6-2020

Ventilator: An Analysis of the History, Design, and Current World Implications

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Ventilator: An Analysis of the History, Design, and Current World Implications

By

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Submitted in partial fulfillment

of the requirements for

Honors in the Department of Mechanical Engineering

UNION COLLEGE

June, 2020

ABSTRACT

SANTOS, JOANA Ventilator: An Analysis of the History, Design, and Current World Implications. Department of Mechanical Engineering, June 2020.

ADVISOR: Professor Rebecca Cortez

Due to the sudden COVID-19 pandemic, many countries around the world were not prepared to provide such large amounts of artificial mechanical ventilators to its citizens, causing a shortage around the globe. Artificial ventilators have been continuously developed and improved since the 1900s, and because of their importance in the healthcare industry, technological advancements continue to be made today. This thesis aims to guide its readers through the history of the artificial ventilator, which leads to some of the designs seen in hospitals and intensive care today. It will also show some of the designs that have been developed in the initial two months of the COVID-19 outbreak in the United States.

Such devices require intensive testing in order to guarantee the safety of the patient, therefore, standards' organizations such as the ISO have established laws and regulations to guarantee the highest level of performance and safety of these machines. Also, due to the enormous importance of the safety and performance of these machines, their design innovation is directly related to the improvements made to the device's user and patient interface.

TABLE OF CONTENTS

LIST OF FIGURES

CHAPTER 1

INTRODUCTION

1.1 BACKGROUND

In order to understand the basic requirements and functionality of a mechanical ventilator, pulmonary physiology must first be explained. During the pulmonary circulation, deoxygenated blood flows from the right atrium and ventricle, through the pulmonary artery, and to the lungs. Inside the lungs, it then flows to the pulmonary capillaries, where the gas exchange of carbon dioxide and oxygen takes place within the alveoli. The alveoli, according to the National Cancer Institute [1], are tiny air sacs at the end of the bronchioles (tiny branches of air tubes) in the lungs. Carbon dioxide in the blood passes into the lungs through the alveoli, and oxygen in the lungs passes through the alveoli into the blood.

Blood flow through the lungs is affected by hydrostatic pressure. In a normal, upright adult, the lung spans a height of approximately 30 cm. This translates into a 23-mmHg vertical pressure gradient, with 15 mmHg above the heart and 8 mmHg below the heart. Due to this pressure difference, the apex of the lung has little flow, while the bottom of the lung has five times as much flow [2]. These differences can be explained by taking a closer look at the three zones of local blood flow inside the lungs [2]. Zone one consists of the capillary pressure never increasing above the alveolar air pressure during the cardiac cycle, which results in blood flowing through this zone in the lung. In zone two, the capillary pressure is greater than the alveolar air pressure only during systole - the time-period when the heart is contracting, specifically, the left ventricle of the heart is

contracting; and the former is less than the latter during diastole - the time-period when the heart is in a state of relaxation and dilatation (expansion). These two stages result in intermittent blood flow in this zone. At last, in zone three, the capillary pressure is always greater than the alveolar air pressure, resulting in continuous blood flow [2]. Figure 1 shows a simplified diagram of pulmonary circulation from the British Lung Foundation.

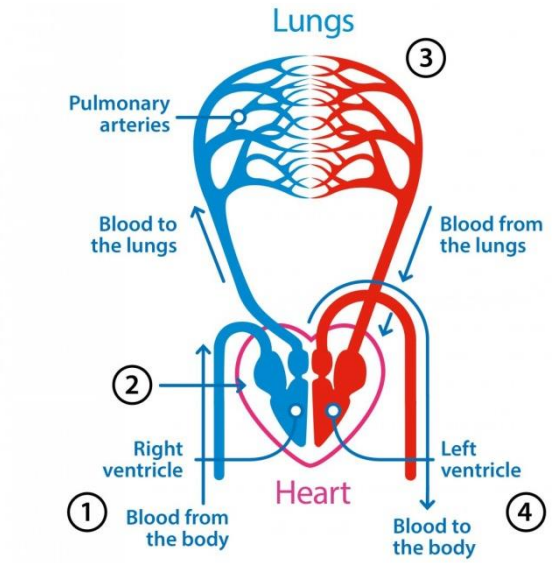

Figure 1. Pulmonary circulation diagram showing the flow of oxygen-rich blood, in red, and the flow of oxygen-poor blood, in blue [3].

The carbon dioxide and oxygen gas exchange takes place within the alveoli in the pulmonary capillaries, however, for it to receive oxygen, air needs to be transported there. This phenomenon is possible through pulmonary ventilation. Pulmonary ventilation is the process of air flowing into the lungs during inspiration, where the contraction of the diaphragm and the expansion of the chest cage pull the lungs downward and outward creating more negative pleural pressure, and out of the lungs during expiration, where the pleural pressure behaves in a reverse manner [4]. Air flows because of pressure differences between the atmosphere and the gases inside the lungs (from higher pressure to lower pressure region). Pulmonary ventilation involves three different pressures: atmospheric

pressure (the pressure of the air outside the body), interalveolar pressure (the pressure inside the alveoli of the lungs), and intrapleural pressure (the pressure within the pleural cavity) [4]. Figure 2 shows a simplified, visual representation of the pulmonary ventilation process described in the section above from Teach Me Physiology.

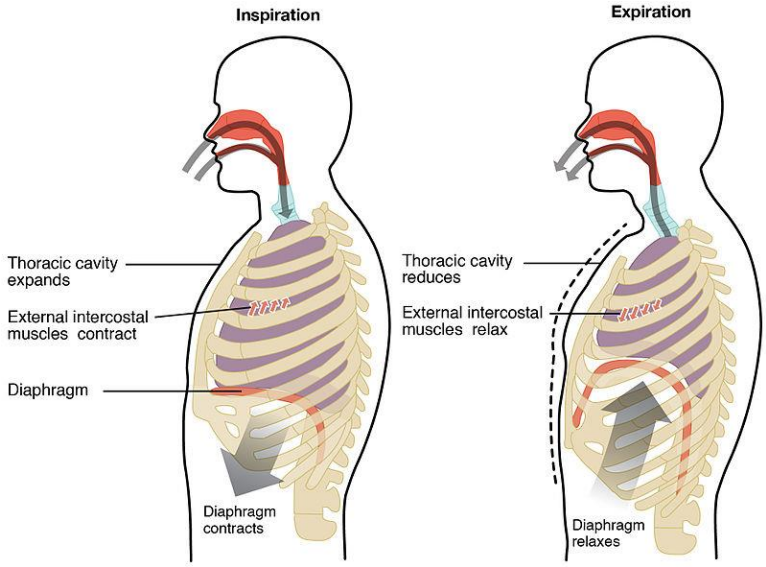

Figure 2. Simplified pulmonary ventilation diagram showing the process of inspiration and expiration at rest [6].

Gas exchange occurs in two places during a respiratory cycle: in the lungs, where there is an oxygen intake and a carbon dioxide release at the respiratory membrane, and at the tissues, where the opposite happens. External respiration is the exchange of gases with the external environment - occurs in the alveoli of the lungs, - and, internal respiration is the exchange of gases with the internal environment - occurs in the tissues [5]. The process of gas exchange is said to be extremely simple due to diffusion [5]. Energy is not required to move oxygen or carbon dioxide across membranes, since these gases follow pressure gradients that allow them to diffuse. At the same time, the anatomy of the lung maximizes the diffusion of gases: high permeable respiratory membrane, very thin respiratory and blood capillary membranes, and large surface area throughout the lungs [5]. Figure 3 shows

the gas exchange process in the external respiration cycle from OpenStax's Anatomy and Physiology.

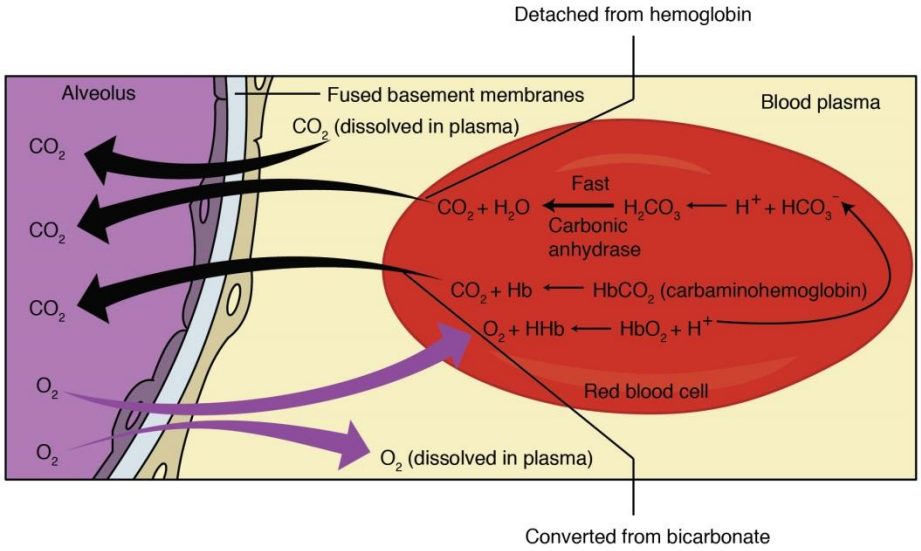

Figure 3. External respiration gas exchange. In external respiration, oxygen diffuses across the respiratory membrane from the alveolus to the capillary, whereas carbon dioxide diffuses out of the capillary into the alveolus [5].

Before analyzing the history and progress of mechanical ventilators throughout the years, a deeper look into the basic requirements and functionality of this machine needs to be made. Some of the most common features of a ventilator are: insufflation phase, expiratory phase, and cycling mode. Regardless of the operating system of a ventilator, it is very often represented as a pressure source developing an internal resistance and connected to the patient by a compressible system with compliance [7]. The lung system's compliance opposes resistance to the transfer of the gas, which is represented by the instantaneous flow. Then, instantaneous flow is equal to the pressure source minus the alveolar pressure (which is a function of both the flow and the total resistance, as discussed above), divided by the total resistance. The flow-pressure relationship, therefore, depends on the total resistance, meaning if the ventilator has a high internal resistance, it will depend slightly on the insufflation pressure. However, a low internal resistance ventilator does not

maintain a constant ventilator flow. Based on this information, the main characterization of ventilators is the value of the driving pressure. Pressure ventilators have low driving pressures, low internal resistance, and their performance is influenced by downstream conditions, while volume generator ventilators have high driving pressures, high internal resistance, and their performance is not influenced by downstream conditions [7].

The first ventilators generally had a change in driving pressure during the insufflation phase, where instantaneous flow varied, but the tidal volume remained constant. More recently, manufactured ventilators have the option of specifying the flow pattern, providing continuous, accelerating, decelerating, or sine wave flows, depending on the patient's need. This regulation is achieved through the electronic control of pressure changes, or, if the driving pressure of the ventilator remains constant, through the modification of the internal resistance of the machine. So, this therapy may be based on pressure, volume, or flow control, and since flow is the derivative of volume and because the three variables are related, only one variable acts as the independent variable for control. A programmed variation of the pressure may be obtained using a step motor or a microprocessor-controlled servo valve [7].

The expiratory phase is very similar to the insufflation phase, with the main difference being the source pressure, which in this case is the atmospheric pressure [7]. In this phase, the ventilator operates in a passive way, unless a resistance needs to be used to maintain a positive end-expiratory pressure (PEEP) greater than atmospheric pressure, which will be discussed later in a later section.

The cycling mode determines the parameters, which control the change from the inspiratory phase to the expiratory phase and vice-versa. Time, volume, pressure, and flow

are the four mechanisms that can act as signals for either expiration or insufflation, while the patient mechanism can only determine the switching from expiration to insufflation. This mechanism is often used in the conventional (or original) artificial ventilation and it is the present assist-controlled mechanical ventilation (assist-CMV or AMV) [7]. These modes will be explained further in a later section.

The ventilator system becomes more complex when there is a switch from inspiration to expiration [7]. Flow controlled cycling is rarely used, and time-cycled high-pressure generators have inspiratory characteristics completely unaffected by what may occur at the patient level, similar to volume cycling systems. In certain ventilator models, however, there is an intermediate cycle where switching from the inspiratory to the expiratory phase is time-controlled. In these cases, inspiration includes two phases: an active period - the true insufflation time, volume-cycled, whose duration varies as a function of flow, - and a passive period - insufflation ended, but expiration has not started. During the passive period, the flow is zero, the intrapulmonary volume remains constant, and the alveolar pressure stabilizes rapidly to the thorax relaxation pressure level (a function of the thorax-lung static compliance) for a given volume. The thorax relaxation pressure level is the pause at the end of the insufflation, also known as a plateau [7].

One of the most interesting facts regarding mechanical ventilators is that, independently of their manufacturing year and manufacturer, the basic features are mostly the same [7]. First, the ventilator must always have a source of gas – the driving system. This source is normally compressed gas being fed from a bottle to the respective ventilator or a mechanical system that compresses a gas mixture being drawn at the atmospheric pressure. Inspiratory tubing is the connection of the gas source to the patient, which

transports the mixture during the insufflation phase. Once the insufflation phase is complete, the expired gases flow through the expiratory tube to be discharged. The inspiratory and the expiratory tubes are connected to a common piece, which allows for an easy connection to the patient's intubation cannula or tracheostomy tube. A system of valves is also part of the ventilator system, as it prevents inspiratory gas from flowing into the expiratory branch during insufflation and the expired gas from entering the inspiratory arm during expiration - rebreathing. It could be placed either inside the ventilator apparatus or near the connecting common piece [7]. Figure 4 shows a diagram with the basic feature of a ventilator [7] to serve as a visual aid following the explanation above.

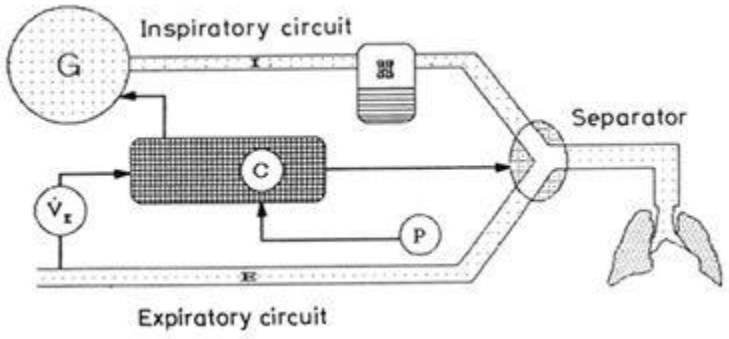

Figure 4. Diagram of a basic ventilator showing its features: G, gas source, I, inspiratory branch, E, expiratory branch, S, separator, H, humidifier, P, pressure manometer, V, flowmeter measuring the instantaneous expired flow, and C, control systems [7].

The control system, similar to the source of gas, is an essential part of the mechanical ventilator [2]. This component is used to determine all the respiratory cycle's parameters: insufflation time, expiration time, duration of an inspiratory pause, percent cycle times, the volume of insufflated gas, minute flow, and mode of ventilation. It is responsible for initiating the respiratory cycle, and for the control of the system of valves. In addition to the basic elements mentioned above, mechanical ventilators require two essential accessories: a humidifier, used to humidify insufflated gases saturated with water vapor, and a system to monitor artificial ventilation [2].

This system to monitor artificial ventilation is especially important because it is responsible for measuring the pressure of the system, which usually only needs to be measured at a site within the ventilator, however, in cases where the intubation tubes are small-size, it may be significant to measure the intratracheal pressure since the resistance to gas flow is higher [7]. This system also measures expiratory instantaneous flow, which is considered essential for the monitoring and safety of the patient, even in ventilators that are only to be used over short periods of time - in the operating room, especially.

In addition to the basic features of a ventilator, there are also basic parameters for artificial ventilators, such as: tidal volume, frequency, and inspired minute volume. Similar to the pressure, volume, and flow control, once two of these parameters are set, the third one is also determined. These parameters can also be selected indirectly by setting the behavior for peak flow during insufflation and expiration times [7]. When a control system is not available, this method of control is hard to use and requires a trial-and-error method before the correct setting is achieved since the frequency is known, but the remaining two parameters are only known after the patient is connected to the machine.

To be able to determine the most appropriate ventilator output waveform, for inspiration and expiration, there needs to be a previous knowledge of flow patterns and their representation [7]. Theoretically, there are four different flow patterns available. The square or continuous flow pattern is where the flow instantaneously reaches its peak value and remains constant throughout the insufflation time. The accelerating flow pattern is where flow increases progressively to a maximum value at the end of insufflation. The decelerating flow pattern is where flow decreases progressively from a peak value to zero at the end of insufflation. The sine wave flow pattern is where the last curve of the figure

simply illustrates any other pattern possibly evoking a "sine wave" variation of flow whose characteristics are adjustable in some advanced sophisticated ventilators [7]. Figure 5 shows the flow patterns described above [7].

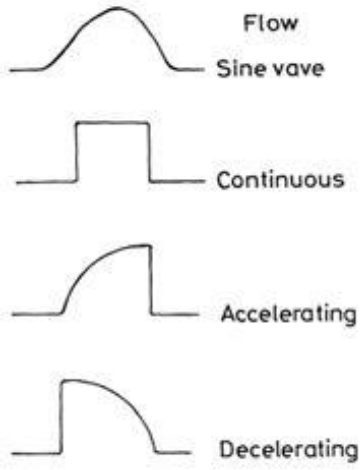

Figure 5. Variation of instantaneous flow during insufflation [7].

Given the basic flow patterns above, and the fact that when a patient receives mechanical ventilation, this therapy may be based on pressure, volume, or flow control, the most appropriate output waveform for the individual patient can be generated [2]. Figure 6 shows several examples of idealized ventilator output waveforms [2].

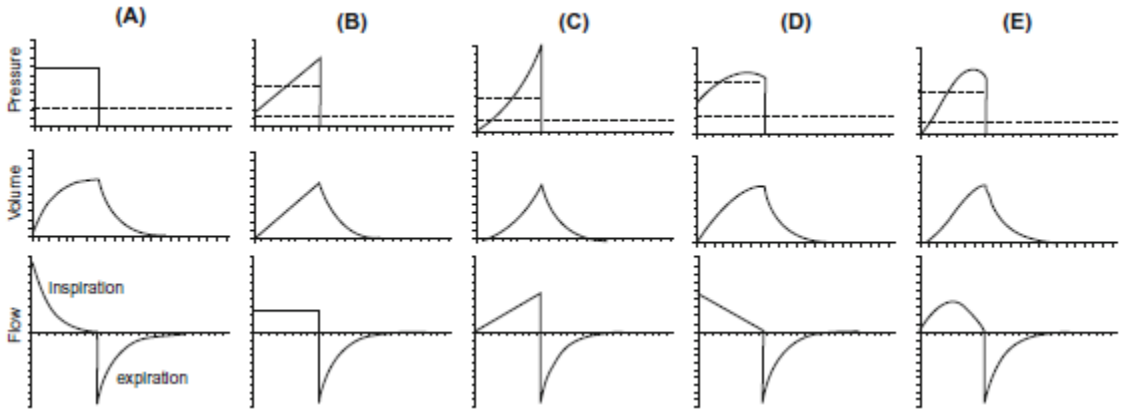

Figure 6. Idealized ventilator output waveforms. A: Constant pressure-controlled inspiration. B: Constant flow-controlled inspiration. C: Ascending-ramp flow-controlled inspiration. D: Descending-ramp flow-controlled inspiration. E: sinusoidal flow-controlled inspiration. In all cases, expiration pressure equals zero [2].

1.2 MODES OF VENTILATION

For artificial ventilators there are two major categories: non-invasive ventilators and invasive ventilators [8]. Within these categories, there are also modes of ventilation. The original assist mode delivers the breath at regular intervals, which are pre-determined by the ventilator controls and the characteristics of the machine. The reason why the industry had to deviate from this mode was that conscious patients often relax when subjected to controlled ventilation, but some may not synchronize with the ventilator unless they are hyperventilated, sedated or paralyzed. This is called desynchronization and it could cause lung damage and circulatory embarrassment. The solution found was the assisted mechanical ventilation, or AMV [8].

The AMV technique delivers breaths whenever the patient's (spontaneous) inspiration activates the machine's trigger. This trigger is able to interpret the patient's breath and tell the machine to produce a small inspiratory flow or reduce the pressure in the respiratory system. The sensitivity of this trigger is of extreme importance. Its response requires speed in order to minimize the patient's work of breathing, but if it were too sensitive, then it would be activated by something other than inspiration such as the movement of the ventilator tubing. Since this technique allows the patient to control each breath, its pattern, and frequency, it slowly became more popular to manufacture. This gave patients a better treatment method, since it became almost impossible for them to oppose the action of breathing of the machine [8].

As science continued to develop, more patient-controlled modes were created. The inspiratory pressure support, or IPS, allows the patient to control the pattern of the breath. Similar to AMV, the machine is triggered by the patient's inspiration – regularity or

reduction in tube pressure. IPS, however, then provides a constant positive pressure throughout inspiration, so that the ventilator supports the spontaneous efforts of the patient [8]. The intermittent mandatory ventilation, or IMV, allows the patient to breathe independently through a breathing system, which provides humidified gases with the appropriate concentration of oxygen, and the machine then delivers a mandatory breath of known characteristics at preset intervals. One of the disadvantages of this model, similar to the original assist mode, is that the mandatory breaths are pre-determined by the machine, and the other disadvantage is that, due to this pre-determination, there is a possibility that the mandatory breath is "stacked" on top the patient's spontaneous inspiration flow [8]. The goal with the development of this model was that the frequency of the mandatory breathes would decrease as the patient's condition improved, however, there were never any conclusive studies that proved this theory over the conventional method of allowing the patient increasing periods off the ventilator [8].

The next developed model was the synchronized intermittent mandatory ventilation, or SIMV [8]. This model incorporated the benefits of the IMV model and improved its disadvantages such that the mandatory breath was synchronized with the spontaneous breath. This was possible by creating a time window where the ventilator can be triggered by a spontaneous breath. The duration of this window is usually the same length as the duration of the mandatory breath cycle and the frequency of its occurrence is determined by the adjustment of the SIMV frequency control [8]. Another advantage of this model and the ones that preceded it is that those ventilators can be set to provide a mandatory breath at predetermined time intervals if the patient is not stable enough and fails to trigger the ventilator during the time window.

The third model, as described by Sykes, is the mandatory minute volume, or MMV [8]. This model was developed to provide a more reliable backup ventilation system when the patient's spontaneous ventilation becomes inadequate. Once this minute volume control is set up, the patient is able to continue the inspiration flow without any disturbance. This control only becomes significant once the patient's minute volume is less than the mandatory minute volume, where the ventilator makes up for the deficit. As the patient's minute volume decreases, the frequency of the mandatory breaths increases. While the MMV model is extremely impressive in theory, it is of very little practical value, since if the mandatory minute volume is inadequate, the patient's respiratory rate will increase, and the tidal volume will tend to decrease. It then leads to an increase in tidal volume ratio, or dead space, ratio so that ventilation becomes less efficient, producing an additional increase in breathing frequency, therefore creating an endless unproductive cycle [8].

The final model discussed is the continuous positive pressure breathing, CPAP, or CPAPB [9]. This model is significantly different from the previous six, since it is used to maintain adequate oxygenation when the patient does not require ventilatory assistance, and it can be used with a facemask or with an endotracheal tube. However, in order to minimize the work of breathing by a patient receiving CPAP, the airway pressure should be maintained at a constant value throughout the respiratory cycle. If this pressure decreases below the set CPAP value during inspiration, the machine's control system adds inspiratory work. If the opposite happens and it increases above the set value in expiration, additional expiratory work is required. Therefore, an important aspect of this model is that the system maintains a constant value of pressure [9]. Another critical component of ventilators that use this mode is the trigger mechanism, since it is necessary to ensure that

the ventilator can generate inspiratory flow rates of up to 1.5 L/s. This is critical because patients in respiratory distress often generate high peak inspiratory flows. In addition, patients are often treated with CPAP in the earlier stages of their disease and may develop CO_2 retention and therefore require some form of ventilatory assistance in the future [9]. Figure 7 shows a graphical representation of the seven different ventilation models as functions of pressure versus time [8].

Using all the information provided in the introduction, there is a better understanding of the basics of the respiratory system, and the basic functionalities of artificial ventilators. The following chapters will describe some of the most historic artificial mechanical ventilators, some of the newer models, models developed by non-ventilator manufacturing companies and individuals to fight the COVID-19 pandemic, and finally the laws and regulations that manufacturers need to optimize their design around.

The goal of presenting multiple old ventilation machines and newer ones is to be able to interpret how this intricate piece of machinery, the artificial ventilator, has improved and adapted over time. This variety of models will provide a deeper understanding of how each manufacturer has improved upon its previous models, and how specific components might have played major roles. Finally, the laws and regulations of artificial ventilators chapter will give an insight into what is currently happening in the world with the COVID-19 pandemic. Such laws have been loosened in order to allow for the necessary mass manufacturing of such machines. However, with changing the laws, comes complications, and this chapter will unravel what those might be.

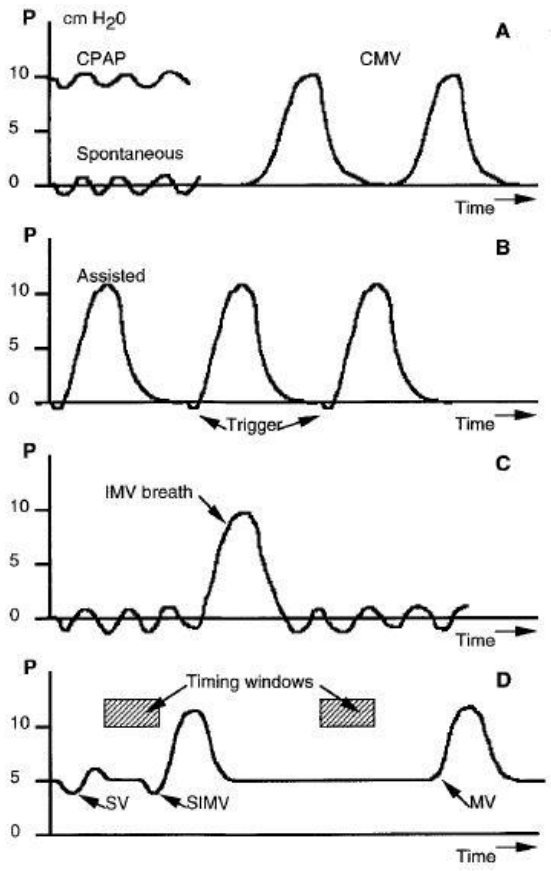

Figure 7. Airway pressure traces, P, with various breathing modes. (a) shows the continuous positive airway pressure breathing, or original assist mode, spontaneous breathing and controlled mechanical ventilation, CMV; (b) shows the assisted ventilation also showing the sub-atmospheric triggering pressure; (c) shows the intermittent mandatory ventilation showing an IMV breath superimposed on spontaneous breathing; and (d) the synchronized intermittent mandatory ventilation showing a spontaneous breath followed by a synchronized mandatory breath, SIMV. Since there was no spontaneous breath during the second timing window, the ventilator delivered a mandatory breath, MV, at the preset interval as mentioned in the paragraphs above [8].

CHAPTER 2

HISTORIC ARTIFICIAL VENTILATOR MODELS

2.1 PULMOTOR

In 1900, a German scientist by the name Heinrich Draeger developed a machine to help miners with asphyxia – the Pulmotor (Figure 8) [10]. It consisted of an oxygen cylinder, a pressure regulator, a pressure gauge, a clockwork-driven valve, and a facemask. The valve cycled every five seconds (12 breaths/min). It was manually operated, so when the valve opened, the facemask was applied, and when the valve closed, the facemask was removed from the victim to allow expiration to occur. For its time, the Pulmotor was a dependable, easy-to-use, and excellent first-aid device. As mentioned above, it started by being used in mines, but it quickly started being used in public-utility businesses and hospitals. Nowadays, the Pulmotor is described as a positive-pressure, time-cycled ventilator [10].

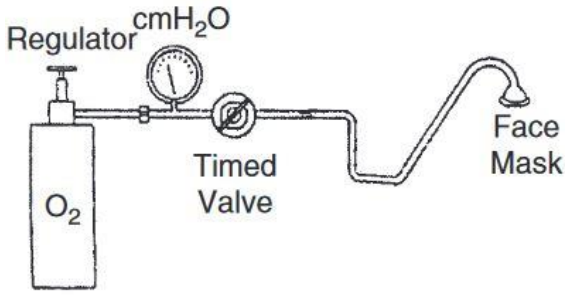

Figure 8. Schematic of Dräger Pulmotor [10].

2.2 IRON LUNG AND EMERSON TANK

Historically, all ventilation was done by negative-pressure ventilators. Negative-pressure ventilators supported ventilation by exposing the chest wall to increased atmospheric pressure during inspiration, with expiration occurring as the pressure around the chest could return to atmospheric levels [11]. The most well-known noninvasive positive-pressure ventilation, or NPPV, is the Iron Lung. Developed in 1928 by Philip Drinker, a Boston engineer, the machine was very popular during the 1950's polio epidemic [11]. Around the same time, J. H. Emerson constructed a simpler, quieter, lighter, and less expensive version of the Iron Lung that could be manually operated in the event of power failure [2]. Figure 9 shows a patient receiving respiratory treatment inside an Emerson tank respirator device [2].

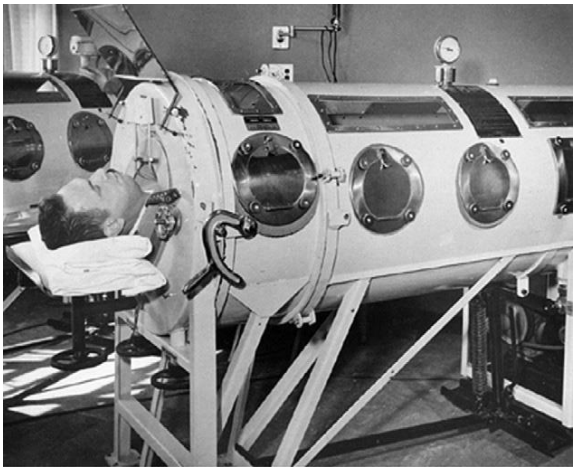

Figure 9. Emerson tank respirator device being used by a polio epidemic patient. The Iron Lung and the Emerson machine were very similar, such that Drinker attempted to sue for patent infringement but was unsuccessful [2].

These machines consisted of a one-ton metal cylinder, where the patient would lay on a mattress with the head sitting right outside one end of the cylinder through an air-tight rubber neck seal. Some disadvantages of these models such as patient discomfort, restrictions on positioning, problems with correct fitting, time-consuming application, lack

of portability, and a tendency to potentiate obstructive sleep apneas, stimulated the development of new techniques [12].

So, around 1960, the majority of ventilatory support was still done through Iron Lung or Emerson machines, and invasive positive pressure ventilation was simply used for administration of anesthesia [12]. Since the polio epidemic was still far from over, an enormous effort was made to provide round-the-clock ventilation to these patients using invasive positive pressure resuscitators. These devices were borrowed from anesthesia suites and powered manually by medical students, nurses, and other volunteers [12]. This new technique proved to increase the survival rates, and with the development of intensive care units, the beginning of the gradual transition to invasive positive pressure ventilation started. Administration of positive pressure ventilation via translaryngeal endotracheal tubes became standard practice for the support of patients with acute respiratory failure [12].

2.3 BIRD MARK 7

With the European implementation of this new standard of treatment came patented devices. The first low-cost, mass-produced mechanical ventilator was manufactured in the United States by Forrest Bird [13]. He named his machine the Bird Mark 7 ventilator (Figure 10) in 1957 after manufacturing six previous prototypes. This ventilator was powered by compressed air or oxygen at multiple pressures. It was a non-rebreathing ventilator designed to deliver the driving gas with or without entrained air to the patient. Air entrainment is the intentional creation of tiny air bubbles. Inspiration in these devices can be cycled automatically or patient triggered, and its functions are either flow or pressure generated. The change from inspiration to expiration is pressure cycled and from

expiration to inspiration by time [13]. Figure 10 shows the apparatus of a Bird Mark 7 artificial ventilator from the Wood Library Museum, and Figure 11 shows a technical drawing of the same machine [15].

Figure 10. Bird Mark 7. On the left is the full apparatus including the compressed air or oxygen mechanism [14], and on the right is the ventilation system with the system's controls [2].

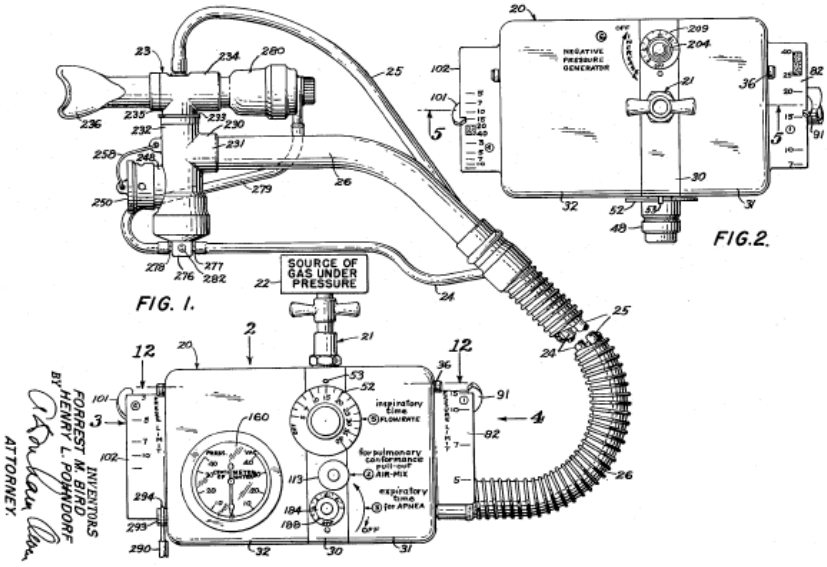

Figure 11. View in elevation of a respirator embodying the principles of the Bird Mark 7 invention with a control assembly shown in front elevation, connected to a breathing head assembly, shown in a side elevation (Fig. 1), and top plan view of the control assembly (Fig. 2) [15].

CHAPTER 3

MODERN ARTIFICIAL VENTILATOR MODELS

3.1 PORTA-LUNG PEGASO V

Modern negative pressure ventilators are, in general, improved versions of the Iron Lung and the Emerson ventilator [16]. The modern tank ventilators are constructed of aluminum and plastic, and are, lighter than the old models. Most modern tank ventilators have windows that permit patient observation and portholes through which catheters and monitor leads can be passed. These holes also allow access to the patient for procedures (arterial blood gas sampling, for example). These models allow the patient's head to be raised so that the aspiration of material from the pharynx into the trachea is prevented [16]. Figure 12 shows the Porta-Lung Pegaso V ventilators from the Porta-Lung website. There is a very clear similarity between these ventilators and the Iron Lung ventilators.

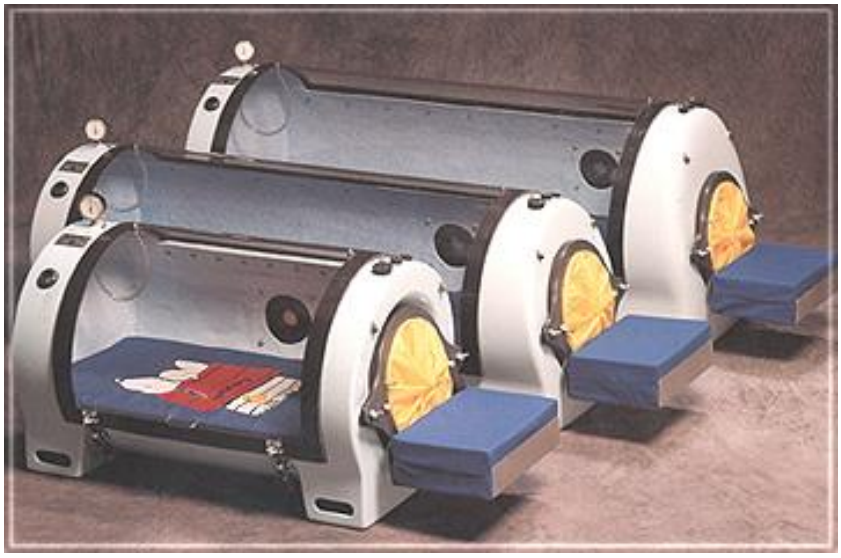

Figure 12. Porta-Lung Pegaso V ventilators [17].

In these models, negative pressure is generated by bellow pumps incorporated into the structure of the ventilator or by separate rotary pumps [18]. These pumps are pressure-cycled, meaning the ventilator continues to develop sub-atmospheric pressure until a predetermined level is reached. The controls were set to deliver the pressure during inspiration and expiration independently. All modern negative pressure ventilators provide control mode; additionally, some provide assist and assist-control modes, therefore patient-generated negative pressure is what triggers the machine's system [18]. Finally, even though they are not suitable for use in general intensive care units, the scientific community has started to use these negative pressure ventilators in pediatric intensive care units to avoid the need for endotracheal intubation [19].

One of the many benefits of the most recent artificial ventilator models is their use of electronic feedback [19]. The breakdown of these ventilators' systems is as follows: the pressurized gas is held in a reservoir and delivered to the patient via an inspiratory valve; the inspiratory valve, and therefore the inspiratory flow, is controlled by the electronic control unit; the airway pressure and flow of gas into the patient are monitored by the pressure and inspiratory flow sensors; and the expiratory flow can also be monitored to check for leaks and disconnection of the patient from the ventilator. Therefore, this design enables the ventilator to be used as either a flow or a pressure generator [19]. The complete ventilator machine is now both the electronic unit and the pneumatic unit, which are often compacted together. Figure 13 shows a diagram of the breakdown described above [19].

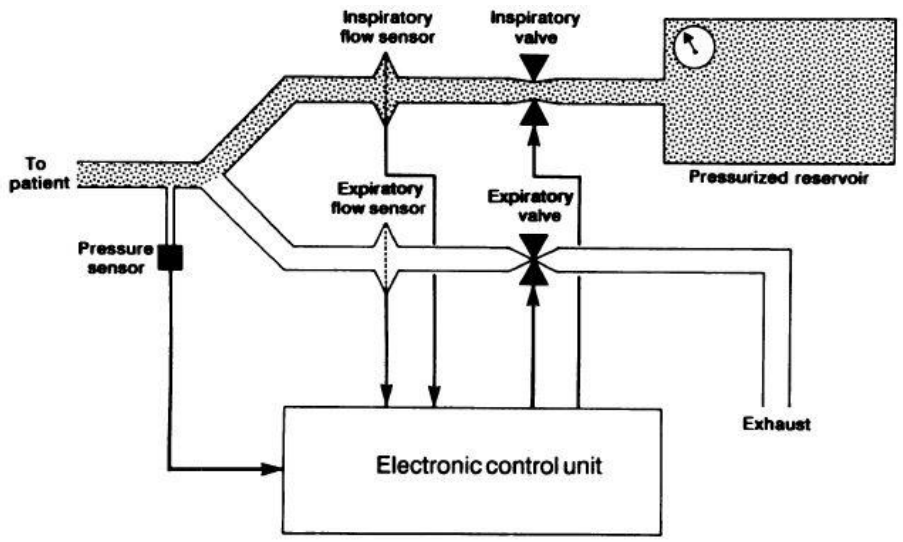

Figure 13. Block diagram of a modern ventilator. The shaded area indicates the gas path during inspiration [19].

3.2 SIEMENS' SERVO 900D

Other improvements in modern ventilator models are the minimum movement of parts inside the machine, the sophisticated alarm systems, small physical scale, and the simplicity of the maintenance and repair [19]. At the same time, if a mode of ventilation is discovered to be more beneficial to the patient, clinically, it can often be added to an existing machine by altering the software in the electronic unit. In fact, the scientific community's ability to produce new modes of artificial ventilation exceeds the ability to test them clinically, by far [19]. An example of such model is the Servo 900D ventilator, manufactured by Siemens [20].

Siemens' Servo 900D ventilator was manufactured for both human and animal use. It aimed to provide a ventilator with a digital control unit which, depending on the parameter (described in the Introduction chapter) and the regulation of the analog control unit, compensates for the coarseness in the regulating function of the control unit so that the predetermined pattern is maintained almost in its entirety [20]. Analog control units

deal with continuous signals that can take a wide range of values; the transfer function is given by the differential equation in s-domain and uses Laplace Transform techniques. So, in this model, the advantages of the rapid analog adjustment and accurate digital adjustment are combined. Since the digital unit only compensates for the fault of the regulation of the analog unit, the ventilator will be insensitive to failures in the digital unit. Should the digital unit fail, the function of the ventilator is not inferior to a system controlled by only an analog control unit, but instead becomes very similar to it. For the Servo 900D, the preferred digital control unit is a microprocessor. Siemens was also able to provide a simple solution to the potential failure of the full digital unit, and therefore the entire ventilator, by simply adding more microprocessors (one for each individual machine function) and by distributing them correctly [20]. Figure 14 shows the system's simplified diagram [20].

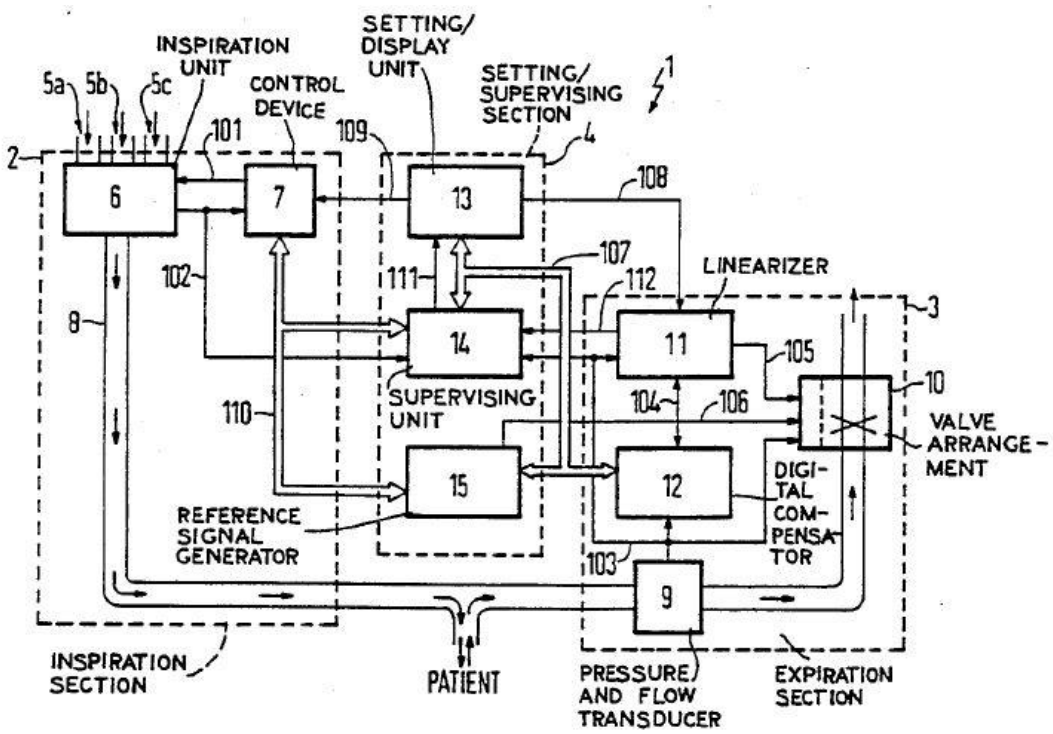

Figure 14. Simplified diagram of the Siemens's Servo 900D ventilator. Includes an inspiration section (2), an expiration section (3), and a setting/supervising section (4) [20].

There have been many more artificial ventilator designs since the Siemens's Servo 900D ventilator was released. These designs have used all the important and most developed components of their competitors and created machines that operated better and safer the ones before.

CHAPTER 4

COVID-19 VENTILATOR INNOVATION

Due to the COVID-19 pandemic, many non-ventilator manufacturing companies had to stop their original production and begin manufacturing artificial ventilators. Original ventilator manufacturing companies allowed companies like Ford and Tesla access to their basic functionality designs, in order to speed up manufacturing processes in this time of need [21].

Ford Motor Company partnered with GE Healthcare to expand the production of artificial ventilators to help with the U.S.'s shortage. GE allowed Ford to produce a simplified version of one of their existing models, the GE/Airon Model A-E ventilator, which was designed to support patients with respiratory failure or difficulty breathing caused by COVID-19 [21]. This design is specifically important during the current pandemic since it operates on air pressure without the need for electricity [22]. Figure 15 shows the Model A-E ventilator and the production projections of this device from the Ford Media Center.

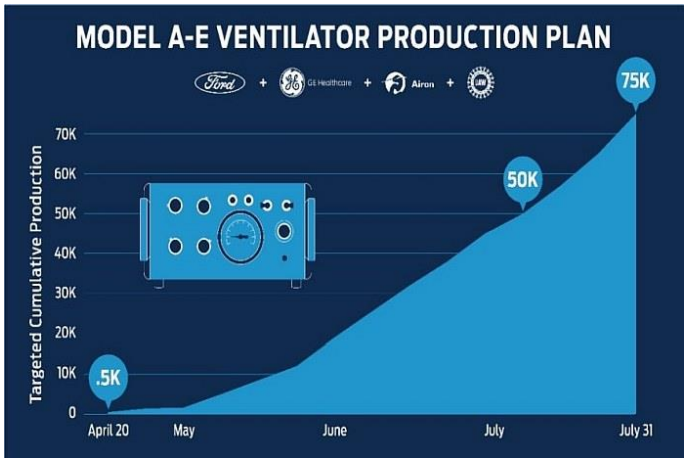

Figure 15. Model A-E Ventilator Production Plan [22].

Tesla, on the other hand, decided to design and manufacture its own artificial ventilator, using components that are usually used in the cars they manufacture, since those are the components the company and its engineers are most familiar with [23]. Tesla had originally manufactured bilevel non-invasive ventilators, known as BiPAP machines, which were extremely criticized by the scientific community [24]. These devices are typically used to treat conditions like sleep apnea, and experts had warned that the machines could cause the virus to spread faster [24]. Despite the BiPAP machines being cheaper, Elon Musk guaranteed that Tesla would start manufacturing and delivering Medtronic invasive ventilators after Medtronic offered to share the design specifications for a basic ventilator model with any company that wanted to help produce them for hospitals racing to treat coronavirus patients [25]. Figure 16 shows the ventilator whose specifications were shared by Medtronic, the PB560 Ventilator, from West Care Medical.

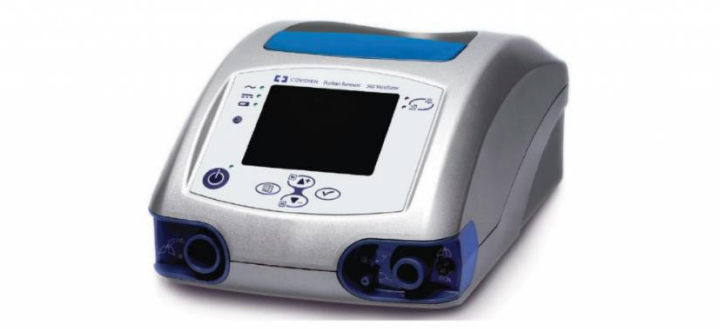

Figure 16. Puritan Bennett 560 portable ventilator [26].

Along with global manufacturing companies, individuals who had prior knowledge of modeling and artificial ventilation offered their expertise, to try to help develop simpler and faster ventilation devices. This happened through different competitions that were started to help fight the artificial ventilation shortage. One of the competitions was started by GrabCAD, which is the largest online community of professional engineers, designers, manufacturers, and STEM students on the planet [27]. This community has almost 6

million members who share CAD files and professional tips, participating in design challenges, or downloading free CAD models from our online library of over 4 million files [27]. One of the winning designs of the CoVent-19 Challenge (Round 1) was the SmithVent, which was designed by a team of Smith College engineering alumni and friends [28]. Their design is a simplified, cost-effective, rapidly manufacturable pneumatic ventilator tailored to COVID-19 patient needs and designed for ease of use by healthcare workers. The components used are mainly readily available off-the-shelf components, thereby reducing complex machining and supporting compatibility with current medical equipment [28]. Figure 17 shows the SmithVent ventilator from the GrabCAD Community Library.

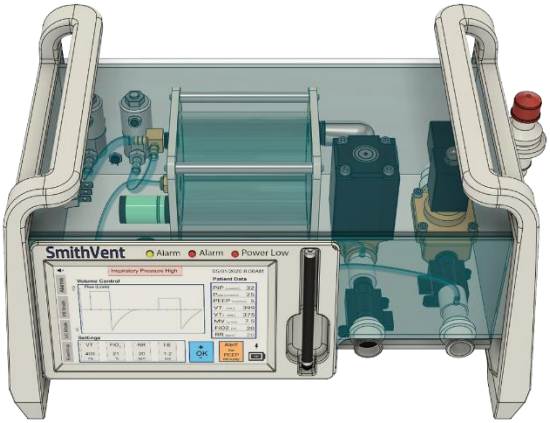

Figure 17. SmithVent ventilator manufactured by a team of Smith College engineering alumni and friends [28].

CHAPTER 5

LAWS AND REGULATIONS' IMPLICATIONS

Because of the incredible advances made in technology in the last century, and because of the variety of manufacturing companies involved with each invention, universal standards became increasingly necessary. These standards provide order in a world where new designs are invented everyday. As the COVID-19 pandemic has shown the world, it is possible to design a piece of machinery in a few weeks [28], when lives are at stake. Furthermore, if the entire world is attempting to defeat a common enemy, there will be thousands of new machines developed each day. Of course, the current world situation is an extreme scenario, but it shows exactly why such standards are necessary.

The International Organization for Standardization (ISO) is an international standard-setting body composed of representatives from various national standards organizations [29]. This organization develops standards that are internationally agreed upon by experts. These standards are described as the appropriate way of doing something. Specifically, for artificial ventilator machines, ISO standards instruct anyone or any company who would like to produce such machines, of the manufacturing and testing restrictions and recommendations.

When it comes to basic requirements for the safety and performance of artificial lung ventilators, the ISO 10651-5:2006 standard [30] contains useful information such as construction requirements. The building materials allowed for the machine components need to consider the chemical and physical properties of any substance that will flow through them. Special attention needs to be paid to the toxicity of materials and their

compatibility with substances and gases with which they enter in contact during normal use of the ventilator (usually compressed air and oxygen). In the controls' system, a single fault condition should not cause any monitoring or alarm system function, and the corresponding ventilation control function to fail such that the monitoring or alarm system function becomes simultaneously ineffective, and thus fails to detect the loss of the monitored resuscitator function. This standard also provides guidance for the machine's performance. Ventilatory requirements include delivered oxygen concentration, and resistance to spontaneous breathing such as, inspiratory resistance during the ventilator expiratory phase, spontaneous breathing with the gas input pressure outside the rated range, and expiratory resistance. There are also performance requirements that focus on the modes of ventilation such as: manually cycled, automatic pressure-cycled, automatic time-cycled or automatic volume-cycled, patient-triggered, and demand valve mode. Each of these modes has many individual requirements, for example: delivered volume and its consistency, pressure limitation under normal use, etc.

For ventilator test conditions, the standard that manufacturers need to comply with is ISO 80601-2-12. This standard is for medical electrical equipment, with a focus on requirements for basic safety and essential performance of critical care ventilators [31]. For artificial ventilators, specifically, the testing needs to be conducted using the gas, and its supplying component, that is pre-determined for the normal use of the machine. The only exception is that industrial-grade oxygen and air may be substituted for the equivalent medical gas, as appropriate, unless otherwise stated. During testing, the gas flowrate and leakage are other functions that need to be in accordance with the standard. The ventilator error notification is also tested. The tolerances are required to be provided by the

manufacturing companies in the technical description, and they are used to adjust the machine's uncertainty.

Even though all the guidelines described above are in place, COVID-19 has required the United States government to take serious measures. On March 24, 2020, the United States Department of Health and Human Services (HHS) Secretary declared that this pandemic created circumstances that justified the authorization of emergency use of medical devices, including alternative products used as medical devices, due to the shortages in the country [32]. The HHS also issued a declaration, which provided liability immunity for activities related to medical countermeasures against COVID-19 [32]. However, even though manufacturing companies and individuals may apply for the Emergency Use Authorization (EUA), many doctors and physicians still have concerns about using newly manufactured machines, since their main priority is to provide adequate care for their patients. The Food and Drug Administration (FDA), is and will continue to work closely with these manufacturers in order to guarantee the quality determined in the ISO standards for all the new machines [32].

CHAPTER 6

THE FUTURE OF ARTIFICIAL VENTILATION

The COVID-19 pandemic has shown that the scientific community is able to achieve amazing results in a short amount of time. As shown in chapter 3, not only did non-ventilator manufacturers stop their original manufacturing efforts to help fight this virus, but also did millions of modeling engineers, individually, provide their services and knowledge to help create fast and efficient artificial ventilation machines. While the current goal is to manufacture enough ventilators for all the hospitals in countries around the world, if this pandemic hadn't fallen upon us, the future of artificial ventilation would be in improving this intricate machine's control settings and display [33].

6.1 OPERATOR INTERFACE

As ventilation modes have become more complex, the operator interfaces on these machines have become more complex as well. Artificial ventilator manufacturers have created multiple options for control settings, which tend to get lost in layers of different screen views [33]. An even more important negative effect is that due to this customization, no two machines will have the same computer layout. This is indeed a scientific advancement, since individual operators will be able to prioritize which controls to see, but it will very quickly become chaotic, since any other operator besides the one who set up the machine will not understand the layout. This has proven to become a problem [33]. There is a lack of studies done on the ease of ventilator display use. The future of artificial ventilator research is in identifying the optimal display for the three basic functions: to

allow input of control and alarm parameters, to monitor the ventilator's status, and to monitor the ventilator–patient interaction status. This is the future of operator interface [33].

6.2 PATIENT INTERFACE

When it comes to the interface between a modern ventilator and the patient, the technologies used have not developed significantly since the 1900s, however, the humidification systems have certainly evolved [34]. Even though this is the case for humidification systems, these machines are still not capable of measuring and directly controlling this primary variable of gas conditioning (humidity). Despite the technological advancements, Solomita, et al. have shown that simple, unheated circuits provide better humidification of inspired gas, through data collected, even though modern devices use heated wires and automatic-temperature control [34]. Another factor that adds difficulty to these components is the fact that the natural way of respiration of a patient degrades the gas flow delivery, requiring the device to have complex mathematical algorithms to adjust it. The solution to this problem is theoretically simple, as it only requires the patient circuit to be a permanent part of the ventilator and treat water molecules the same way as it treats molecules of oxygen, nitrogen, helium, and nitric oxide. To do this, however, ventilator manufacturers and humidifier manufacturers would have to combine efforts and design a system together, instead of having the patient circuit (plastic tubing) and the humidifier as separate machines. As mentioned before, in theory, it sounds like an easy design collaboration to accomplish, but when there is monetary value involved, nowadays, collaborations like this one are extremely rare [33]. Projects like this are the future of

patient interfaces. Figure 18 shows the location of a humidifier on a Bellavista 1000 ventilator, from Vyaire Medical, connected to the ventilator, but not incorporated.

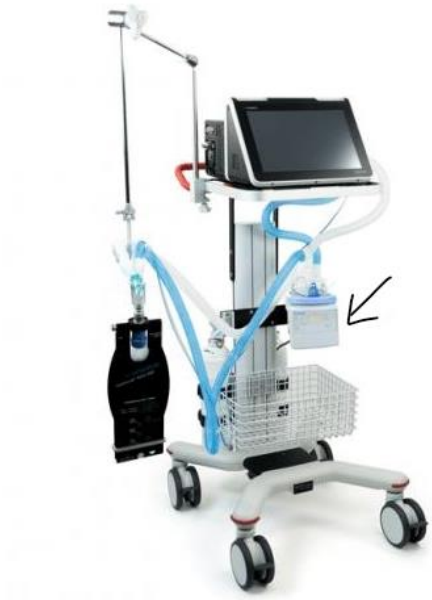

Figure 18. Bellavista 1000 ventilator showing the location of the humidifier machine (arrow pointing at it), connected but not incorporated in the ventilator [35].

Another major area of improvement in the artificial ventilator industry is the sensors and the software algorithms required to make sense of the data they are receiving from the patient, and, therefore, providing [33]. The future of targeting systems, in both basic research and commercial applications, is to develop “closed-loop” targeting systems based on mathematical models of physiologic processes, or artificial intelligence, or combinations thereof, with the goal of automating the moment-to-moment adjustment of ventilator output to patient needs. The main example of such a system is the INTELLiVENT-ASV, the Hamilton Medical G5 ventilator, which is currently only available in Europe [33].

The INTELLiVENT-ASV mode is an upgraded version of the ASV mode. ASV, or adaptive support ventilation, evolved as a form of mandatory minute ventilation (MMV)

implemented with adaptive pressure control. It is considered a closed-loop controlled ventilatory mode, which is designed to ensure the optimization of the patient's work of breathing [36]. Like ASV, INTELLiVENT-ASV is a form of pressure control intermittent mandatory ventilation using adaptive-pressure. It's targeted to automatically adjust inspiratory pressure to maintain a target tidal volume, which, in turn, is selected by an optimization model. Optimal targeting systems attempt to maximize or minimize a predetermined performance factor. For ASV, this factor is the tidal volume and frequency that minimizes the work rate of ventilation for the patient's specific state of lung mechanics [33]. The goal of this system is to work in parallel with the patient, so as the lung mechanics change, the ventilatory pattern changes. This system requires the operator to input certain variables, one of these being the patient's weight so that the minute ventilation requirement is calculated for the specific patient. For INTELLiVENT-ASV, the system also requires input data from end-tidal CO_2 monitoring and pulse oximetry. This extra input data along with the improved targeting software, allow the ventilator to automatically select and adjust minute ventilation [33]. This specific piece of programming is what makes the Hamilton Medical G5 ventilator (Figure 19), the first-ever closed-loop ventilation system that offers automatic adjustment of oxygenation and ventilation [33].

6.3 VENTILATOR DISPLAY

Alongside the innovative targeting system, the Hamilton Medical G5 ventilator was also developed to provide an improved operator interface [33]. As mentioned before, the future of artificial ventilation will depend on the improvement of the ventilator display. The G5 ventilator was developed to facilitate understanding of complex information in a visually intuitive way. So, this display still contains all the important data displays, such

as: data on lung mechanics, end-tidal carbon dioxide, and pulse oximetry [33]; but all this important and sometimes confusing information if presented in a much simpler form. Figure 20 shows what Hamilton Medical called the "Ventilation Cockpit," since it provides the operator with all the data he or she would ever need to interpret [33].

Figure 19. Hamilton Medical G5 ventilator. the first-ever closed-loop ventilation system that offers automatic adjustment of oxygenation and ventilation [33].

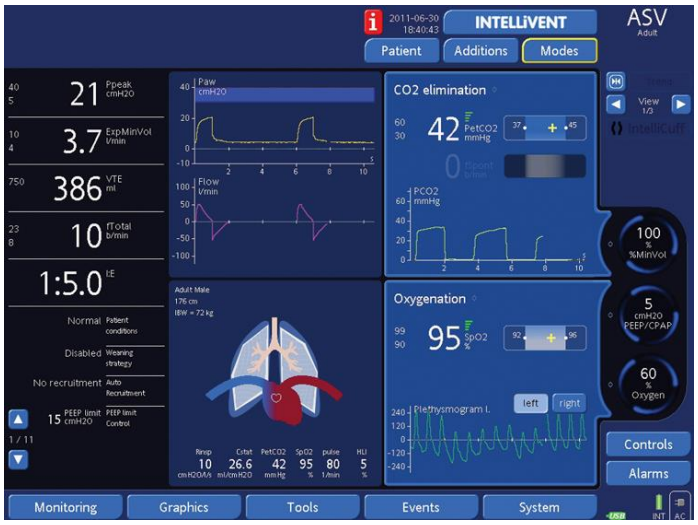

Figure 20. Ventilation Cockpit. The operator interface of the Hamilton G5 ventilator with the INTELLiVENT-ASV mode option [37].

Early studies performed by Hamilton Medical on their INTELLiVENT mode have shown that, when compared to the original ASV mode, patients spend more time with optimal ventilation and less time with nonsecure ventilation [38]. It also shows that this mode delivered lower volumes and pressures for equivalent gas exchange. Ventilation control modes such as INTELLiVENT, which closed-loop ventilation systems that offer automatic adjustment of oxygenation and ventilation, are the future of artificial ventilation. However, one of the challenges a system so complex as this one will face is the development of standardized vocabularies, taxonomies, and data transfer protocols in order to assure higher levels of accuracy, security, and usability [33].

CHAPTER 7

CONCLUSION

The artificial mechanical ventilator has come a long way since the 1900s. Being a medical device, the ventilator has been a piece of technology that has been continuously updated to guarantee the highest level of comfort to its users, the patients. Since its main function is to control the patient's breathing, such machine needs to be highly tested and have a very low probability of failure, hence the strict laws and legislations established by the ISO. The COVID-19 pandemic has required standard organizations like the ISO to loosen their requirements, but it has also generated an amazing sense of comradery. Companies like General Electric and Medtronic have allowed non-ventilator manufacturers like Ford and Tesla to use their patented designs to help fight the artificial ventilator shortage in the US and around the world.

Even though the world has been hit with this pandemic, the future of ventilators for the long term remains the same – to make its interface more accessible. The Hamilton Medical G5 ventilator with the INTELLiVENT-ASL mode is just the beginning of a closed-loop and automated artificial ventilation.

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