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Title:

Design and Simulation of a Mechanical Respirator

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Abstract

The aim of this project is to design and simulate a mechanical ventilator, which is used to maintain, or improve the functional capabilities of persons with disabilities in breathing period.

This project presents a mathematical model of Pressure Controlled Ventilator (PCV) signal. This mathematical model represents the respiratory activities and some important controlled parameter during mechanical ventilation are considered. The mathematical model is expressed and modelled using periodic functions with inequalities to control the beginning of inspiration and expiration durations. The created mathematical model of PCV signal is combined with an existing multi compartmental model of respiratory system that is modified and developed in the internal parameters - compliances (C) to test the created mathematical model.

Moreover, in order to verify the mathematical model, a simulation using MATLAB and its SIMULINK library is done. The obtained simulation results as pressure, flow and volume agree well with the real and experimental curves, which confirms the accuracy of our model.

Dedication

We, Zeineb Boudjeba and Khaoula Khelifa, would like to dedicate our work to our beloved parents and siblings, our friends and supportive teachers, every person that helped and supported us, everyone who shared the journey with us.

Without all your help, support and patience, we wouldn't be where we are today.

Acknowledgements

We would like to thank and appreciate the great effort which was done by our supervisor ***Dr. Abderrahmane OUADI*** *and the Control/Computer Engineering departments at the Institute of Electrical and Electronic Engineering for providing us with the opportunity and the privilege to realize this project.* ***Dr. OUADI*** *expertise, valuable guidance and orientation have strongly contributed to our work.*

We would also like to thank all of the teachers and professors who have worked unconditionally to give us a high quality education during our studies.

Last but not least, special thanks to everyone who had an impact on this work.

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List of Abbreviations

ICU Intensive Care Unit

AC Alternating Current

DC Direct Current

MV Mechanical Ventilation

PEEP Positive End Expiratory Pressure

PP Plateau Pressure

PIP Peak Inspiratory Pressure

MIP Maximal Inspiratory Pressure

BVM Bag Valve Mask

FiO2 Fraction of inspired Oxygen

BLDC Brushless DC

BLDCM Brushless DC Motor

HME Heat and Moisture Exchanger

MCU Microcontroller Unit

PWM Pulse Width Modulation

NI National Instruments

ADC Analog to Digital Converter

PCV Pressure Controlled Ventilation

Introduction

Recently, considerable attention has been attracted to mechanical ventilation due the increasing need of such devices during these hard times (Covid-19 pandemic).

The ventilator is a pneumatic and electronic system designed to monitor, assist, or control pulmonary ventilation, and respiration intermittently or continuously. It can also be used to control human body oxygen levels, for example during surgery where blood loss can result in hypoxia, or lack of sufficient oxygen in the patient´s body. Mechanical ventilation is designed to maintain an adequate exchange of gases, even through diminished breathing rates.

The respirator is basically made of a compressed air reservoir, air and oxygen supplies, a set of valves and tubes, and a disposable or reusable patient circuit. The air reservoir is pneumatically compressed several times a minute to deliver room-air or in most cases an air/oxygen mixture. The lungs elasticity allows releasing the overpressure, this is called passive exhalation, and the exhaled air is released usually through a one-way valve within the patient circuit.

The aim of this thesis is to derive and evaluate concepts for a mechanical ventilation device, to simulate basic human lung behavior and test different pulmonary therapies without connecting a lung to the device.

In the beginning of writing this report, numerous papers and books about mechanical ventilation have been read and studied to get a better understanding on how the respiratory system functions.

This report is structured as follows: Chapter 1 has established a basic understanding of mechanical ventilator fundamentals. The chapter's final section additionally

covers the different control modes.
A system hardware description is then introduced in Chapter 2, which illustrates different hardware types of ventilators as well as system components and functions.
Based on the previous chapter, the hardware design is finally done in Chapter 3 with the full components functionalities and schematics.
After that, a simulation based on the developed mathematical model of Pressure Controlled Ventilation is built using MATLAB and SIMULINK in Chapter 4, simulation results are analyzed and compared. Lastly, the conclusion and recommendations for future work are presented.

Chapter 1

Mechanical Ventilator Fundamentals

1.1 History

Mechanical ventilators first appeared in the early 1800s, in the form of negative-pressure ventilation [1]. Positive-pressure machines began to be used around 1900, and the standard intensive care unit (ICU) ventilator of nowadays was not introduced until the 1940s.

The first mechanical apparatus providing Non Invasive Ventilation, a bag and mask manual ventilator, was introduced in 1780 by Chaussier. A more sophisticated blower with a mask was adopted in 1887 by Fell, and in 1911 Dräger's Pulmotor was first introduced.

The negative-pressure ventilator was the predominant device used to provide ventilatory support in the 19th century and the first half of the 20th century. There have been 4 distinct generations of ICU ventilators from the original 1940s until today, each having features different from those of the previous generation [1, 2].

Ventilators designed for positive-pressure invasive ventilation became available in the 1940s and 1950s. These first-generation ICU ventilators were not equipped with patient-triggered ventilation. The sophistication range of these ventilators, though, was very wide.

The second-generation of ICU ventilators differed from the first in a multitude of ways. Patient-triggered inspiration was the most defining characteristic of this ventilator generation. But still only volume ventilation was applicable.

Figure 1.1: Dräger Pulmotor 1911. [1]

The third-generation ICU ventilators were then introduced. Microprocessor control was its single most powerful feature. These ventilators were considerably more responsive to patient demand than any of the earlier generations of mechanical ventilators.

The fourth-Generation ICU Ventilators is the current generation of ICU ventilators, and are the most complex of any mechanical ventilators ever manufactured.

Figure 1.2: Current-generation intensive care unit ventilators. [1]

1.2 Natural Lung Mechanics

Lung Structure

The lungs are a pair of air-filled, spongy organs placed on either side of the chest (thorax) as illustrated in figure 1.3. The trachea (windpipe) conducts inhaled air through its tubular branches, called bronchi, into the lungs. Then the bronchi split into smaller and smaller branches (bronchioles), becoming microscopic at the last.

The bronchioles ultimately end up in clusters of microscopic air sacs called alveoli. Oxygen from the atmosphere is absorbed into the blood by the alveoli. Carbon dioxide, a metabolism waste substance passes from the blood to the alveoli, where it can be exhaled.

The lungs are covered by a thin tissue layer called the pleura. The same kind of thin tissue makes the inside of the chest cavity - also called pleura [3].

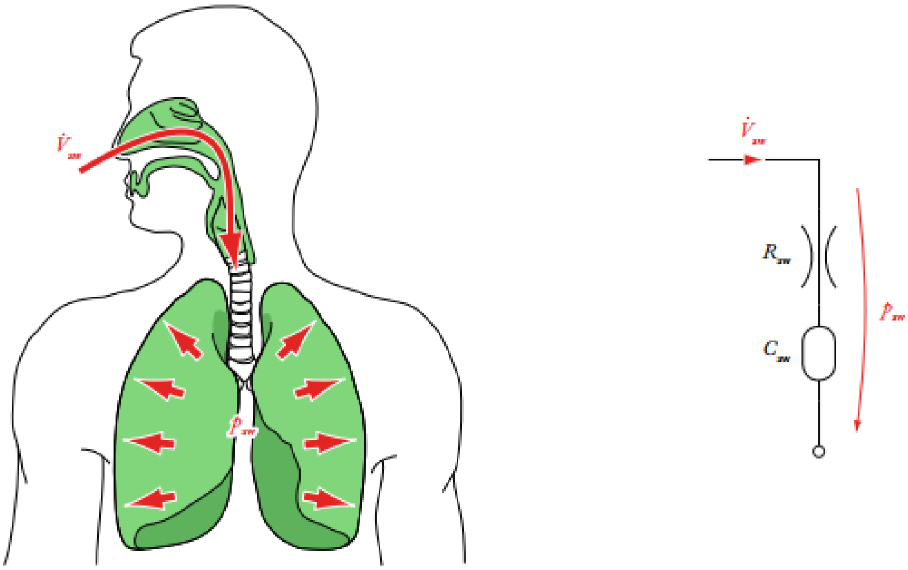


Figure 1.3: Basic lung structure and RC model. [4]

Natural pulmonary ventilation results in airflow between the atmosphere and lung alveoli of the lungs due to pressure variations produced by contraction and respiratory muscle relaxation.

Breathing is usually automatic and is subconsciously controlled at the base of the brain by the respiratory centre. Breathing continues during sleep and usually even when a person is unconscious. People can also control their breathing however they want. Sensory organs in the brain monitor the blood and sense oxygen and carbon dioxide levels. Increased carbon dioxide concentration is normally the strongest trigger for deeper and more intense breathing. On the other hand, when the carbon dioxide concentration in the blood is small, the brain reduces the frequency and depth of breaths. During breathing at rest, the average adult inhales and exhales about 15 times a minute [3].

Basically, when the pressure within the lungs is less than ambient pressure, inspiration happens as air travels into the lungs. Conversely, as air passes out of the lungs, expiration occurs when the pressure within the lungs is greater than the ambient pressure.

The lungs can be modeled with a simple RC model like shown in figure 1.3

1.3 Functional System Description of Mechanical Ventilators

As mentioned above, a normal healthy person can breath on their own. However, artificial respiration is necessary in abnormal circumstances such as surgery, or critical diseases, in which natural breathing fails. This is done by a mechanical ventilator.

A mechanical ventilator is simply a device designed to fully or partially assist a patient's own respiratory efforts in order to meet the respiratory needs of the body. The patient is connected to the ventilator with a hollow tube that goes in their mouth and down into their main airway (trachea) in the case of Invasive Ventilation, or simply via a face mask in Non-Invasive Ventilation and therefore eliminating the need of an endotracheal airway. They remain on the ventilator until they get well enough to not need it anymore [5, 6].

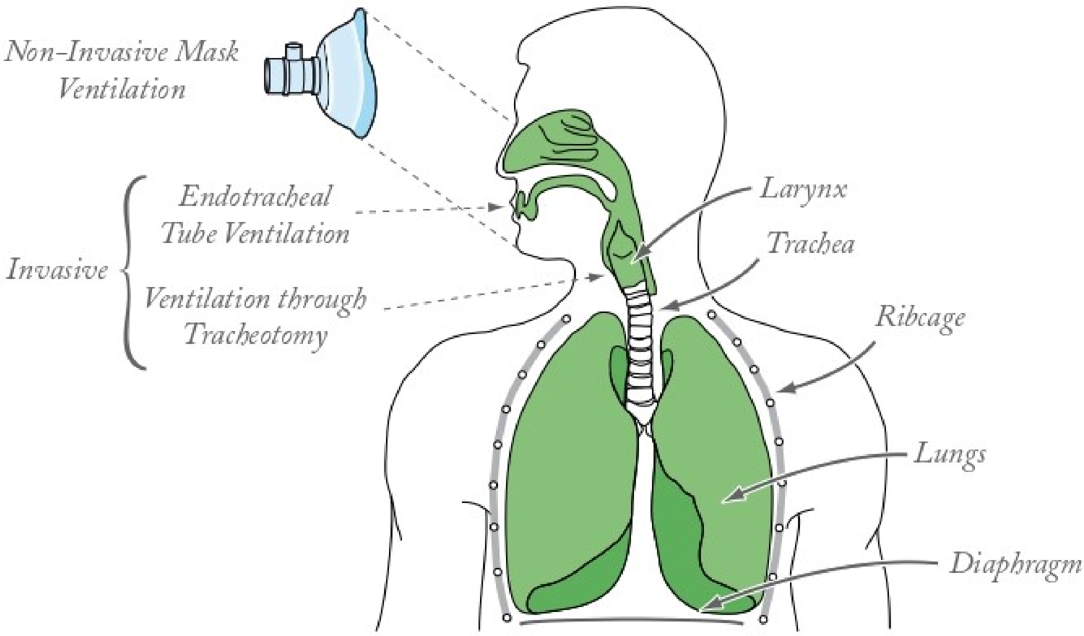


Figure 1.4: Invasive and Non-Invasive Ventilation. [4]

1.3.1 Mechanical Ventilation Parameters

- **Positive End Expiratory Pressure (PEEP)** is the residual lung pressure which maintains the airway open at the end of the expiration and is an important parameter in MV therapy [7].

- **Plateau Pressure (PP)** is the pressure applied to small airways and alveoli in end inspiratory pause to avoid volutrauma [7].

- **Peak Inspiratory Pressure (PIP or P_{peak})** is the highest pressure applied to the lung during inhalation, or it is the sum of plateau pressure (pressure used to sustain air in the lungs) and pressure used to overcome airway resistance (elastic lung and chest wall recoil, friction, etc.) [7].
  In other words:

$$P_{peak} = P_{plat} + P_{resistance}$$

  Consequently, P_{plat} can never be more than P_{peak}, since intrinsic resistance that must be overcome by $P_{resistance}$ will always exist (see figure 1.5).

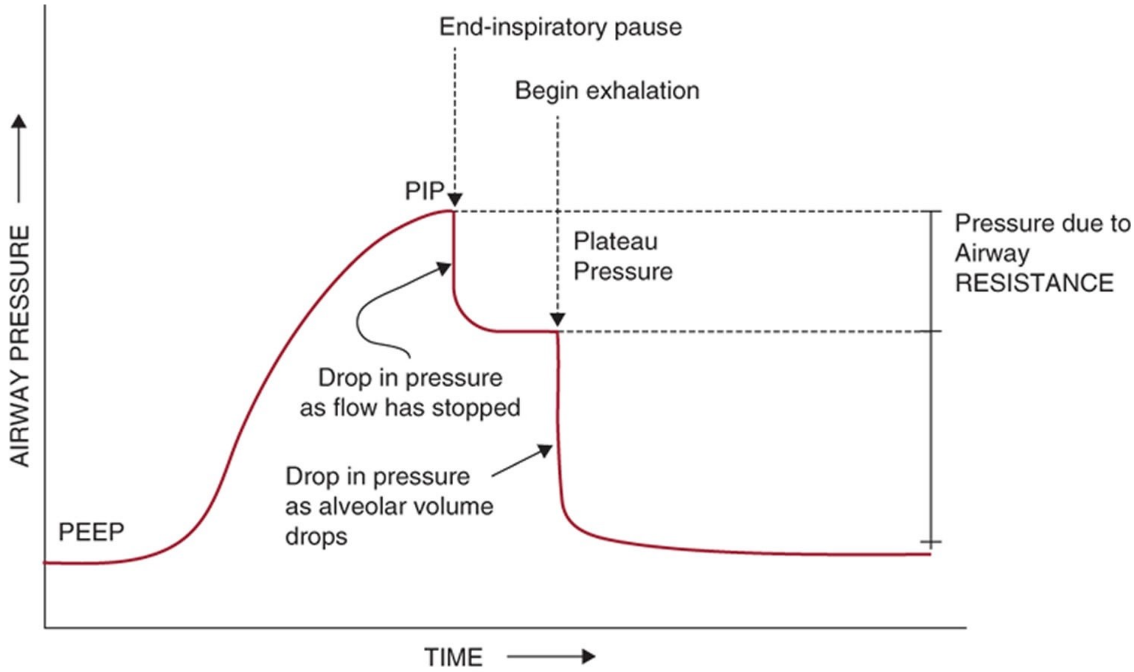


Figure 1.5: Peak inspiratory pressure and plateau pressure. [8]

- **Maximal Inspiratory Pressure (MIP)** is the maximal negative pressure generated during inhalation.

- **Tidal Volume (V_T)** is the net volume of air that is inhaled/exhaled with each normal breathing cycle and is shown schematically in figure 1.6 [7].

- **I:E Ratio** is the length of the inhalation time relative to exhalation. A 1:3 ratio, for example, means that the exhalation phase lasts three times longer than the inhalation phase. It usually ranges from 1:1 to 1:3 [7].

- **Flow Rate** is the speed at which the tidal volume is transmitted to the lungs [7].

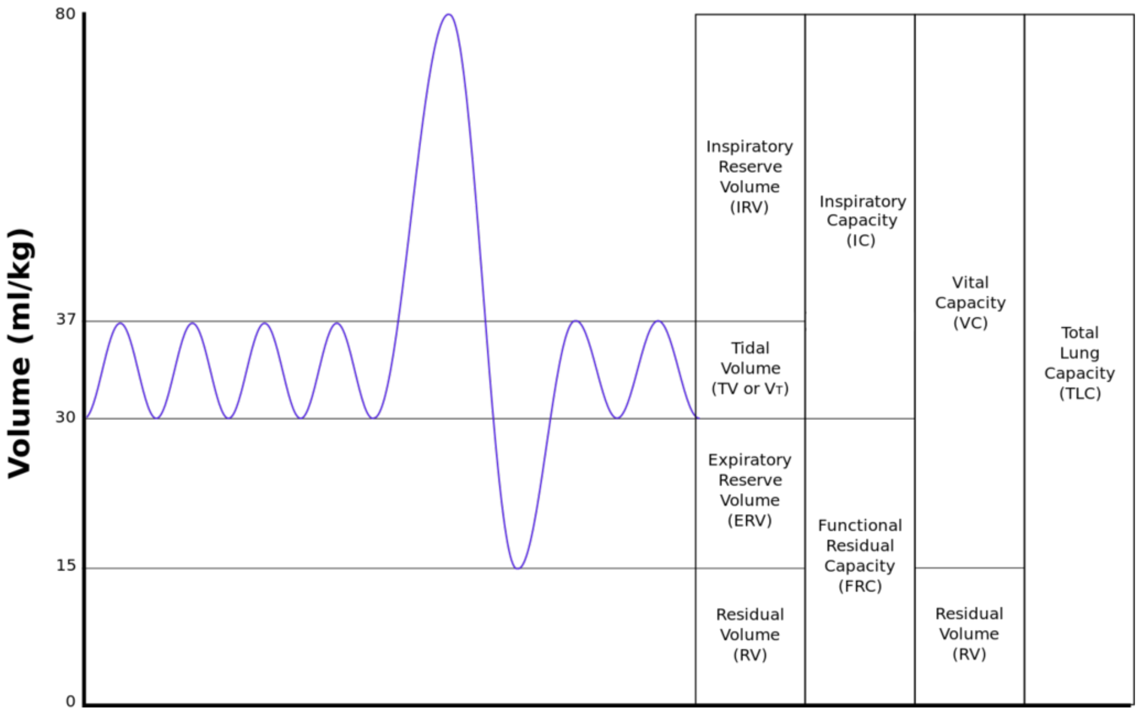


Figure 1.6: Lung volumes and capacities with associated terminology. [8]

1.4 Mechanical Ventilation Control Modes

1.4.1 Volume Modes

- **Volume Assist/Control (A/C) mode**
  Volume Assist/Control mode is one of the most common methods of mechanical ventilation in the intensive care unit. It's a volume-cycled mode of ventilation. It requires the PEEP and the Vt to be set and fixed directly by the clinician, which will then be supplied by the ventilator at a set of predefined time intervals or when the patient initiates a breath [9].

- **Volume Synchronized Intermittent Mandatory Ventilation (SIMV) mode**
  Synchronized Intermittent Mandatory Ventilation is a ventilator mode that offers partial mechanical support and is typically used to help wean patients out from the ventilator. This mode will provide a predetermined number of breaths at a fixed volume, but the volume can be defined by patient effort. One consideration when using SIMV is that it can result in increased work

of breathing. However, it can be balanced by adding pressure support to spontaneous breathing [10, 11].

1.4.2 Pressure Modes

- **Pressure Assist/Control (A/C) mode**
  Although volume control is the most common mode of ventilation, patients may be more likely to tolerate pressure control ventilation. Pressure Assist/Control mode requires the PEEP and the PIP (instead of tidal volume) to be set directly by the clinician. Thus, the change in volume during inspiration is a passive process [9].

- **Pressure Support Ventilation (PSV) mode**
  Pressure support ventilation is an assisted ventilatory mode that is patient-triggered. It is accessible with invasive or non-invasive mechanical ventilation. This ventilation mode is the most comfortable for patients and is convenient for weaning from invasive ventilation and for providing supportive care with non-invasive ventilation [10, 11].

1.5 Pressure vs Volume Loops

A pressure vs Volume Loop is a graphical representation of the relationship between pressure and volume during inspiration and expiration. Pressure is on the bottom axis and volume is on the side axis. This loop is generated every time there is a breath generated, either by the ventilator or by the patient [9].

A normal shape of a PV loop should be positioned in the centre as shown in figure 1.7. Notice that as the pressure increases initially (during inspiration) while there is not a great rise in volume. At the top of the curve we have the peak inspiratory pressure (PIP) and achievement of the tidal volume (V_T).

The curve then moves down as the exhalation phase takes place. Both pressure and volume fall back to the initial state.

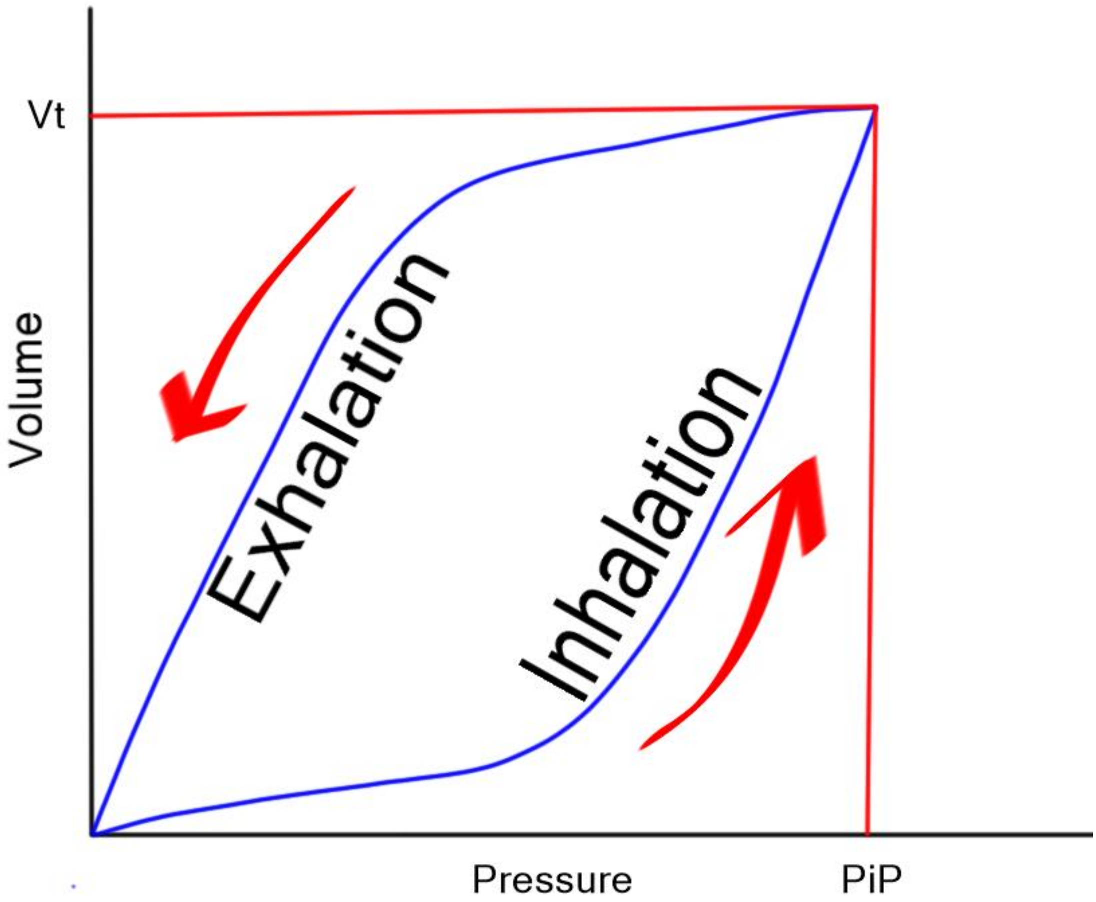


Figure 1.7: Pressure vs Volume loop. [12]

1.6 Types of Breath

There exists three types of breath: spontaneous, mandatory and assisted.

A **spontaneous** breath is entirely determined by the patient. **Mandatory** breath is determined solely by the mechanical ventilator. During **assisted** breath, the ventilator plays a role in the breathing process (e.g. Pressure Support Ventilation, Continuous Positive Airway Pressure).

In continuous spontaneous ventilation (CSV), all breaths are spontaneous and are initiated by the patient. In the case of intermittent mandatory ventilation (IMV), spontaneous breaths are permitted between mandatory breaths. A variant of IMV, synchronised-IMV (SIMV) allows for the mandatory breaths to be started by the patient [13].

Chapter 2

Mechanical Ventialor Hardware System

2.1 System Description

A ventilator is a control system where the output is a flow of oxygen-enriched air with certain specifications. A mechanical ventilator is essentially a machine that helps a patient breathe during surgery or in cases where they can not breathe on their own due to a serious illness. Mechanical ventilators are supplied with either electricity or compressed gas energy or both. The energy is transmitted or converted (by the ventilator drive mechanism) in a predetermined manner (by the control circuit) to improve or replace the patient's muscles in the breathing process (the desired output) [14].

Any mechanical ventilators design must include three basic functions, which are:

- A source of input energy to drive the device.
- A means of converting input energy into output energy in the form of pressure and flow to regulate the timing and size of breaths.
- A means of monitoring the output performance of the device and the condition of the patient.

Since ventilators deliver gas to the patient, they must have a pneumatic component. Current-generation ventilators have two components: (a) Pneumatic Subsystem, (b) Electronic System based on a digital processor [15].

2.1.1 Pneumatic System

The pneumatic system is responsible for supplying the gas mixture to the patient. Room air and 100% oxygen are supplied to the ventilator at 50 lb/in^2. The ventilator decreases this pressure and combines these gases for a prescribed FiO2 and flow into the ventilator circuit. The ventilator circuit not only provides the patient with gas, but also filters, warms and moisturizes the inspired gas.

The pneumatic system may be either a single circuit or a double circuit. For single-circuit ventilators, the gas that drives the ventilator is the same gas that is supplied to the patient. For double-circuit ventilators, the gas supplied to the patient is independent from the gas on which the pneumonic system operates.

2.1.2 Electronic System

The electrical power is used to control capacitors, solenoids, inspiratory and expiratory valves and the regulation of gas flow. These functions are usually controlled by a microprocessor, which is a single chip made up of integrated circuits.

The microprocessor is used to store pre-programmed ventilator modes as well as temporary data, such as pressure, flow measurements and airway resistance and compliance.

These two systems make up a combined-power ventilator that uses gas power to provide the driving force, or flow, to the patient. It uses an electrically powered microprocessor to monitor special valves that control the supply of gas for inspiration and expiration. This ventilator could not work without both a high-pressure gas and electrical power supply.

To sum up, **the majority of existing Intensive Care Unit (ICU) ventilators are classified as electrically powered, pneumatically driven, microprocessor controlled ventilators**, as shown in figure 2.1.

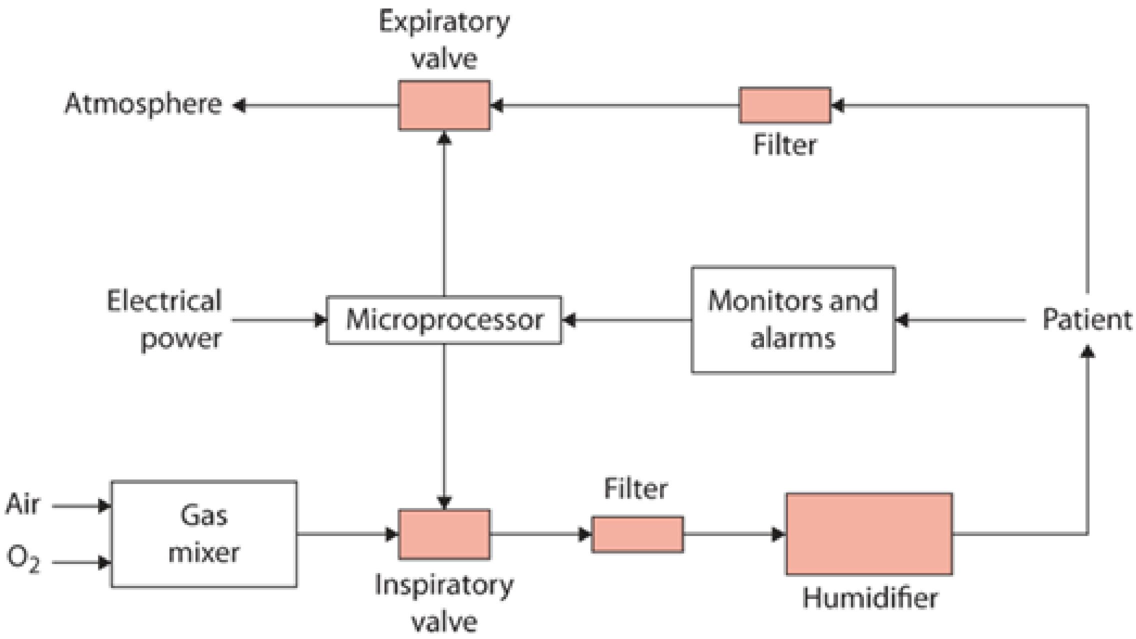


Figure 2.1: General block diagram of ventilator system. [16]

2.2 Different Hardware Types of Ventilators

2.2.1 Bag Valve Mask (BVM) Ventilation

Bag-Valve-Mask (BVM) ventilators are also known as manual and self-inflating resuscitators used in situations like hypoventilation when the rate of spontaneous breathing drops.

In BVM ventilation, a self-inflating bag is attached to a nonrebreathing valve and then to a face mask that sticks to the soft tissues of the face. The opposite end of the bag is attached to an oxygen source and usually a reservoir bag as shown in figure 2.2. The mask is manually held tightly against the face, and squeezing the bag ventilates the patient through the nose and mouth. Unless contraindicated, airway adjuncts are used during BVM ventilation to assist in creating a patent airway. Positive end expiratory pressure (PEEP) valves should be used if further assistance is needed for oxygenation [17].

Successful BVM ventilation requires technical competence and is used following these instructions:

- High flow oxygen is attached to the system and it is attached to a mask or tube.
- Appropriate mask size should cover the nose and mouth without gaps.

- Maintaining proper patient and airway position
- the bag is used to deliver oxygen to a spontaneously breathing patient or the bag compressed to manually ventilate them via a mask or tube, while avoiding over-ventilating and hyperventilating the patient.

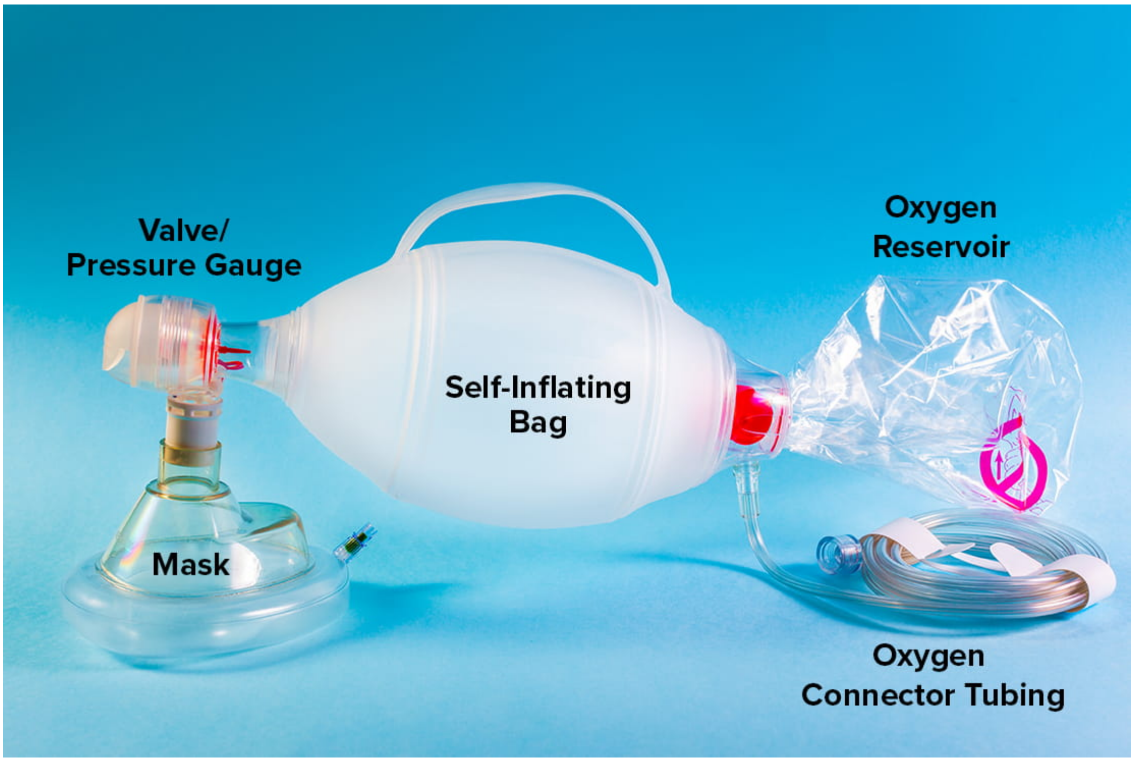


Figure 2.2: Bag Valve Mask ventilation.

2.2.2 Electro-Pneumatic System

Pneumatically powered ventilators use compressed gas as an energy source for their operation. Medical gases are anhydrous (without water), and oil-free at a pressure of 50 psi.

Ventilators may also be electrically powered, utilizing 120/220 Vac 50/60 Hz alternating current (AC) or 12 V direct current (DC) for a power source. The electrical power can be used to run electric motors to drive mechanical devices that generate gas flow.

Electro-pneumatic ventilators are powered by a combination of both pneumatic and electric power sources. They require both an electrical (for microprocessor-controlled systems) and pneumatic power source [15].

2.2.3 Blower Based System

When a patient is undergoing general anesthesia, they will most likely be ventilated by a bag valve mask. This bag valve mask is used during the time from when the patient loses consciousness and the ability to support their own airway to the time that they are intubated and placed on a mechanical ventilator [18].

However, the bag valve method is a difficult task to master and requires full attention of the person performing the ventilation. Compressing the bag too much endangers the patient and risks lung damage. Not enough pressure will result in lowered levels of gas exchange and if prolonged will lead to injury of the patient.

Electric blower based ventilators are a replacement device for the bag valve masks. This device will effectively provide constant positive airway pressure to the patient thus maintaining an open airway. It will give the operator the ability to deliver specific pressures and volumes of gas to their patient at regular intervals as desirable. It can also be portable and powered by a rechargeable battery pack contained within the casing [19].

The use of variable speed blowers or fans is a good method of controlling the gas flow rate. The speed of such blowers can be rapidly increased or decreased to impart a desired rate of flow. This allows greater flexibility in controlling each inspiration and exhalation. The rapid rate of change in the gas flow allows the ventilator to vary the rate of flow multiple times or even continuously within the time span of a single breath. Furthermore, a ventilator needs to make rapid and repeated adjustments to respond accordingly. A blower based ventilator is capable of such gas flow control.

Conventional flow control in a blower based ventilation system uses an electronic feedback controller to control the gas flow rate.

Figure 2.3 shows a schematic outline of the blower system.

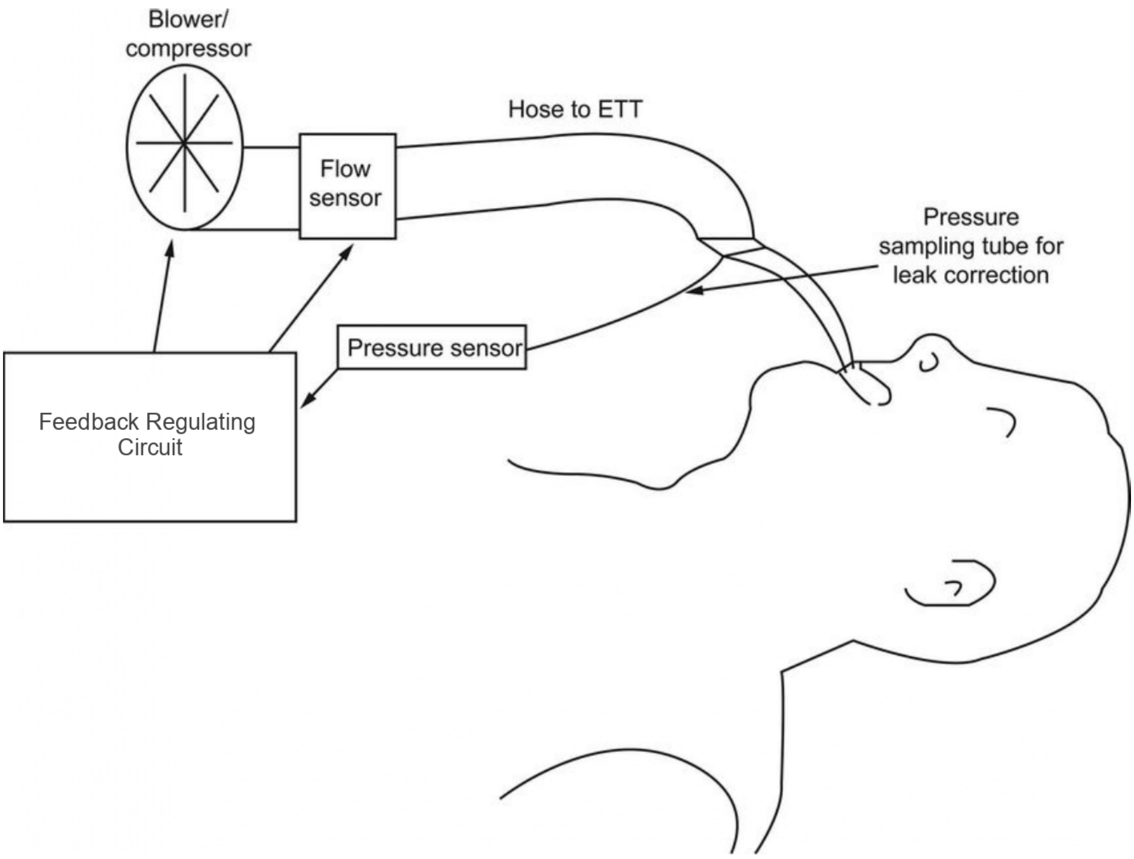


Figure 2.3: Diagram of a blower based system.

2.3 System Components and Functions

An electronics system for a medical respirator can be complex because of the variety of components and functions that must be accurate [20]. In general, the following components are usually seen in a modern mechanical ventilator:

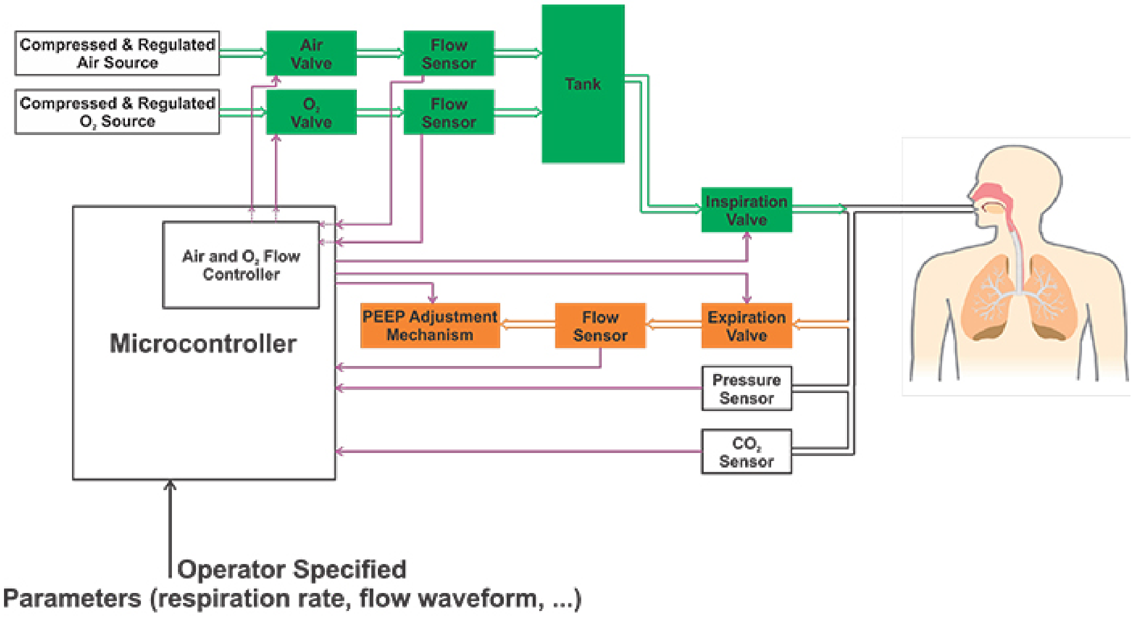


Figure 2.4: General diagram of system components. [21]

2.3.1 Electrical Supply

Almost all key functions of a modern ventilator are electrically powered, including valves, microprocessors, sensors, alarms, and displays. Electricity to a ventilator is comparable to blood in a living body. If the electrical supply stops, the ventilator system is immediately paralysed.

Most ventilators have two electricity sources: AC (alternating current) power is the primary one, and an integrated battery is the secondary one (figure 2.5); while the ventilator is connected to AC power, this battery charges. If AC power is unexpectedly interrupted, the running ventilator should switch immediately and automatically to battery power so that mechanical ventilation continues. Typically the ventilator also alarms to alert the operator to the abnormal condition. If AC power resumes, the ventilator switches back to AC power and recharges the battery [22].

The internal battery is, however, just a temporary electrical source with limited capacity. Should a ventilator switch to its battery, it is important that the AC supply be restored as quickly as possible, because mechanical ventilation discontinues if the battery is depleted.

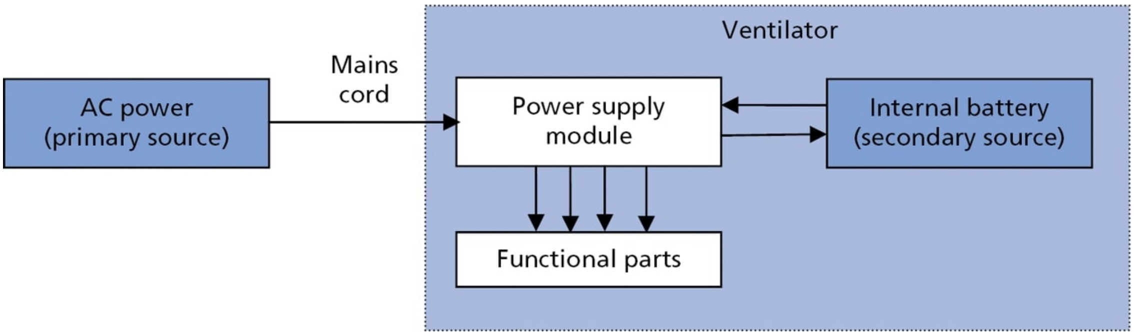


Figure 2.5: Simplified diagram of ventilator electrical supply. [23]

2.3.2 Gas Supply

Most ventilator systems found in the hospital setting require supplies of both **compressed air** and **compressed oxygen**. The gas supplies serve not only as the source of the fresh gas delivered to a ventilated patient, but also as the ultimate force to propel the gas throughout the gas passageway [23, 24].

- **Compressed oxygen supply**
  The oxygen concentration of inspiratory gas is defined as **FiO2** or **fraction of inspired oxygen**. FiO2 can be freely set or adjusted between 21% and 100% in most ventilators.
  If one gas supply fails, a ventilator typically continues mechanical ventilation with the remaining gas supply, and sets off an alarm to alert the clinician of the abnormal situation. In this case, the ventilated patient receives either pure air or pure oxygen only.
  The typical oxygen source is either the hospital's central oxygen supply delivered through a wall pipeline network, or oxygen cylinders. A newer oxygen source, for use in hospitals, is a special oxygen concentrator. It is still in its testing phase.

- **Compressed air supply**
  Compressed air is traditionally provided from an air compressor, air cylinders, or from a central supply delivered through a wall pipeline network. Some ventilators use internal turbines as the internal air supply. In this case, an external air supply is not required.

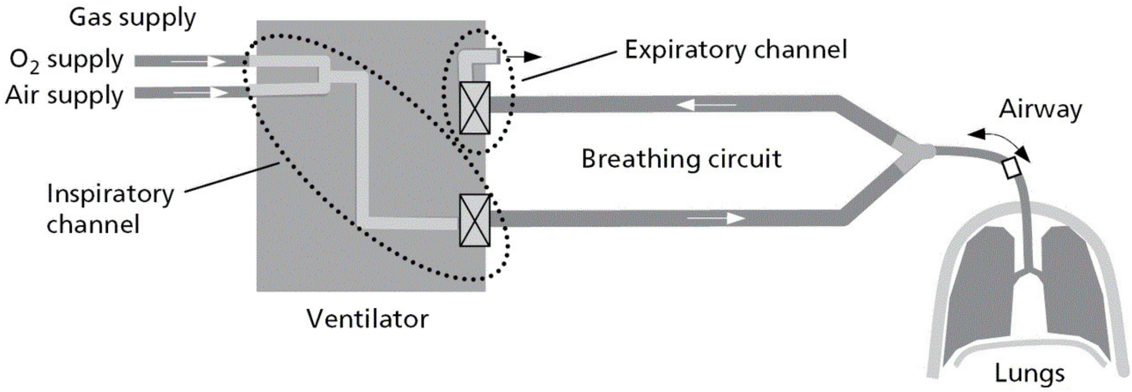


Figure 2.6: The gas passageway of a ventilator system. [23]

2.3.3 BLDC Based Blowers

BLDC motors are replacing brush motors in numerous applications as they offer significant energy efficiency improvements, lower acoustic noise and better reliability to name a few advantages. Especially in the field of mechanical ventilation, BLDC motors are much more widely used than ever.

Brushless DC motors are a suitable fit for medical technology applications. In optimized, continuous operation, the motors satisfy the requirements of ventilator applications by delivering high-performance bearing and cogging-free running with a linear speed and torque curve [19]. The brushless motors with an integrated speed controller operate to provide precise speed control for any desired air flow rate. Thanks to their robust construction, they are also suitable for applications actuating high loads [25].

2.3.4 Pressure Measurement

The use of pressure sensors in ventilators and respirators is similar to their use in CPAP and BiPAP. Essentially, when the lungs need assistance taking in air, a machine blows the required amount of air into the lungs. Pressure sensors play a critical role when it comes to monitoring and controlling the required amount of air, or positive airway pressure [26].

Assessment of respiratory pressure-volume curves permits analysis of the static mechanical properties of the respiratory system. In critically ill patients, measurement of pressure-volume curves has been suggested as a method for assessing the severity of lung injury and for monitoring the evolution of lung disease. It can also guide the ventilatory adjustments to optimize mechanical ventilation.

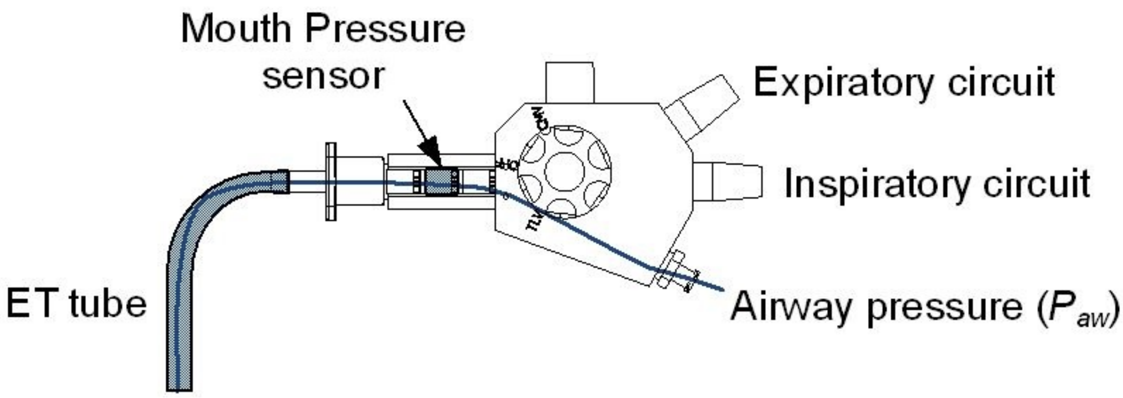


Figure 2.7: Pressure sensor connected to the endotracheal tube. [27]

2.3.5 Flow Measurement

Accurate monitoring of flow rate and volume exchanges is essential to minimize ventilator-induced lung injury. Mechanical ventilators employ flowmeters to estimate the amount of gases delivered to patients and use the flow signal as a feedback to adjust the desired amount of gas to be delivered and to avoid common side effects related to uncorrected ventilation, such as volutrauma or barotrauma (too high amount of gas delivered to patients) [28].

Flowmeters must fulfill strict requirements in terms of both dynamic and static characteristics.

2.3.6 Inspiratory Flow Unit

The green blocks in figure 2.4 correspond to the Inspiratory Flow System. This path is connected to regulated sources of air and oxygen. Ventilators may have internal regulators as well. These internal regulators are used to reduce the gas pressure provided by the air and oxygen source to a lower level suitable for the ventilator [21].

As shown in Figure 2.4, the air and oxygen sources are connected to valves and flow sensors. The microcontroller monitors the output of the flow sensors and controls the valves accordingly. In this way, the ventilator produces oxygen-enriched air with a specified oxygen concentration and conducts it to the Tank.

During inspiration, the ventilator opens the Inspiration Valve and closes the Expiration Valve. The Inspiration Valve is controlled in a way that the patient receives breaths consistent with the desired predefined waveforms.

As you can see, the key factor is adjusting the valves so that different system parameters (such as oxygen concentration, flow, and pressure) meet the specifications. Figure 2.4 shows the control loop for adjusting the air and oxygen valves.

2.3.7 Expiratory Flow Unit

During expiration, the ventilator opens the Expiration Valve and closes the Inspiration Valve. The exhaled air goes through the Flow Sensor and the PEEP Adjustment Mechanism. The output of the flow sensor is monitored by the microprocessor to achieve the sophisticated ventilator–patient interactions that were discussed above [21].

With the example waveforms depicted in figure 2.8, the pressure at the end of expiration is zero (PEEP is zero); however, we sometimes prefer to keep the airway pressure above the atmospheric level during expiration (non-zero PEEP). This keeps the patient's lungs from collapsing.

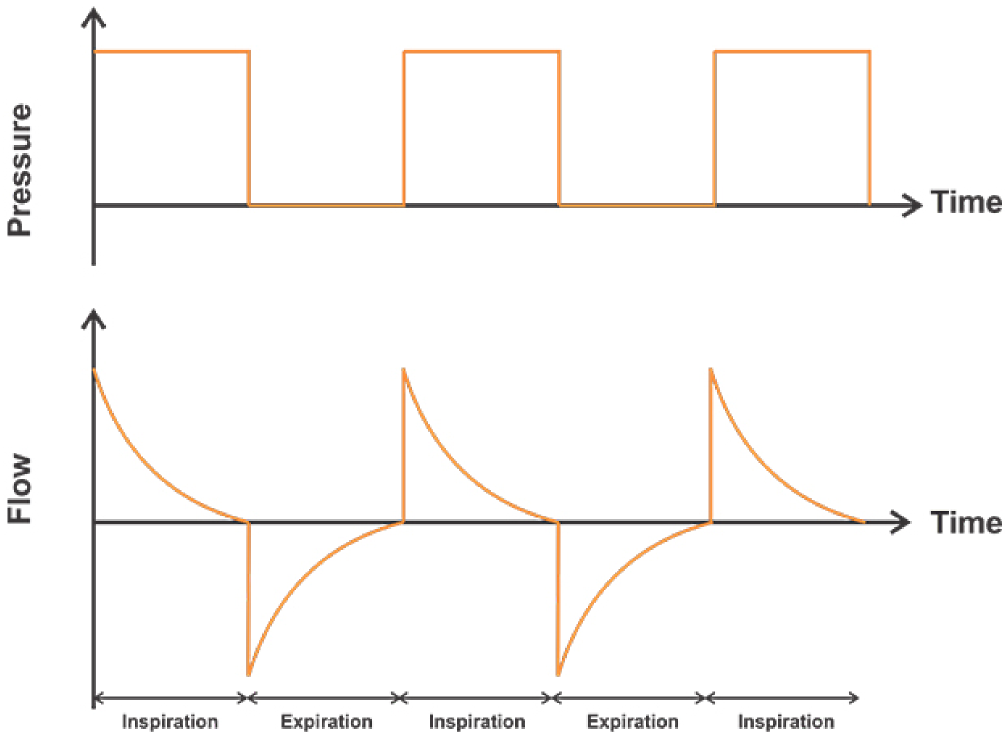


Figure 2.8: Waveforms for describing the pressure and flow of air through a ventilator. [21]

2.3.8 Oxygen Measurement

Oxygen management is a critical element of intensive care in mechanically ventilated patients. Oxygen is added to the air that is used to provide mechanical ventilation. When we breathe in, we bring fresh oxygen to our lungs. Normally, oxygen makes up 21% of all of the gases in the air that we breathe in. In normal health, this is enough incoming oxygen to keep the oxygen level in the blood at a level that meets the needs of all cells. However, most patients need a higher concentration of oxygen than is present in normal atmospheric air [29].

In most modern ventilators, some sort of feedback mechanism exists which monitors the inspired oxygen concentration and relates it to the proportioning valve in a feedback loop.

2.3.9 Monitoring

Ventilator monitoring includes functions that continuously sense the status of the ventilator system. The monitoring results may be displayed numerically and graphically. Clinicians use this information to check and understand the status of mechanical ventilation. These monitoring signals also serve as the inputs to the alarm system and may be used for some automatic regulation mechanisms. These signals include pneumatic ones (pressure, flow, and volume), airway CO2, FiO2, and oxygen saturation. If we think of ventilation functions as the 'doing'; the monitoring functions are the 'seeing' and 'knowing' [23].

2.3.10 User Interface or Graphical User Graphic Interface (GUI)

The ventilator user interacts with the ventilator through the user interface. This allows the user to control and manipulate the ventilator operation as they desire. The user interface typically includes an operating panel with elements such as LCDs (liquid crystal displays), knobs, switches, and keypads. The panel separates control settings such as breath rate, tidal volume, O2 concentration, and inspiration time from alarm settings, such as high-pressure limit, low-pressure limit, and minimum volume limits. There are also selections for setting the ventilator mode and breath parameters. Alarm conditions may trigger specific messages to display in the display window on the control panel and trigger an auditory alarm as well [23].

Control selections are fed to the microprocessor control unit which takes the input selections and adjusts the settings for control valves in the circuit that produce the desired ventilation waveform and profile for the patient.

Nowadays, touch screens are an increasingly popular means of input and the importance of the ventilator user interface is increasingly recognized.

Figure 2.9: Graphical User Interface (GUI).

2.3.11 Alarm and Safety Mechanisms

Ventilator alarms use audible and visual signals to alert the clinician to predefined abnormal conditions that require special attention or corrective actions. Ventilator alarms are based on both monitoring inputs and the alarm criteria set by the manufacturer or operator. If designed and used appropriately, alarms are a major contributor to the safety of the ventilated patients. They may create difficulties of their own if they are not [23].

2.3.12 Humidifiers

The airflow generated by a ventilator is often greater than what the body is used to and thus it must be warmed and humidified, using a humidifier, to avoid serious complications related to dry gases.

Basically, humidifiers (also called heaters) are devices that add molecules of water to gas. They can be divided into active heated humidifiers, which are devices heated by external sources of heat and water, and passive humidifiers such as heat and moisture exchangers (HMEs), which use the patient's own temperature and hydration to achieve humidification in successive breaths [23].

2.3.13 Gas Filters

The filters of a mechanical ventilator play several important roles. Use of these filters can potentially minimize the risk of infection and cross-contamination to patients, caregivers and visitors in the critical care setting, they are used to prevent undesirable particles being delivered to patients through inspired gases and to potentially remove particles from exhaled gases.

A filter may be mounted in one of three positions in a breathing circuit for different purposes like shown in figure 2.10. The first position is the artificial airway. This is the position of an HME filter for short-term airway humidification [23].

The second position is the beginning of the inspiratory limb. The inspiratory filter is intended to separate the ventilator and the ventilated patient. On one hand, use of an inspiratory filter can prevent the ventilator from being contaminated. Bear in mind that disinfection or sterilization of the gas passageway inside the ventilator is almost impossible. In addition, it can prevent particles from the source gases or ventilator from reaching the patient.

The third position is the end of the expiratory limb. An expiratory filter can safeguard the medical staff by filtering the expired gas. This is particularly valuable if the patient has infectious diseases. It also protects delicate sensors.

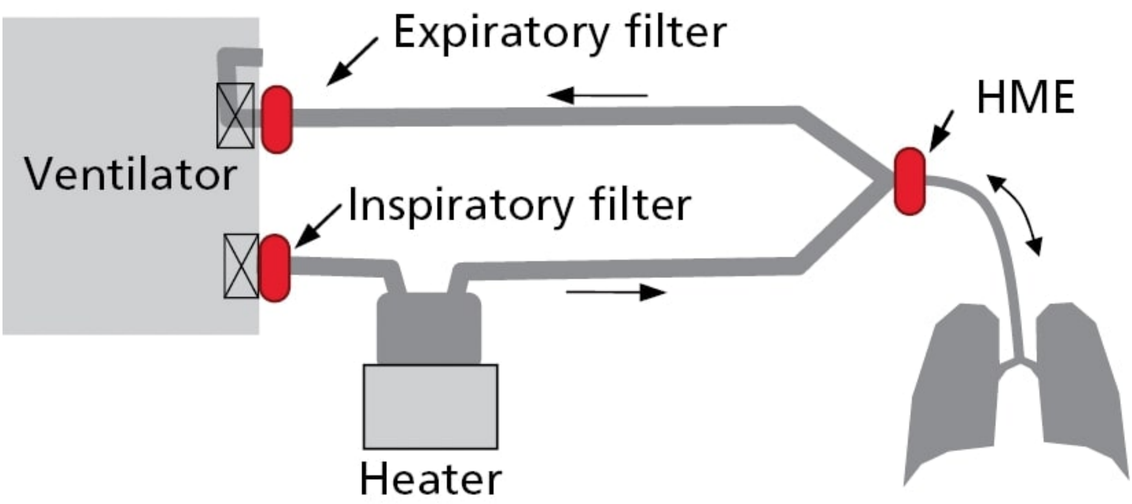


Figure 2.10: Three possible positions for a gas filter. [23]

2.3.14 Patient Circuit

The patient circuit is another essential part of a ventilator system. Its primary function is to provide a flexible, gas-tight channel connecting the ventilator and the artificial airway. It is also the location where an active humidifier, and gas filter can be mounted [30].

Do not underestimate the importance of the breathing circuit, because it is the source of many ventilation problems, including disconnection, leakage, occlusion, and circuit rainout.

The breathing circuit has three parts: the inspiratory limb, the expiratory limb, and the Y-piece or wye as shown in figure 2.11.

Both the inspiratory and expiratory limbs are flexible tubes that connect the Y-piece to the inspiratory or expiratory port of a ventilator.

Under normal conditions, gas moves freely in the breathing circuit and airway, driven by the pressure gradient. In each of these limbs, gas moves in one direction only.

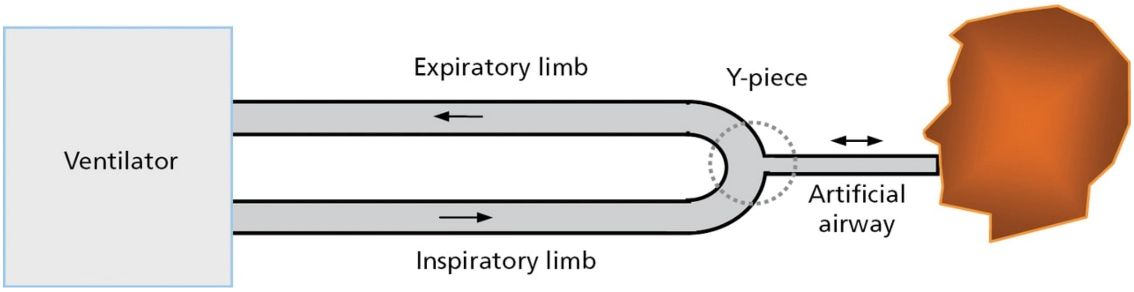


Figure 2.11: The patient circuit. [23]

Chapter 3

Mechanical Ventilator Hardware Design

3.1 System Background

A ventilator is a machine designed to mechanically move air in and out of the lungs to intermittently or continuously assist or control pulmonary ventilation. The most common indicators of the ventilation are the absolute volume and changes of volume of the gas space in the lungs achieved during a few breathing maneuvers. The ventilator is constantly monitored and adjusted to maintain appropriate pressure, volume, and flow [31].

This system requires a set of sensors for pressure, volume, and flow. The information from the sensors modulates the operations in the microprocessor (MCU). This MCU receives information from the airways, lungs, and chest wall through the sensors, and decides how the ventilator pumps.

Figure 3.1 shows a functional diagram for a ventilation system.

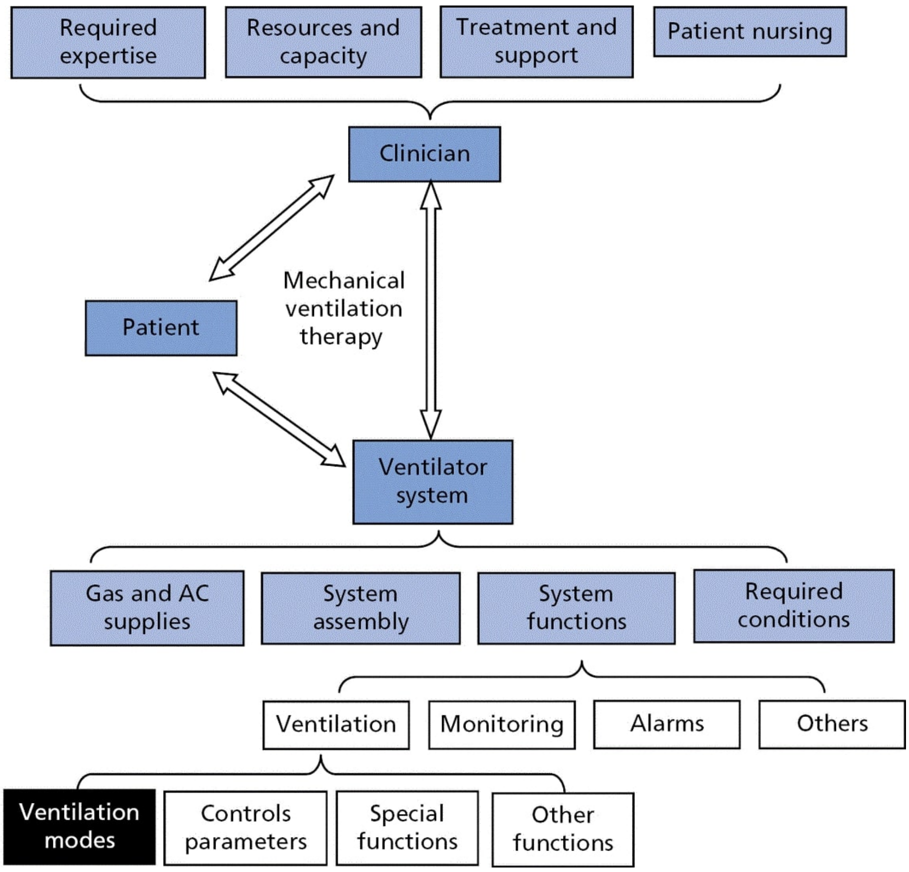


Figure 3.1: Functional diagram for a ventilation system. [23]

3.1.1 Input Power

The power source for a ventilator is what generates the force to inflate the patient's lungs. It may be either electrical energy or compressed gas. An electrically powered ventilator uses AC voltage from an electrical line outlet. In addition to powering the ventilator, this AC voltage may be reduced and converted to direct current. This DC source can then be used to power delicate electronic control circuits. Some ventilators have rechargeable batteries to be used as a back-up source of power if AC current is not available. A pneumatically powered ventilator uses compressed gas. This is the power source for most modern intensive care ventilators. Ventilators powered by compressed gas usually have internal pressure reducing valves so that the normal operating pressure is lower than the source pressure [32].

3.1.2 Power Transmission and Conversion

The power transmission and conversion system consists of the drive and output control mechanisms. The drive mechanism generates the actual force needed to deliver gas to the patient under pressure. The output control consists of one or more valves that regulate gas flow to and from the patient. The ventilator's drive mechanism converts the input power to useful work. The type of drive mechanism determines the characteristic flow and pressure patterns the ventilator produces. Drive mechanisms can be either: (1) a direct application of compressed gas through a pressure reducing valve, or (2) an indirect application using an electric motor or compressor [32].

The output control valve regulates the flow of gas to and from the patient. It may be a simple on/off exhalation. An example would be the typical infant ventilator. The valve in the exhalation manifold closes to provide a periodic pressure waveform that rises to a preset limit during inspiration (forcing gas into the lungs) then opens to allow pressure to fall to another preset limit during exhalation (allowing gas to escape from the lungs). Alternatively, there can be a set of output control valves that shape the output waveform.

3.1.3 System Sensors

The signal that shows lung volume is a differential signal, but this is not the signal measured directly from the lungs. To get this signal, it is necessary to transduce the pressure to voltage. This is done by a pressure sensor.

There are a variety of sensors that use integrated circuits for signal conditioning. This is an advantage because external components are not necessary. However, it is necessary to check the sensor and ADC resolution. If the ADC resolution is greater than the sensor resolution, amplifying the signal is recommended. Some sensors provide differential outputs to pass the signal through an instrument amplifier, when necessary. The sensor that fits volume measurement is a differential pressure sensor that accepts two sources of pressure simultaneously. The output is proportional to the difference of the two sources. It is important to mention that the normal pipeline gas source of a hospital is 50 PSI. This is a measurement that can be taken by Freescale pressure sensors.

3.1.4 Flow Measurement

The **Venturi effect** is used to measure flow. This is the reduction in fluid pressure that results when a fluid flows through a constricted section of a pipe. This effect is called a jet effect because the velocity of the substance increases on the way from the wide section to the narrow section. The pressure also increases over a smaller surface area; the same force applied to a smaller area equals a higher pressure in that area. According to fluid dynamics, a fluid's velocity must increase as it passes through a constriction to satisfy the conservation of mass, while its pressure must decrease to conserve energy [33].

Equation 3.1 refers to the Venturi effect

$$P_1 - P_2 = \frac{\rho}{2}(v_1^2 - v_2^2) \tag{3.1}$$

Where ρ is the density of the fluid, $v1$ is the (slower) fluid velocity where the pipe is wider, $v2$ is the (faster) fluid velocity where the pipe is narrower (as seen in figure 3.2). This assumes the flowing fluid (or other substance) is not significantly compressible, even though the pressure varies, the density is assumed to remain approximately constant.

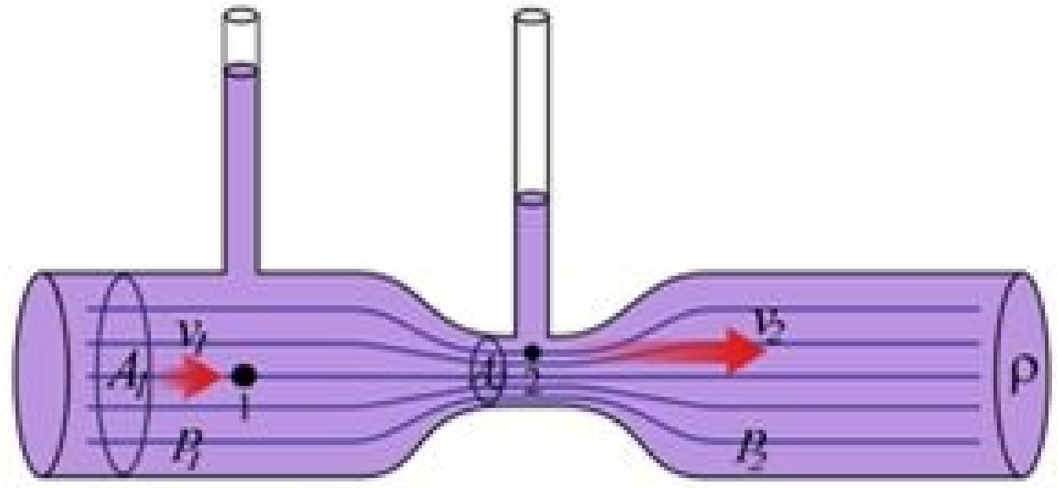


Figure 3.2: Venturi effect graphical demonstration. [33]

Figure 3.3 shows Venturi pipes used in industrial and in scientific laboratories for measuring liquid flow. This case system uses a reusable medical Venture pipe, called a D-Lite sensor.

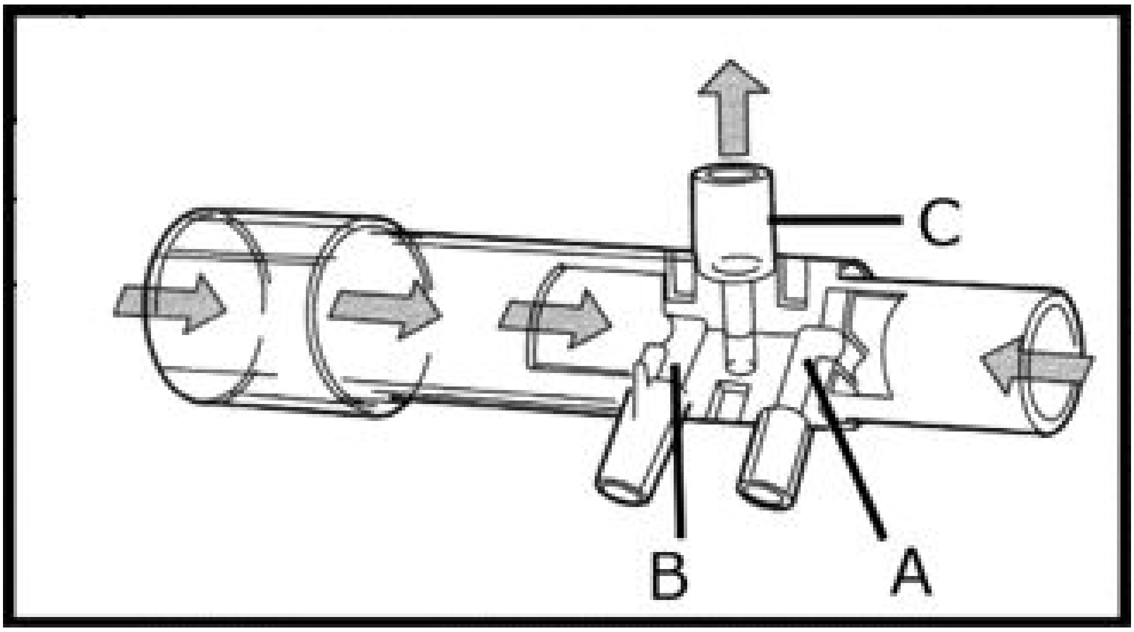


Figure 3.3: Venturi real pipe application for a D-lite sensor.

Figure 3.3 shows the sensor with three sampling ports. During inspiration the gas flows from the ventilator to the patient, A measures the total pressure, B measures the static pressure. The difference between the two gives a dynamic pressure, which is proportional to the velocity of gas flow. During expiration the process is reversed. C measures CO2, O2, and anesthetic gas concentration.

A Venturi can be used to measure the volumetric flow rate Q.

Since:

$$\begin{cases} Q = v_1 A_1 = v_2 A_2 \\ P_1 - P_2 = \frac{\rho}{2}(v_1^2 - v_2^2) \end{cases} \tag{3.2}$$

Then:

$$Q = A_2 \sqrt{\frac{2(P_1 - P_2)}{\rho(1 - (\frac{A_2}{A_1})^2)}} = A_2 \sqrt{\frac{2}{\rho(1 - (\frac{A_2}{A_1})^2)}} * \sqrt{\Delta P} \tag{3.3}$$

Divide as follows:

$$Q = A_2 \sqrt{\frac{2}{\rho(1 - (\frac{A_2}{A_1})^2)}} * \sqrt{\Delta P} = K_{sensor} * P_{gas} * \sqrt{\Delta P} \tag{3.4}$$

Finally, the pneumatic system has a constant value relating pressure with flow, you can measure directly the differential pressure from the venture, square root it and multiply it for your system constant to obtain the instant volumetric flow.

3.1.5 Alarm System

An important part of this application is the alarm that indicates different patient parameters such as exhaled volume or airway pressure. The ventilation system must be able to detect whether a breath has been taken. The MCU measures changes in aspiratory flow and pressure by using sensors. If no inspiration is detected within a certain period of time, the monitor sounds an alarm. The conditions programmed depend on each system. PWM cycles can be programmed to sound alarms. Sometimes, the ventilation system uses different alarms for different situations [23].

3.2 Hardware Design

An electronics system for a medical respirator can be complex because of the variety of components and functions that must be accurate.

It must have the following modules:

3.2.1 National Instruments USB-6009 Data Acquisition Board

Data acquisition is the procedure of collecting information from the process. With analog instruments it is difficult to collect data. There are also a lot of problems such as noise, drift, instability and high energy consumption. In digital systems it is easier to control noise and they provide easy transfer and storage of data.

Measurement and acquisition system consists of three parts:
- acquisition
- analysis
- data presentation

Acquisition card National Instruments USB-6009 allows data acquisition for mobile measurements, practice and lab measurements etc. It connects to the computer via the USB interface and that is a big advantage. The card is good for some complex measurements [34].

The external appearance of the card is shown in figure 3.4.

Figure 3.4: Acquisition card NI USB-6009. [35]

3.2.2 Pressure sensors

To measure accurately the control system, it is important to have the correct sensors. The following pressure sensor is used:
MPX2010 — 0 to 50 kPa integrated on-chip pressure sensor, temperature compensated, and calibrated [36]

Key features:
- Temperature Compensated over 0°C to +85°C
- Ideally suited for microprocessor or microcontroller-based systems
- One analog output voltage (0 - 27 mV), CASE 344.

Figure 3.5 shows output voltage versus the pressure graph.

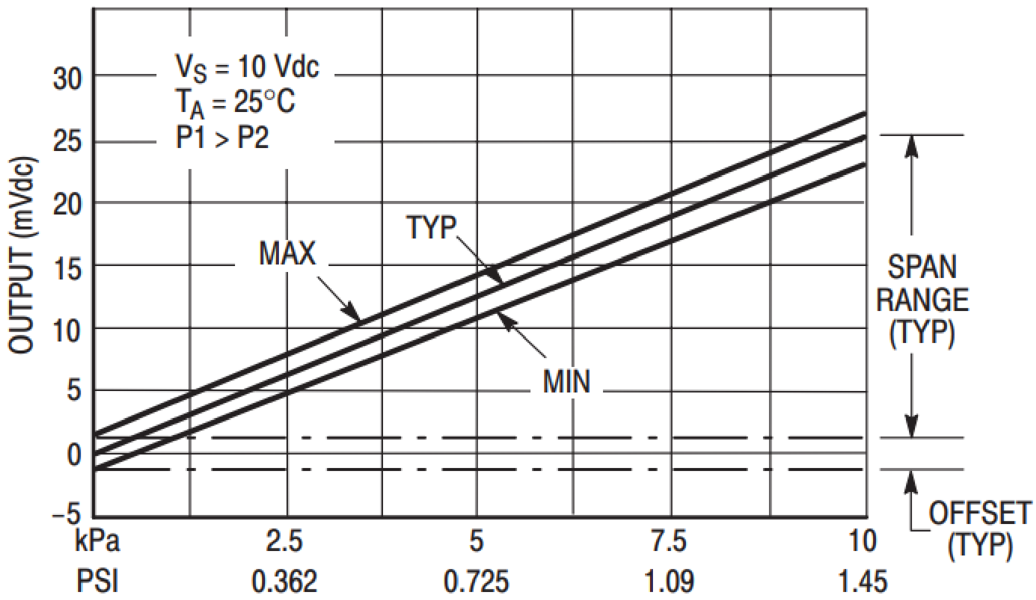


Figure 3.5: Output versus Pressure graph. [36]

3.2.3 Power Supply

The entire ventilator system is powered by a 12 V 15 A power supply connected to a 120/220 V AC power line that may be switched according to the power supply. The system also will have a 5 V regulator at 2 A used to drive four 200 mA electro valves, a differential pressure (flow) sensor, and the NI USB-6009 Data Acquisition Board. And finally, a 3 V at 500 mA regulator to power an 80mA buzzer, and the pressure sensor.

3.2.4 Analog to Digital Converter (ADC)

To measure sensor analog signals for pressure calculating, the system needs an accurate ADC. The NI USB-6009 provides 8 channels with 14-bit ADC that can be used to read different sensors [34].

3.2.5 Signals Processing and Sensors Measurement Circuits

Reliable measurement is a critical factor for this application. Signal treatment and decoupling are important factors to consider. The system must avoid electrical noise using capacitors and inductors for decoupling. This version does not use external analog treatment, it uses internal op amps to amplify small signals like differential pressure between the simulated lung and ambient without a pumping system.

3.2.6 Human Interface

A user interface gives users a way to interact with the source code. It allows the user to change the values passed to the source code and see the data that the source code computes. In LabVIEW, the user interface is the front panel.
Inputs are acquired from a device such as a data acquisition device. Outputs are displayed with indicators such as graphs, charts, or LEDs.

3.2.7 BLDC Blower

A blower-based ventilator will effectively provide constant positive airway pressure to the patient thus maintaining an open airway. It will give the operator the ability to deliver specific pressures and volumes of gas to their patient at regular intervals as the operator sees fit.

The blower is shown in Figure 3.6. Figure 3.7 shows a schematic outline of the blower system. The most important task is to build the feedback control circuit that will drive the blower to create the desired pressures within the mask and in the patient's airway.

Figure 3.6: Photos of a blower model.

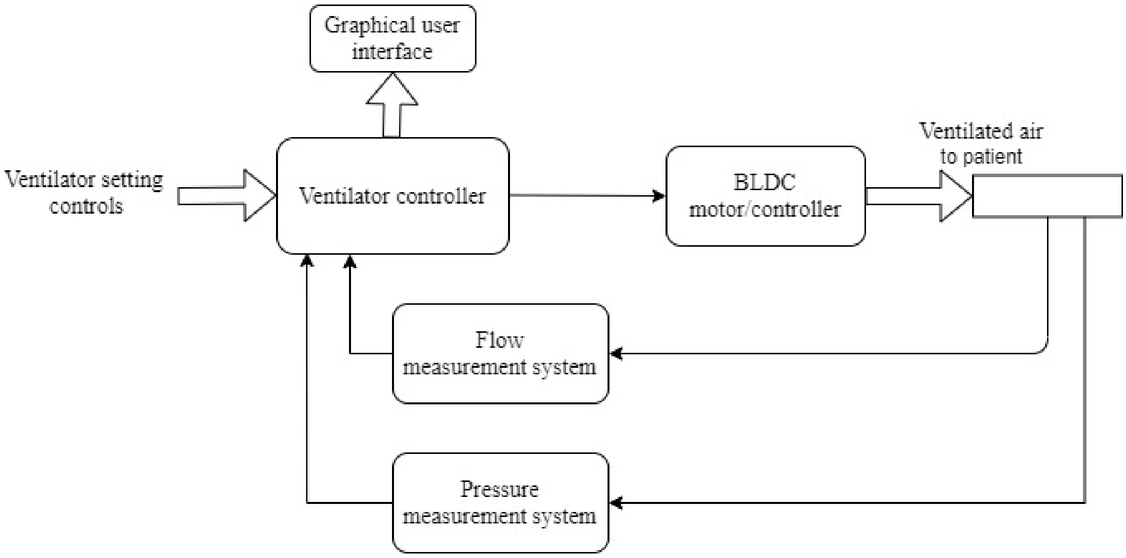


Figure 3.7: Block diagram for the blower based system.

3.3 Schematics

3.3.1 NI USB-6009 DAQ

It has 8 analog inputs, 2 analog outputs, 12 digital inputs/outputs and 32- bit counter. Maximum sample speed by each analog input is 48 kS/s. Sample speed on analog outputs is 150 S/s and it can't be changed. Analog inputs have 14-bit resolution and analog outputs have 12-bit resolution. USB interface allows better transferability and easier connecting with the PC. Pin layout of the card is shown in Appendix.

Work with the acquisition card NI USB-6009 is possible with 32-bit program package Matlab and its library Simulink. Inside of Simulink there is a toolbox called Data Acquisition Toolbox which reads the card and makes it possible to use its inputs and outputs in schemes inside of Simulink. That makes measurement easier because it enables to work with data inside the program.

3.3.2 Power Supply

Power supply connections will have some decoupling capacitors, separated 3.3 V, and a separated ground for analog ground. It is protected against inverse current and transients with a diode and a TVS diode. It is important to have test points and LEDs for debugging the power supply. Figure 3.8 shows the actual power supply schematic diagram.

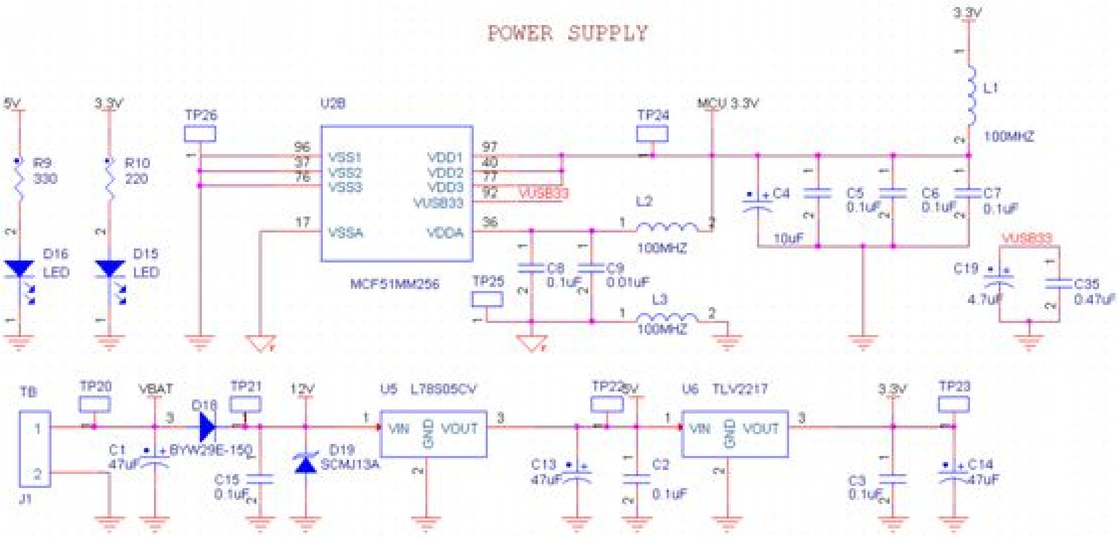


Figure 3.8: Power supply.

3.3.3 Alarms

Output alarms are those that are triggered by an unacceptable state of the ventilator's output. More specifically, an output alarm is activated when the value of a control variable (pressure, volume, flow, or time) falls outside an expected range.

Mechanical ventilation system has two types of alarms for the patient, visual, and audio. Visual is a bright LED, and audio is the buzzer, which uses the same switching circuit as valves to ensure correct working.

Figure 3.9 shows two types of alarms.

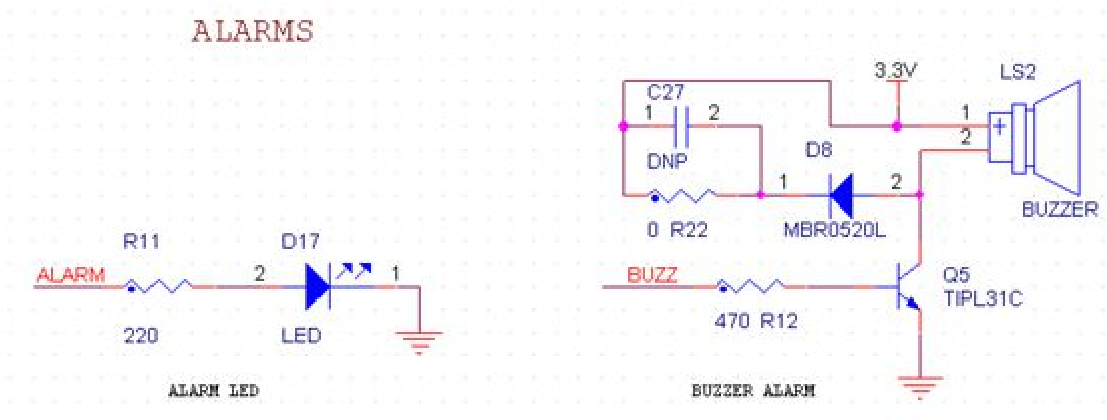


Figure 3.9: Alarms.

3.3.4 Pressure Sensors

For adequate and stable signal gain and output flexibility, a two-stage differential op-amp circuit with analog or switch output is used, as shown in figure 3.10. The four op-amps are packaged in a single 14 pin quad package. There are several features to note about the circuitry.

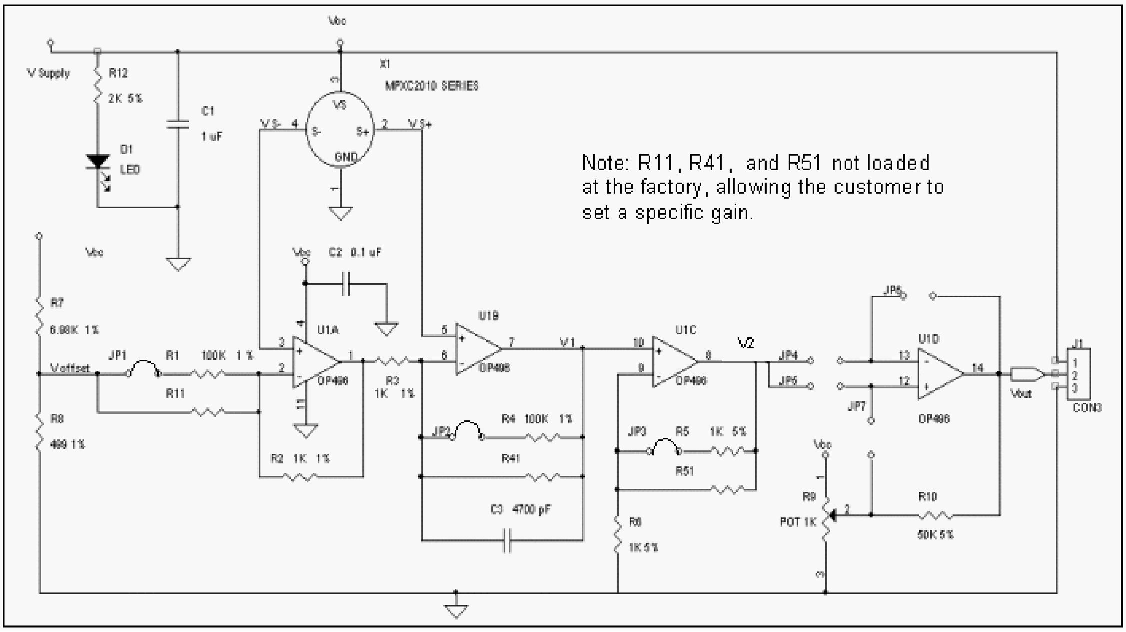


Figure 3.10: Pressure sensors. [37]

The first gain stage is accomplished by feeding both pressure sensor outputs ($VS-\&VS+$) into the non-inverting inputs of operational amplifiers. These op-amps are used in the standard non-inverting feedback configuration. With the condition that Resistors $R2 = R3$, and $R1 = R4$ (as closely as possible), this configuration results in a gain of $G1 = R4/R3 + 1$.

The default gain is 101, but there are provisions for easily changing this value. The signal V (op-amp Pin 7) is then calculated as:

$$V1 = G1 * (VS + -VS-) + Voffset \tag{3.5}$$

Voffset is the reference voltage for the first op-amp and is pre-set with a voltage divider from the supply voltage. This value is set to be 6.7 percent of the supply voltage. It is important to keep this value relatively small simply because it too is amplified by the second gain stage. It is also desirable to have resistors $R7$ and $R8$ sufficiently large to reduce power consumption.

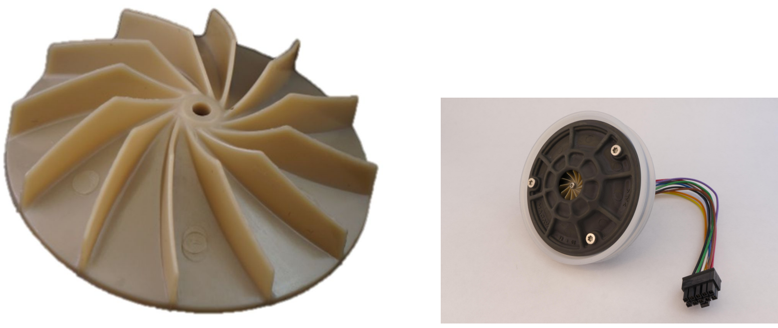
The second gain stage takes the signal from the first gain stage, V, and feeds it into the non-inverting input of a single op-amp. This op-amp is also configured with standard noninverting feedback, resulting in a gain of $G2 = R5/R6 + 1$. The default value is set to 2, but can easily be changed.

The signal produced at the output of the second stage amplifier, V (op-amp pin 8) is the fully amplified signal. This is calculated as

$$V2 = G2 * V1 \tag{3.6}$$

3.3.5 BLDC Blower

The blower consists of an impeller (figure 3.11a) inside a housing (figure 3.11b). When the impeller is rotating air is forced to the outside out of the blower by centrifugal forces. The mixing chamber is a volume where air and oxygen is coming together and mixed before the gas is going to the patient. The check valve is a valve that permits the gas to flow in only one direction. When the gas flows in the desired direction the valve opens, while back-flow closes the valve. (See figure 3.12.)



(a) Impeller. (b) Total blower

Figure 3.11: The blower (b) consists of an impeller (a) inside a housing.

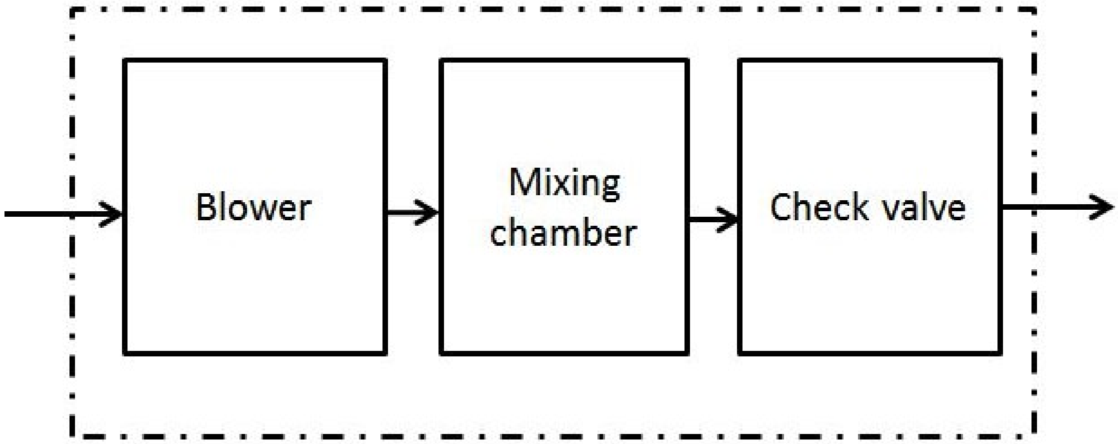


Figure 3.12: Block diagram of blower, mixing chamber and check valve.

BLDC Motor Mathematical Model

For a six-state three-phase star-shaped BLDCM with two-phase conduction, certain assumptions are considered when modeling the motor and analyzing and controlling the electromagnetic torque [38].
1. The three-phase winding is symmetrical, the air gap of the magnetic field is a square wave, and the stator current and the magnetic field distribution of the rotor are symmetrical.
2. The effects of the slot, commutation process, and armature reaction are ignored.
3. The armature windings are evenly and continuously distributed on the inner surface of the stator.
4. The magnetic circuit is not saturated regardless of eddy current and hysteresis loss.
An equivalent circuit of BLDCM is shown in figure 3.13.

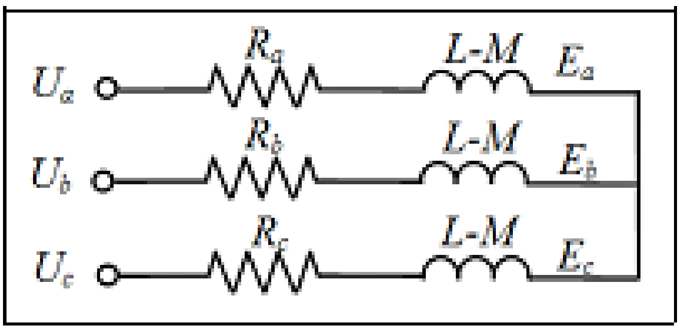


Figure 3.13: Equivalent circuit of BLDCM. [38]

In the figure, U_a, Ub, and U_c are three-phase voltages; I_a, I_b, and I_c are three-phase stator currents; E_b, E_a, and E_c are the back EMFs of the three phases;

$R_a = R_b = R_c = R$ denotes the stator resistance of the three phases; L is the identical self-inductance of the phase stator winding; and M is the identical mutual inductance between any two phase windings. The inductance of each phase winding is defined as $L - M$.

The voltage equation of the three-phase winding of BLDCM is

$$\begin{cases} U_a = R_a I_a + (L-M)dI_a/dt + E_a \\ U_b = R_b I_b + (L-M)dI_b/dt + E_b \\ U_c = R_c I_c + (L-M)dI_c/dt + E_c \end{cases} \tag{3.7}$$

Then, the matrix form of the phase voltage equation on the BLDCM can be written as (considering approximations of current)

$$\begin{bmatrix} U_a \\ U_b \\ U_c \end{bmatrix} = \begin{bmatrix} R & 0 & 0 \\ 0 & R & 0 \\ 0 & 0 & R \end{bmatrix} \begin{bmatrix} I_a \\ I_b \\ I_c \end{bmatrix} + \begin{bmatrix} L-M & 0 & 0 \\ 0 & L-M & 0 \\ 0 & 0 & L-M \end{bmatrix} \frac{d}{dt} \begin{bmatrix} I_a \\ I_b \\ I_c \end{bmatrix} + \begin{bmatrix} E_a \\ E_b \\ E_c \end{bmatrix}$$

At any moment, the BLDCM stator has only two phase conduction. When phases A and B are conducted, $I_a = -I_b = I_s$ and $E_a = -E_b = E_s$.
Thus, equation 3.7 can be rewritten as

$$U_a - U_b = 2RI_s + 2(L-M)dI_s/dt + 2E_s \tag{3.8}$$

The electromagnetic torque of BLDCM is generated by the interaction between the current in the stator winding and the magnetic field produced by the permanent magnet of the rotor

$$T_e = \frac{E_a I_a + E_b I_b + E_c I_c}{w} = \frac{2E_s I_s}{w} \tag{3.9}$$

where w is the mechanical angular speed of the motor.

According to the relationship between the single phase back EMF of BLDCM and the rotational speed

$$E_s = K_e w \tag{3.10}$$

where K_e is the back EMF coefficient. By using equations 3.9 and 3.10, the following expression can be obtained

$$T_e = 2K_e I_s \tag{3.11}$$

According to equation 3.11, the electromagnetic torque of BLDCM is proportional to the current under a given back EMF coefficient. $U_d = U_a - U_b$ is defined as the line voltage, which is the power supply voltage, and it can be obtained from equation 3.8 as follows

$$U_d - 2(L - M)dI_s/dt = 2RI_s + 2E_s \tag{3.12}$$

The rotor dynamic equation of a motor is expressed as

$$T_e = T_m + Bw + J\frac{dw}{dt} \tag{3.13}$$

where T_m is the load torque, J is the motor inertia, and B is the motor damping coefficient.
The speed of the BLDCM is given as

$$T_e - T_m = J\frac{dw}{dt} + Bw \tag{3.14}$$

3.4 PWM in Blower-Based Mechanical Ventilation

BLDCM can be driven by two techniques. They are Pulse Amplitude Modulation (PAM) and **Pulse Width Modulation (PWM)** technique [39].

3.4.1 PWM Working Principle

Pulse width modulation control works by switching the power supplied to the motor on and off very rapidly. The DC voltage is converted to a square wave signal, alternating between fully on (nearly 12v) and zero, giving the motor a series of power “kicks”. Pulse width modulation technique (PWM) is a technique for speed control which can overcome the problem of poor starting performance of a motor. PWM for motor speed control works in a very similar way. Instead of supplying a varying voltage to a motor, it is supplied with a fixed voltage value (such as 12v) which starts it spinning immediately. The voltage is then removed and the motor ‘coasts. By continuing this voltage on/off cycle with a varying duty cycle, the motor speed can be controlled.

Pulse-width modulation (PWM) or duty-cycle variation methods are commonly used in speed control of DC motors [**?**]. The duty cycle is defined as the percentage of digital ‘high’ to digital ‘low’ plus digital ‘high’ pulse-width during a PWM period.

Chapter 4

Simulation and Results

4.1 Pressure Controlled Ventilation (PCV)

The pressure signal generated by PCV in reality takes the same breathing behavior as the respiratory activities. The suggested mathematical model for the PCV pressure signal should also represent the breathing behavior and activities.

4.1.1 The Mathematical Model for Pressure Controlled Ventilation Signal

The pressure signal of PCV depends on setting parameters in a real PCV device that formulates the typical waveform shown in figure 1.5.Therefore, the mathematical model was formulated to represent the inspiration and the expiration activities and reflect the main input parameters during pressure support using PCV. The periodic functions method was used to express the mathematical model of PCV pressure signal as seen in equations 4.1.

$$P(t) = \begin{cases} P_{aw}\frac{t}{\tau} + PEEP, 0 \leqslant t \leqslant \tau \\ P_{aw} + PEEP, \tau \leqslant t \leqslant T_{in} \\ PEEP, T_{in} \leqslant t \leqslant T_{ex} \end{cases} \tag{4.1}$$

where

- $P(t)$ - the pressure signal of PCV
- $PEEP$ - positive end-expiratory pressure
- P_{aw} - the pressure in respiratory airway

- T - time equal to $(T_{in} - \tau)$
- T_{ex} - expiration time

As shown from these equations, the ventilation pressure or supported pressure of PCV was represented as time based function P(t) to reflect the respiratory activities—inspiration and expiration during ventilation. These activities are generated from a change in the pressure amount in lungs during inspiration and expiration processes. The inspiratory pressure (IP) and expiration pressure (EP) were represented in mathematical model by P_{aw} and $PEEP$.

The parameter values of the mathematical model can be clarified as follows:

- The total cycle time (TCT) equals inspiratory time plus expiratory time ($T_{in} + T_{ex} = TCT$), then the TCT for one breath could be written by Equation 4.2.

$$1sec(T_{in}) + 2sec(t_{ex}) = 3sec \tag{4.2}$$

- The respiratory rate (RR) or frequency is obtained by dividing the number of breaths per minute to TCT as shown in Equation 4.3.

$$RR = \frac{1min}{TCT} = \frac{60sec}{3sec} = 20breaths/min \tag{4.3}$$

- The rise time of pressure (τ) is one of the setting variables in PCV This time is taken by the ventilator to reach the set pressure at the beginning of inspiration. In ventilators, the τ controls the adjustment of the inspiratory flow delivery by clinician. It has proportional relation with flow i.e. a fast rise time associated with high flow at the beginning of inspiratory and vice versa.
- The IP is based on the setting pressure value that considers the working pressure on the supplying system. Practically, it is determined by looking for a V_T of 5 - 6 ml/kg

Knowing all parameters related to time (T_{in}, T_{ex}, RR, ratio I/E), and other setting variables will determine the beginning of each breathing cycle as well as the beginning of inspiration and expiration using shown Inequalities 4.4, 4.5 and 4.6.

$$(3n - 3) \leqslant t \leqslant (3n - 3)T + \tau \tag{4.4}$$

$$(3n-3)T+\tau \leqslant t \leqslant (3n-3)T_{in} \tag{4.5}$$

$$(3n-3)T_{in} \leqslant t \leqslant 3nT_{ex} \tag{4.6}$$

where n is the number of breathing cycle and T equals to 1 sec

4.1.2 The Multi Compartment Model for Lungs

The suggested mathematical model for PCV was integrated as a pressure that is applied to a multi compartment model for lungs. It demonstrates the electric model for respiratory system as a multi compartment model shown in figure 4.1. This model has been converted to the mathematical model using transfer functions that represent the airflow (output) changes as a function of input pressure shown in equation 4.7.
Where: all parameters refer to normal lungs:

- The current flow (I) represents airflow, and the voltage source (V) demonstrates the applied pressure produced by the ventilator.
- $R_C = 1cmH2O/L/s$, shows the airflow resistance of the central airways.
- $R_P = 0.5cmH2O/L/s$, demonstrates the resistance of the peripheral airways.
- $C_L = 200ml/cmH2O$, represents the capacity of the alveoli.
- $C_W = 200ml/cmH2O$, indicates the chest wall capacity, which is in series with the alveoli.
- $C_S = 5ml/cmH2O$ represents a shunt capacitance known as "dead space" of air, which does not participate in the exchange of oxygen and carbon dioxide between air and blood.
- C_T demonstrates the total compliance of airways and has variation values that depend on C_L and C_W.

$$\frac{I(s)}{P(s)} = \frac{s^2 + (\frac{s}{R_P * C_T})}{R_C * s^2 + (\frac{1}{C_S} + \frac{R_C}{R_P * C_T})s + (\frac{1}{R_P * C_S})(\frac{1}{C_L} + \frac{1}{C_W})} \tag{4.7}$$

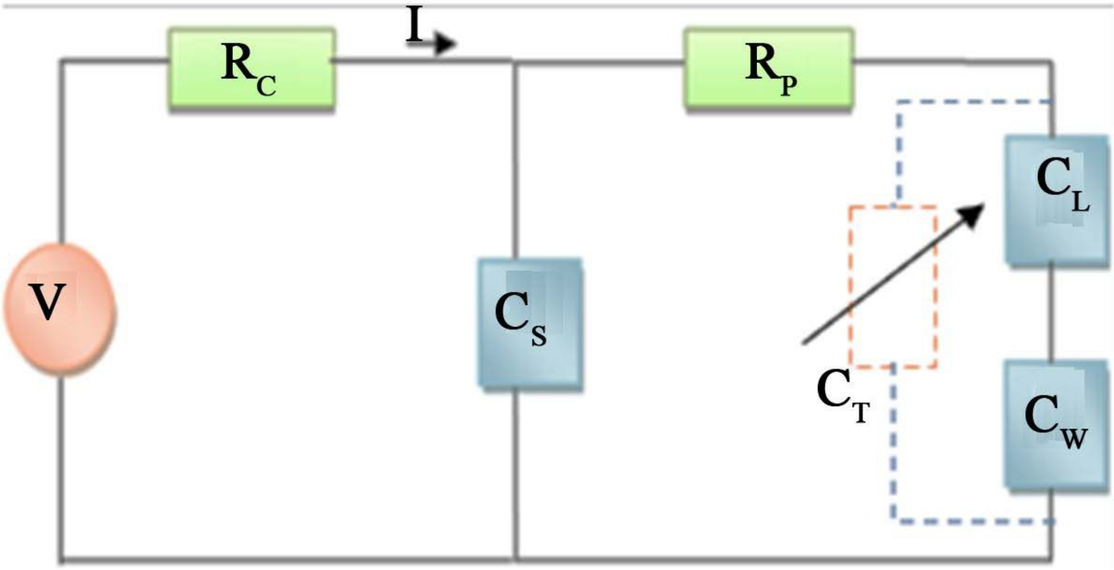


Figure 4.1: The equivalent electrical model of respiratory system—multi compartments.

Therefore, the transfer function has expressed the airflow output as a function of pressure input I(s)/P(s) using the transform function in equation 4.8. This equation is assumed to represent a healthy respiratory system with above stated parameters as well as the total capacity of lungs equal to

$$\frac{I(s)}{P(s)} = \frac{s^2 + 420s}{s^2 + 620s + 4000} \tag{4.8}$$

4.2 Simulation

A complete SIMULINK diagram of the proposed mathematical model of the respiratory system is shown in figure 4.2. The created mathematical model of PCV signal was modelled and simulated as a generator for the pressure, which acts as an input for the transfer function block. Output of the transfer function is the flow that can be integrated to produce volume.

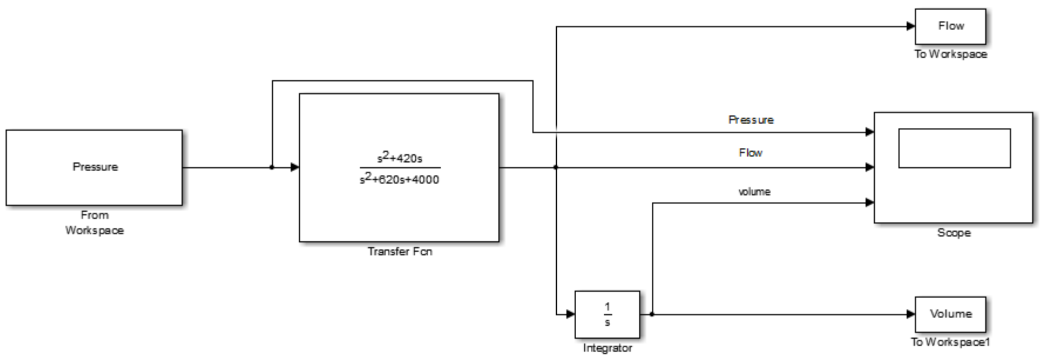


Figure 4.2: SIMULINK model of mechanical ventilation.

4.3 Results

All three signals (pressure, flow and volume) are displayed using MATLAB plot function as shown in figures 4.3, 4.4 and 4.5.

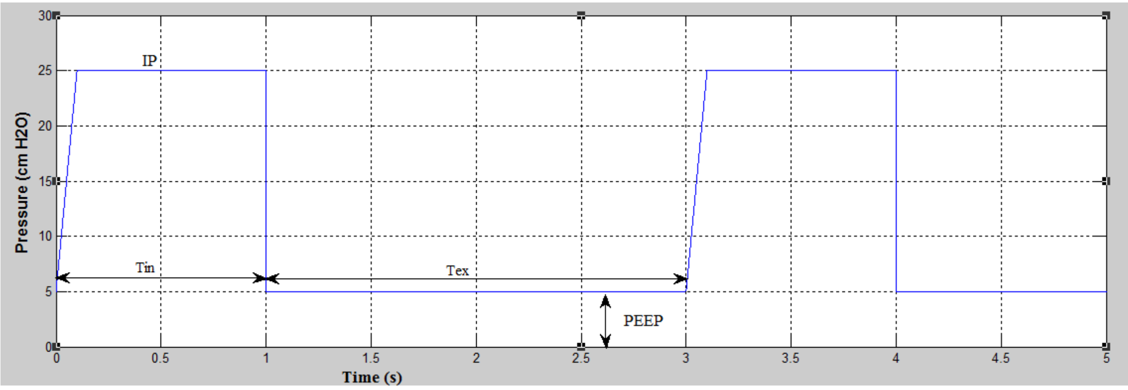


Figure 4.3: Pressure graph.

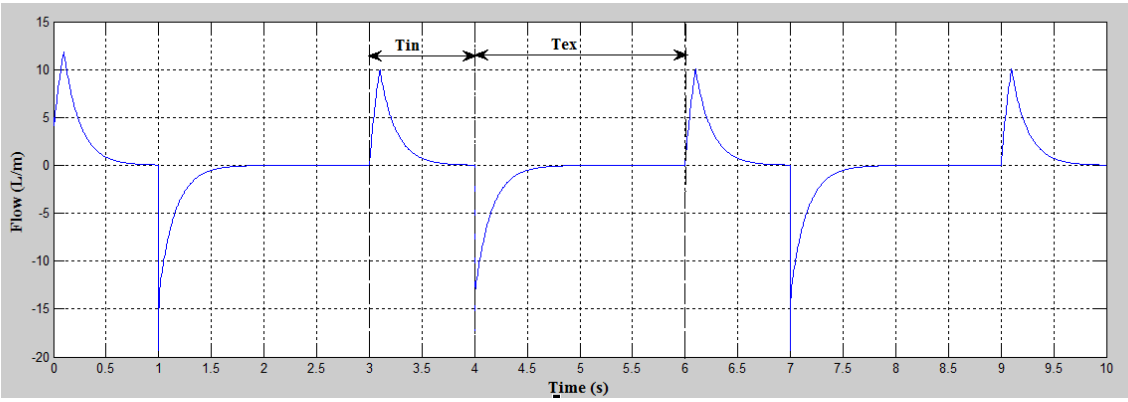


Figure 4.4: Flow graph.

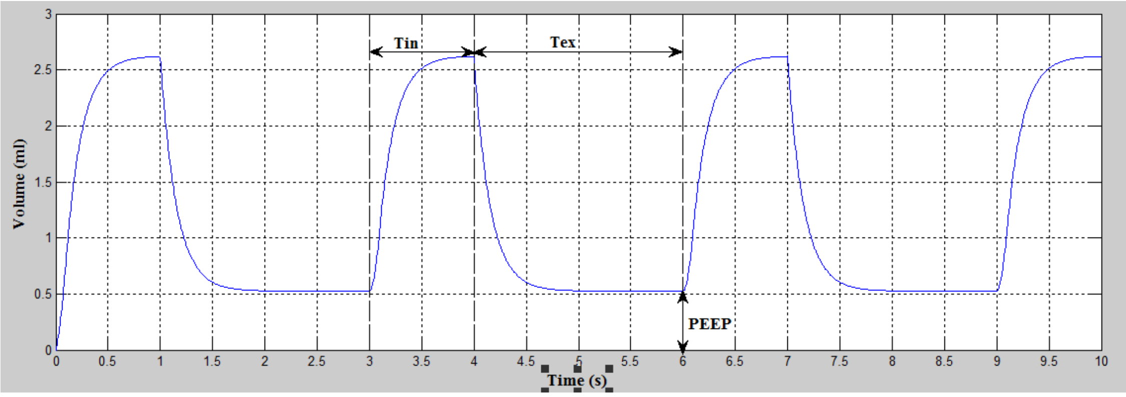


Figure 4.5: Volume graph.

Figure 4.6 illustrate graphically the simulation PCV pressure, flow and volume waves of 4 respiratory cycles. The setting parameter values used are set according to respiratory system status. T_{in} and T_{ex} are unchanged as well as the complete cycle time (3 seconds), whereas the IP and PEEP can be changed in a way that P_{plat} does not exceed the peak pressure.

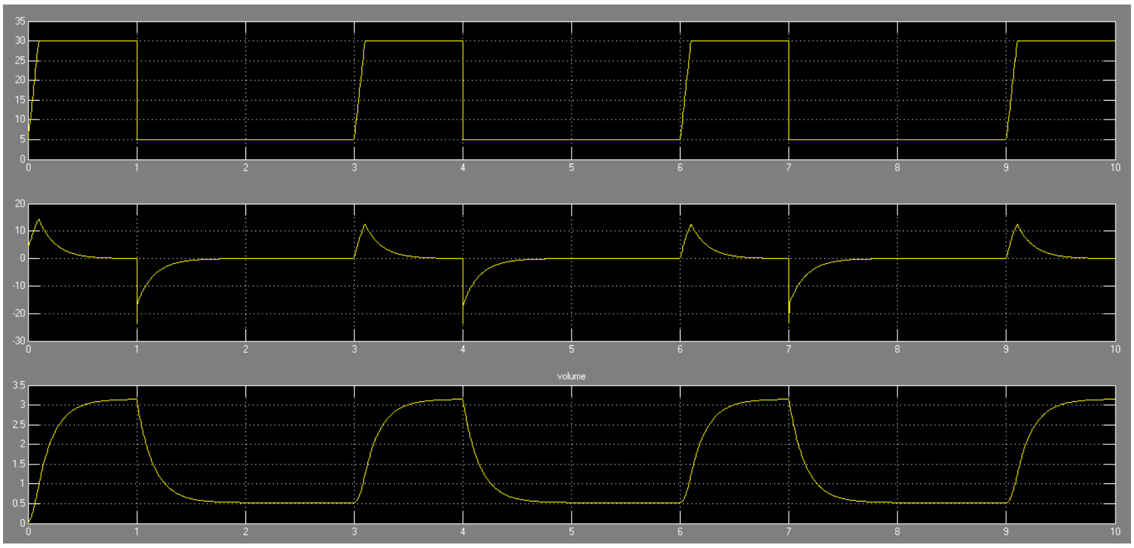


Figure 4.6: Pressure, flow and volume graphs.

Figures 4.3, 4.4 and 4.5 illustrate the waveforms of P, F and V, respectively, obtained at parameter values listed in section 4.1.2. In these figures, time duration equals approximately 12 seconds, for 4 complete respiratory cycles. The first waveform of figure 4.3 represents the pressure waveform of the created mathematical model, which has constant pressure at 25 cm H2O and constant width of inspiration and expiration time during the entire course ($T_{in} = 1sec$ and $T_{ex} = 2sec$).

The general view of flow waveform in figure 4.4 shows that the flow rapidly returns to zero at the beginning of plateau time then the flow at the end of the pause expiratory time takes the shown course. The flow must have the highest numeric value at the beginning of inspiration.

Figure 4.5 shows volume waveform, which does not have inspiratory pause, and the expiration follows immediately after inspiration.

The obtained result of P, F, V waveforms showed corresponding in their shapes and characteristics with standard reference curves.

Conclusion

In this work, we designed the essential hardware components for a mechanical ventilation system by establishing a general system background and functional hardware design along with the schematics. A BLDC motor is introduced in our design due to its significant energy efficiency improvements and better reliability.

We also developed the mathematical model of the PCV pressure signal. This mathematical model was simulated using the MATLAB environment and its SIMULINK library which provides access to numerous computational toolboxes. The mathematical model is able to represent setting parameters and its limit values, and therefore represents the real PCV device by simulator. This simulator is able to monitor the input and output signals as continuous waveforms to follow the real artificial ventilation process.

The results show a promising aspect of the system and its use in modern mechanical ventilation. The proposed work can be improved further more by using LABVIEW to build the Human Machine Interface, and also by the hardware implementation of the system which was impossible due to the current Covid-19 situation.

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Appendix: NI USB 6009 - Pinout

The following figure shows the pinout of the NI USB-6008/6009. Analog input signal names are listed as single-ended analog input name, AI x, and then differential analog input name, (AI $x+/-$).