DESIGN AND FEBRICATION OF COST EFFIECTIVE MECHANICAL VENTILATOR

A

PROJECT REPORT

Submitted in the partial fulfillment of the requirements of Degree in Bachelor of Technology

By

Anurag Vashishth	**1708240016**
Abdul Kabir	**1708240002**
Aryan Kumar	**1708240017**
Subhash Chandra Panday	**1708240055**

Under the guidance of

Mr. Ganesh Kr. Sharma (Assistant professor)

In Pursuit of Excellence

MORADABAD INSTITUTE OF TECHNOLOGY

Ram Ganga Vihar, Phase-II, Moradabad-244 001 (U.P.)

(2021)

Dr. Munish Chhabra
Professor & Head
Deptt. of Mechanical Engg.
Moradabad Institute of Technology
Moradabad - 244001

ACKNOWLEDGEMENT

I would like to extend my heartfelt gratitude to the people who supported and guided me to the completion of my project successfully. I feel very grateful to have such good mentors at my side who have helped me at every instance when it was required. **Project guide** **Mr. Ganesh kr. Sharma** have been very helpful during the entire period of the completion of the presentation and the following report.

I also extend my gratitude towards my college mates who provided me with various ideas, which was a great help to me. I would also acknowledge my friends for supporting me for the entire duration of the presentation.

Dr. Munish Chhabra
Professor & Head

MORADABAD INSTITUTE OF TECHNOLOGY, MORADABD

CERTIFICATE

This is to certify that the Project entitled **DESIGN AND FEBRICATION OF COST EFFIECTIVE MECHANICAL VENTILATOR** Submitted by **Anurag Vashishth, Abdul Kabir, Subhash Chandra Panday, Aryan Kumar** Roll No. **17082400 16/02/55/17** in partial fulfillment of the requirement of the Degree of B.Tech in Mechanical Engineering, embodies the work done by him under my guidance.

Signature of Co-Guide

Name Dr. Munish Chhabra

Designation Prof.

Date: 09.08.21

9/8/21

Signature of the Guide

Name : Ganesh Kr. Sharma

Designation : Assistant Professor

Dr. Munish Chhabra
Professor & Head
Deptt. of Mechanical Engg
Moradabad Institute of Technology

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Dr. Munish Chhabra
Professor & Head
Deptt. of Mechanical Engg.

ABSTRACT

The worldwide medical community currently faces a critical shortage of medical equipment to address the COVID-19 pandemic. In particular this is the case for ventilators, which are needed during COVID-19 related treatment at onset, during the intensive care phase and during the very extended recovery times. Companies are scaling up production, but this will not be sufficient to meet the demand according to the current forecasts [1] [2]. There is a wide spectrum of devices, ranging from highly sophisticated through to simpler units useful in the milder phases of illness. Amid pandemic crisis in India we were motivated to cater the local shortage of this medical equipment and develop a somewhat semi-automatic mechanical ventilator that could be used in emergency medical units in hospital as well as mobile medical units such as ambulances to provide emergency ventilation using an Ambu bag based setup to patients affected with COVID-19 [3] [4].

This paper describes the design and prototyping of a low-cost portable mechanical ventilator for use in mass casualty cases and resource-poor environments. The ventilator delivers breaths by compressing a conventional bag-valve mask (BVM) with a pivoting cam arm, eliminating the need for a human operator for the BVM. An initial prototype was built out of acrylic, measuring 11.25 x 6.7 x 8 inches (285 x 170 x 200 mm) and weighing 9 lbs (4.1 kg). It is driven by an electric motor powered by a 14.8 VDC battery and features an adjustable tidal volume up to a maximum of 750 ml. Tidal volume and number of breaths per minute are set via user-friendly input knobs. The prototype also features an assist-control mode and an alarm to indicate over pressurization of the system [5] [4]. Future iterations of the device will include a controllable inspiration to expiration time ratio, a pressure relief valve, PEEP capabilities and an LCD screen. With a prototyping cost of only $420, the bulk-manufacturing price for the ventilator is estimated to be less than $200. Through this prototype, the strategy of cam-actuated BVM compression is proven to be a viable option to achieve low-cost, low-power portable ventilator technology that provides essential ventilator features at a fraction of the cost of existing technology [6].

We propose the design of a ventilator which can be easily manufactured and integrated into the hospital environment to support COVID-19 patients. The unit is designed to support standard ventilator modes of operation, most importantly SIMV-PC (Synchronized Intermittent Mandatory Ventilation) mode. The design is under prototyping stage. Our proposed ventilator is also capable of Advanced Synchronized intermittent mandatory ventilation (SIMV) mode and a basic non-invasive operation mode where a fixed pressure is made available to the patient. Our design also provides PEEP (Positive End-Expiratory Pressure), which is not a ventilatory mode in itself but is designed to support steady low positive pressure to the lungs [5].

Dr. Munish Chhabra
Professor & Head
Deptt. of Mechanical Engg.
Moradabad Institute of Technology

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Dr. Munish Chhabra
Professor & Head
Deptt. of Mechanical Engg.
Moradabad Institute of Technology
Moradabad - 244001

CHAPTER 1

INTRODUCTION

The last few months have seen an increased demand for ventilators in the treatment of patients with COVID-19, a fact that led to a ventilator shortage worldwide. The consequence of this shortage is calamitous, especially in deprived areas [1]. Even well-equipped hospitals have developed protocols for sharing the same ventilator between two patients a dubious practice (e.g.)—because it opens up the possibility of not only sharing bacterial and viral load among patients, but also provoking untoward harm. As an attempt to face the worldwide problem of ventilator shortage, researchers have started an initiative of producing low-cost, open-source ventilators. This paper contributes to this initiative [2] [3].

Researchers agree that mechanical ventilation can harm lungs, provoking a condition known as ventilation-induced lung injury (e.g.). The two most common types of damage are polytrauma and atelic trauma. Vol trauma appears when the ventilation in excess distends the airways and alveoli, causing over-stretching of the corresponding lung parenchyma. Polytrauma causes an inflammatory reaction, which eventually leads to the rupture of the alveolar walls and edema. Atelic trauma, in contrast, seems to be caused by insufficient ventilation; poor ventilation allows alveolar units to collapse and reopen, in a repetitive, sequential movement, which may lead to injury as well [2]. To avoid atelic trauma, most researchers recommend clinicians doing mechanical ventilation with positive end-expiratory pressure (PEEP). PEEP has become a ubiquitous tool to prevent atelic trauma, gaining evidence after the seminal results in.

Respiratory diseases and injury-induced respiratory failure constitute a major public health problem in both developed and less developed countries. Asthma, chronic obstructive pulmonary disease and other chronic respiratory conditions are widespread. These conditions are exacerbated by air pollution, smoking, and burning of biomass for fuel, all of which are on the rise in developing countries1,2 Patients with underlying lung disease may develop respiratory failure under a variety of challenges and can be supported mechanical ventilation. These are machines which mechanically assist patients inspire and exhale, allowing the exchange of oxygen and carbon dioxide to occur in the lungs, a process referred to as artificial respiration [4].

Fig – 1.1 Mechanical Ventilator

Dr. Munish Chhabra
Professor & Head

CHAPTER 2

LITRATURE REVIEW

Mechanical ventilation is an essential life-saving technology. There are, however, numerous associated complications that influence the morbidity and mortality of patients receiving intensive care. Therefore, it was essential to use the safest and most effective form of ventilation for the shortest possible duration. Because of the potential complications and costs of mechanical ventilation, research to date have focused on accurate weaning readiness assessment, methods and organizational aspects that influence the weaning process [2].

In early 2005, the literature was reviewed from 1986 to 2004 by accessing the following databases: Medline, ProQuest, Science Direct, CINAHL, and Blackwell Science. The keywords mechanical ventilation, weaning, protocols, critical care, nursing role, decision-making and weaning readiness were used separately and combinations [3].

Two authors (J.M.K. and N.S.) searched MEDLINE, PubMed, and Google Scholar from the year 2000 through July 2018 using the following key words: mechanical ventilation education, mechanical ventilation training, graduate medical education, housestaff, resident, and fellow. No additional publications were identified after reviewing the references from identified articles. Studies were included if they were published in English-language, peer-reviewed journals and the abstracts described instruction, assessment, or opinions pertaining to MV in GME. Articles were excluded if they focused on non-GME learning groups or if no description of educational intervention or assessment was included [5]. The full text of each article was reviewed by the coauthors to confirm inclusion.

Controversy exists in weaning practices about appropriate and efficacious weaning readiness assessment indicators, the best method of weaning and the use of weaning protocols. Arguably, the implementation of weaning protocols may have little effect in an environment that favors collaboration between nursing and medical staff, autonomous nursing decision-making in relation to weaning practices, and high numbers of nurses qualified at postgraduate level. In summary, this paper presents two main findings: (i) construction of a low-cost, open-source mechanical ventilator for patients with COVID-19, and (ii) deployment of a method that monitors the pressure in the patient's lungs and acts on the machinery accordingly. Experimental data support the potential of these findings, as detailed in the sequence [4].

Further research is required that better quantifies critical care nurses' role in weaning practices and the contextual issues that influence both the nursing role and the process of weaning from mechanical ventilation.

Dr. Munish Chhabra
Professor & Head
Deptt. of Mechanical Engg.

CHAPTER 3

METHODOLOGY

3.1. DEVICE DESIGN

a). Air Delivery Technique-: Two main strategies were identified for the ventilator's air delivery system. One strategy 3 uses a constant pressure source to intermittently deliver air while the other delivers breaths by compressing an air reservoir. The latter approach was adopted as it eliminates the need for the continuous operation of a positive pressure source. This reduces power requirements and the need for expensive and difficult to repair pneumatic components. Where most emergency and portable ventilators are designed with all custom mechanical components, we chose to take an orthogonal approach by building on the inexpensive BVM, an existing technology which is the simplest embodiment of a volume-displacement ventilator. Due to the simplicity of their design and their production in large volumes, BVMs are very inexpensive (approximately $10) and are frequently used in hospitals and ambulances. They are also readily available in developing countries. Equipped with an air reservoir and a complete valve system, they inherently provide the basic needs required for a ventilator [4] [5] [7].

b). Compression Mechanism-: The most obvious means to actuate a BVM is to mimic the hand motion for which the bag was designed. This requires the use of linear actuation mechanisms (e.g., lead screw or rack and pinion) which despite being simple to implement, require linear bearings and extra space. Other compression techniques were sought to take advantage of the cylindrical BVM shape. However, since BVMs were designed for manual operation, their compressible outer surface is made from high-friction material to maintain hand-contact with minimal slippage. This eliminates the option of tightening a strap wrapped around the bag as a means of actuation. To avoid the problems associated with high surface friction, the two main candidates for actuation were a roller chain and cam compression. These options employ rolling contact with the bag rather than sliding contact, eliminating losses due to kinetic friction between the actuator and the bag [5].

c). Roller-chain Concept-: The roller-chain concept utilizes roller-chains with roller diameters larger than link width. (Figure 3.1A) The chain wraps around the circumference of the bag, and as a result is very space efficient. A sprocket connects to the motor shaft; its clockwise/anticlockwise rotation compressing and expanding the bag for breath delivery [6].

While this idea seemed initially feasible, preliminary experiments revealed that radial compression of a BVM requires significantly higher force than the vertical compression for which the bag was designed. Additionally, its operation was noisy, and the bag crumpled under radial compression, inhibiting the desired pure rolling motion, and preventing an accurate and repeatable tidal volume from being delivered.

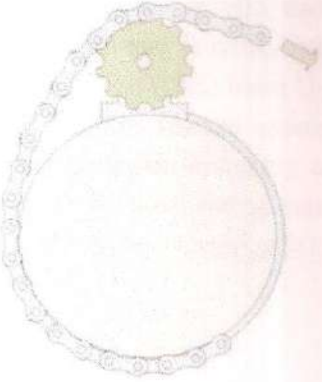

Fig – 3.1A Roller Chain Device [6]

A trade-off was encountered; while small pitch/roller-diameter chains are more space efficient and yield higher angular resolution for compression, bag crumpling becomes an issue. On the other hand, a higher pitch/roller-diameter chain overcomes crumpling but takes up more space and decreases angular resolution. In either case, the use of roller chains added a significant amount of weight to the system suggesting the need for a more effective mechanism needing a smaller contact area [7].

d). CAM Concept-: The cam concept utilizes a crescent-shaped cam to compress the BVM, which allows smooth, repeatable deformation to ensure constant air delivery (Figure 3.1B). As it rotates, the cam makes a rolling contact along the surface of the bag and unlike the roller-chain, achieving low-noise of operation. By controlling the angle of the cam's shaft, the amount of air volume delivered can be accurately controlled. The cam mechanism was found to be more space efficient and have a lower power requirement than the roller chain concept, and was therefore the method of choice [8].

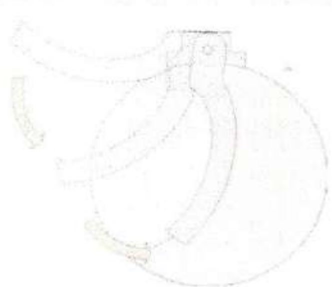

Fig – 3.1B CAM System [9]

3.2 PROTOTYPE DESIGN

With the cam concept selected as the best method for BVM compression, an initial prototype was built to measure force and power requirements. The enclosure's frame consists of four vertically mounted sections of ½" (12.5 mm) clear acrylic, each mounted on the bottom section of the frame with interlocking mates. The material is easily laser cut and allows for visibility of the internal components. The two inner sections (ribs) have U-shaped slots made to conform to BVM's surface contour, while the end sections feature openings to accommodate the BVM's valve neck and oxygen reservoir. The pivoting cam assembly was mounted on top of the hinged aluminum lid, and consisted of two 2.5" (63.5 mm) radius crescent-shaped pieces attached to a 5/16" (8 mm) aluminum shaft mounted with nylon bushings [4].

3.3 CONTROLERS

A. **Parameters-:** The operator adjusts tidal volume, breath rate, and inspiratory to expiratory time ratio using three continuous analog knobs mounted to the outside of the ventilator. The prototype has a range of 200-750 mL tidal volume and 5-30 breaths per minute (bpm). This yields a maximum minute ventilation (Ve) of 21L and a minimum Ve of 1.5L. However, these values do not reflect the limits of the final design only the settings of the prototype. Theoretically, the ventilator is able to deliver anywhere from 0L minute volume to 60L minute volume. However, this has not been fully tested. I:E ratio was not implemented on the prototype but theoretically could have any desired range within the limits determined by the other parameters. The ranges on the final design will be determined in consultation with respiration specialists to allow for the broadest range of safe settings [8].

B. **Controlling Circuit-:** An off-the-shelf Arduino Demilune microcontroller board was selected to control our device. The microcontroller runs a simple control loop to achieve user-prescribed performance. The control loop is triggered by the internal timer set by user inputs, with the inspiratory stroke initiated at the beginning of the loop. Once the prescribed tidal volume is reached, the actuator returns the cam back to its initial position and holds until the next breath [2] [4]. The loop then repeats to deliver intermittent breaths. If the loop is interrupted by a breath attempt by the patient (sensed through the pressure sensor), the ventilator immediately delivers a breath, interrupting the loop and resetting the timer. A diagram showing this loop is found in Figure 3.3A [2] [3] [6].

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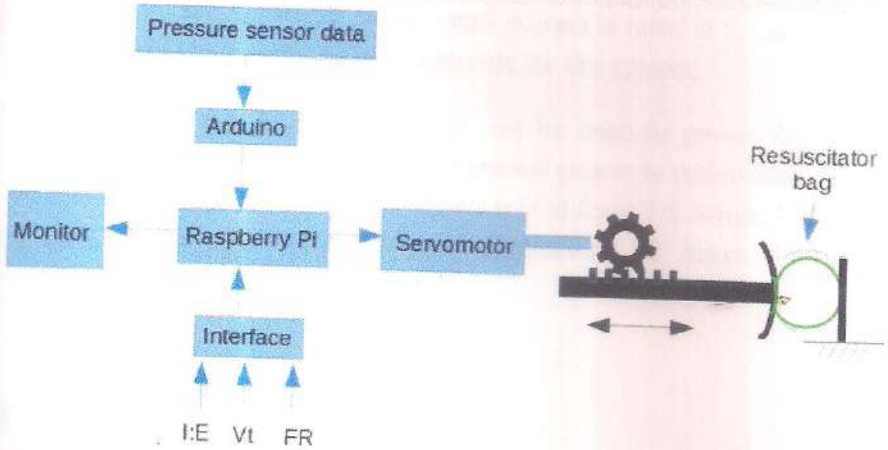

Fig – 3.3A Ventilator Control Circuit [5]

C. Motor-: According to initial experiments, a maximum torque of 1.5 Nm was required for maximum volume delivery. A PK51 DC gearmotor with a stall torque of 2.8 Nm was selected for the prototype. Despite the lower torque value measured in our experiment, we found that this motor did not provide quite enough torque to effectively drive the cam at the slower inhalation cycle rates prescribed to some patients. While a larger motor will be necessary to achieve better speed control, this motor functioned acceptably at the proof-of-concept phase. It was desirable for its gear reduction ratio of 51:1, and an operating speed in the required range of 50-70 rpm [1] [3].

Fig – 3.3B DC Servo Motor

D. Motor Driver-: The motor driver comprises of two H-Bridge circuits. These circuits direct current through the motor in opposite directions, depending on which set of switches on the circuits are energized. Speed of the motor is signaled with a pwm pin. The power is supplied directly from the battery, so the only limit is the current capability of the chip and battery. We opted to use the Solarbotics® motor driver, which is capable of supplying 5 amps of current to the two circuits [1] [2] [3]. The PK51 motor's stall current is rated at 5.2 amps which means the motor driver will be able to handle the requirements for the system.

E. **POWER SUPPLY-:** An AC/DC converter can be used to power the ventilator directly from a wall outlet or a vehicle inverter. When external power is unavailable, the ventilator can run off of any battery capable of delivering 12-15 volt at least 3.5 Amps. For the prototype, we used a 14.8 volt, four-cell Li-Ion battery pack capable of 4.2 Amps (limited by protective circuitry), with a capacity of 2200 mA-hr.

3.4 EXPIREMENTS

A. **Frist Expirement Setup-:** A bench level experiment was conducted on the first prototype to determine performance characteristics. Data was collected using our prototype's cam mechanism to compress an adult sized BVM. The test apparatus included a spirometer to measure flow-rate (and volume, by integrating over time), a hand dynamometer to measure force exerted on the cam, a rotary motion sensor to measure cam angular displacement, and a pressure sensor to measure internal air pressure. The experimental setup is depicted in Figure 3.4A.

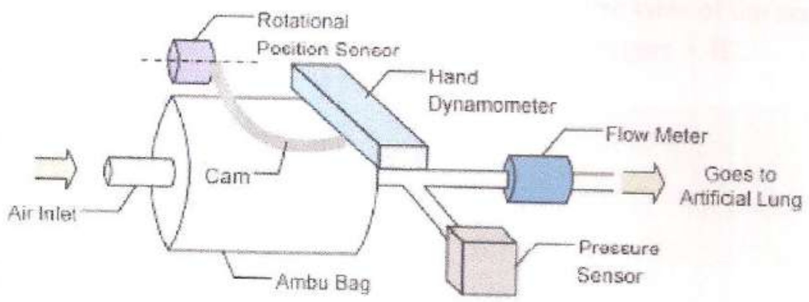

Fig – 3.4A CAM System Expiremental Setup [1] [3]

The air volume delivered was measured as a function of cam angle by integrating the flow rate over time. Results indicated that the volume delivered versus cam angle relation is approximately linear (Figure 3.4B). Data analysis showed that the peak power required was 30 W, and maximum torque was 1.5 Nm. The maximum volume delivered per stroke was approximately 750 mL. The target tidal volume is 6-8 ml/kg for adult human use, so this is adequate for most clinical situations.

Dr. Munish Chhabra
Professor & Head
Deptt. of Mechanical Engg.
Moradabad Institute of Technology
Moradabad - 244001

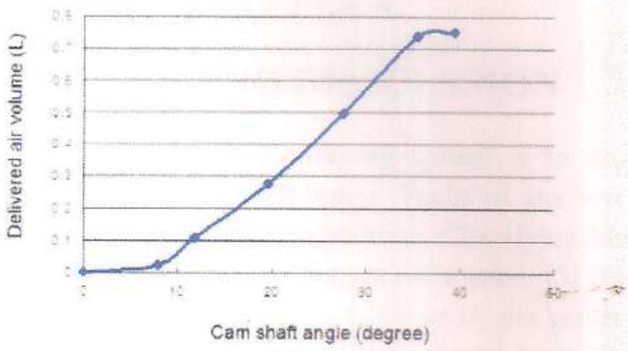

Fig – 3.4B Air volume and CAM angle graph

B. **Second Expirement Setup-:** A second prototype was built in which all the moving components were moved inside the enclosure. Enclosure dimensions were increased to accommodate the cam arm's range of motion, and to make space for the motor, microcontroller and battery pack. The enclosure's lid was made of acrylic, and hinges from the side of the unit to better constrain the top of the bag. The support ribs inside the enclosure also serve as mounting blocks for camshaft bushings. A potentiometer was coupled to the end of the shaft for use as a position feedback sensor. An isometric view of the second prototype's CAD model is shown in Figure 3.4C, and the built device in Figure 3.4D.

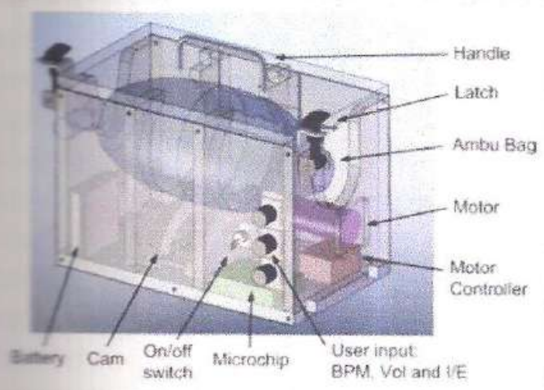

Fig – 3.4C 3D Drawing of Second setup

Fig – 3.4D Second Prototype

Dr. Munish Chhabra
Professor & Head
Deptt. of Mechanical Engg.
Moradabad Institute of Technology
Moradabad - 244001

CHAPTER 4

RESULTS AND DISSCUSSION

The main objective of a pulmonary health monitoring system is to determine whether the processed data comes from healthy or unhealthy lungs. Faults of any sort in the equipment's operation should also be classified as an unhealthy condition. Classifying whether the data comes from a healthy or unhealthy condition is vital for the patient's safety [5] [9]. A trained clinician usually does this classification and does take action when he or she perceives something went wrong. However, this practice is prone to errors, since he or she might be under job stressors like work-life imbalance, sleep deprivation, burnout, among others. One likely reason for this misinterpretation is that signals collected from healthy and unhealthy patients resemble the same. To take a grasp of how this similarity works, we carried out experiments that emulated the mechanical ventilator working under both circumstances, i.e., the healthy and unhealthy conditions. The unhealthy conditions were emulated by adding small objects upon the bag. Two scenarios were considered, say faulty cases A and B[4] [9]. They correspond to two distinct objects kept attached to the middle of the bag's top part while it was inflated and deflated. Faulty case A corresponds to a car-key case shell that weighs about 48 grams, and faulty case B corresponds to a 2-Euro coin that weighs about 8.5 grams. For both cases, by adding weights in the top of the bag, we intended to mimic unhealthy lungs, which required increased pressure to keep the tidal volume as previously set in the equipment.

All experiments performed in **the laboratory OF SAI HOSPATIAL, MORADABAD** were taken for the I:E fixed at 1:2, Vt at about five breaths per minute, and FR at about 350 mL. A sample of the experimental data is shown in Figure 4.1. As can be observed in Figure 4.1, the signals from the healthy and unhealthy conditions seem similar, a fact that may mislead clinicians when trying to make a classification. To improve the classification of what those signals are about required carefully-designed algorithms. This paper contributes to this path—which is yet far from complete—showing a novel algorithm for mechanical ventilators. Additionally, In Figure 4.1 it is noticed that there is about 0.02 Volts variability between consecutive breaths. From the engineering point of view, this variability may be related to a low frequency response to a periodic excited elastic-system, the lung. This also validates the sensitivity of the used pressure sensor [8].

The main idea behind our classification method is as follows. The simplest information that can be drawn from a signal is its average (i.e., statistical mean value). The average can be calculated in real-time, during the equipment's operation, through a simple linear low-pass filter. A side effect it produces, however, is that known as drift—drift means that the signal's magnitude goes to infinity as the time evolves [8] [9]. A strategy to eliminate this drift effect is by resetting the low-pass filter from time to time, a sequential procedure inspired in the Clegg-integrator philosophy. The Clegg-integrator system is a stable system with the output set to zero—when required. Here the discrete-time version of the Clegg integrator reads as

$$x(k+1) = \begin{cases} 0, & |d(k)| > th \\ x(k) + h[-a(x(k) - d(k))], & |d(k)| \leq th, \end{cases}$$

[1] [5]

where d(k) denotes the one-dimensional input data to be processed, x(k) stands for the filter output, and a, t, and h denote positive filter constants. In this paper, d(k) accounts for the data collected from the pressure sensor

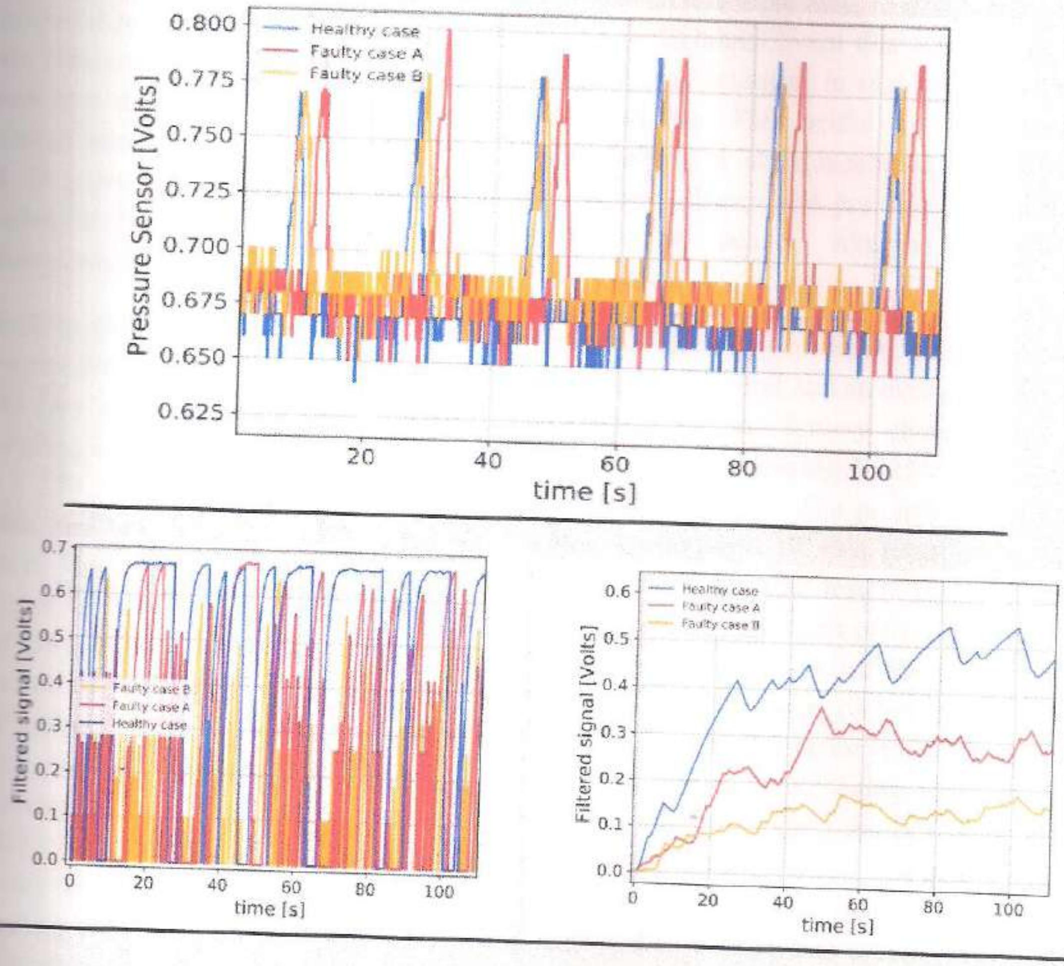

Fig – 4.1 Result of different expirement

CHAPTER 5

CONCLUSION

Since the start of the COVID-19 pandemic, researchers have been striving to help society face many problems caused by this pandemic. Among the recent initiatives, one has drawn the authors' attention: producing low-cost, open-source mechanical ventilators. The motivation comes from the worldwide shortage of mechanical ventilators in the treatment of COVID-19 patients—mechanical ventilators keep severely ill patients alive. This paper contributes to this initiative [5] [7]. This paper has detailed the construction of a functional, low-cost, and open-source mechanical ventilator. The authors' contribution to this topic aims to mitigate the effects of this worldwide ventilator's shortage—a shocking, unfortunate event that hits hard deprived areas. This paper has shown a numerical method that can monitor, in real-time, whether the patient has a healthy or unhealthy pulmonary condition. This useful yet straightforward numerical method opens up the possibility of applications in other mechanical ventilators as well. In summary, this paper contributes to both fronts—theory and practice. Alarms can be included in this project, using either an alarm screen or speakers, like the ones that alert clinicians when the pressure reaches some threshold values.

A working prototype that can be operated on a test lung has been developed. The prototype has user-controlled breath rate and tidal volume. It features assist control and an over-pressure alarm. It has low power requirements, running for 3.5 hours on one battery charge at its most demanding setting. It is portable, weighing 9 lbs (4.1 kg) and measuring 11.25 x 6.7 x 8 inches (285 x 170 x 200 mm) , and has a handle and easy to use latches. The prototype can display settings and status on a computer screen. Further development of this proof-of-concept is planned. Future iterations will incorporate changes prompted by the results of our prototype testing. It will incorporate an adjustable inspiratory to expiratory ratio, an option missing in this prototype due to its underpowered motor. We will investigate the effects that changing the motor will cause to cost, weight and battery life. We will also incorporate add_on features including a PEEP valve, a humidity exchanger and a blow-off valve. [5] [6] Since BVM infrastructure already supports commercial add_ons, these components can be easily purchased and incorporated. Ways to minimize deadspace will be explored, including the option of using a Laerdal® brand BVM whose valves can be placed at the patient end of the tubing. In later iterations we hope to be independent from Laerdal® by manufacturing our own bags or contracting their production. The design will be changed to be injection molded such that the mass-produced a version would cost less than $200 to produce. Weight will be minimized and battery-life extended. Consideration to a pediatric version will also be given. Cam arm shape will be optimized to ensure the use of the most efficient rolling contact embodiment. An LCD screen will be included, and alarms programmed for loss of power, loss of breathing circuit integrity and low battery life. Extensive testing of the ventilator's repeatability will be conducted. Finally, we will test the ventilator on a lung model to meet ventilator standards and market the product.

Dr. Munish Chhabra
Deptt. of Mechanical Engg.
Institute of Technology

APPENDIX-A

Python program for Resberry pie [10]-:

```
import time

import RPi.GPIO as GPIO from os

import _exit import serial

import numpy as np

import matplotlib.pyplot as plt

GPIO.setmode(GPIO.BOARD)

#LE

GPIO.setup(7,GPIO.IN)

#FreR

GPIO.setup(3,GPIO.IN)

GPIO.setup(5,GPIO.IN)

GPIO.setup(11,GPIO.IN)

GPIO.setup(13,GPIO.IN)

#VT

GPIO.setup(15,GPIO.IN)

GPIO.setup(19,GPIO.IN)

GPIO.setup(21,GPIO.IN)

GPIO.setup(23,GPIO.IN) k=0 h=0.15

GPIO.setup(32,GPIO.OUT)

freqPWM=50 pwm=GPIO.PWM(32,freqPWM)

pwm.start(4) ser=serial.Serial("/dev/ttyACM0",9600)

time.sleep(2)

xs=[]
```

Dr. Munish Chhabra
Professor & Head
Deptt. of Mechanical Engg.
Moradabad Institute of Technology
Moradabad - 244001

```
ys=[]
f=open("Datos3.txt","w")
j=0
x1a=0.5
while True:
x=ser.readline()
x1=x.rstrip()
if x1==b'':
x1=x1a try:
float(x1) x1=x1
except: x1=x1a i
if float(x1)>1: x1=x1a
x1a=x1
x2=float(x1)
#print('Presión= ',x2)
xs.append(k*h)
ys.append(x2)
f.write(str(round(k*h,3))+'\t')
f.write(str(round(x2,3))+'\n')
if(GPIO.input(7)==True):
aux1=3
if(GPIO.input(7)==False):
aux1=2
#Vt
```

Dr. Munish Chhabra
Professor & Head
Deptt. of Mechanical Engg.
Moradabad Institute of Technology
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```
if(GPIO.input(3)==False&GPIO.input(5)==False&GPIO.input(11)==False\
&GPIO.input(13)==False):

aux2=0.7

if(GPIO.input(3)==True&GPIO.input(5)==False&GPIO.input(11)==False\
&GPIO.input(13)==False):

aux2=0.8

if(GPIO.input(3)==True&GPIO.input(5)==True&GPIO.input(11)==False\
&GPIO.input(13)==False):

aux2=0.9

if(GPIO.input(3)==True&GPIO.input(5)==True&GPIO.input(11)==True\
&GPIO.input(13)==False):

aux2=1

#FR

if(GPIO.input(15)==False&GPIO.input(19)==False&GPIO.input(21)==False\
&GPIO.input(23)==False):

fr=5

fr=5 if(GPIO.input(15)==True&GPIO.input(19)==False&GPIO.input(21)==False\
&GPIO.input(23)==False):

fr=10

if(GPIO.input(15)==True&GPIO.input(19)==True&GPIO.input(21)==False\
&GPIO.input(23)==False):

fr=15

if(GPIO.input(15)==True&GPIO.input(19)==True&GPIO.input(21)==True\
&GPIO.input(23)==False):

fr=20

if(GPIO.input(15)==True&GPIO.input(19)==True\
&GPIO.input(21)==True&GPIO.input(23)==True):

fr=5
```

Dr. Munish Chhabra
Professor & Head
Deptt. of Mechanical Engg.
Moradabad Institute of Technology
Moradabad - 244001