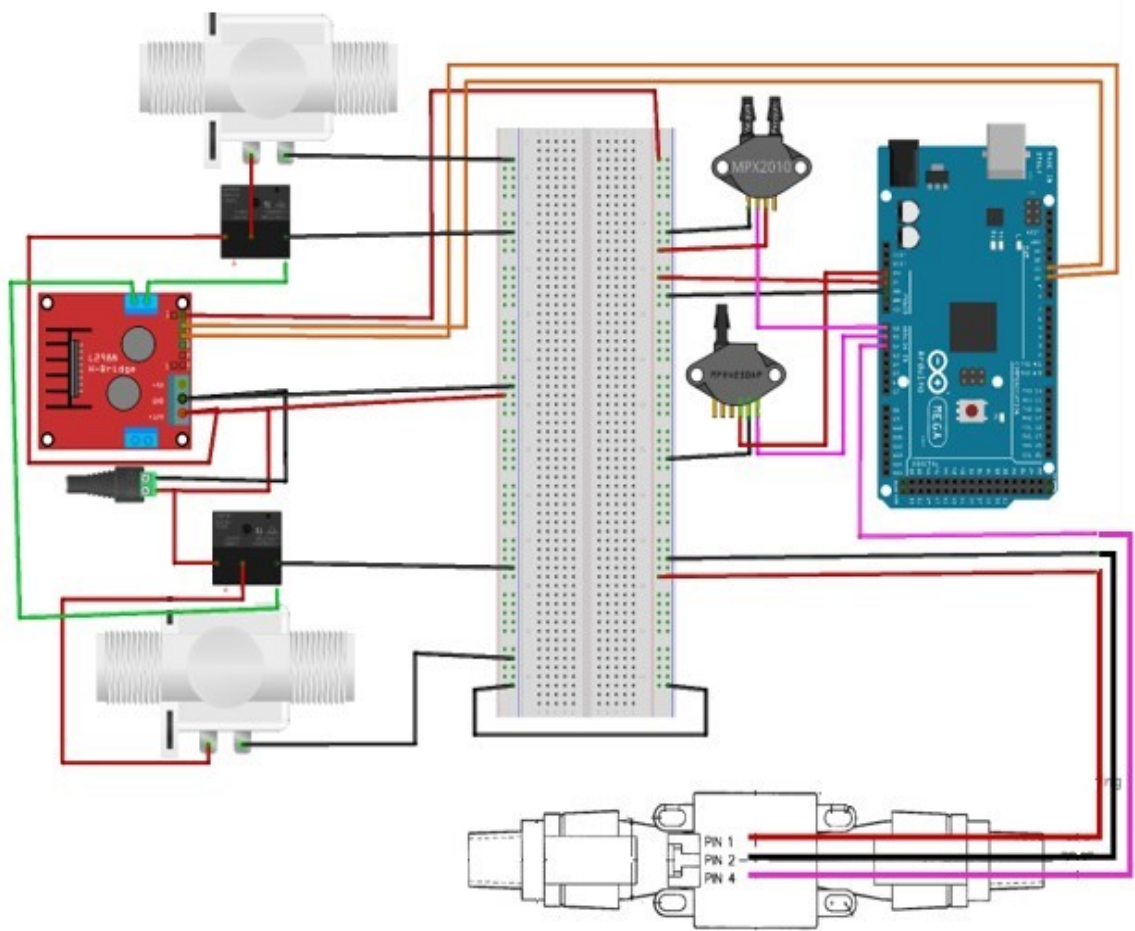


Figure 5.14: System circuit design.

Principle of Operation As the actuator motor rotates at a constant set angular speed, the slider mechanism will move up and down periodically. This will deliver fixed volume of air to the lung simulator through the inspiration pneumatic line, all the sensors start to give the physically measured data to the controller (Arduino mega). As the signal from inspiration line pressure sensor increases, the normally closed airway solenoid valve will be actuated. This will allow air into the lung simulator (inspiration period). Inspiration will continue until airway pressure is equal to or slightly greater than the set pressure (20 cmH₂O) or tidal volume is equal to or slightly greater than the reference value set by the operator. When the airway pressure or tidal volume reading value is greater than the set point, the controller sends a signal to make the airway valve close and the normally closed expiratory valve open. This will let extra volume and pressure of air into the atmosphere. A decreased inspiration line pressure sensor data will allow to open the airway solenoid valve and keep the expiration line open (expiration cycle). Any sudden increase in flow rate above 80 ml/se will terminate the inspiration (set for safety). A visible change above the accepted range in flow rate is not happening in breathing.

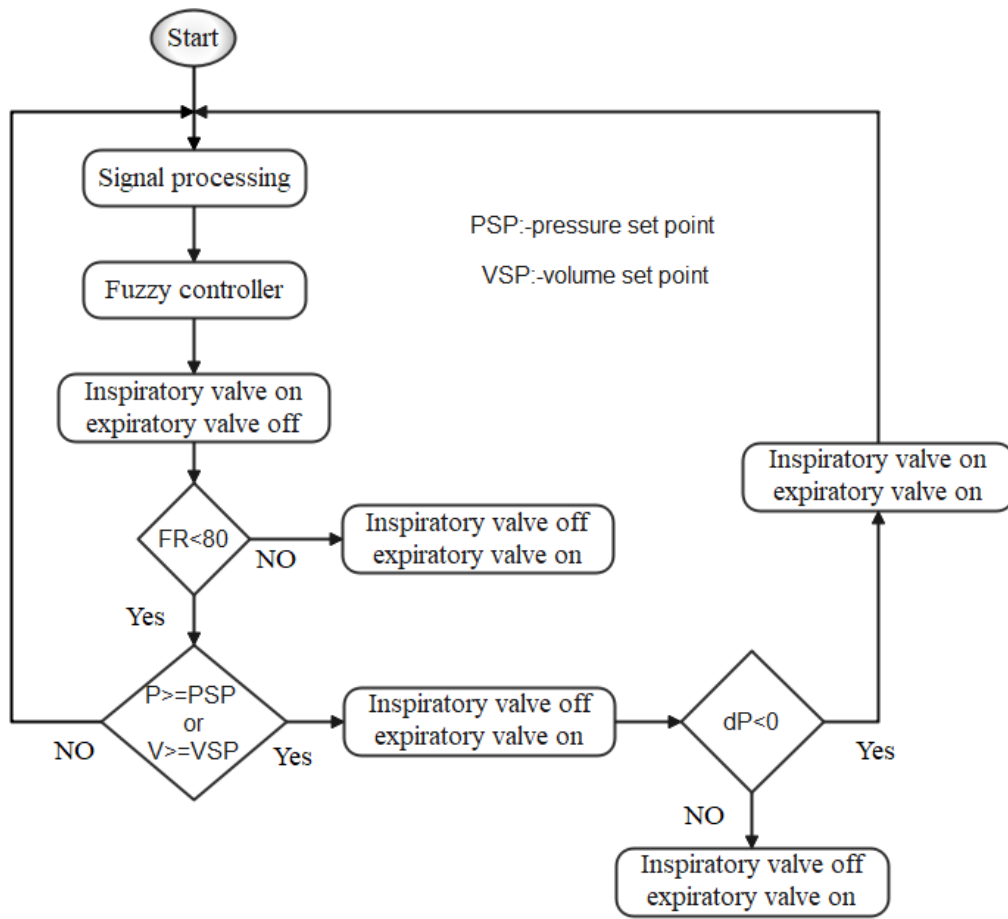


Figure 5.15: Flowchart describing the operating principle.

Chapter6

RESULT AND DISCUSSION

6.1 RESULT

6.1.1 Simulation Result Without Controller

Open loop simulation result shows the maximum delivering capacity of the designed mechanism. The designed system can deliver 24 cmH₂O and 600 ml of air at 32 ml/se at maximum capacity. Both lung (airway) and outside the lung pressure is same at open loop simulation. Figure 6.1 shows lung pressure, flow rate, volume and pressure outside of the lung sensor reading.

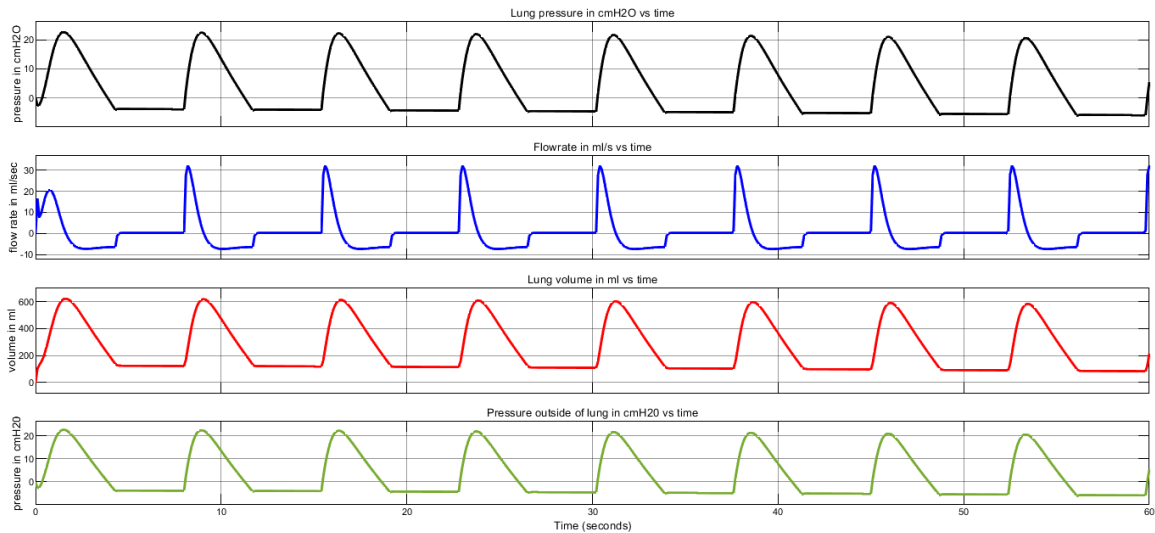


Figure 6.1: Simulation result without controller.

6.1.2 Simulation Result With Controller

Taking reference values 70 kg estimated body weight and 20 cmH₂O at constant input of 8.5 Figure 6.2 and constant input of 12 (increasing RR /rate of respiration) Figure 6.3 with equal simulation time 60 second. In closed loop simulation result lung pressure is maintained at 20 cmH₂O, and volume is 550 ml. Pressure outside of the lung is 22 cmH₂O which is above the lung pressure.

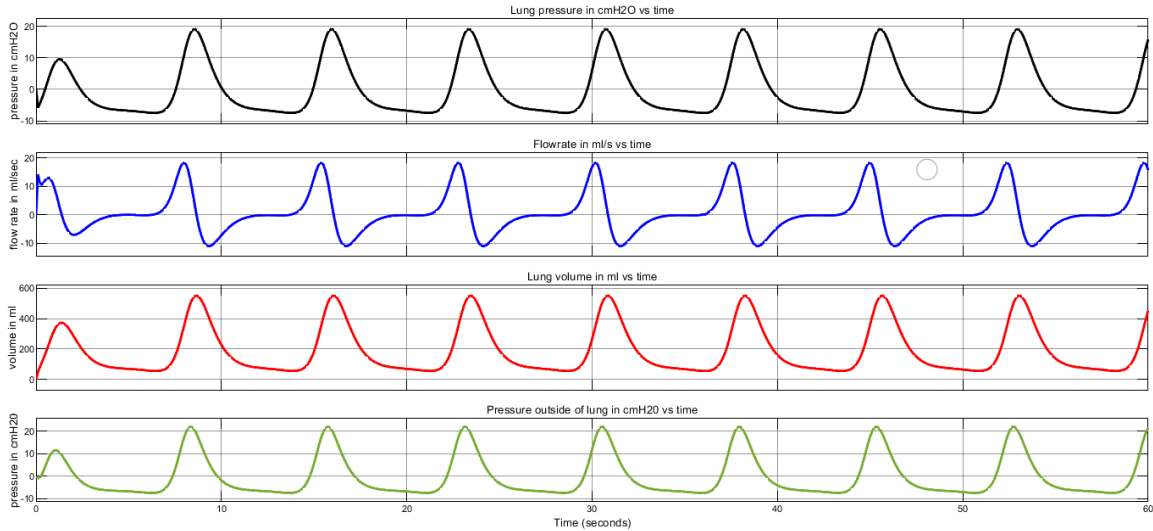


Figure 6.2: Closed loop simulation result at 70 kg and 20 cmH₂O set point.

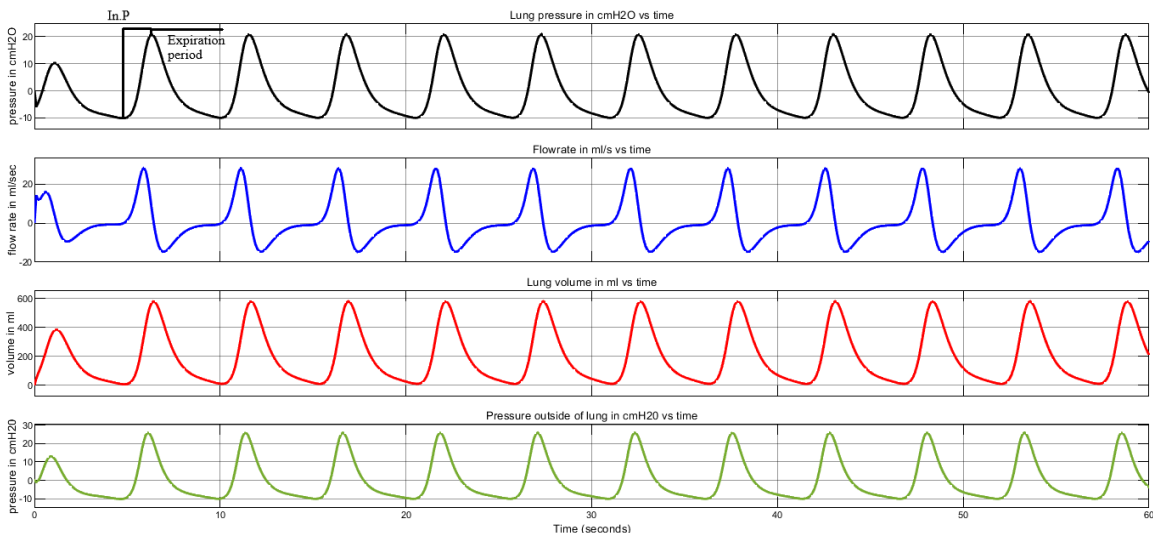


Figure 6.3: Closed loop simulation result at 70 kg and 20 cmH₂O set point.

Taking reference values of 30 kg estimated body weight and 15 cmH₂O at a constant input of 8.5. When the set point is the delivering capacity their will be a pause at the end of

inspiration until the pressure and volume is lower than the predefined value. Lung pressure plot shows peak and plateau pressure will also be created at the end of inspiration as plotted in Figure 6.4.

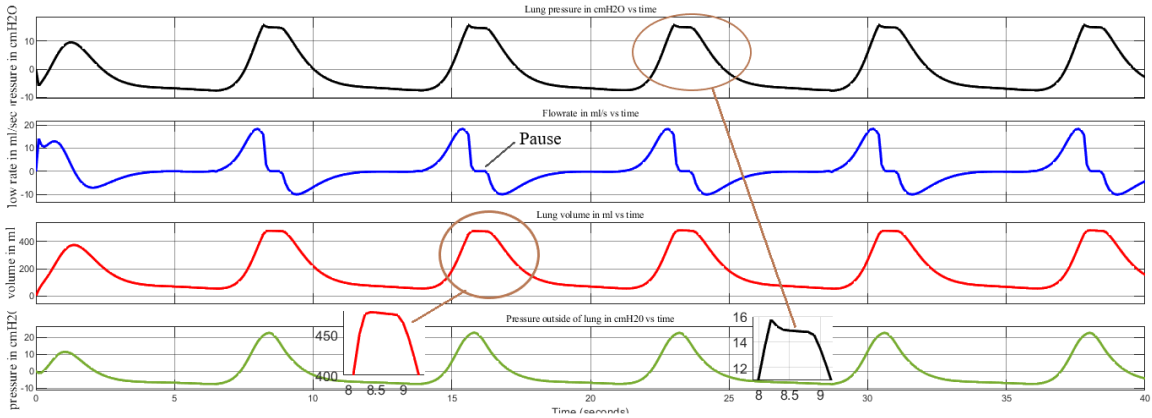


Figure 6.4: Closed loop at 30kg and 15 pressure set point.

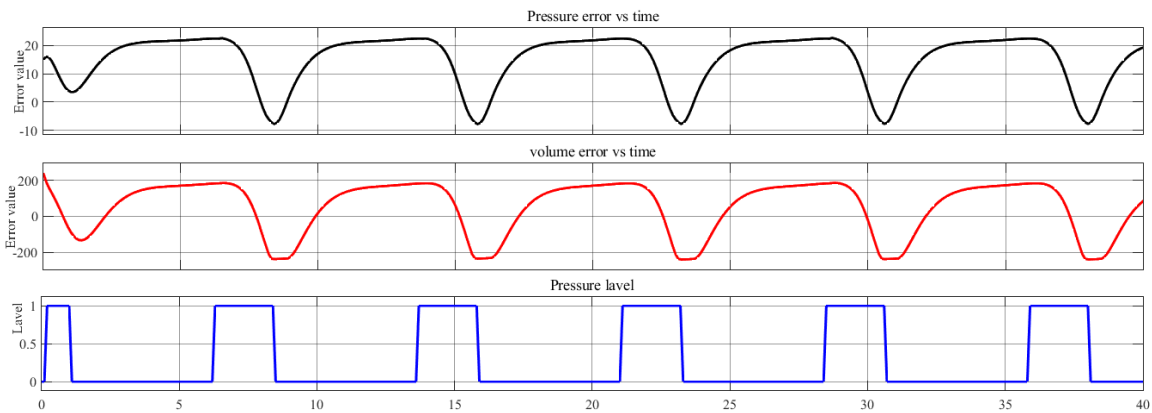


Figure 6.5: Controller input signal (error) for Figure 6.4.

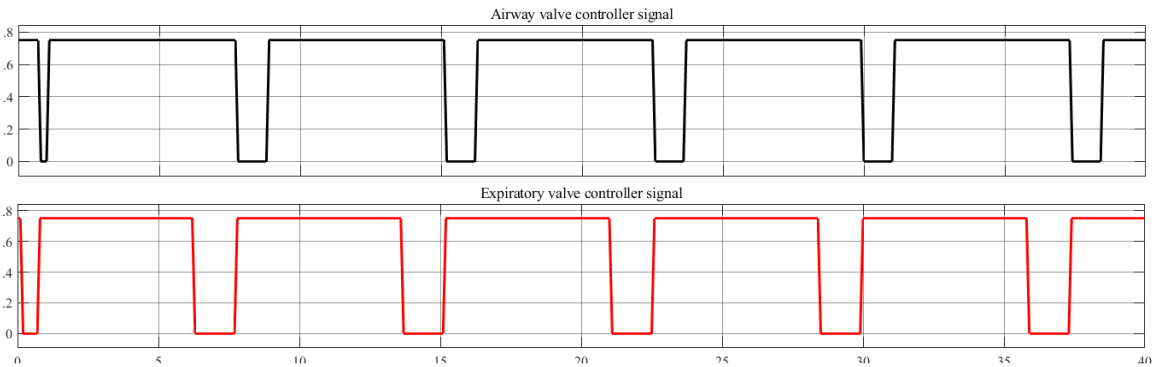


Figure 6.6: Controller output signal for Figure 6.4.

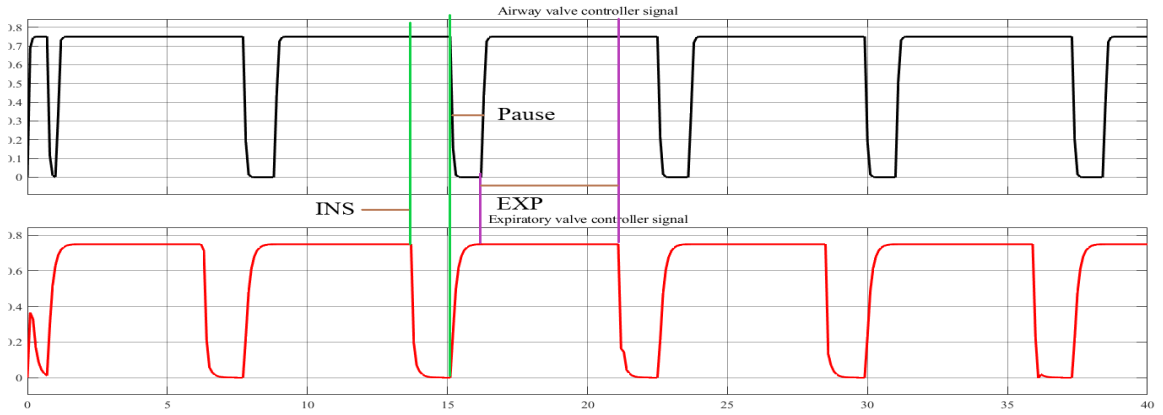


Figure 6.7: Valve response signal for Figure 6.4.

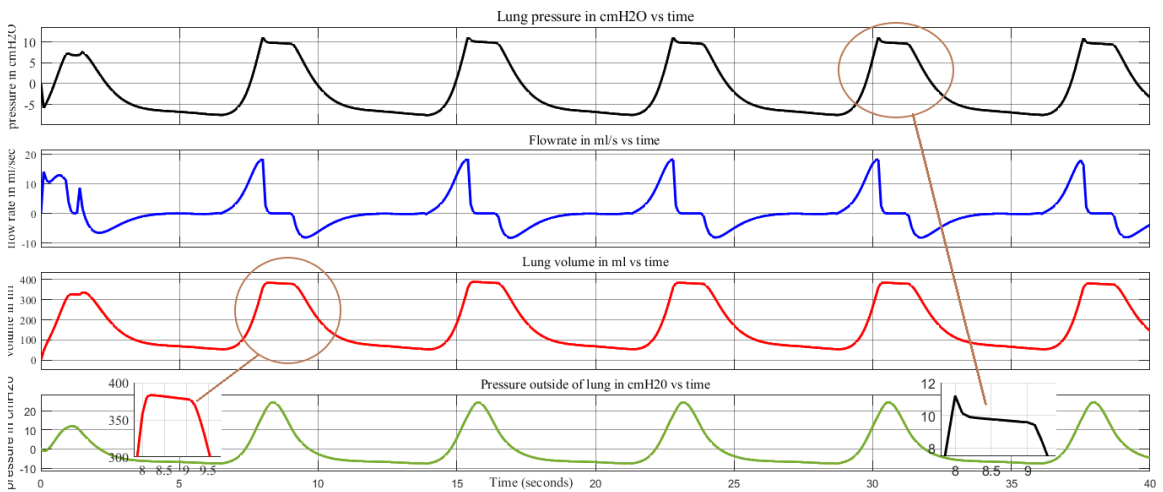


Figure 6.8: Closed loop at 20kg and 10 pressure set point.

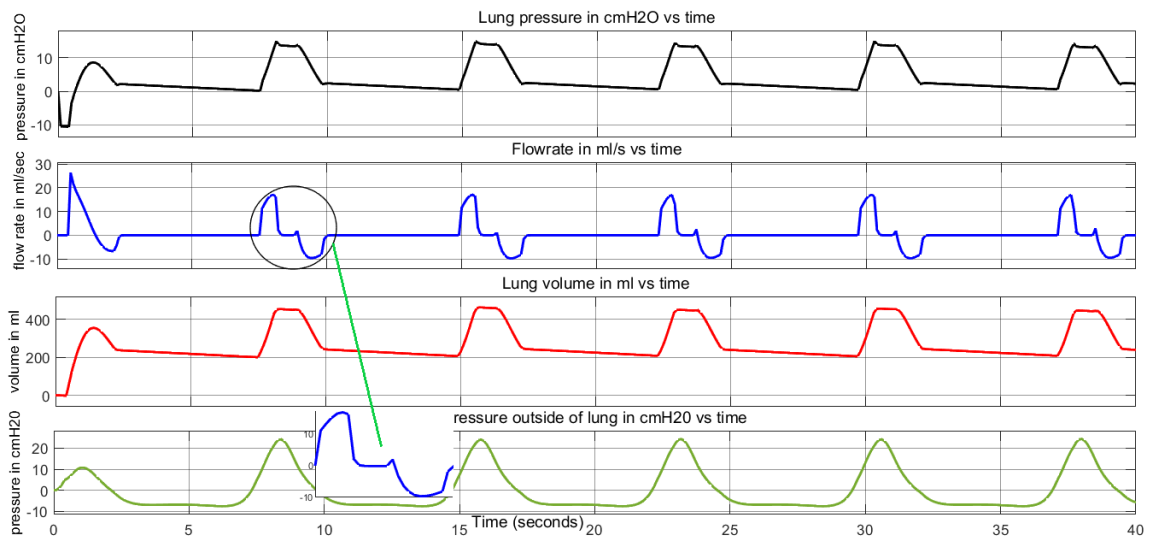


Figure 6.9: Closed loop at 30kg and 15 pressure set point with PEEP.

6.1.3 Prototype Result

A inactive lung simulator was linked to the ventilator air exchange lines to evaluate the functioning of the controller and the ventilator system.

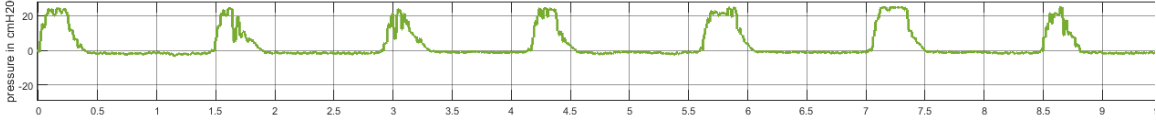


Figure 6.10: Uncontrolled pressure reading experimental result.

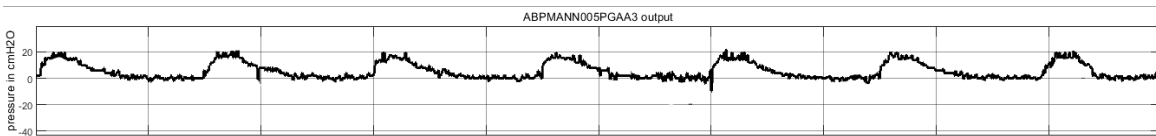
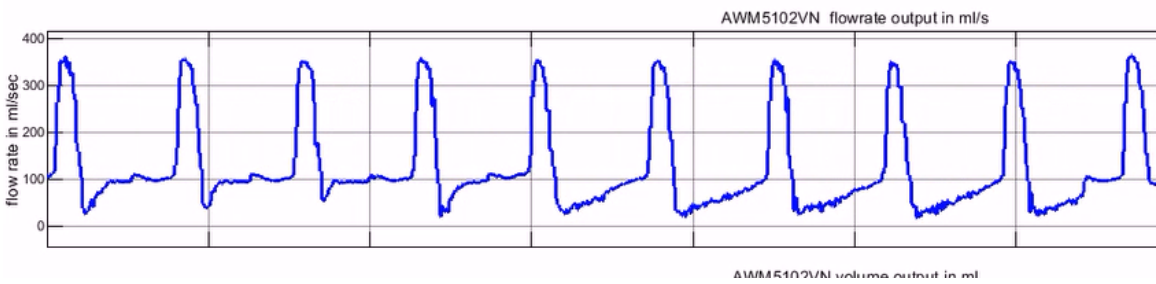
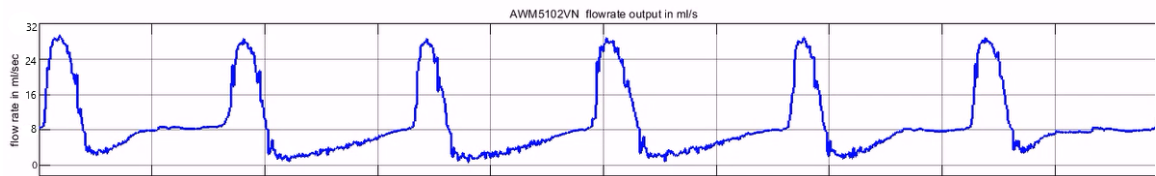


Figure 6.11: Controlled Pressure experimental result.



(a)



(b)

Figure 6.12: Flow rate experimental result for (a) un scaled or direct sensor output (b) scaled sensor output in ml/sec.

6.2 Discussion

The system model's open loop and closed loop simulations are depicted in sections 6.1.1 and 6.1.2 respectively. The closed loop response shows the mechanism delivering capacity, and the closed loop result shows the controlled pressure and volume output of the modeled

system. From open loop results, lung pressure (lung pressure in cmH₂O vs time graph) and pressure outside the lung (pressure outside of the lung in cmH₂O vs time graph) are equal, and the ventilator's delivering capacity is 600 ml and 24 cmH₂O at a flow rate of 32 ml/s. From closed loop results, lung pressure is less than pressure outside of the lung depending on preset values, and volume is maintained at 550 ml as shown in Figure 6.2 and 6.3. Both the experimental and simulation output indicate the maximum capacity of the system is 24 cmH₂O and the flow rate is 32 ml/s. Depending on the patients, lung compliance and resistance volume and airway pressure may be different. Figure 6.10 shows uncontrolled pressure reading in the ventilator system. Figure 6.11 shows scaled experimental result of airway pressure from ABPMANN005PGAA3 pressure sensor after the installed patient airway controlled valve. To prevent the collapse of the lung, pressure in the airway must be above atmospheric pressure (to keep the alveoli on) so that PEEP must be maintained at the end of expiration. PEEP response is presented in Figure 6.9 Direct output (non scaled) from volume flow rate experimental result is shown in figure 6.12a. Standard flow rate in ml per second can be done by multiplying the sensor output signal with scaling factor ($V_o \times (1/204.6) \times (1000/60)$). The resulting maximum and zero value of the sensor reading in standard ml per second is 28.51 and 8.15 respectively. Scaled standard flow rate in ml/sec is shown in figure 6.12b.

Chapter7

CONCLUSION, SUMMARY, RECOMMENDATION AND FUTURE RESEARCH

7.1 Conclusion

A prototype and system simulation model of a fuzzy-based pressure and volume-controlled (hybrid) ventilator device that is appropriate for mechanical ventilation have been developed in this study. The designed mechanism can supply 600 ml of air at a low flow rate of 28 ml/se. Simulation results show the ventilator can be used for different patients at different weights and age (volume and pressure requirement). The volume and pressure required for the patient is set by the clinician or the operator by considering the estimated body weight in kg/ml requirement. The flow rate and pressure reading from test and simulation results are the same. The flow rate and pressure reading from the test show an achievement of the problem statement and objective of the study.

7.2 Summary

Respiratory diseases such as pneumonia, ARDS, tuberculosis, acute asthma, lung cancer, and covid-19 are the main causes of death, especially in the developing world, and require immediate treatment, although a mechanical ventilator is costly and easily unavailable machine. The aim of this thesis is to design and develop a prototype of a fuzzy-based pressure and volume controlled (hybrid) easily available, low-cost mechanical ventilator for pandemic emergencies and scarcity. Different literature has been reviewed to understand the anatomy and mechanics of the lung, and control strategies. DC motor at constant angular speed is used to actuate the mechanism and deliver a high quantity of air at a low flow rate. Sensors, actuators, linear test lung simulators are connected to the air supplying mechanism. The result is tested by directly deploying the designed fuzzy controller and Simulink

model on Arduino mega. Finally, we got an acceptable pressure, volume, and flow rate controlled result from the prototype test.

7.3 Recommendation

In this study, a new combination of mechanisms and air pumps for CPAP mechanical ventilator simulation model and prototype is done. The system is tested using a breathing LTL (linear compliance) as a lung simulator, and this can be used only to simulate passive patients. However, the system is not tested with a lung simulator that has standard resistance and compliance.

7.4 Future Research

This thesis work has successfully achieved the objective of the study. System analysis is done at 8 RPM (8 breaths/minute) motor speed. Though, due to the application of the system, it might require a further control strategies for the breathing rate (rate of respiration per minute) to make it applicable for patients at different respiratory rate.

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