

Article

Low-Cost, Open-Source Mechanical Ventilator with Pulmonary Monitoring for COVID-19 Patients

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Abstract: This paper shows the construction of a low-cost, open-source mechanical ventilator. The motivation for constructing this kind of ventilator comes from the worldwide shortage of mechanical ventilators for treating COVID-19 patients—the COVID-19 pandemic has been striking hard in some regions, especially the deprived ones. Constructing a low-cost, open-source mechanical ventilator aims to mitigate the effects of this shortage on those regions. The equipment documented here employs commercial spare parts only. This paper also shows a numerical method for monitoring the patients' pulmonary condition. The method considers pressure measurements from the inspiratory limb and alerts clinicians in real-time whether the patient is under a healthy or unhealthy situation. Experiments carried out in the laboratory that had emulated healthy and unhealthy patients illustrate the potential benefits of the derived mechanical ventilator.

Keywords: mechanical ventilator; low-cost ventilator; COVID-19; pressure sensor; artificial ventilation; health monitoring

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 [1–3]. The consequence of this shortage is calamitous, especially in deprived areas [4–7]. Even well-equipped hospitals have developed protocols for sharing the same ventilator between two patients [8]—a dubious practice (e.g., [9–11])—because it opens up the possibility of not only sharing bacterial and viral load among patients [12], but also provoking untoward harm [9]. As an attempt to face the worldwide problem of ventilator shortage, researchers have started an initiative of producing low-cost, open-source ventilators [13–17]. This paper contributes to this initiative.

Researchers agree that mechanical ventilation can harm lungs, provoking a condition known as ventilation-induced lung injury (e.g., [18–24]). The two most common types of damage are volutrauma and atelectrauma [20]. Volutrauma appears when the ventilation in excess distends the airways and alveoli, causing over-stretching of the corresponding lung parenchyma [21,22]. Volutrauma causes an inflammatory reaction, which eventually leads to the rupture of the alveolar walls and edema. Atelectrauma, 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 [18,20,21,24]. To avoid atelectrauma, most researchers recommend clinicians doing mechanical ventilation with positive end-expiratory pressure (PEEP). PEEP has become a ubiquitous tool to prevent atelectrauma [18,19,24], gaining evidence after the seminal results in [18]. However, using PEEP

to reduce lung injury is debatable since recent findings suggest that PEEP causes other harm like lung inflammation and edema formation [23]. These investigations together indicate that the full understanding of the mechanical ventilation process requires more research (e.g., [25])—yet mechanical ventilators undoubtedly help patients suffering from acute respiratory problems [26,27].

Since mechanical ventilators potentially expose the patient's lungs to damage, all initiatives of constructing low-cost mechanical ventilators must provide the regulation of not only the lung's pressure but also the positive end-expiratory pressure (PEEP)—two points for concern [28]. The first point involves regulating the machine to prevent excessive pressure, which is a side effect of the surplus of energy from the ventilator machinery [29]. We developed a novel method that monitors the patient's pulmonary condition to mitigate the chance of occurring those undesired spikes in the lung's pressure. The sensor of pressure, which is attached in the inspiratory limb, feeds a carefully-designed algorithm. This algorithm combines the Clegg-integrator system [30] with a low-pass filter in discrete-time. This algorithm applied in a mechanical ventilator represents a novelty. Experimental data support the usefulness of this algorithm.

The second point for concern on the ventilator's design is that of assuring PEEP. A limitation of the mechanical ventilator shown here is that it does not account for PEEP. Yet PEEP valves are commercially available and can be readily adapted in the inspiratory limb [31]. For this reason, the no-PEEP-monitoring condition does not prevent the use of this low-cost ventilator in clinical trials.

This paper's main contribution is to describe the construction of a low-cost, open-source mechanical ventilator for patients with COVID-19. Recall that technology can be considered as low cost if the corresponding deployment can be produced by using inexpensive spare parts [32–34]. An electronic device that has been largely used in these low-cost initiatives is the Raspberry-Pi, a versatile small-size single-board computer [35–37]. The Raspberry-Pi is a critique tool in this paper since it acts as the “brain” of the machinery, as detailed in the sequel.

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.

2. Materials and Methods

To be used in intensive care units, mechanical ventilators must comply with the following three guidelines (e.g., [38,39]): they should allow users to set (i) the respiratory frequency (FR); (ii) the ratio of inspiration-to-expiration at each respiratory cycle (I:E); and (iii) the air volume supplied to the patient (V_t).

Those guidelines were fulfilled by the mechanical ventilator described in this paper, as indicated in the schematic shown in Figure 1. As can be seen, the ventilation apparatus hinges upon the parameters FR, I:E, and V_t , which should be set by the clinician. The air and oxygen supply flows through the corresponding limb and reach the patient's lungs.

Remark 1. *When clinicians increase the I:E ratio in mechanical ventilators, they indeed increase the inspiration time within the respiratory cycle, which means that the lung receives more oxygen. Prolonged high levels of oxygen can cause alveolar overstretching, which triggers volutrauma.*

Figure 2 depicts the electronics and mechanical actuators elements. They are connected with the parts shown in the scheme of Figure 1. Finally, Figure 3 shows a photo of the mechanical ventilator assembled in the laboratory.

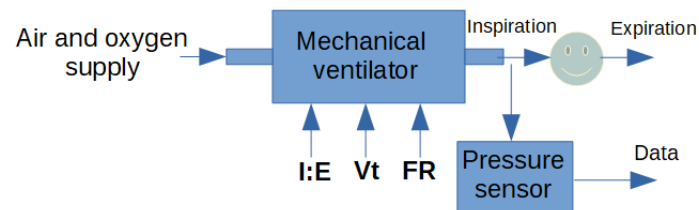


Figure 1. Scheme of the mechanical ventilator. The system depends on the signals I:E, Vt, and FR, provided by the clinician. The device adjusts the airflow to the patient's lungs accordingly.

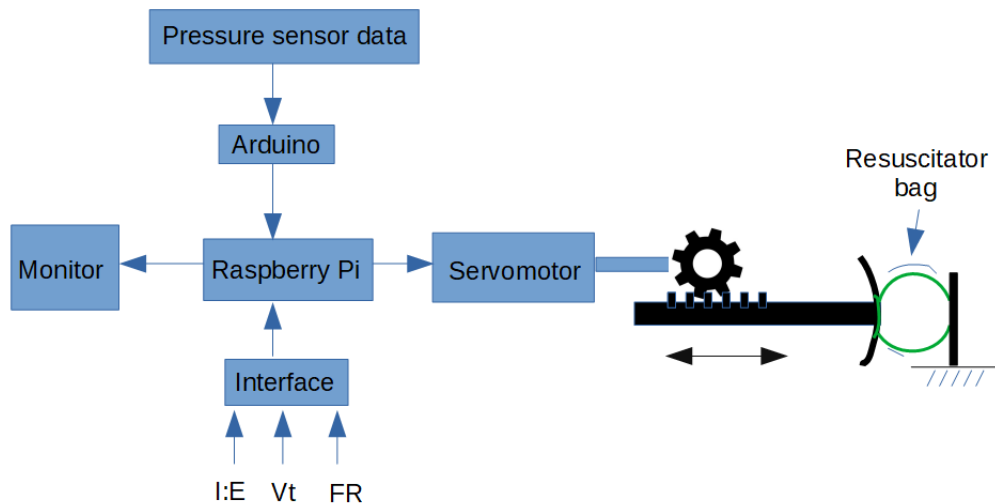


Figure 2. Electronic and mechanical scheme. An Arduino board collects data from the pressure sensor and sends them to the Raspberry Pi. The Raspberry Pi commands the actuator, and its shaft compresses the resuscitator bag accordingly.

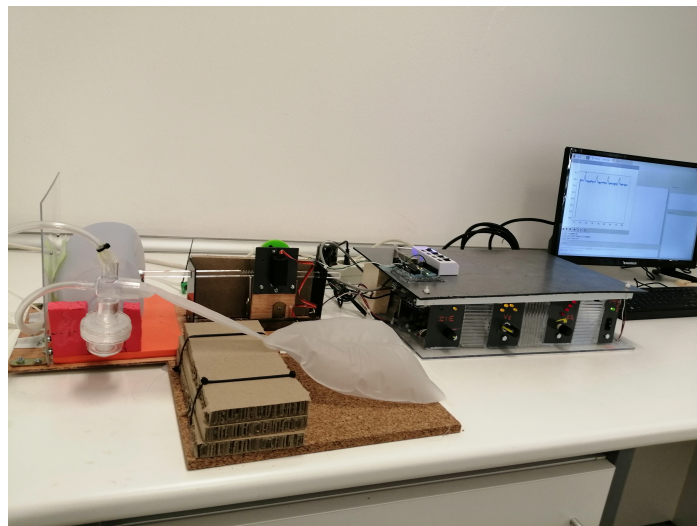


Figure 3. A picture of the experimental mechanical ventilator.

2.1. Instrumentation

Figure 4 presents the electronic instrumentation for the pressure sensor. According to the manufacturer, this pressure sensor can measure maximum differential pressure of up to 70 cm H₂O. This sensor produced measurements that were proportional to voltage, according to the next equation.

$$\text{Measured_Pressure}[cmH_2O] = 21.8(\text{Output_Pressure_Sensor_Voltage}[\text{Volts}] - 0.4). \quad (1)$$

It is worth pointing out that the pressure sensor was located at the mechanical ventilator's expiratory limb, as suggested in the literature [33].

The pressure sensor was connected in the circuitry in such a way that it sent measurements to an Arduino Uno board, which processes them and transmits the processed data to the Raspberry Pi. The Raspberry Pi takes these measurements and feeds the pulmonary monitoring algorithm, as described in the sequence.

Figure 5 shows the electronic circuit interface from the Raspberry Pi to the servomotor. Recall that a servomotor is a high-precision mechanical-rotary position system that works by controlling an internal DC-motor [40]. The servomotor used in this project is the FS5109M from the Feetech manufacturer. It worked with input signals given in the form of pulse-width modulation (PWM). According to its manufacturer, the servomotor operation frequency was fixed at 50 Hz.

The gear was homemade using polychloride vinyl. The gear was then attached to the servomotor rod. The rod was made by a Plexiglas bar. The radius of this gear is about 2.5 cm, and each gear's teeth was separated from each other in about 1 degree. The Plexiglas bar was displaced in about 7.8 cm, enough to compress the airbag at 'full' capacity. The total length of the rod bar is about 30 cm.

Figure 6 shows the electronic circuit interface to Raspberry Pi for the user command I:E (the ratio of inspiration and expiration). The clinician can set the I:E value through a rotary switch positioned at either 1:2 or 1:3.

Figure 7 shows the electronic circuit interface to Raspberry Pi for the user respiratory frequency command FR. The clinician can select the following FR values: 5, 10, 15, or 25 breaths per minute. Some commercial mechanical ventilators allow clinicians to select this value from the continuous interval [5, 25]. However, for sake of simplicity, we decided to let the FR values be selected within those discrete values only.

Figure 8 shows the electronic circuit interface to Raspberry Pi for the user command V_t (air volume supplied to the patient). The V_t value depends on the patients' age, body weight, and other factors [8].

Appendix A shows the Python code programmed into the Raspberry Pi.

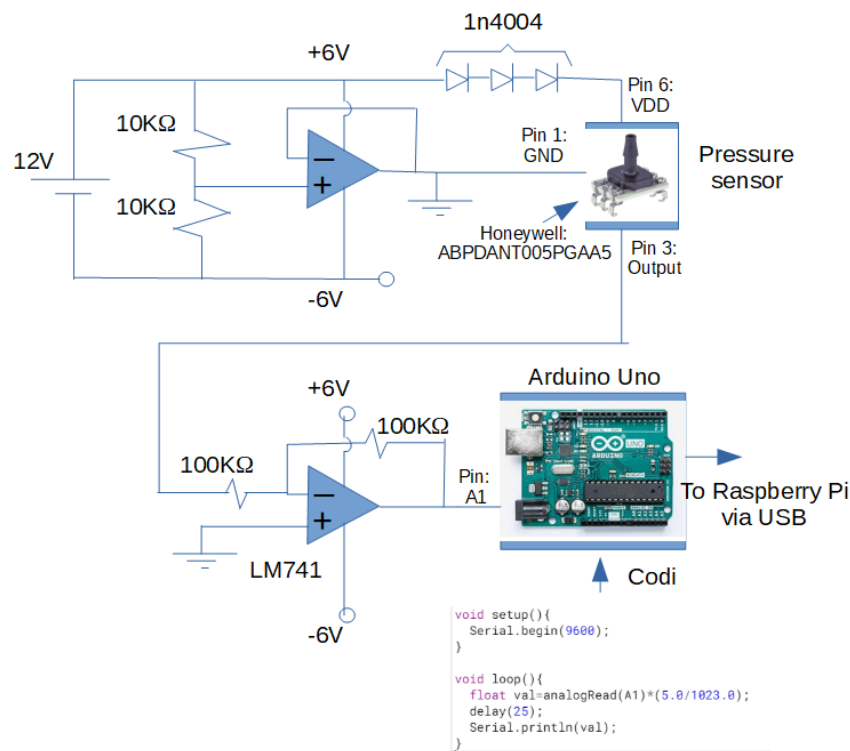


Figure 4. Electronic circuit: pressure sensor instrumentation.

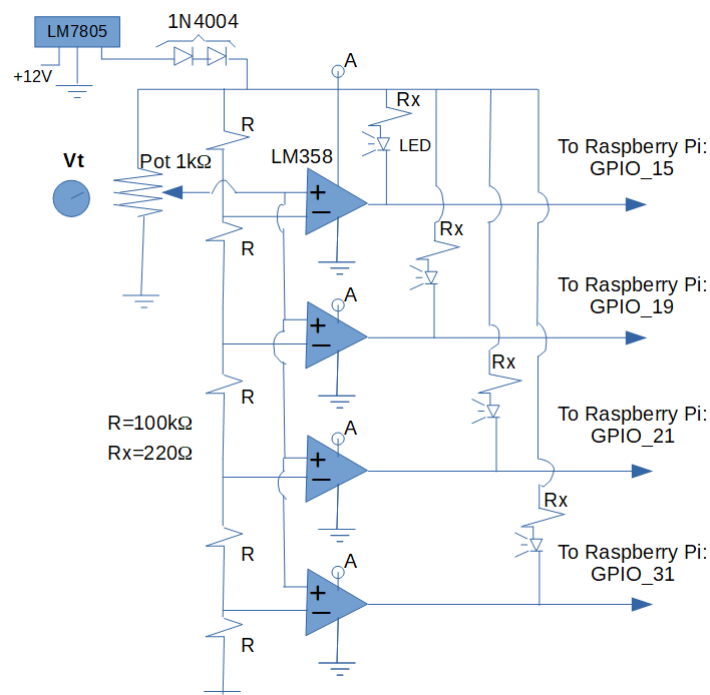


Figure 8. Electronic circuit: Vt user interface.

Remark 2. This low-cost mechanical ventilator uses as a key element the self-inflating bag (i.e., air-bag resuscitator) known as emergency “Ambu bag”. The manual of the Ambu bag informs that it contained an air volume of 1475 mL, but measurements made in the laboratory indicated that it contained a volume up to 800 mL.

3. Results and Discussion

This section presents experimental data and discusses their implications for patients’ treatment.

3.1. Challenges of Pulmonary Health Monitoring

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.

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 [41,42]. 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. 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. A video showing this experiment was recorded and is available at www.youtube.com/embed/E_m5AOQadDg.

All experiments performed in the laboratory were taken for the I:E fixed at 1:2, V_t at about five breaths per minute, and FR at about 350 mL. A sample of the experimental data is shown in Figure 9. As can be observed in Figure 9, 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 9 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.

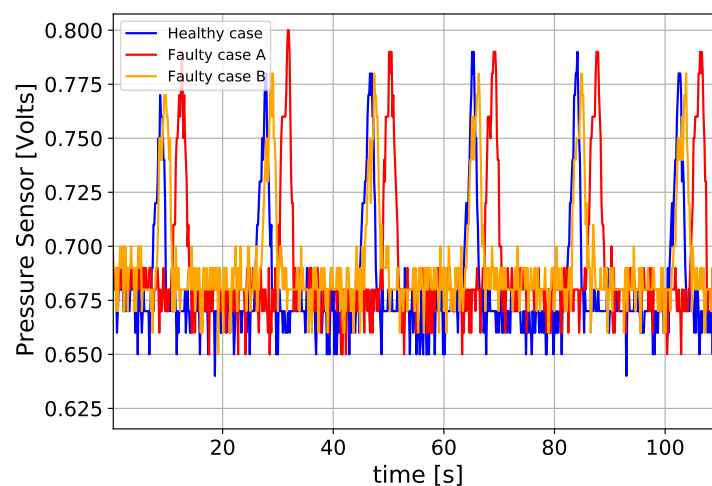


Figure 9. Experimental data from the pressure taken from the mechanical ventilator. The curve in blue represents data from the healthy condition. The equipment was adjusted to emulate healthy and unhealthy (faulty) conditions. The measure of pressure in cmH2O can be obtained through the equation in (1). The voltage range [0.625, 0.80] corresponds to the cmH2O range [4.905, 8.72].

3.2. Numerical Method for the Classification of Healthy and Unhealthy Conditions

The main idea behind our classification method is as follows. The most simple 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.

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 [43,44]. 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

$$x(k+1) = \begin{cases} 0, & |d(k)| > th \\ x(k) + h[-a(x(k) - d(k))], & |d(k)| \leq th, \end{cases} \quad (2)$$

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.

The data shown in Figure 9 were applied in the system (2) with $a = 1$, $t = 4.5333$, and $h = 0.15$, and the corresponding outcome is depicted in Figure 10. The jumps are due to the resetting actions of the Clegg integrator. This reset strategy is a crucial part of our lung monitoring system. Yet we recognize that Figure 10 does not give us a clue about what those signals really mean.

To extract a meaning from the data depicted in Figure 10, we applied them in the next first-order, low-pass filter

$$w(k+1) = w(k) + h(-a_1(w(k) - x(k))), \quad (3)$$

with $a_1 = 0.05$, which resulted in the data shown in Figure 11.

Figure 11 now clarifies the hidden meaning of Figure 10—the curves in Figure 11 allow us to discern what measurements came from the healthy condition from those that came from unhealthy conditions. Note in Figure 11 that the pressure associated with the healthy condition seems to reach a periodic, steady-state behavior around 0.5V—far above the others. In summary, the upper curve (in blue) indicates the healthy reference signal. Each novel signal, collected during the equipment's operation, should be compared with this reference value around 0.5 V. This simple comparison allows clinicians assess whether the novel, collected signal represents a healthy or unhealthy condition.

Remark 3. In practical terms, if real-time measurements show values diverging from the reference value, fixed at 0.5 V, the system should trigger an alarm to alert clinicians that an unhealthy condition has just started.

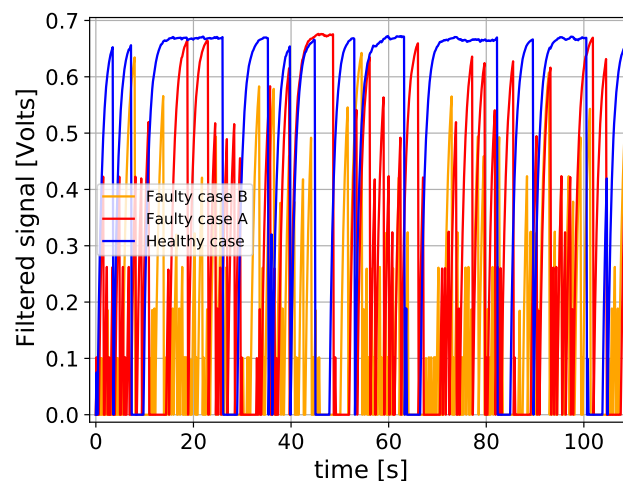


Figure 10. Curves depict data processed in the Clegg integrator of Equation (2). The curve in blue represents data from the healthy condition.

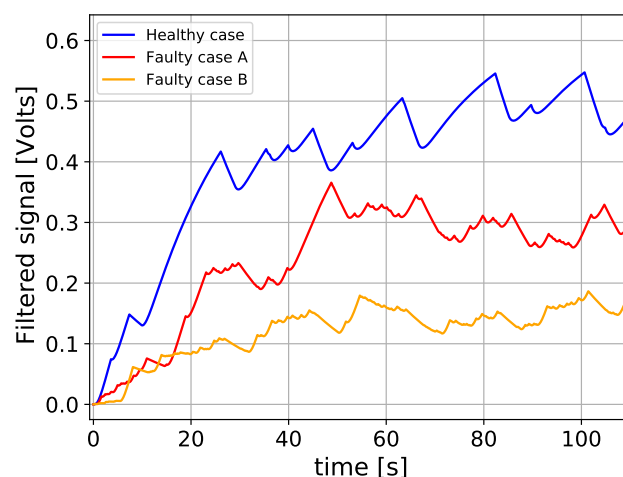


Figure 11. Final data sensor analysis stage coming from the low-pass filter. The curve in blue represents data from the healthy condition.

3.3. Limitations

This study has some intrinsic limitations, as described next.

Recent progress has been made on measuring—in real-time—how healthy are the lungs in mechanically ventilated patients [45–47]. The underlying idea is to continuously measure the lung's condition using sensors and other devices (e.g., [46,47]). These measurements then feed physiological models, which drive the mechanical ventilator in a way that it rearranges itself to control the patient's ventilation—all tasks done automatically, with no clinician's intervention at all. Even though automatic ventilation seems a future trend for the patient's safety, we did not include physiological models into the design of this low-cost mechanical ventilator—more sensors and devices would be required to do so. Moreover, the involved costs would increase drastically. We hypothesize that this inclusion could be done in our equipment in the future.

This low-cost, open-source mechanical ventilator has never been used in animals. For this reason, the results presented in this paper preclude their interpretation for in vivo lungs. Anatomy and physiology of the lungs cannot be oversimplified by an Ambu bag. For this reason, it is difficult to predict whether our algorithm will work in experiments involving in vivo lungs.

Moreover, recent research indicates that critically ill patients require tidal volume adjusted at around 6 mL per kilogram of predicted body weight [48]. To date, it is unknown whether this low-cost mechanical ventilator can supply this tidal volume in patients.

Remark 4. *The experiments indicate that the three main variables, i.e., the respiratory frequency (FR), the ratio of inspiration-to-expiration at each respiratory cycle (I:E), and the air volume supplied to the patient (Vt), can be efficiently controlled in the laboratory. However, it is unclear whether this mechanical ventilator complies with the regulations issued by regulatory agencies like MHRA, UK, and FDA, USA. No attempts to enforce compliance with regulations have been made so far.*

In summary, the experimental data presented here were taken in a laboratory environment that considered electronic and mechanical parts only. Understanding the potential and consequences of using this low-cost mechanical ventilator in the lungs requires more research. Attempts to use it in lungs should be made with care.

4. Concluding Remarks

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 [1]. This paper contributes to this initiative.

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 ventilators shortage—a shocking, unfortunate event that hits hard deprived areas [4–7].

This paper has shown a numerical method that can monitor, in real-time, whether the patient has a healthy or unhealthy pulmonary condition (see Section 3.2 in connection). 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 ([15], section 2.5).

The team's knowledge gained while working in this project—like complete schematics and documentation—is made free at "www.anvargas.com/blog".

Author Contributions: The authors contributed equally to this work. L.A.—This author contributed to the initial research topic and was involved in the electronic system and review of the paper redaction, including the state of the art on low cost mechanical ventilators for Covid-19 patients; A.N.V.—This co-author was involved in the electronic circuits realization, review process on the state of the art for the Covid pandemia, and on the wiring and review of the document. Main issues on the reviews' comments were also directed by him for a correct and efficient reply to them; G.P.-V.—This co-author reviewed the medical issues of the paper presentation and the

experimental set-up to pressure sensing and mechanisms of the final part of the proposed mechanical ventilator. She also participated in the writing process. All authors have read and agreed to the published version of the manuscript.

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Conflicts of Interest: The authors declare no conflict of interest.

Appendix A

Table A1. Python Program in Raspberry Pi.

```
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)
#I:E
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=[]
ys=[]
f=open("Datos3.txt","w")
j1=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
if float(x1)>1:
x1=x1a
x1a=x1
x2=float(x1)
```

Table A1. Cont.

```

#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
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
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
#Servo-Motor
j1=j1+h
if j1<=(60/(aux1*fr)):
pwm.ChangeDutyCycle(8*aux2)
if j1>(60/(aux1*fr)) and j1<=(60/fr):
pwm.ChangeDutyCycle(3.0)
if j1 >(60/fr):
j1=0
time.sleep(h)
k=k+1
if k>10000*h:
pwm.ChangeDutyCycle(3.0)
pwm.stop()
GPIO.cleanup()
ser.close()
f.close()
break
pwm.ChangeDutyCycle(3.0)
plt.plot(xs,ys)
plt.show()
f.close()
-exit()

```

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