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Ventilator sharing for two patients: individualisation of tidal volumes in volume and pressure-controlled ventilation modes

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Abstract

The COVID-19 outbreak leads to a rapid global increase in the number of patients requiring mechanical ventilation. A potential mismatch between the number of ventilators available and the number of patients requiring one could be avoided by using the same ventilator for ventilation of two patients. This is what is called ventilation sharing.

By adding medical-grade filters, valves, and sensors, to this ventilation sharing set-up, it could be possible to provide individualised ventilation. This master thesis aims to test this assumption. Specifically, this work investigates the possibility of individualising tidal volumes for two patients undergoing differential multi-ventilation in volume and pressure-controlled ventilation mode. The ultimate goal is to answer to the medical validation of this way of ventilation sharing.

This work contributed to the characterisation of the volume distribution between patients. Firstly, the multi-ventilation differential circuit was set up and volume and pressure sensors were implemented. Secondly, characterisation of different valves, to select the one to be used in the set-up, was performed. Thirdly, a study on the distribution of volumes between two patients was carried out for two different ventilation modes, to determine if volume individuation was possible. A first clinical simulation was also performed.

The results of the tests show that the individualisation of tidal volumes is possible in both ventilation modes. However, the use of the pressure-controlled ventilation mode is safer.

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Introduction

Ventilators are extremely complex machines that play a crucial role in critical care conditions, surgical intervention, and procedures requiring anaesthesia in replacing the patient's lung. These complex machines are produced by a small number of highly specialised companies [1].

Towards the end of 2019, a respiratory infectious disease caused by an emerging virus, the coronavirus SARS-CoV2 appears in China. Very quickly this virus spreads from country to country and finally becomes the largest infectious pandemic since the Spanish flu of 1918[2].

Due to the damage caused to the respiratory system by this virus, this leads to a rapid global increase in the number of patients requiring mechanical ventilation, sometimes exceeding the number of ventilators available in health care facilities [3]. This has caused severe disruption to hospital infrastructure. In this context of a potential mismatch between the number of ventilators available and the number of patients requiring one, one solution receiving considerable interest is to use the same ventilator for several patients connected in parallel. This solution is usually referred to as "ventilation sharing".

This concept has already been addressed in 2006 by Neyman and colleagues who have proven its feasibility [4]. However, the many efforts made to characterise it at the onset of the pandemic led the medical associations to issue a joint statement explicitly warning medical professionals not to share mechanical ventilators with this current approach. One of the main reasons raised for this concern is that ventilator sharing lack the ability to individualise ventilation to each patient.

Subsequently, a new design, an improved version of the ventilation sharing, was developed by a team of Belgian researchers. By combining different modules, it would be possible to individualise the ventilation parameters for each patient. This new way of sharing the ventilator is called "differential multi-ventilation".

The tests performed on differential multi-ventilation are relatively few and always performed for one ventilation mode only, pressure-controlled ventilation. The interest of this thesis is therefore to replicate a differential multi-ventilation system and afterwards to analyse whether this system allows individualised ventilation for each patient in both pressure and volume-controlled modes.

This paper will be structured in the following way. The first part is first made of the state of the art about ventilation sharing. It gives all the theoretical background needed to understand the working principle of respiration and mechanical ventilation. There is then a very short chapter dedicated to explaining in detail the research question of this master thesis. The second part of this paper is devoted to the study of the differential multi-ventilation way of sharing a ventilator. The first chapter is dedicated to the description of the materials and methods used in this work. The following chapter presents all the results of the differential multi-ventilation study. The next chapter discusses the results of the differential multi-ventilation study, their interpretations and limitations. Finally, a global conclusion summarizing the findings of this master thesis will be made

Part I

State of the art

Chapter 1

Context and current knowledge

1 The Covid-19 epidemic and its consequences on the supply of medical equipment

At the end of 2019, an epidemic caused by an emerging virus, the coronavirus SARS-CoV-2 (for Severe Acute Respiratory Syndrome Coronavirus 2) appeared in China. The SARS-CoV-2 is a newly discovered virus closely related to bat coronavirus, pangolin coronavirus, and SARS-CoV [5].

Due to its high contagion rate, the virus rapidly spread from region to region, then from country to country. The number of infected patients is growing day by day all over the world, as shown in figure 1.1, which shows a timeline map of cumulative confirmed cases per million people. On March 11, 2020 the World Health Organization designated Covid-19 a global pandemic (*CO* for corona, *V* for virus, *D* for disease and *19* for when the outbreak was first identified, i.e. end 2019). The Covid-19 pandemic became the largest infectious pandemic since the Spanish flu of 1918[2].

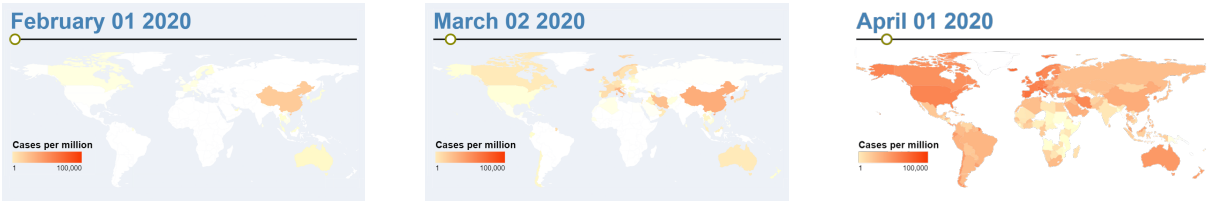
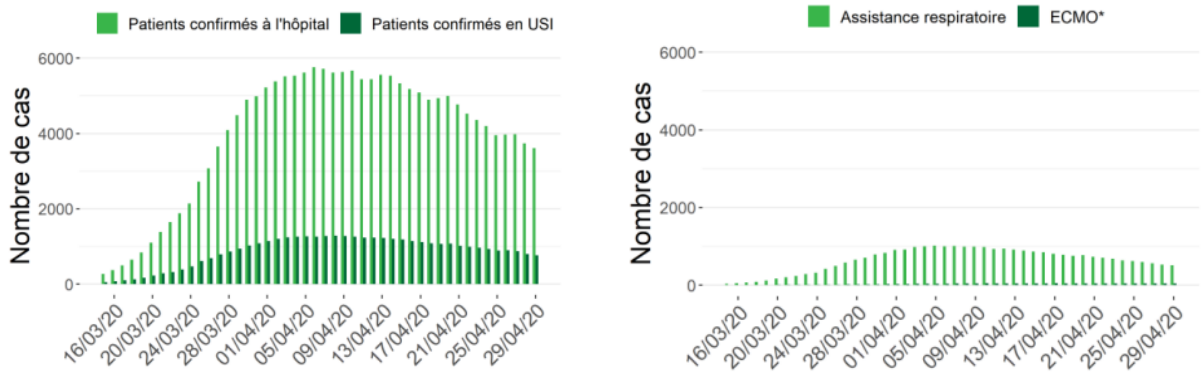


Figure 1.1: Timeline map of cumulative confirmed Covid-19 infected patients per million people.

In addition to being highly contagious, the SARS-CoV-2 virus has a significant impact on human respiratory health, because it is more active in environments with high humidity levels. When the virus enters the human body, it comes into contact with the mucous membranes that line through the nose, mouth, and eyes. Like other viruses, this one infects a healthy cell and exploits it to produce new viral components. As it replicates, adjacent cells are infected by the new viruses. It then travels to the lungs, where it has the most severe consequences. The inflammation due to the presence of the virus in the lung can indeed lead to an acute respiratory distress syndrome (ARDS), resulting in a decrease in oxygen saturation [6, 7, 8, 9]. In up to 17% of cases, severely affected patients require mechanical ventilation support [7]. The COVID-19 outbreak required therefore hospital and intensive care unit (ICU) capacity to dramatically increase to handle the sudden rise of patients needing respiratory assistance.

Due to the damage caused by this new virus, strict sanitary measures in most of the affected countries have been taken. These include social distancing, wearing masks, using hydro-alcoholic gel and other personal protective equipment. The goal of this practice is to flatten the curve of new infections, thereby avoiding a surge in demand on the health care system. However, the effects of these emerging sanitary measures took weeks to appear and the need for these medical equipments is only growing. Panic buying, massive disruption of the global supply chain and quarantine measures prevented the flow of equipment and production of medical devices, causing a worldwide shortage of medical equipment.



(a) Evolution of the number of hospitalised people

(b) Severity of inpatient cases

Figure 1.2: Hospitalisations due to Covid-19 in Belgium from 16th march 2020 to 31st april 2020. On April 29, 3609 hospital beds, including 769 intensive care unit beds, were occupied by confirmed COVID-19 patients, and 517 patients required ventilatory assistance[10].

This shortage particularly affects mechanical ventilators, as illustred by figure 1.2. This leads to more drastic consequences , forcing ICU staff to choose which patients to give respiratory assistance to and which patients to let die of respiratory distress[3]. Ventilators are extremely complex machines that play a crucial role in replacing the patient's lungs. As a result, they are produced by a small number of highly specialized companies at a high cost, which makes its supply even more complex.

The issue of how to bridge the gap between demand and supply for ventilators soon became a concern for many of us. The following two possible solutions have been highlighted:

1. Manufacture emergency/low tech respirators that provide basic functionality, at a lower cost and in a shorter time frame;

2. Increase the capacity of existing ventilators by connecting multiple patients to a single ventilator.

These two solutions will be developed later in this work, with a major interest for the second solution since it is the basis of this work.

It should be noted that in addition to being essential in case of health crisis as during the Covid-19 crisis, these solutions could benefit other causes. For example to be used in case of natural disaster or war, when the number of people requiring respiratory assistance suddenly increases. It is also useful for countries that do not have enough respirators to provide health care for their entire population. An article in the 'Le journal du Médecin' [3] published in April 2020 states that the Gaza Strip has about 20 artificial respirators for two million inhabitants; the Central African Republic has a total of three respirators for a population of five million and Burkina Faso has only 12 intensive care beds for 20 million citizens.

2 Overview of the respiratory system

We can survive for some time without food and water, but we absolutely cannot do without oxygen. The thousands of billions of cells in the body need a continuous supply of oxygen to perform their vital functions. This supply of oxygen is provided by the respiratory system, which gets oxygen from the environment and releases its waste products into it.

External respiration [1] refers to the movement of gases between the environment and the cells of the body and serves a variety of purposes. The most well-known is the supply of oxygen ($O2$) for aerobic metabolism in human cells and the elimination of carbon dioxide ($CO2$) generated. But it also helps to keep blood parameters like paCO2, paO2, SaO2, and pH in a healthy range and we may also consider its function in speech.

There are four processes that underlie external respiration[11]:

1. Air exchange between the atmosphere and the lungs. This process is commonly called ventilation or breathing. The mechanisms by which ventilation takes place are called ventilatory mechanics.
2. Exchange of O2 and CO2 between the lungs and the blood by diffusion.
3. Transport of O2 and CO2 in the blood.
4. Exchange of gases between blood and cells, in the purpose of cellular respiration

2.1 Anatomy of the respiratory system

The respiratory system includes the nose and nasal cavities, the pharynx, the larynx, the trachea, the right and left primary bronchi and their ramifications, as well as the lungs which contains the alveoli (see figure 1.3). This system is therefore a set of airways that filters the air and transports it to the inside of the lung where the gas exchanges will take place at the level of the alveolar sacs.

Functionally, the respiratory system can be divided into a *conducting zone* (also called conducting airways) and a *respiratory zone*. Organs and structures not directly engaged in gas exchange are included in the conducting zone of the respiratory system. The respiratory zone is where gas exchange takes place, and is composed exclusively of microscopic structures, namely the respiratory bronchioles as well as the alveoli[11, 13].

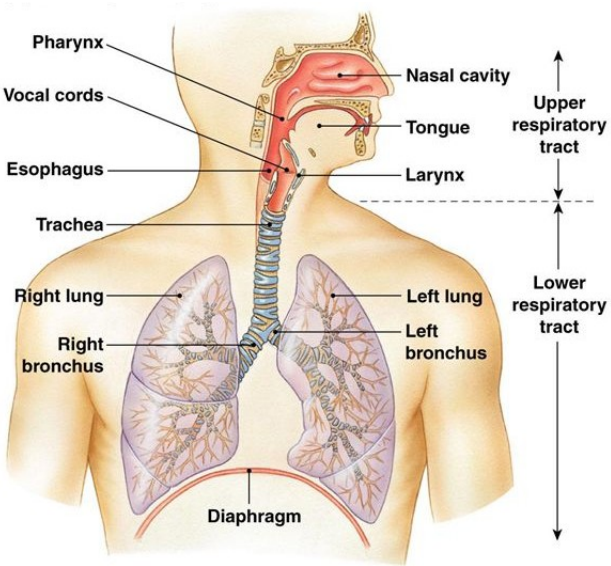


Figure 1.3: Main respiratory structures[12]

[1] In the context of human physiology, we can talk about either internal or external respiration. Internal respiration refers to cellular respiration.

2.1.1 Conducting zone

The conducting zone's main duties are to provide a passage for incoming and outgoing air, remove debris and pathogens from entering air, and warm and humidify it. Air enters the respiratory system through the mouth and the nose and passes into the pharynx. From the pharynx, the airflow goes into the trachea passing through the larynx. The trachea is the main conducting airway. It descends into the thoracic cavity, where it divides into two main bronchi, one for each lung. In the lungs, the bronchi gradually divide into smaller bronchi, the bronchioles. The bronchioles then divide again into terminal bronchioles, allowing air to enter alveoli, which is part of the respiratory zone.

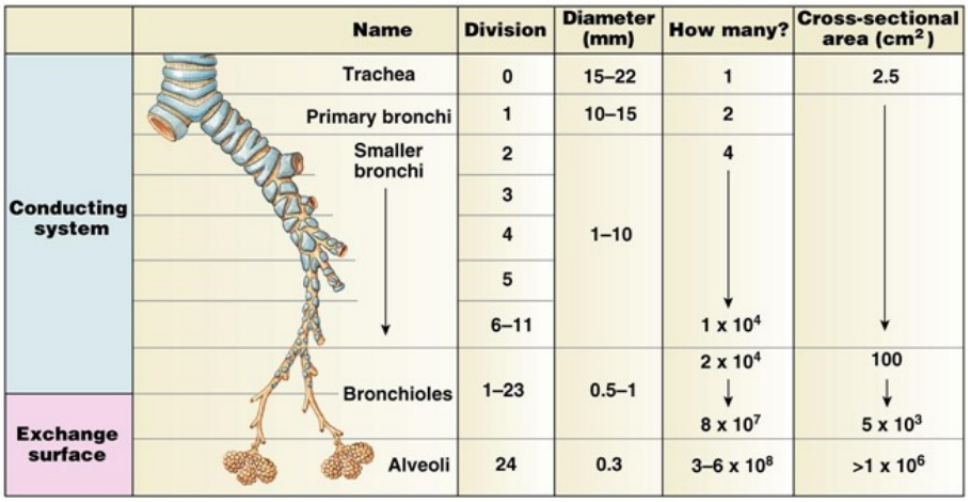


Figure 1.4: Bronchial tree[12]

Ultimately, there are 20 to 25 airway divisions in the lungs, and the whole is often referred to as the bronchial tree. As illustrated in figure 1.4, the diameter of the airways gradually decreases from the trachea to the terminal bronchioles. But with each division, the number of airways increases, so the total cross-sectional area increases with each division of the airways

2.1.2 Respiratory zone

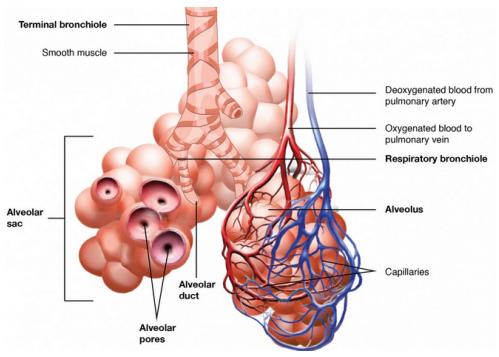


Figure 1.5: Respiratory zone of the respiratory system. Highlight on the terminal and respiratory bronchiole, alvolar duc, alveolar sac and the alveoli[14].

The respiratory zone, in contrast to the conducting zone, involves structures that are directly engaged in gas exchange. It begins where the terminal bronchioles join a respiratory bronchiole, the smallest type of bronchiole, which then leads to an alveolar duct, opening into a cluster of alveoli (see figure 1.5). Alveoli are thin-walled air-filled sacs that make up most of the lung tissue. Indeed, the lungs contain 3–600 million alveoli with a total surface area, for gas exchange, of approximately 140 m^2 [14]. Their main role is to ensure gas exchange (section 2.3). Within these alveoli, the air breathed in and the blood are separated only by a thin membrane called the alveolar-capillary mem-

brane.

2.1.3 Lungs and pleura

The two lungs, which contain all the components of the bronchial tree beyond the primary bronchi, occupy most of the space in the thoracic cavity. The lungs are soft and spongy because they are mostly air spaces surrounded by the alveolar cells and elastic connective tissue. The right lung is shorter, wider, and has a larger volume than the left. It is split into three lobes, with one of the secondary bronchi supplying each lobe. The left lung is longer and narrower than the right lung. It features an indentation on its medial surface for the apex of the heart, known as the cardiac notch. The left lung is divided into two lobes.

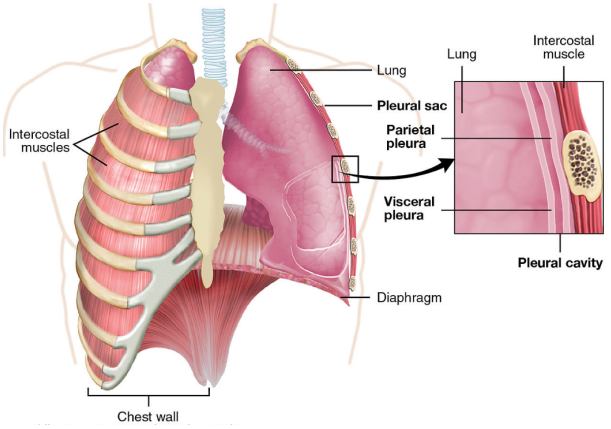


Figure 1.6: Pulmonary pleura[14]

The pleura is a double-layered serous membrane that surrounds each lung. The side of the pleura that covers the lung is referred to as the visceral pleura and the side of the pleura which covers the chest wall is called the parietal pleura. These two sides are continuous and meet at the hilum of the lung. The pleural cavity is the tiny gap between the visceral and parietal pleurae, as illustrated on figure 1.6. The pleura produces a thin coating of serous fluid, which is contained inside it. As the lungs expand and deflate, the fluid works as a lubricant to minimise friction as the two layers move against each other. The serous fluid also generates surface tension, which pulls the visceral and parietal pleura adjacent to each other and avoid lungs to collapse during breathing (see also section 2.2.4). Indeed, there are two forces that tend to move the visceral pleura and the parietal pleura away from each other.

1. The lungs tend to recoil inwards due to elastic properties of lung tissues;
2. The chest wall tends to recoil outwards;

However, due to the high cohesive force between the two pleura, these two forces balance each other and this prevents the lungs from collapsing [15, 16, 17].

2.2 Mechanics of ventilation

2.2.1 Gas laws principles

The air movement during pulmonary ventilation is governed by the basic principles of gas laws like Boyle-Mariotte's law. A reminder of the basic principles of gas law is available in Appendix A 1.

2.2.2 Pulmonary ventilation

Pulmonary ventilation is a periodic phenomenon that consists of a succession of inspiratory movements during which a certain volume of air is inhaled and expiratory movements during which a certain volume of air is exhaled. There are three major pressures that drive pulmonary ventilation [18]:

- **Atmospheric pressure (Patm)**
- **Intra-alveolar pressure (Palv).** It is the pressure of the air within the alveoli, which changes during the different phases of breathing. Because the alveoli are connected to the atmosphere via the tubing of the airways, the interpulmonary pressure of the alveoli always tends to equalize with the atmospheric pressure. At rest, in between breaths, Palv = Patm.
- **Intrapleural pressure (Pintra).** It is the pressure within the pleural cavity. Similar to intra-alveolar pressure, intrapleural pressure also changes during the different phases of breathing. However, due to certain characteristics of the lungs, the intrapleural pressure is remains approximately −4 mm Hg (ralative pressure) throughout the breathing cycle [18]. This negative pressure acts like a suction to keep the lungs inflated. If this becomes zero, such as in the case of pneumothorax, then the pleural cavity has the same pressure as the atmosphere and the lung would collapse.

Inspiration

The diaphragm and external intercostal muscles contract during inspiration, extending the thoracic cavity and lungs (figure 1.7(b)). As a result of the increased volume, the intra-alveolar pressure decreases by approximately 1mm Hg (relative pressure), allowing outside air to flow into the lungs until Palv = Patm. Deep breathing requires more forceful contractions of the diaphragm, intercostal muscles, and involves additional muscles to produce larger changes in the thoracic volume.

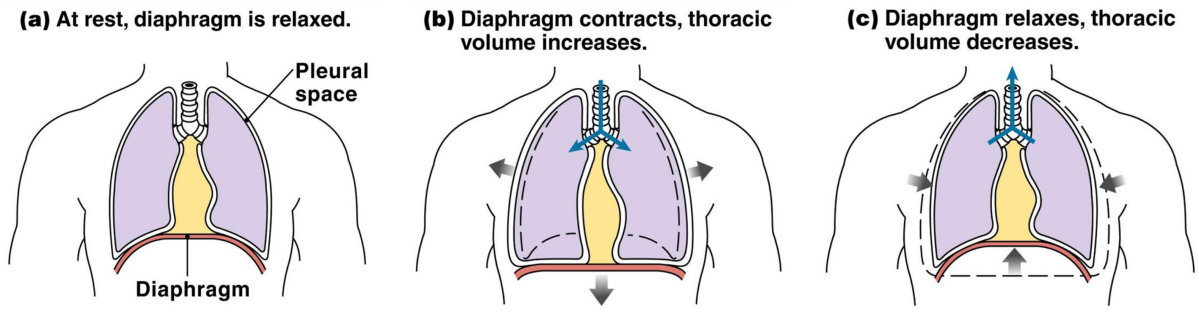


Figure 1.7: Changes in volume of the thoracic cavity during breathing cycles[12]

Expiration

While inspiration requires muscular contraction, and hence energy expenditure, expiration during quiet breathing is a passive process that relies on the elasticity of the lungs and the rib cage, i.e. their ability to spring back to their original dimensions. As the diaphragm returns to its original position and other intercostal muscles relax, thoracic and lung volumes decrease (see figure 1.7(c)). As a result, Palv increases and becomes higher than Patm, pushing air out of

the lungs. Deep expiration, unlike quiet expiration, is an active process during which specific muscles contract to increase Palv.

2.2.3 Alveolar ventialtion is a measured of pulmonary ventilation efficiency

The tidal volume inspired at every breath (V_T) during pulmonary ventilation contains two parts. The part that participates in alveolar gas exchange is the alveolar tidal volume (V_{ALV}). The other part that does not participate in the gas exchange is dead space. Dead space volume (V_{DS}) is the volume of air that fills the airways when breathing and that does not reaches the alveoli.

The efficiency of pulmonary ventilation, whose main role is to provide a sufficient amount of oxygen to the body, depends on the alveolar volume and frequency of exchange cycles. That's what is called alveolar ventilation and is measured by :

$$MV_{ALV} = V_{ALV} \cdot RR = (V_T - V_{DS}) \cdot RR \tag{1.1}$$

where MV_{ALV} is the alveolar ventilation over a minute and RR is the respiratory rate.

2.2.4 Physical factors affecting pulmonary ventilation

We already know that the respiratory muscles consume energy. Other factors of the respiratory system also influence the effort that must be expended to ventilate. Those two factors are airway resistance and lung compliance. They vary from person to person and also vary with illness.

Resistance of the airways

[18] Resistance is a force that slows motion, in this case, the flow of gases. The size of the airway is the primary factor affecting resistance. A small tubular diameter forces air through a smaller space, causing more collisions of air molecules with the walls of the airways. The following formula helps to describe the relationship between gas flow (Q), airway resistance (R) and pressure difference (ΔP = Patm-Palv):

$$Q = \frac{\Delta P}{R} \tag{1.2}$$

The equation states that the gas flow is inversely proportional to the resistance. In other words, this demonstrates that as resistance increases, the pressure gradient must also increase to maintain the same rate of flow into the alveoli.

Hagen-Poiseuille equation states that: .

Hagen-Poiseuille equation

Consider a tube of length L [m] and radius r [m]. In this tube, a fluid of viscosity η [Pa.s] flows due to a pressure difference ΔP [Pa]. If the flow is laminar, then the flow rate Q [$m^3 s^{-1}$] is given by :

$$Q = \frac{\pi \Delta P r^4}{8 \eta L}$$

During quiet breathing, airflow through most branches of the bronchial tree is laminar and steady [19]. Therefore, Hagen-Poiseuille formula can be used to higlight the parameters influencing R:

$$R = \frac{8\eta L}{\pi r^4} \tag{1.3}$$

Given this equation, it is clear that the radius is the most important factor in airway resistance and that small changes in radius can lead to significant changes in airway resistance. For example, if the radius of the tube doubles, the resistance decreases by a factor of 16.

Compliance

Compliance is the measure of distensibility of matter and specifies the ease with which matter can be stretched or distorted. Compliance is the opposite of elastance (E) or elastic recoil.

Lung compliance (C_L) measures the change in lung volume (ΔV) as a function of the change in transpulmonary pressure (ΔP = $Palv - Pintra$):

$$C_L = \frac{1}{E} = \frac{\Delta V}{\Delta P} \quad [\frac{ml}{cmH_2O}] \tag{1.4}$$

and is usually expressed in millilitres per centimetre of water (ml/cmH2O).

The compliance of the lungs is often described by examining the pressure-volume characteristics of the lung under static conditions: when there is no flow of air and the respiratory muscles are relaxed. The transpulmonary pressure under these conditions reflects the magnitude of the elastic recoil pressure of the lungs. Representative static compliance curves for different (normal vs diseased) lungs are shown in figure 1.8. Compliance is determined from the slope of the pressure-volume curve of the lung. Normal respiratory compliance in adults is approximately 150–200 ml/cmH2O [20].

Both overly high and overly low compliances are abnormal. 'Soft lung' refers to the situation where lung compliance is abnormally high (i.e. one with low elastic recoil). Even a small pressure gradient results in a sizeable lung volume change. Typical clinical examples are chronic obstructive pulmonary disease (COPD) and emphysema patients. 'Stiff lung' refers to the situation where the lung compliance is abnormally low (i.e. one with high elastic recoil). Even a large pressure gradient results in a small lung volume change. Typical clinical examples are acute respiratory distress syndrome (ARDS), acute lung injury (ALI) and fibrosis [21, 22].

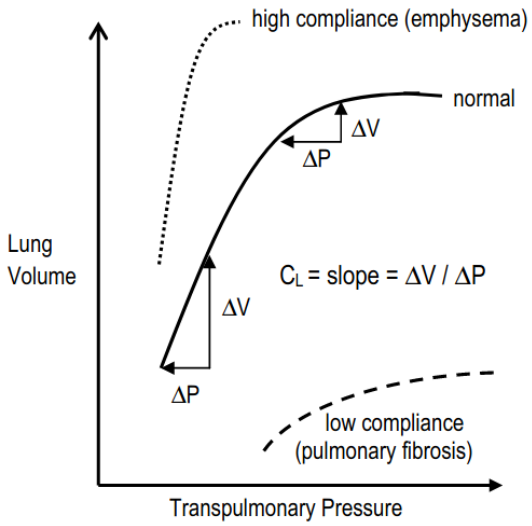


Figure 1.8: Static lung compliance curves

2.2.5 Respiratory volumes and capacities

Lung volumes are also known as respiratory volumes. It refers to the volume of gas in the lungs at a given time during the respiratory cycle. Lung capacities are derived from a summation of different lung volumes. These volumes change depending on the depth of breathing, ethnicity, gender, age, body composition, and the presence of specific respiratory diseases [23]. Tables 1.1 and 1.2 summarises the different respiratory volumes and capacities, figure 1.9 illustrates them.

Measure	Average value for men	Average value for woman	Description
Tidal volum (V_T)	500 mL	500 mL	Amount of air that normally enters the lungs during quiet breathing
Inspiratory reserve volume (IRV)	3100 mL	1900 mL	Amount of air that can be inhale with a effort past a normal inspiration
Expiratory reserve volume (ERV)	1200 mL	700 mL	Amount of air that can be exhale with a effort past a normal tidal expiration
Residual volume (RV)	1200 mL	1100 mL	Amount of air remaining in the lungs after maximal exhalation

Table 1.1: Lung volumes for men and women [13]

Measure	Average value for men	Average value for woman	Description
Total Lung Capacity(TLC)	6000 mL	4200 mL	Maximum amount of air a person can hold in the lungs after a forceful inhalation: TLC = V_T+IRV+ERV+RV
Vital capacity (VC)	4800 mL	3100 mL	Maximum amount of air that can be exhaled after maximal inhalation : VC = V_T+IRV+ERV
Inspiratory capacity(IC)	3600 mL	2400 mL	Maximum amount of air that can be inhaled past a normal tidal expiration : IC = V_T+IRV
Function Residual Capacity(FRC)	2400 mL	1800 mL	Amount of air that remains in the lung after a normal tidal expiration : FRC = EVR + RV

Table 1.2: Lung capacities for men and women [13]

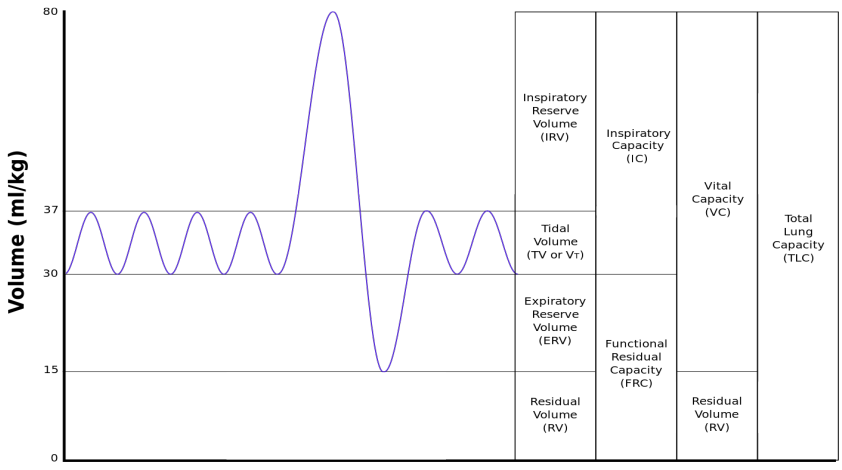


Figure 1.9: Respiratory volumes and capacities[24]

2.3 Gas exchange

The purpose of the respiratory system is to perform gas exchange. Pulmonary ventilation provides air to the alveoli for this gas exchange process. The Appendix A 2 provides an explanation of this process.

3 Overview of mechanical ventilation

Mechanical ventilation is a form of life support that is used when a person is unable to breathe sufficiently on their own, which may be the case during disease or injury of respiratory system. It uses mechanical ventilator [2], i.e. a machine that successfully aids, supports, or replaces deficient natural lung ventilation, artificially meeting the critical demands of breathing when used correctly.

3.1 Different types of mechanical ventilation

There are two types of mechanical ventilation.

- *Negative-pressure ventilation*: it sucks the air into the lungs by making the chest expand and contract. In negative pressure ventilation, the surface of the thorax is exposed to a subatmospheric pressure during inspiration. The subatmospheric pressure causes thoracic expansion and an attendant decrease in pleural and alveolar pressures, thereby creating a pressure gradient for air to move from the airway opening into the alveoli, like during natural breathing. When the pressure surrounding the thorax increases and becomes equal as the atmospheric pressure, expiration occurs passively owing to the elastic recoil of the lung and chest wall[25, 26]. Figure 1.10 illustrates a ventilator used for negative-pressure ventilation.
- *Positive-pressure ventilation*: it pushes air into the lungs. In positive-pressure ventilation, the pressure of the air going into the lungs is raised above atmospheric pressure. This creates a pressure gradient, which allows air to enter the lungs. Again, expiration of air happens passively. Figure 1.11 illustrates a modern ventilator used for positive-pressure ventilation. Positive pressure ventilation can be provided in *non-invasive* way, involving various types of face masks or in *invasive* way, involving an endotracheal tube and therefore requires deep sedation.

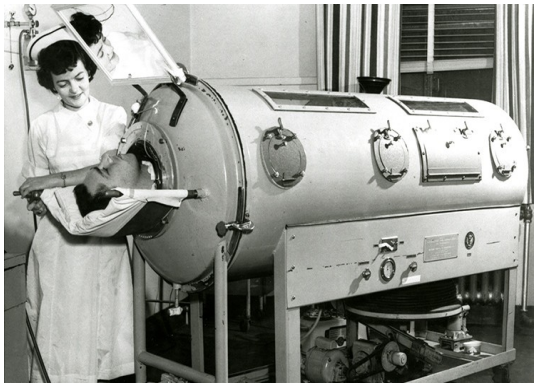


Figure 1.10: Ventilator used for negative-pressure ventilation[27]

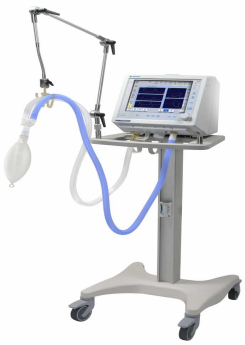


Figure 1.11: Modern ventilator used for Positive-pressure ventilation[28]

Currently, invasive positive pressure ventilation is the most common form of mechanical ventilation in hospitals. That's why the rest of this work will only focus on this form of ventilation.

[2] Ventilator, respirator, or breathing machine are all names for a mechanical ventilator.

3.2 The respiratory system's equation of motion

The respiratory system's equation of motion may be used to describe the forces at work during mechanical ventilation at any given moment in time.

To understand this equation, the lung-ventilator unit can be thought of as a pipe with a balloon at its end, as represented on figure 1.12. The pipe represents the airways, the endotracheal tube, and the ventilator tubing while the balloon represents the alveoli.

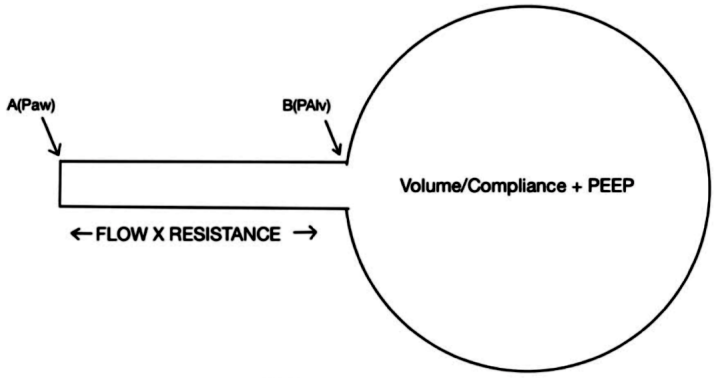


Figure 1.12: Simplification of the lung-ventilator unit

Pressure at point B is equivalent to alveolar pressure and represents the sum of elastic pressure (in the balloon, P_{elast}) and $PEEP$ (the alveolar pressure at the beginning of inspiration).

Pressure at point A is equivalent to airways pressure measured by the ventilator and is the sum of the resistive pressure (in the pipe, P_R) and P_{elast}. It is important to understand that the pressure measured by the ventilator does not correspond to the pressure in the alveoli.

In a more stringent way, the respiratory system's equation explain how pressure, volume, and flow vary during ventilation whilst compliance, resistance and PEEP are constant (see figure 3.2):

$$P_{aw} = P_0 + P_R + P_{elast} \tag{1.5}$$

$$= P_0 + (R \cdot Q) + (V_T \cdot \frac{1}{C_L}) \tag{1.6}$$

where :

- P_{aw} is the airway pressure $[cmH_2O]$;
- P_0 is the alveolar pressure at the beginning of inspiration, i.e. the PEEP $[cmH_2O]$;
- R is the resistance to flow$[cmH_2O \cdot s \cdot L^{-1}]$;
- Q is the flow $[L \cdot s^{-1}]$;
- V_T is the tidal volume $[L]$;
- C_L is the compliance of the respiratory system$[L \cdot cmH2O^{-1}]$;

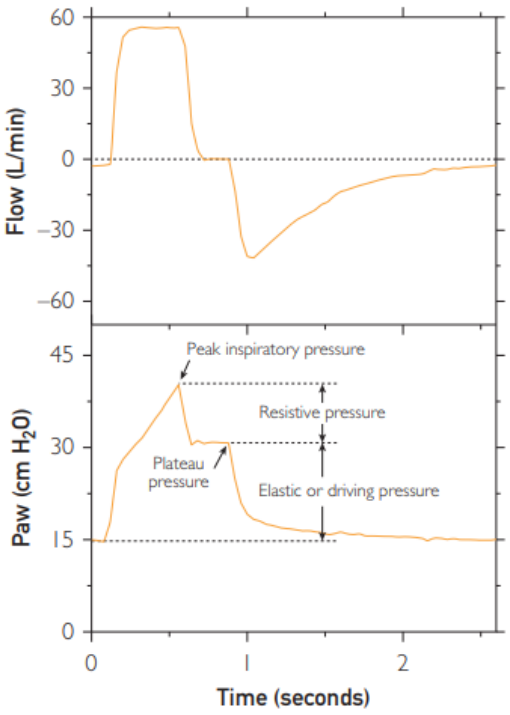


Figure 1.13: Breathing curves derived from the respiratory system's equation [29]

3.3 Ventilator modes and parameters

Ventilators are the centerpieces of this work. Understanding how it works and how to adjust the ventilation parameters is therefore an essential step in the realization of this work. This is what this section is dedicated to.

3.3.1 Ventilation modes

Looking at the relationships between pressure, flow, and volume in equation 1.6, we understand that by setting one, the other two become constant. It also shows that it is not possible to preset more than one variable at a time. Ventilatory assistance can therefore be delivered either in volume or pressure. This is referred to as:

- *Volumetric (volume mode)* where a certain pre-adjusted inspiratory flow/volume is kept constant uncontrollably varying lung pressure conditions;
- *Barometric (pressure mode)*, where airway pressure is adjusted and remains constant uncontrollably varying flow/volume conditions [30].

In addition to that, we distinguish three different types of mechanical ventilation [31]:

- Controlled ventilation: no inspiratory effort by the patient, the whole ventilation is provided by the ventilator;
- Assisted ventilation: assistance delivered by the ventilator in response to a request from the patient (trigger), sharing the work of breathing between the patient and the ventilator. The assisted modes can therefore only be used if the ventilated patient has a spontaneous respiratory activity;
- Assisted-controlled ventilation: associates the two previous modalities, i.e. ventilation is provided by the ventilator but the patient can also trigger a breath.

In the framework of this work, only controlled ventilation modes will be considered.

Volume-controlled mode (VC)

Priority is given to the delivery of a tidal volume (V_T), which corresponds to the basic set point: this is the parameter to be adjusted in first intention. This V_T is administered to the patient at a set respiratory rate (RR), regulated in order to get a suitable minute volume. This RR determines the number of breath cycle per minute. During each breath cycle, V_T is delivered during a preset inspiratory time (t_i). When V_T is delivered, the pressure in the airway progressively rises until it reaches a peak pressure. This peak pressure is affected by the V_T administered (it rises with Vt) as well as the patient's airway, which provides some resistance (the peak pressure increases if the airway is narrowed or obstructed). It is important to note that in this mode, the pressure is not controlled but must be monitored in order to avoid barotrauma (see section 3.4). This can be done by using alarms.

However, peak pressure reflects the pressure in the airways (P_{aw}), not in the alveoli. This is why the V_T administration phase is usually followed by a inspiration pause (T_{pause}), during which no volume is administered. This allows time for the pressure in the airways to equalize

with the pressure in the alveoli. Indeed, there is no air flow and therefore no resistance ($P_{aw} = P_{alv}$) [32, 33]. The pressure during this pause is the plateau pressure (P_{plat}).

Expiration is triggered when V_T has been administered or when the maximum alarm pressure (p_{max}) has been reached. In the second case, the device maintains the p_{max} value until the end of the inspiratory time and then switches to expiration. It is therefore possible that the set tidal volume is not fully applied if the maximum ventilation pressure (p_{max}) is reached during inspiration [32]. Figure 1.14a illustrates the typical curves we get in a volume-controlled mode.

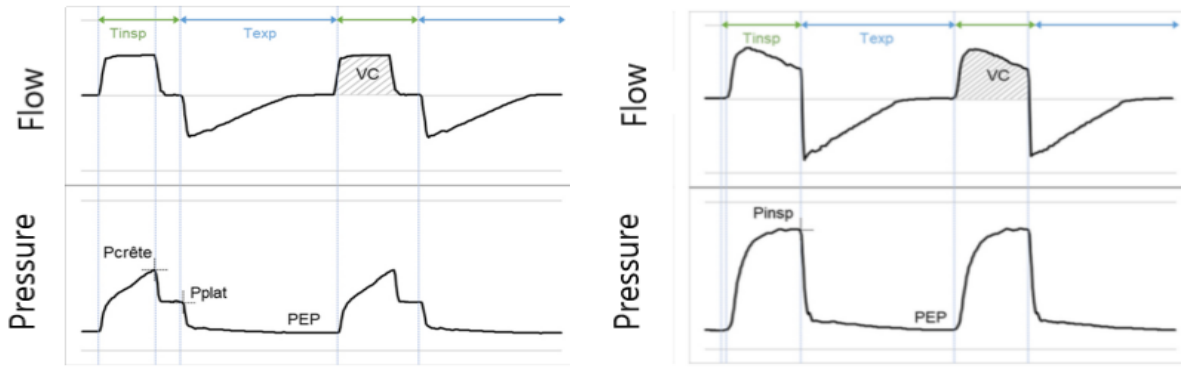
Pressure-controlled mode (PC)

During pressure-controlled ventilation, a constant peak inspiratory pressure (PIP) is applied to the airways during the inspiratory phase, at a set frequency (RR). Again, this RR determines the number of breath cycle per minute. During each breath cycle, the PIP is maintained during a preset inspiratory time (t_i). The peak pressure is constant in the lungs while the volume varies. Pressure-controlled ventilation avoids therefore dangerous peak pressures and thus protects the lungs from possible ventilation-induced injuries [34].

The volume inspired or expired is the result mechanical properties of the patient's respiratory system (i.e. inspiratory resistance, compliance and intrinsic PEEP) and the pressure applied. This volume must be monitored.

Between the expiration and the inspiration, the airway pressure corresponds to the positive expiratory pressure ($PEEP$). The inspiratory flow is highest at the start of inspiration and then decreases rapidly (i.e. decelerating flow pattern). The rate at which the ventilator pressurises the airways is determined by the slope of the pressure increase, i.e. the inspiratory rise time (IRT). The faster the slope, the greater the assistance delivered.

At equal V_T, the peak pressure is lower in PC mode than in VC mode, due to the decelerating flow which reduces the resistive pressure during inspiration. However, the P_{plat}, which reflects the maximum alveolar pressure, is not modified[32]. Figure 1.14b illustrates the typical curves we get in a pressure-controlled mode.



(a) **(b)**

Figure 1.14: Basic waveform for volume (a) and pressure-controlled (b) ventilation modes. *$P_{crête}$ is PIP, PEP is $PEEP$ and VC is V_T* [35]

3.3.2 Ventilators settings

Many parameters can be adjusted on a medical ventilator, which allows it to be able to adapt to each patient and each disease. However, the parameters that the operator is offered to set one ventilator or another can vary.

Table 1.3 gives a short description of all the parameters and the relationships between them. It also highlights the most common range of values, or values not to be exceeded when adjusting each parameter on the ventilator, in order to use protective ventilation.

Parameter	Units	Description	Ranges of values
Tidal Volume (V_T)	ml	Volume inspired and expired with each breath	4-6 ml/ kgPBW [3]
Respiratory rate (RR)	$breaths/min$	Number of breaths (1 breath = 1 inspiration + 1 expiration) a patient take per minute	8-30
Minute volume (MV)	L/min	Volume delivered to a patient over a minute. Minute volume is what determines the patient's paCO2. $MV = V_T \times RR$	7-12
Inspiratory time (T_i)	s	Duration of the insufflation. Ventilator switches to expiration when the set T_i is over	0.8 - 1
I:E ratio	/	Ratio between inspiratory time and expiratory time. $I = RR \cdot T_i$ and $E = (60 - I)$.$I = RR \cdot T_i$ and $E = (60 - I)$. Or, I:E = $\frac{T_i}{T_e}$	1:2, 1:3 , rarely 1:1 , 1:4
Inspiration pause (T_{pause})	%	Pourcentage of T_i durig wich no volume is administered, allowing the pressure in the airways to equalise with the pressure in the alveoli	5-15
Inspiration rise time (IRT)	%	Pourcentage of T_i that determines the time period tore-ach the set airways flow/pressure. reach the set airways flow/pressure. Figure 1.15 illustrates it	5-10
Peak inspiratory flow rate (PIF)	L/min	Used in VC ventilation. Maximum flow at which a set tidal volume breath is delivered. $V_T = T_i \cdot PIF$ (Need to remove T_{pause} and IRT from T_i to use this equation)	30-75
Flow pattern shape	/	Either constant or decelerating, see figure 1.16	/
Fraction of inspired oxygen (F_iO_2)	%	Oxygen concentration in the inspiratory gas	21-100
Positive end-expiratory pressure ($PEEP$)	cmH_2O	Pressure at end-expiration, which prevent lung from collapsing. Patients have a Auto-PEEP (Intrinsic PEEP) $\rightarrow$PEEP$_{tot} = Auto - PEEP + PEEP$	≥ 5
Peak inpiratory pressure (PIP)	cmH_2O	Used in PC ventilation. Inspiratory pressure above PEEP	10-35 ($\rightarrow$ In order to deliver desired V_T)
Plateau pressure (P_{plat})	cmH_2O	Pressure after a inspiratory pause, roughly represents the pressure in the alveoli	< 35

Table 1.3: Main Ventilator settings, ranges and relations for VC and PC ventilation

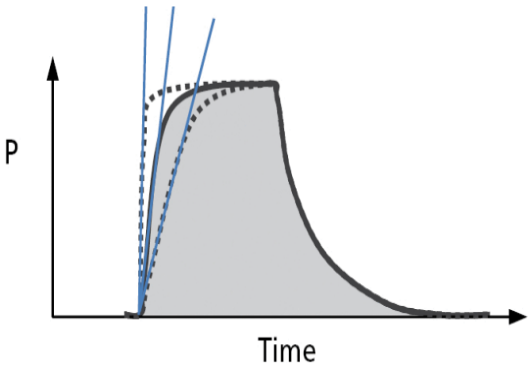


Figure 1.15: Ventilator inspiration rise time effect

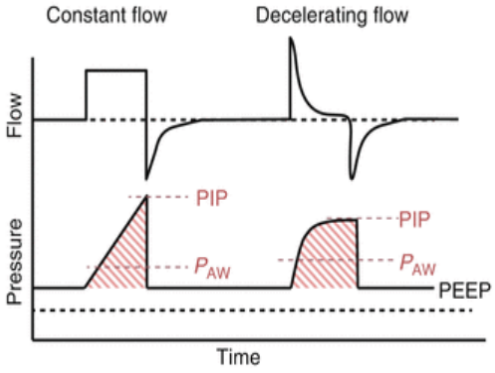


Figure 1.16: Comparison between constant and decelerating inspiratory flow patterns[37]

3.4 Risks and complications of mechanical ventilation

As mentioned in Mayo Clinic Proceeding paper [35], Mechanical ventilation is "a necessary evil": a lifesaving technique but with important potential complications. It can indeed cause damage to the lung if misused. It is more currently refered as VILI (ventilator induced lung injury). Common VILI are the following ones:

- Infections. The longer the patient are on the ventilator, the more they are likely to be infected;
- Collapsed lung (Pneumothorax). Sometimes, a hole that appeared in a part of the lung becomes the weakness of that part. This lets air leak out and causes a collapsed lung.
- Barotrauma/Volotrauma. It is related to overpressure in the alveoli during mechanical ventilation leading to overdistension of the alveoli. This overdistension can lead to alveolar rupture under the effect of the pressure or volume generated by mechanical ventilation. This rupture can then cause a pneumothorax [38].

In order to avoid those injuries, protective ventilation should be applied. The ranges of values mentioned in table 1.3, give a good basis of the value not to exceed to perform such protective ventilation. However, depending on ethnicity, gender, age, body composition, the disease being treated or the state of recovery, these values vary and must be adjusted on a case-by-case basis. However, in patients with ARDS, since the alveoli are partially filled with fluid, it is suggested to decrease the V_T to 4 mL/kg PBW and keeping P_{plat} < 30 cmH_2O to avoid barotrauma [36, 39].

4 Solutions to ventilators shortage

As mentioned above, the Covid-19 emergency brought to light two major problems concerning the medical ventilator fleet. The first one is the impossibility for a country to considerably increase its stock in a short period of time because of the complexity of these machines, whose few companies know how to produce them. The second, for the same reasons, is the unequal distribution of these devices within the different countries in the world.

To bridge the gap between demand and supply for ventilators, two solutions have been highlighted :

1. Manufacture emergency/low tech respirators that provide basic functionality, at a lower cost and in a shorter time frame;

2. Increase the capacity of existing ventilators by connecting multiple patients to a single ventilator, i.e. ventilator sharing.

The second solution is the one that will be studied in this work.

4.1 Ventilator sharing

4.1.1 Dual patient ventilation, the basis for ventilator sharing

The splitting of both inspiratory and expiratory tubes using two Y connectors, to allow two or more patients to be ventilated by the same ventilator is called 'Dual Patient Ventilation'. It is the most basic and simple way to perform ventilator sharing. Moreover, it may be used right away to increase the capacity of ventilators that doctors are already familiar with. It just requires the use of easily accessible tubing and ventilatory equipment.

Interest in dual patient ventilation (DPV) has been stimulated by the Covid-19 pandemic. However, this concept is not new. In 2006, Neyman and colleagues published a study demonstrating the feasibility of ventilating 4 adults lung simulators with the same respiratory requirements, using one ventilator for a limited time [4]. In 2008, Paladino and colleagues published a trial of ventilating four adult sized sheep with same respiratory requirements using one ventilator for 12 hours [40]. In 2012, a study similar to Neyman's was published by Branson and his colleagues but with less hopeful conclusions: it is not possible to control the tidal volume for each subject if different respiratory requirements are needed. In their view, despite the attractiveness of the concept, it could not support mass-casualty respiratory failure[41].

4.1.2 Limitations of dual patient ventilation and its joint statement

Recently, at the onset of the 2020 pandemic, many efforts have been made to characterize dual patient ventilation. Published studies reveal that compliance and resistance of each patient's pulmonary system's were integrated into a single circuit that controlled the balance of airflow. Dual patient ventilation relies therefore on careful matching of patient characteristics. However, many safety issues are raised by this patient interdependence. For example, the potential for cross-flows between patients is a serious concern, and the ability to precisely control pressure and or/flow to each patient is lost if changes in patient compliance (what is bound to happen during recovery).

For these reasons, medical associations including the American Association for Respiratory Care issued a joint statement explicitly warning medical professionals not to share mechanical ventilators with current approaches. Indeed the joint statement states that the current approaches to ventilator sharing lack the ability to individualize ventilation to each patient, measure pulmonary mechanics, accommodate rebalancing of the airflow when one patient improves or deteriorates, potential cross-contamination, lack of alarms, insufficient monitoring, and inability to adapt to sudden changes in patient status have prevented widespread acceptance of ventilator sharing [42, 43].

As a result, it has been suggested that ventilator circuit adjustments and monitoring be used to offer individualized ventilation and enhance the safety profile. This improved device is called 'Differential multiventilation'.

4.1.3 Differential multiventilation, a more comprehensive solution for ventilator sharing

Even though the use of a shared ventilator in its simplest form is potentially too dangerous to be applied in real life, there are several potential changes that might be made to this setup that could make its applicability in clinical practice more feasible. These modifications could allow for individualization and monitoring of ventilation for each patient. This is referred to as 'Differential multiventilation'.

The modules required to set up a differential multiventilation system are the following one:

- **Y Connectors.** It splits the airflow cirucit between the patients. There are two Y connectors in the setup: one at the inlet and one at the outlet of the ventilator.
- **Flow restrictor.** It titrates flow and therefore pressure delivered to each patient (see section 5.1). There are two flow restrictors in the setup: one on each inspiratory circuit.
- **Respiratory filters.** Heat and Moisture Exchanger Filter (HMEF) and High Efficiency Particulate Air filte (HEPA) filters should be inserted for each patient to avoid cross-contamination. Moreover, filtration prevents the inhalation of harmful pathogens and helps protect hospital equipment from contamination with bacteria and viruses.

  HMEF allows to humidify and warm the possibly dry and cold gases coming from the ventilator before entering the patient's lungs. This is essential because the endotracheal tube used for positive mechanical ventilation bypasses the natural humidification mechanisms of the nose and pharynx. There are one HMEF at the end of each circuit, before entering patient's lungs.

  HEPA are set on the expiratory circuit of each patient.
- **Capnogram.** Each patient's exhaled CO2 pressure is measured by the capnogram, which gives information on the patient's oxygenation.
- **In-line PEEP valve.** It allows to set individual PEEP values for each patient. There are two in-line PPEP valves in the setup : one on each expiratory circuit.
- **One-way valves.** It ensures there is no backflow between the patient's circuits and therefore no cross-contamination. There are four one-way valves on the setup : two on each inpiratory circuit and two on each expiratory circuit.
- **Individual Pressure and/or Volume Measurement.** Individual monitoring of pressure and flow/volume waveforms considerably improves the level of patient safety.

The complete system is presented in the figure 1.17.

5 Flow restrictors for differential multiventilation

Flow restrictors is an adjustable device used to regulate the pressure and tidal volume given to each patient in the case of differential multiventilation. To operate properly flow restrictor should be placed on the inspiratory circuit of each patient, as illustrated in figure 1.17. To be able to monitor the changes in tidal volume and pressure brought on by flow modifications, each patient has to have their pressure and tidal volume meter.

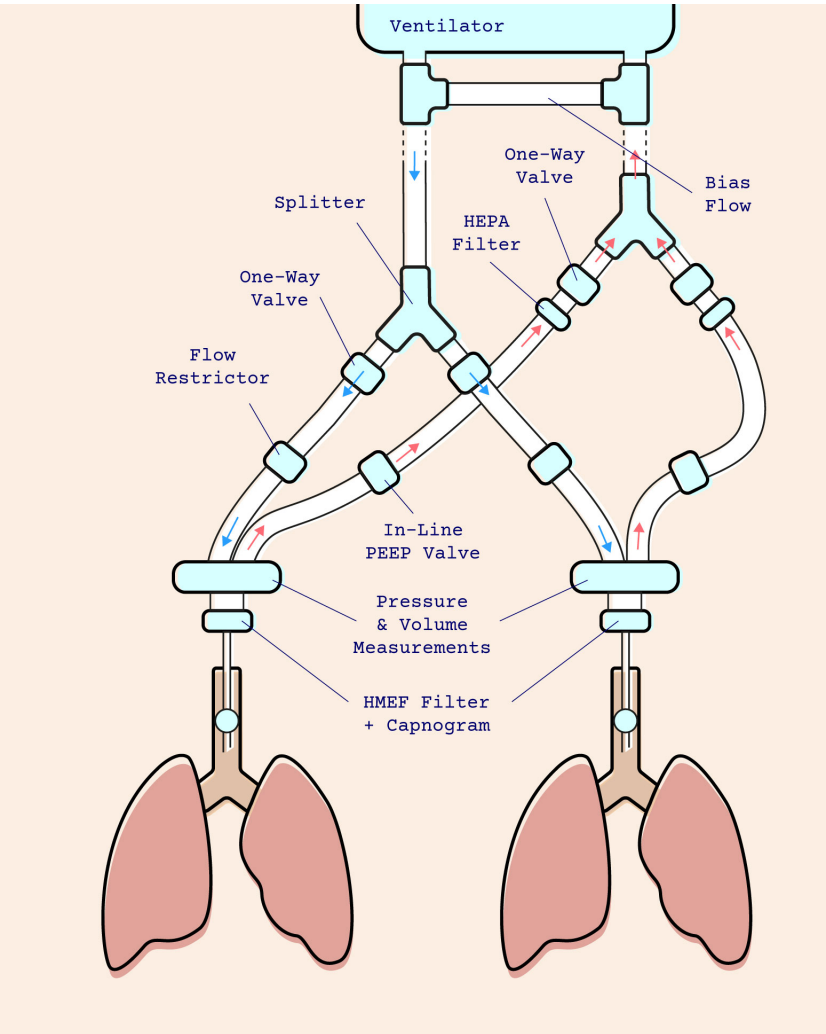


Figure 1.17: Differential multiventilation setup for two patients

5.1 Working principle

Insufflation time is set on the ventilator. This is the time needed for the ventilator to reach peak inspiratory pressure (PIP) in the case of PC ventilation and the time needed for the ventilator to deliver the tidal volume (V_T) in the case of VC ventilation. But it is also the time wherein there is a flow passing through the restriction. By adjusting the flow restrictor, the cross-sectional area of the air passage, i.e. the size of the restriction, is more or less reduced. A larger restriction increases the resistance to air flow, which reduces the volume of air that can pass through the restrictor during the given insufflation time. As a result, the pressure downstream of the restrictor will also be decreased.

The 'Differential Multiventilation International Working Group' website promoted the fact that "at least one of the circuits should have a low resistance, meaning a circuit where the internal pressure measurement of the ventilator at the inspiratory and expiratory end is almost the same, is needed for the ventilator to work properly"[44]. Therefore, the flow restrictor of one of the circuits should be left open.

5.2 Important properties

Not all flow restrictors are created equal in terms of performance. They all have advantages and disadvantages depending on their use. Being able to compare these different valves is important in order to select the most appropriate valve to use in the case of differential multiventilation application. The 'Differential Multiventilation International Working Group' resources on its website [44] various properties to take into account to make this comparison:

- **Safe materials.** It is strictly necessary that the materials used are not toxic, nor without any danger for breathing purposes. Considering that the valve will be in direct contact with the inspired air, it is also important for the materials being used to be resistant to humidity and high oxygen concentrations.
- **Disinfectable or sterilizable.** Disinfection is part of the general framework for the treatment of reusable medical devices. It allows to eliminate a part of the undesirable micro-organisms. It reduces their number without reaching zero and in no case allows to reach the sterile state. It is an essential step to limit contamination between patients. [45]

  However, in order to be sterile, and therefore to avoid contamination between patients, the medical equipment must undergo an autoclave process, i.e. a process in which the material is subjected to moist heat under high pressure. The valves must therefore be heat resistant and water resistant.
- **Sensitivity.** Sensitivity is a measure of the applied pressure/tidal volume changes when making a change to the flow restrictor. It can be expressed in change per degree of rotation (e.g. 10 ml/degree). Sensivity is therefore depends on the rotation range (number of turns) of the valves.

  If dealing with non-linear flow restrictor (see futher), sensitivity is measured as a mean of the sensitivity within the most linear regions.
- **Linearity.** The flow restrictor is considered linear if it maintains the same sensitivity throughout a broad range of adjustments. The use of linear flow restrictors will be simpler. Figure 1.18 illustrates different linearity in flow restrictor.

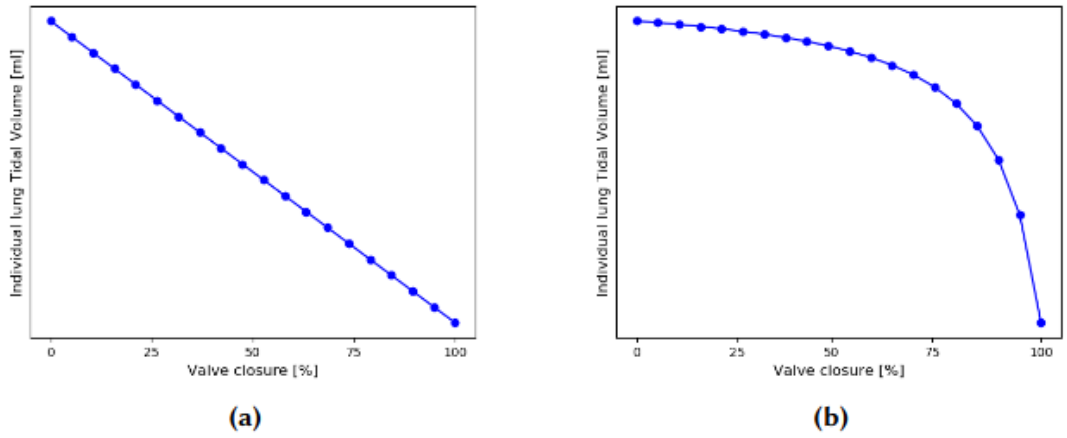


Figure 1.18: Linearity of flow restrictors. Plot (a) represents a linear flow restrictor, i.e. the same sensitivity throughout the whole adjustement while plot (b) represents a non linear flow restrictor.

- **Weight.** A heavy flow restrictor will increase chance of disconnection if not supported.

5.3 Choosing the right flow restrictor

The 'Differential Multiventilation International Working Group' website has done a remarkable job of putting together a set of flow restrictors that can be used in the case of differential multiventilation.

Some of the flow restrictors are commercially available, from a local plumbing/HVAC supplier. Others were created from scratch during the Covid-19 pandemic and are intended to be 3D printed. This was done because commercially available flow restrictor have been on backorder since the epidemic began and are not typically provided in hospitals.

It should be noted that in addition to the flow restrictor as such, another family of valves, the in-line peep valve, seems to have a similar behavior in terms of flow/pressure titration . They can therefore be used as flow restrictors.

5.4 Current statues

At the time of writing, very few studies have been done to characterize the flow restrictor. The only study available is the one done by Tormod Martinson and Christian Tronstad, from Oslo University Hospital, Norway [44]. They compared a diaphragm valve (Bürkert type 3232) and an adjustable pressure limiting (APL) valve in term of sensitivity and linearity. The result of their study can be seen in figure 1.19 : APL valve have a more linear behavior than the diaphragm valve, but in terms of sensitivity per degree of rotation on the linear interval, the diaphragm valve is better.

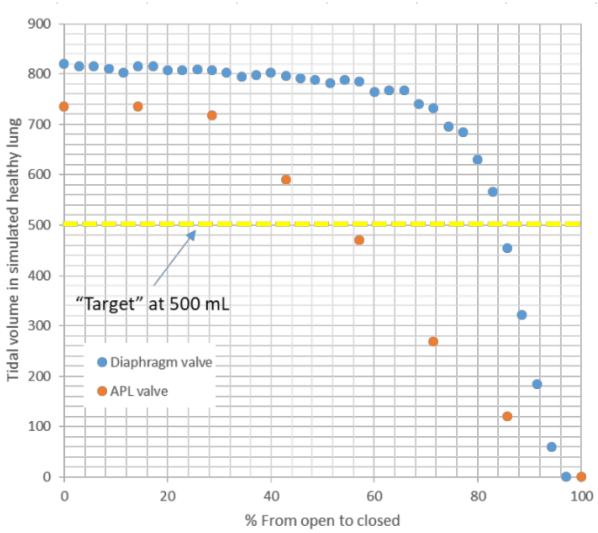


Figure 1.19: Adjustments in test lung tidal volume by diaphragm valve and APL valve

Chapter 2

Research question

Dual patient ventilation is the best-known way to duplicate the use of a ventilator. However, this way of sharing a ventilator was met with a joint statement explicitly warning medical professionals against such a practice. The majority of the concerns raised in this joint statement may theoretically be addressed by a dual ventilation solution that uses medical-grade filters, valves, and sensors to provide individualised ventilation. This way of sharing a ventilator is known as differential multi-ventilation.

Which ventilation mode should be preferred? Which modules are the best suitable to use? Is it possible to individualise the tidal volume for each patient? To what extent? These are all questions that arise when one is interested in differential multi-ventilation. Hence, the purpose of this work is to provide answers to these questions.

Specifically, the research question of this work is the following one.

Investigation into the possibility of individualising tidal volumes for two patients undergoing differential multi-ventilation in volume and pressure-controlled ventilation mode, to provide answers as to the medical validation of this way of ventilation sharing.

Three main parts make up the effort done to address this research question:

1. **Setting up a test bench and doing preliminary testing.** .
    - Implementation of sensors for data acquisition;
    - Setting up a monitoring screen;
    - Determining the compliances of test lungs;
2. **Study of different flow restrictors for differential multiventilation**
    - Characterisation;
    - Comparison;
3. **Analysis of tidal volume distribution when sharing a ventilator between two patients)**
    - Volume-controlled ventilation VS Pressure controlled-ventilation;
    - Effect of ventilation parameters;

- Effect of lung compliance;
- Impact of airway resistance;
- Individualisable?
- First clinical simulation.

It should be noted that in the course of this work, the emphasis was put on individualisation of the tidal volume only since individualisation of the positive end-inspiratory pressure (PEEP) and capnogram are the subject of two other master thesis realized by Justine Pironnet and Rodolphe Vanhoeter.

Part II

Differential multiventilation study

Chapter 3

Materials and methods

As its name suggests, this part discusses the materials and procedures employed in all of the tests carried out throughout this work. It consists of three parts:

1. Setting up a bench test;
2. Analysis of tidal volume distribution when sharing a ventilator between two patients
3. Comparison of different flow restrictors for differential multiventilation

1 Setting up a benchtest

1.1 Experimental setup

The bench test was set up using basic medical equipment, including a ventilator, tubes, medical-grade filters and test lungs and more specific equipment, including pressure and flow sensors, Y-connectors, flow restrictors and one-way valves. A more detailed description of this hardware is as follows:

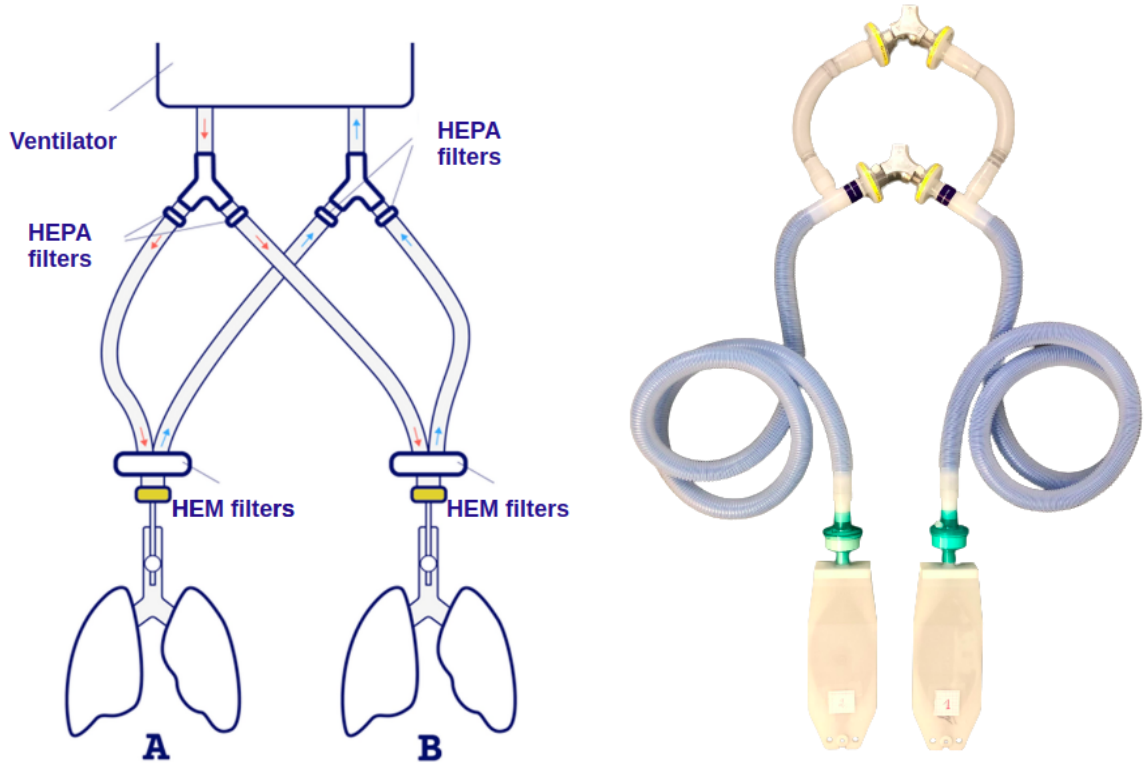
- **Ventilator.** The Siemens Servo Ventilator 300A (SV300A) is a lung ventilator intended for adult, pediatric and neonatal patients. The ventilator has two main units: a control unit and a patient unit. In the SV300A, flow measurements and all preset and indicated volumes are referenced to standard conditions (1013 mbar, 760 mm Hg) [46]. The ventilator is made up of two gas modules which constitute the air inlet and the oxygen inlet. The modules shall be connected to a medical pipeline system, gas tanks or an air compressor with an outlet pressure between 2 and 6.5 bar. In our setup, we connected it to a Kaeser CLASSIC mini 210/10 W piston compressor, with outlet pressure set to 4 bar.

  We connected the SV300A to a Siemens Servo Screen 390, which is a computer unit that reads information from the ventilator, makes calculations, and presents the information to the operator clearly and logically including respiratory curves.

  The SV300A allow a flow range between 0.1 ml/s and 3l/s and allows to ventilate a patient in controlled ventilation, assisted ventilation and assisted-controlled ventilation [46]. It is therefore an appropriate ventilator for the tests carried out in this work although it is a ventilator from the 2000s and more recent, state-of-the-art ventilators have emerged.

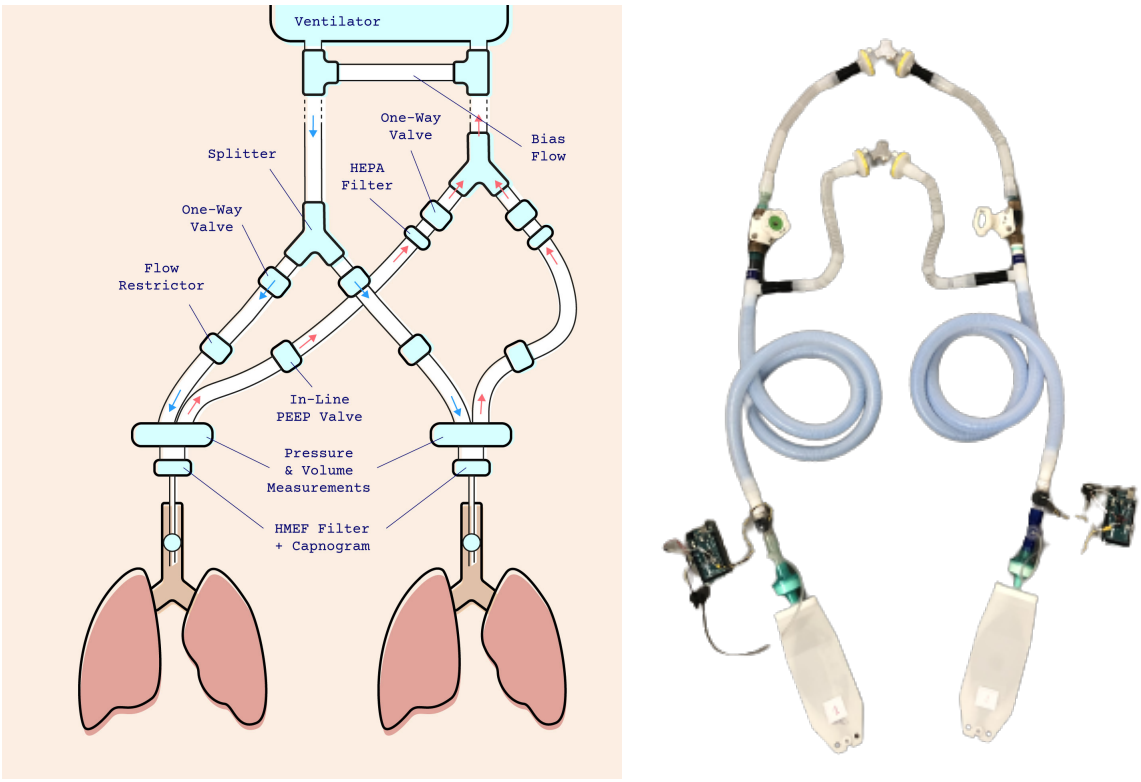
- **Tubing and respiratory filters.** Tubing consists of standard double lumen corrugated tubing (22-mm outer diameter) (figure B.1b) and standard single lumen tubing (22-mm outer diameter). Medical filters consist of standard HEPA filters (Iso-Gard®HEPA S, Gibeck-Teleflex, High Wycombe, England; DS=26mL) (figure B.3) and standard HMEF filters (DAR™ Adult – Pediatric Electrostatic Filter HME (small), Medtronic, Dublin, Ireland) (figure B.4).
- **Test lung.** Three standard test lung (Test lung 190, Maquet critical care AB, Solna, Sweden; volume=1L) (figure B.5) whose compliance is determined below.
- **Y connector.** A Y connector made of surgical-grade stainless steel to split respiratory flows, designed by Jean Ruwet and Grégoire Le Brun (MultiVentY, Belgium) (figure B.6).
- **Flow restrictor.** A Sisto diaphragm valve (PN16 3/4" DN20) was used as a flow restrictor (figure B.7). The tips of HEMF filters were cut and then heated to adapt the valves to the standard 2-mm–outer diameter corrugated flex tubing. Leaks were sealed with Teflon. This flow restrictor will be used by default throughout the testing because it is commercially available and hence functional. If a different valve is employed, it will be made clear.
- **One-way valves.** Aslo called non-return valve. It consist 3D printed devices B.8. The design looks like this: two cylinders with different diameters are placed end to end, and a ball with an outer diameter that matches the smaller cylinder's diameter is inserted into it. Depending on the direction of airflow, the ball will either fit into the larger diameter cylinder, allowing air to travel through, or it will fit into the smaller diameter cylinder, stopping airflow.
- **Pressure and flow sensors.** They are described in section 1.2.

Throughout this work, testings were carried out on two different ventilation circuits: the Dual Patient Ventilation circuit and the Basic Differential Multiventilation circuit. These circuits are illustrated through a diagram and a picture in Figure 3.1 and 3.2.



(a) (b)

Figure 3.1: Illustration of the dual patient ventilation circuit. Figure (a) is a diagram that allows us to understand the path of the ventilatory flows, and figure (b) is a picture of the circuit used in the tests.



(a) (b)

Figure 3.2: Illustration of the basic differential multiventilation circuit. Figure (a) is a diagram that allows us to understand the path of the ventilatory flows, and figure (b) is a picture of the circuit used in the tests.

1. **Dual Patient Ventilation circuit .** This is the circuit used for dual patient ventilation, i.e. when two patients are connected to the same ventilator in the simplest way.

   For this circuit, Y connectors (MultiVentY Y connector) are used to connect both inspiratory and expiratory channels on the Siemens Servo Ventilator 300A. Each inspiratory channel consists of a HEPA filter, a standard double lumen corrugated tubing (22-mm outer diameter), a pressure sensor and a mass-flow sensor in series. The whole is connected to an HME filter which in turn is connected to the test lung (Test lung 190, Maquet). Each expiratory channel consists of the same HME filter, a part of the standard double lumen corrugated tubing (22-mm outer diameter), standard single lumen tubing (22-mm outer diameter) and a HEPA filter in series. A diagram and a picture of the circuit are visible in Figure 3.1.

2. **Basic Differential Mltiventilation circuit.** This is the circuit used for basic differential multiventilation, i.e. for connecting two patients on the same ventilator and allowing for tidal volume individualisation.

   This circuit as a whole is similar to the Dual patient ventilation circuit, with some additional modules. A flow control valve is added to each inspiratory channel to allocate the appropriate tidal volume to each patient. Moreover, to complete a medically usable system, the HEPA filters of each expiratory channel were paired with one-way flow valves to ensure pressures do not equilibrate across the circuits and to limit risks of cross-contamination. A diagram and a picture of the circuit are visible in Figure 3.2.

   The term 'Basic' is used because this circuit only allows the individualisation of the current volume and not of the PEEP, which requires an additional in-line PEEP module.

When the in-line PEEP is added to this circuit, we have a 'complete' multiventilation differential circuit.

Throughout the tests, both volume-controlled and pressure-controlled ventilation modes will be used. Depending on the test being conducted, the ventilation parameters for these two modes set on the ventilator may vary.

All parameter sets used in the experiments are summarised in the tables 3.1 and 3.2. These defined parameter sets have been chosen to cover a wide range of ventilation parameters while remaining within suggested values for protective ventilation (see 3.3.2).

Parameters	Ventilator Settings					Units
	1	2	3	4	5	
V_t	600	900	900	750	600	ml
RR	15	15	8	20	20	breaths/min
$PEEP$	5	5	5	15	15	cmH_2O
$I:E$ ratio	1:2	1:2	1:1	1:2	1:3	/
T_{pause}	10	10	10	10	10	%
IRT	5	5	5	5	5	%
FIO_2	21	21	21	21	21	%

Table 3.1: Ventilators settings used for tests in VC mode

Parameters	Ventilator Settings					Units
	1	2	3	4	5	
PIP	20	20	20	30	30	cmH_2O
RR	20	8	15	15	8	breaths/min
$PEEP$	5	5	15	15	10	cmH_2O
$I:E$ ratio	1:2	1:1	1:2	1:2	1:3	/
IRT	5	5	5	5	5	%
FIO_2	21	21	21	21	21	%

Table 3.2: Ventilators settings used for tests in PC mode

1.2 Data acquisition and visualisation

The Siemens Servo Ventilator 300A and associated Servo Screen 390 deliver precise flow information when just one patient is being ventilated. When dealing with ventilation of two patients, however, it is required to add flow and pressure sensors to each patient's breathing circuit to enable individual monitoring of pressure, flow and volume.

1.2.1 mass-flow sensors for monitoring respiratory flow

To monitor per-patient respiratory flow I used an inline flow sensor (SFM 3300-AW, Sensirion AG, Switzerland) represented in Figure 3.3.

It is a digital flow meter that measures bidirectional flow volumes of up to 250 standard litres per minute (slm) [1]. It withstands washing and autoclaving procedures, and it features medical cones for pneumatic connection to standard breathing circuits (standard ISO5356-1:2004). Therefore, the SFM3300-AW is extremely well suited for proximal flow measurements in medical ventilation and other respiratory applications. Its main specifications are the following:

- Flow range : [-250;+250] slm;
- Typical accuracy : 3-7 % of measured value;
- Dead space : < 10 ml
- Operating pressure range : [0.54;1.1] bar (absolute);
- Operating temperature range : [+5;+50] °C;
- Interface: communication between the master and the SFM3300 series sensor runs via the digital I2C interface.
- Update time : 0.5 ms
- Supply voltage: 5V±5%

The SFM3300-AW flow sensor was calibrated using the information provided by the ventilator servo screen. The value measured by the flow sensor must be divided by a factor of 8.3 to match the values on the servo screen.

It should be noted that other options might have been used to monitor a flow because this sensor is not very inexpensive ($194.87 at the time of writing). One option may have been a pressure sensor coupled to Bernouilli's theorem. But in my instance, the SFM330AW sensor appeared a no-brainer due to two reasons: (1) these sensors were made especially for medical applications, making their usage relevant to my work; and (2) these sensors had previously been employed for a study at UCL. Since there was already a sizeable supply of these sensors in the UCL buildings, it was possible to avoid significant delivery delays and keep the expenses associated with my work to a minimum.

[1] mass flow measured in litres per minute at standard conditions (T = 20 °C, p = 1013.25 mbar)

Figure 3.3: Sensirion SFM3300-AW flow sensor

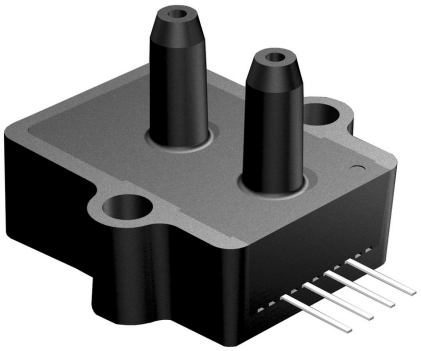


Figure 3.4: ADCA Series Amplified Low-Pressure Sensors

1.2.2 Pressure sensors for monitoring respiratory pressure

To monitor per-patient respiratory pressure, differential low-pressure sensors (ADCA Series Amplified Low-Pressure Sensors) were used, represented in Figure 3.4. According to what is written in its datasheet (see Appendix **??**)it is a pressure sensor that is based upon a proprietary technology to reduce all output offset or common modes errors. Therefore this sensor gives an accurate and stable output over a wide range.

Since this is a differential sensor, a pressure difference is measured between the two ports of the sensor. To connect one of the ports to the breathing circuit, I used a T-connector on top of which is attached a 22mm to 6mm straight connector. Then, a standard 6mm oxygen tubing is used to connect the pressure sensor to the T-connector.

Justine Pironnet defined the features of this sensor (in terms of precision and repeatability) in her master thesis. She concluded that the repeatability and accuracy of this sensor are suitable for usage in this project.

1.2.3 Sensors-Arduino Interfacing

Both SFM3300 sensor and pressure sensor use an Arduino Mega 2560 for communication. To make it as easy as possible to use the sensors, the same Arduino is used to connect both flow and pressure sensors of the same breathing circuit. This allows having only two Arduino to control both the flow and the pressure of each patient when ventilating two patients.

Connections between the sensors and the Arduino are shown in the figure 3.5 and 3.6. For this purpose, 4 cables were welded on the flow sensor's connector surface contact lands (figure 3.7). For the pressure sensor, Jumper wires were used for connection with the Arduino. Communication between the Arduino micro-controller and the SFM3300 sensor occurs via the digital I2C interface.

1.2.4 Graphical interface for individual monitoring of pressure, flow and volume waveforms

The programming of the Arduino board is largely inspired by a part of the code realised during the 'Breath4Life' project at UCL, available on the 'UCL Crypto Group Gitlab' of Gaëtan Cassiers,

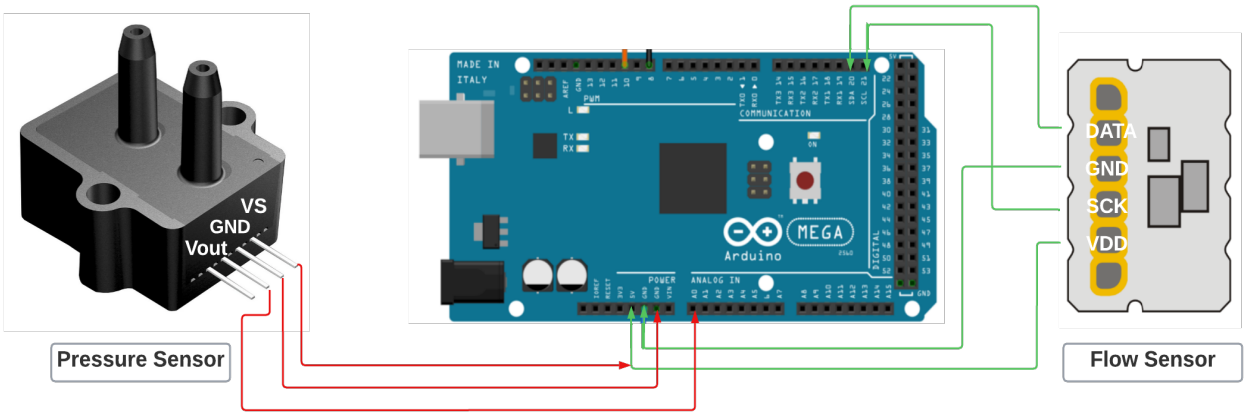


Figure 3.5: Connection diagram of the flow and pressure Sensor to the Arduino Mega 2560

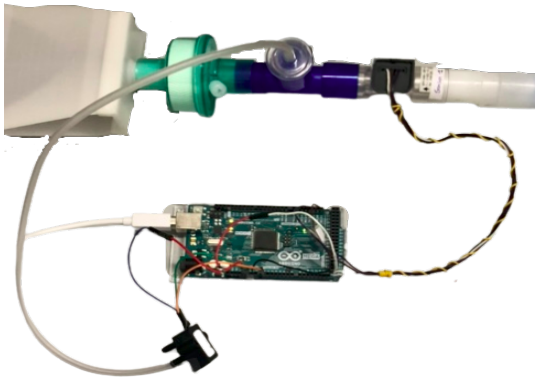


Figure 3.6: Picture of the connection of the flow sensor and the pressure sensor to the Arduino

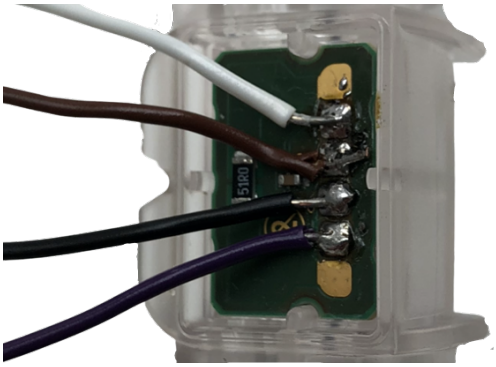


Figure 3.7: Flow sensor's connector surface contact lands

under the name 'b4l-databench' [2].

The key modifications made to the code were (1) simplification and cleanup, (2) adjustments to the graphical data visualization, and (3) the implementation of scripts to enable connection with two Arduino devices at once, enabling the use of a single computer to track the pressure, flow, and volume waveforms of both patients.

The code was run on an HP Spectrum X360 13-ac019nb, with a Linux operating system. The Arduino Mega 2560 is connected to the PC via a USB-B to USB-A cable.

The programming language used is C, especially for FreeRTOS and UART. The graphical representation of the data, however, is done via a python script (Matplotlib). Figure 3.8 provides an example of the graphical interface created to monitor each patient's pressure, flow, and volume waveforms. Volume waveforms were computed integrating flow over time. This interface may be changed to show just one of the waveforms or to show both patients' curves on a single graph 3.9.

[2] `https://git-crypto.elen.ucl.ac.be/cassiersg/b4l-databench`

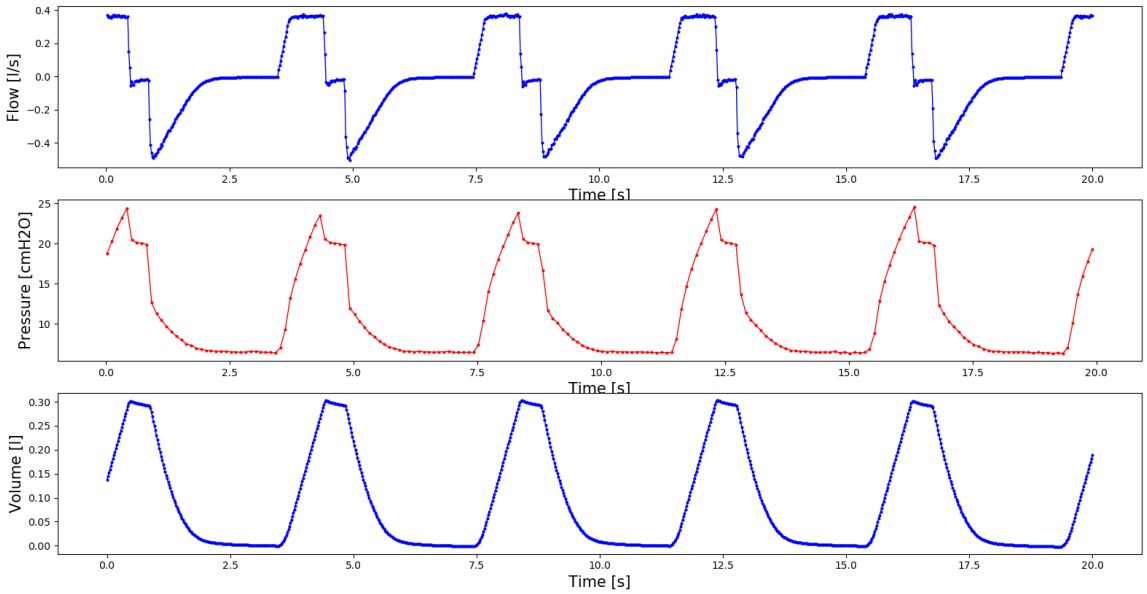


Figure 3.8: Illustration of the graphical interface for monitoring the pressure, flow and volume respiratory curves.

2 Preliminary testing of the bench test: assessment of test lung compliance

The purpose of this bench test is to determine the compliance of the test lungs at my disposal, as this is a key parameter for further testing.

Three patient models are simulated, each with particular compliance using a test lung (Maquet Siemens Adult 1 Liter Test Lung 190). Different values of compliance are obtained by adding a rubber band around the test lungs. This allows us to obtain 3 different test lungs:

- Test lung 1: Maquet Test Lung 190 without any rubber band;
- Test lung 2: Maquet Test Lung 190 with one rubber band;
- Test lung 3: Maquet Test Lung 190 with two rubber bands.

The method used to evaluate the three static lung compliance is the one Apoorva S Kulkarni and Sheela N explain in [47]. In this method lung compliance (C_L) is estimated by a static approach. The value can be estimated both during inhalation as well as the exhalation phase. The equation for calculating C_L is given by:

$$C_L = \frac{V_T}{P_{plat} - PEEP} \tag{3.1}$$

where, V_T: Tidal Volume (which is assumed to be equal to volume at the end of inspiration minus volume at the start of inspiration), P_{plat}: Plateau pressure and $PEEP$: Positive end-expiration pressure.

In this work, C_L is estimated during the inhalation phase and the dual patient ventilation circuit (see 1) is used. For each of the three lungs, equation 3.1 was computed for the five different ventilator settings in both VC and PC mode. The volume and pressure sensors linked to each breathing circuit measure the three values needed to compute these equations.

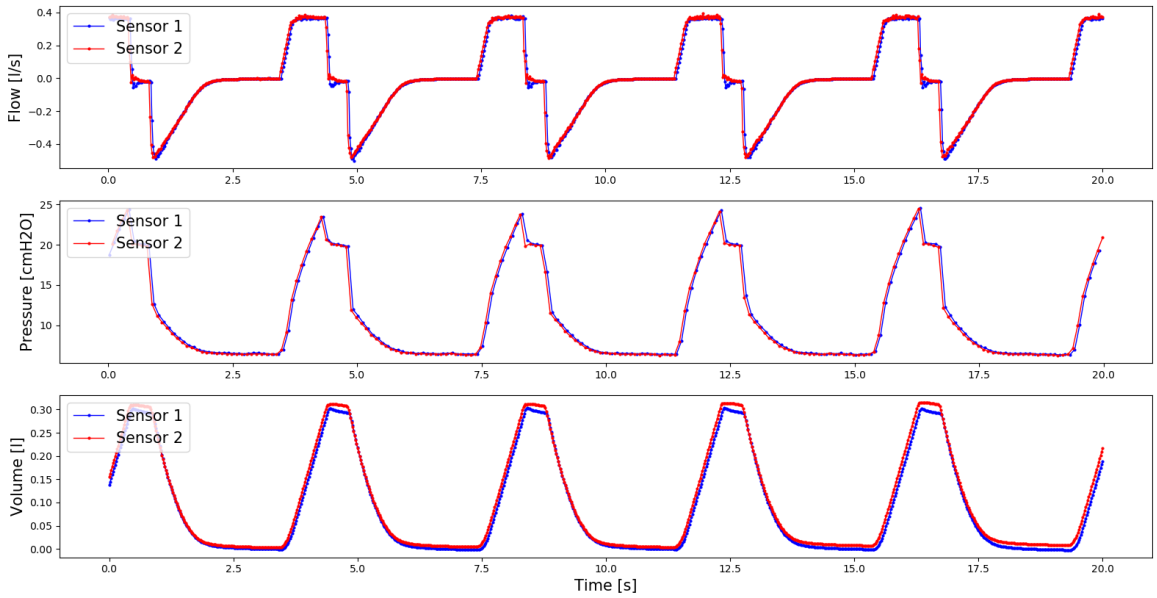


Figure 3.9: Illustration of the changed graphical interface to show both patient's pressure, flow and volume curves on a single graph.

However, since P_{plat} is estimated under zero flow conditions, the measurement will vary depending on ventilation modes VC or PC. In the case of VC mode, P_{plat} corresponds to the pressure during the inspiratory pause (set to 10%), no additional manoeuvres are required to calculate it. In the case of PC mode, however, a two-second end-inspiratory pause manoeuvre is performed on the ventilator, which allows for an exact static measurement of the end-inspiratory pause pressure, i.e. P_{plat}.

3 Study of different flow restrictors for differential multiventilation

In this second part of the work, several tests have been performed to compare and evaluate various flow restrictors that can be used for differential multiventilation.

This study has two main goals:

1. Characterisation of several flow restrictors for differential multiventilation
2. Choosing the flow restrictor that will be used for in vitro testing.

Five different flow restrictors will be tested. Some are commercially available, and others are 3D prints.

3.1 Design and conception of different flow restrictors

Five flow restrictors were selected for this study, see Figure (3.10). The selection was made to cover different types of models available in the literature. Two of these five valves were readily accessible on the market, while three were made via 3D printing.

Figure 3.10: Five tested flow restrictors for differential multiventilation. From left to right: Diaphragm valve, Panda flow restrictor, Needle flow restrictor, Panda APL valve, and Commercial APL valve.

Diaphragm flow restrictor

The diaphragm flow restrictor is a Sisto diaphragm valve (PN16 3/4" DN20). The tips of HEMF filters were cut and then heated to adapt the valves to the standard 2-mm–outer diameter corrugated flex tubing. This valve is easy to obtain from a local plumbing/HVAC supplier. Illustration of this valve is available in Appendix 1 (Figure B.7).

Panda Flow restrictor

The Panda flow restrictor is a 3D printed valve. It is made of two parts: the base and the cover. The base is the path through which the air passes and is made of 22mm connectors (female and male) to fit standard tubing. The cover is the part that is screwed onto the base to adjust the restriction. Room for a sealing O-ring is provided in the design. This valve is shown in Figure 3.11

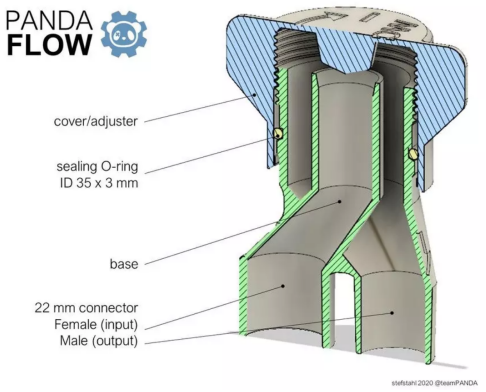


Figure 3.11: Panda flow restrictor

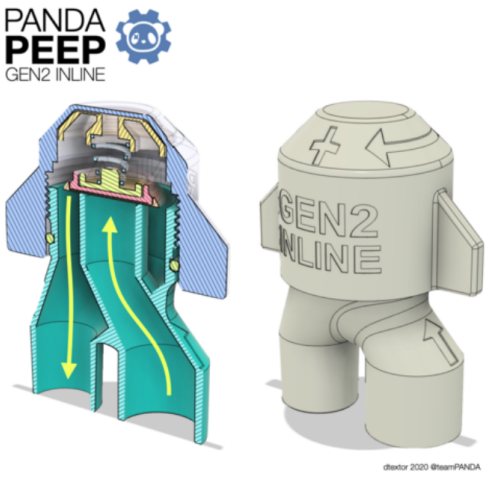


Figure 3.12: Panda APL valve

This open-source flow restrictor was designed by team PANDA (Zurich, Switzerland). All the files .stl files are available on 'thingiverse.com' under the name 'Pandaflow- flow restrictor valve'.

It was printed by the 3D printer 'Ultimaker 3' and the printing was set up using the 'Ultimaker Cura' software. The used print settings are the following:

- *Materials* : PLA
- *Layer tickness*: 0.15 mm
- *Infill density*: 50 %.
- *Filling pattern*: triangle
- *Printing temperature :* 200 ℃
- *Printing speed :* 70 mm/s
- *Generate support*: No

Needle flow restrictor

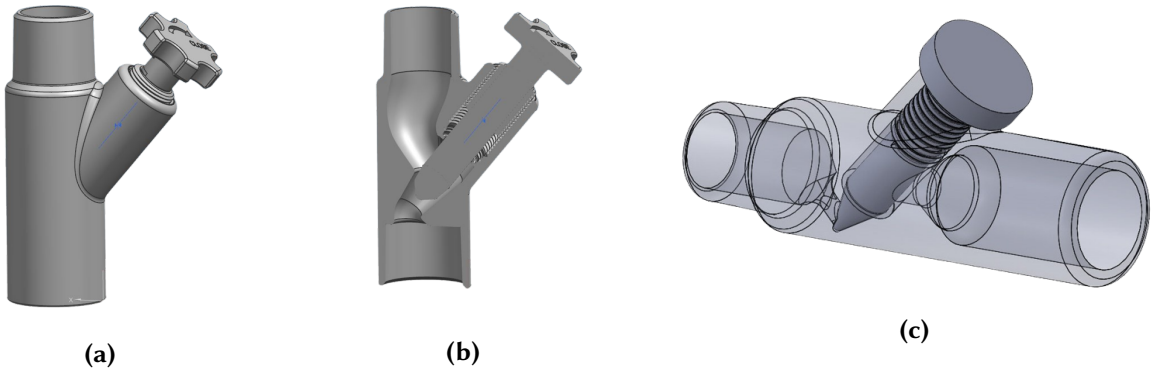
The Needle restrictor is a 3D printed valve. It is made of two parts: the body and the needle. The body, in the shape of a Y, is the path through which the air passes and it has an opening in which the needle is screwed. The more the needle is screwed into the body, the greater the restriction. Connections between restrictor valves and tubing are the iso standard 22mm. A representation of this valve is shown in figure 3.13.

This open-source flow restrictor was designed by Tobin Greensweig and his team for the project 'SpitVent'. All the .slt files are available on their Github "`https://github.com/jkoberg/splitvent`"

It was printed by the 3D printer 'BCN3D Sigma R17' and the printing was set up using the 'BNC3D Cura ' software. The suggested print settings are the following:

- *Materials* : PLA
- *Layer tickness*: 0.1 mm
- *Infill density*: 50 %. The more the infill density, the best it is to prevent air leaks.
- *Filling pattern*: triangle
- *Printing temperature :* 200 ℃
- *Printing speed :* 70 mm/s
- *Generate support*: Yes

With all those settings, the print takes 10h42min and weighs 42 grams.



(a) (b) (c)

Figure 3.13: Needle flow restrictor. Overall view (a) and cross-sectional view (b), (c).

Panda APL valve

Panda APL [3]valve is an in-line PEEP valve also developed by team PANDA (Zurich, Switzerland). It consists of two main parts: the base and the cover Pressure is regulated by the use of a conical bike hub spring, that fits between the body and the cover. The more the cover is screwed on, the greater the force exerted on the spring, so a greater pressure will be required upstream of the valve for flow to occur. Theoretical PEEP values vary between 5 and 20 cmH2O. Sealing is ensured by a 3mm O-ring 35mm ID.

All the printed recommendations and additional information as well as the .stl files are available on 'thingiverse.com' under the name 'PANDApeep Gen2 Inline'. Figure 3.12 illustrates the Panda APL valve

Commercial APL valve

The commercial APL valve is an adjustable Magnetic PEEP Valve (BE 142 Magnetic PEEP Valve) from Instrumentation Industries. PEEP ranges are adjustable from 3 to 20 cm H2O. It is designed for use with ventilators and is reusable. It provides a 22mm O.D outlet x 22mm I.D. inlet.

3.2 Characterisation of the different flow restrictors

The set of characteristics tested is:

- Sensitivity;
- Linearity;
- Leakage;
- Sealing at total closure;
- Ergonomics;
- Reusability

These are the state-of-the-art characteristics [**?**], to which other characteristics were added that were deemed necessary in light of the tests conducted in the first parts of this work.

Sensitivity and linearity

The objective of this test is to evaluate the sensitivity and the linearity, and therefore the control, of the various adjustable valves used to balance tidal volumes.

[3]APL for Adjustable pressure-limiting valve.

The basic differential multiventilation circuit (see 2) is plugged into the Siemens Servo Ventilator 300A ventilator. That is the circuit with the flow restriction inserts in each inspiratory channel but without individual PEEP modules.

The tests are performed for two test lungs with the same compliance (C_L=32mL/cmH2O i.e. the case without rubber bands around the test lungs), and under controlled pressure only. Pressure-controlled ventilation was therefore applied to both lungs with fully open adjustable valves, using an inspiratory maximum pressure of 20 cmH20 and a PEEP of 5 cmH20 at a respiratory rate of 20 breath/min (Ventilator setting 1, see 1.1).

The predetermined ventilation settings provide at first both test lungs the same tidal volume. The tidal volume given to one of the test lungs is then decreased by gradually adjusting the valve on its inspiratory circuits. The other valve is not touched in any way, allowing constant ventilation of about 300 ml of tidal volume for the other test lung.

During the tests, tidal volume was recorded using the SFM3300-AW flow sensor. However, since not all valves have the same number of turns between fully closed and fully open, the number of measurements varies from one valve to another. The number of turns from open to close for the different valves is:

- Diaphragm flow restrictor: 2.5 turns. Adjustment is made by 22.5 degrees or a 16th of a turn;
- Needle flow restrictor: 18 turns. Adjustment is made by 60 degrees or a one 6th of a turn;
- Panda flow restrictor:
- Panda APL valve: 4 3/4 turns. Adjustment is made by 45 degrees or 1/8th of a turn.
- Commercial APL valve: 4 3/4 turns in its working area. Adjustment is made by 60 degrees or 1/6th of a turn.

It should be noted the valves are always adjusted in the smallest possible increments while making certain that adjustments can be executed precisely (smaller increments would have been more relevant but difficult to do with precision).

Leakage and sealing at full closure

The objective of this test is to determine the tightness of the valves. Two types of tightness were determined: operating tightness (leakage) and tightness when the valve is completely closed (sealing at full closure).

Leaks were identified in the following way: the valve is connected at both ends to two medical tubes (standard 22mm tubing). This is then plunged into a basin of water, and the tube is blown through by plugging the other end. The intensity of the bubbles represents the extent of the leakage. If no bubbles are detected, there is no leak. If leaks were identified, they were sealed with Teflon and hose clamps, so that further tests could be carried out with these valves (see Appendix 3 for an example of how hose clamps and rubber adaptors were used to seal leaks) However, not all leaks could be sealed.

To evaluate sealing at full closure, the differential multiventilation circuit was plugged into the Siemens Servo 300A ventilator. Ventilation in a pressure-controlled mode (ventilator settings 1, see 1.1) of two test lung of same compliance (C_L =32mL/cmH20) was then performed. The evaluated valve is then closed to its maximum, and the volume at the outlet of this valve is measured via the mass-flow sensor.

Ergonomics

The objective of this test is to assess the ergonomics of the various valves that have been tested. This assessment is the outcome of a personal subjective evaluation rather than having any scientific foundation.

The comparison criteria for the ergonomics are as follows:

- Size of the adjusting element. A larger adjusting element generally allows for more accurate adjustments;
- Ease of turning the adjusting element. An element that requires almost no force to adjust is more comfortable than one that requires a lot of effort, but is also more likely to loosen/tighten itself;
- Easy to change O-rings when necessary.

Each of these criteria is scored out of 5 and the average is used as the final result for ergonomics characterisation.

Reusability

The objective of this test is to assess the reusability of the different valves. Reusability of flow restrictor most often requires to be disinfectable or sterilizable. To be considered as such, valves must be heat resistant and water resistant;

Therefore, the assessment of reusability is mostly based on the characteristics of the components that make up the valve.

3.3 Choosing of a flow restrictor for differential multi-ventilation

This study aims to answer a problem that people who develop a differential multi-ventilation device are faced with. Namely: which flow restrictor should be used in the setup?

This study is closely related to the characterisation made in the previous section (see 3.2 from this chapter), as the results of that section are used as the basis for this study.

At first, each valve is given a score ranging from 1 (poor) to 5 (excellent) for each characteristic examined in section 2.1 of the result chapter. This allows for drawing a spider web-like chart.

Then to choose only one flow restrictor, a weighted average of the scores given to each valve for each of the criteria examined in section 2.1 of the result chapter.

The Weighted Mean Formula used is the following:

$$\bar{x_w} = \frac{\sum_{i=1}^{n} w_i x_i}{\sum_{i=1}^{n} w_i;} \tag{3.2}$$

where $\bar{x_w}$ is the weighted mean, n the number of values, w_i the allocated weighted value and x_i the observed values, i.e. the score for each criterion in our case.

Allocated weighted values are 1 by default. Characteristics considered more relevant have a greater weight. except for some characteristics considered crucial. The weighted values chosen in this work are the following:

- Sensitivity: w_i =1.5;
- Linearity: w_i = 1;
- Leakage: w_i = 1;
- Sealing at total closure: w_i = 1;
- Ergonomics: w_i =0.5;
- Safe materials w_i= 2;
- Reusable w_i =1.5;

4 Analysis of tidal volume distribution when sharing a ventilator between two patients

This whole section is devoted to the study of the volume distribution between two patients sharing a ventilator. The aim is to investigate whether the distribution is equal between two patients and if not, what factors affect this unequal distribution. This analysis is therefore done step by step, starting with the study of the volume distribution between the two patients in a simple case, i.e. without a flow restrictor on the circuit, then adding a flow restrictor.

4.1 Case without added flow restriction for volume and pressure controlled ventilation modes

4.1.1 Effect of ventilator parameter settings on tidal volume distribution

This study aims to determine whether the following factors could have an effect on the distribution of tidal volume between two test lungs with equal compliance:

1. The positive inspiratory pressure (PIP) ventilator setting, in the case of pressure-controlled ventilation mode;
2. The tidal volume (V_T) ventilator setting, in the case of volume-controlled ventilation.

Effect of the tidal volume ventilator setting

The dual patient ventilation circuit (see 1) is plugged into the Siemens Servo 300A ventilator. Flow sensor and pressure sensors were added upstream of the HEM filter, to record flow, volume and pressure measurements in each of the two test lungs. Flow restrictors have not been included in the circuit. The two ventilated test lungs have equal compliance (C_L = 32mL)

The ventilator is set in volume control mode under settings 1 and 2 (see 1.1) except for the tidal volume.

Next, from 500 ml to 1400 ml, the tidal volume is discreetly changed in steps of 50 ml. Each test lung's tidal volume is collected for 20 seconds (which corresponds to 5 breathing cycles). The ventilator is given time to stabilise between each adjustment before the measurement is taken.

To get statistical data, this experiment is run three times. The average of all tidal volumes recorded during the 3 x 20 seconds is used as a reference.

Effect of the PIP ventilator setting

The dual patient ventilation circuit (see 1) is plugged into the Siemens Servo 300A ventilator. Flow sensor and pressure sensors were added upstream of the HEM filter, to record flow, volume and pressure measurements in each of the two test lungs. Again, flow restrictors have not been included in the circuit.

The ventilator is set in a pressure-controlled mode under settings 1 (see 1.1) except for the PIP.

Next, from 10 cmH2O to 40 cmH2O, the PIP is discreetly changed in steps of 2 cmH2O. Each test lung's tidal volume is collected for 20 seconds (which corresponds to 5 breathing cycles). The ventilator is given time to stabilise between each adjustment before the measurement is taken.

To get static data, this experiment is run three times. The average of all tidal volumes recorded during the 3 x 20 seconds is used as a reference.

PIP is not set higher than 40 cmH2O, since it is not recommended to ventilate with higher pressures than that.

4.1.2 Effect of compliance on tidal volume distribution

This study's objective is to assess how the distribution of tidal volume and pressure between two patients is influenced by compliance differences and how this dependency relates to the mode of ventilation (pressure or volume controlled)

The dual patient ventilation circuit (1) is plugged into the Siemens Servo Ventilator 300A ventilator. Flow sensor and pressure sensors were added upstream of the HEM filter, to record flow, volume and pressure measurements in each of the two lungs.

The study is divided into four tests: two performed under pressure control ventilation and two under volume control ventilation. For each test, the setting parameters were changed according to 1.1 :

- **Test 1:** in volume control mode for ventilator setting 1;
- **Test 2:** in volume control mode for ventilator setting 4;
- **Test 3:** in pressure control mode for ventilator setting 1;
- **Test 3:** in pressure control mode for ventilator setting 5.

Each of these tests is divided into three experiments (Experiment 1, 2, and 3), listed in the table 3.3 These cover three different combinations of test lung compliance. One of the test lungs has constant compliance of $C_L = 32$ throughout the three experiments, but the compliance of the second lung varies throughout the three experiments from $C_L = 29$ to $C_L = 27$ mL/cmH2O.

Experiment	1		2		3	
Use case	Balanced compliance		Unbalanced compliance		Unbalanced compliance (severe)	
Patient	A	B	A	B	A	B
Medical diagnostic	Normal	Normal	Normal	ARDS-Mild	Normal	ARDS-Severe
Compliance (ml/cmH2O)	C_A	C_{B1}	C_A	C_{B2}	C_A	C_{B3}

Table 3.3: Model of the three different experiments performed for each of the four tests aimed at studying the influence of compliance in the distribution of volumes and pressures during dual ventilation. The model covers different combinations of test lung compliances: balanced compliance, unbalanced compliance and severely unbalanced compliance.

Each experiment of the three tests was run three times lasting 20 seconds each to perform a statistical analysis of the measured values. During each experiment, the variable of interest (i.e. the tidal volume (V_T), the positive-end-expiratory pressure (PEEP), positive inspiratory pressure (PIP), and for ventilation in volume control mode the Plateau pressure (P_{plat})) were recorded. Next, the standard deviation (σ^2) and mean for each variable of interest were computed.

The absolute difference between the two values for any variable was divided by the mean of the two values to compute the difference between the two test lungs (Δ). This difference was then highlighted using a column chart.

4.2 Case with added flow restrictor for volume and pressure controlled ventilation modes

Assessment of the impact of a flow restriction insertion in a ventilation-sharing circuit is the goal of the following tests. The ultimate aim is to investigate whether the use of a differential multiventilation setup allows for individually adjusting the volumes delivered to two test lungs having various mechanical properties (different compliances).

4.2.1 Effect of a restriction on the flow distribution

This study aims to analyse the distribution of tidal volume between two patients sharing a ventilator when flow restrictors are added to the respiratory channel of each patient.

For this experiment, two test lungs are ventilated through the means of the basic differential multiventilation circuit (see 2 in the materials and method chapter). Both pressure and volume-controlled ventilation (PCV and VCV) modes are tested under ventilator setting 1 for both of them. To simply assess the impact of the change in resistance, the tests are further carried out on two test lungs with the same compliance.

At the beginning of the experiment, both patients are ventilated with both flow restrictors fully open. Then, the valve of test lung A is closed gradually by 1/16th of a turn while the flow restrictor of test lung B remains open. This is done until the valve of patient A is completely closed, ending the experiment. After each restriction adjustment, pressures and tidal volumes for patients A and B are recorded for 20 seconds.

The difference in percentage between maximum and minimum tidal volume delivered to one test lung was calculated as the absolute difference of these two values divided by the mean of these two values.

4.2.2 Individualisation of the tidal volume with different lungs compliance

The goal of the following tests is to test the ability to separately control tidal volume and pressure delivered for two patients sharing the same ventilator.

Experiments were done using the basic dual multi-ventilation circuit (Fig. 3.2) connected to the Siemens Servo 300A ventilator, for both pressure-controlled and volume-controlled ventilation modes for two test lungs of varying compliance (C_L=32mL/cmH2O, and C_L=27 mL/cmH2O). At the beginning of the experiment, both flow restrictors of patients A and B are opened to the maximum. Then, depending on the ventilation mode (VCV or PCV) the approach is different.

In the case of PCV, the ventilation parameters are set to adequately ventilate the test lung with the lowest compliance (test lung B). Then the valve of test lung A (with the highest compliance) is adjusted to regulate the flow delivered to it. Several valve adjustments are made, allowing test lung A to be ventilated by a different tidal volume.

In the case of VCV, the tidal volume is set on the ventilator so that it is equal to the sum of the tidal volume requirements of test lungs A and B (V_T = [4-6] mL/kg Predicted body weight (PBW)). Then the flow restrictor of test lung A (with the highest compliance) is adjusted to regulate the flow delivered to both test lungs. Several valve adjustments are made, allowing the lung to cover different coverage ratios.

4.3 First simulation to asses the clinical feasibility of using a differential multiventilation system to ventilate two patients using the same ventilator

The goal of the following simulation is to test the ability to connect and disconnect one patient from the differential multiventilation circuit and measure the effects on the other patient.

Experimentation was done using the basic dual multi-ventilation circuit (Fig. 3.2) connected to the Siemens Servo 300A ventilator, in a pressure-controlled ventilation modes for two test lungs of varying compliance (C_L=32mL/cmH2O, and C_L=27 mL/cmH2O).

At the beginning of the experiment, both flow restrictors of patients A and B are opened to the maximum. Then, the valve of patient B was closed at 100%, shutting down its flow. Patient B is then reconnected to the circuit, and by opening his valve, he is assigned the desired tidal volume. During the whole experiment, patient A is monitored and the volume changes are observed.

Chapter 4

Results

1 Preliminary testing of the bench test: assessment of test lungs compliance

Three test lungs with varying levels of compliance were created by placing a specific quantity of rubber bands around them. The purpose of this experiment is to ascertain these test lungs' static compliance values.

Compliance is a characteristic that provides information about the elastic lung tissue's resistance to expansion. Lower compliance results from greater resistance. Static compliance (C_L) is the ratio between a change in tidal volume (V_T) and the corresponding change in transpulmonary pressure (P). The change in transmural pressure can be calculated as the difference between Plateau pressure (P_{plat}) and the Positive end-expiration pressure ($PEEP$) :

$$C_L = \frac{V_T}{P_{plat} - PEEP} \tag{4.1}$$

The resulting pressure-volume graph is shown in Figure 4.1.

Equation 4.1 indicates that compliance, for a certain value of pressure and volume, is the slope of the pressure-volume curve on that point. In our case, we notice that this curve is not quite linear and that the compliance is therefore not constant. Hence, the linear trend line for each of the curves has been drawn, and we associate for each lung a constant compliance value, equivalent to the slope of the linear trend line.

This results in the following approximate compliance values:

Test lung 1:$C_L \cong 32\frac{mL}{cmH_2O}$ Test lung 2:$C_L \cong 29\frac{mL}{cmH_2O}$ Test lung 3:$C_L \cong 27\frac{mL}{cmH_2O}$

As excepted, since adding rubber bands makes the test lung stiffer, the first lung test is the most compliant, followed by the second test lung and then the third test lung.

The reason for using plateau pressure rather than positive inspiratory pressure (PIP) to calculate compliance is as follows. The transpulmonary pressure used in the definition of compliance is the difference between the alveolar pressure and the intrapleural pressure in the pleural cavity, considered constant. Hence, it is the alveolar pressure that determines the transpulmonary pressure and therefore compliance. However, the alveolar pressure is different from the PIP (=maximal airway pressure). Indeed:

$$P_{aw} = P_R + PEEP + P_{alveolar}$$

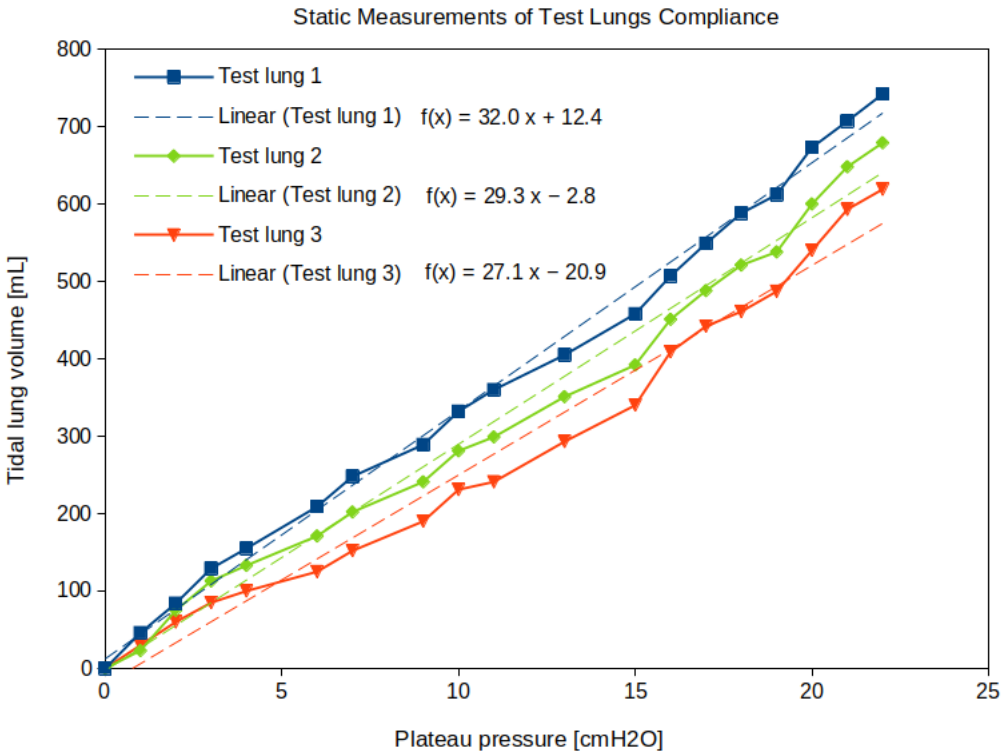


Figure 4.1: Pressure-volume curve (PV) of the three test lungs used for testing. The continuous curves correspond to the change in volume caused by the application of a plateau pressure, recorded without positive end-expiratory pressure (PEEP). The plateau pressure is obtained by performing a 3-second inspiratory pause at the end of inspiration when airflow is zero. Each of these continuous curves corresponds to one of the test lungs: test lung 1 (case without rubber band), test lung 2 (case with one rubber band), and test lung 3 (case with 2 rubber bands. The linear trend of each of these curves is represented by an associated discontinuous line, whose equation is mentioned in the legend. The compliance of each test lung corresponds to the slope of the PV curve. In this work, this value will be approximated by the slope of the linear trend line associated with each lung.

where P_R represent the resisting pressure (resistance to flow x flow rate). To compute the alveolar pressure, we need to remove P_R from the equation. This can be done by performing an end-inspiratory pause, allowing the flow rate to drop to zero, and therefore assuring P_{aw} to equal $P_{alveolar}$. At that moment P_{aw} equal $P_{plateau}$.

2 Study of different flow restrictors for differential multi-ventilation

During the COVID 19 crisis, a wide variety of flow restrictors for use in differential multi-ventilation have been designed. There is, however, very little information available regarding any testing that could have been carried out to determine these valves' features.

This section is therefore dedicated to the comparison of several flow restrictors [1] for differential multi-ventilation, with the ultimate goal of choosing which valve should be employed for in-vitro testing.

Five flow restrictors were studied, among which two were commercially available (Diaphragm valve and Adjustable pressure limiting (APL) valve) and three were 3D printed (Needle valve, Panda valve and Panda APL valve). The use of commercial valves is justified by the need to have a basis for comparison (since the design of these commercial valves has already been validated).

2.1 Characterisation of the different flow restrictors

The different comparison criteria are :

1. Sensitivity/linearity;
2. Leakage;
3. Sealing at total closure;
4. Ergonomics;
5. Reusability/the possibility for autoclaving.

Analysis of sensitivity and linearity

The objective of this study is to evaluate the sensitivity and the linearity, and therefore the control of the five adjustable valves used to balance tidal volumes.

Figure 4.2 shows the results obtained.

First of all, The Panda flow restrictor is not shown in this diagram. Restricting this valve to the maximum result in a volume variation of only about 18%. It was therefore deemed unnecessary to display it. The graph in Fig.4.2 shows that not all valves allow for flow control within the same valve closing range. The diaphragm valve for example could only control the tidal volume when it was almost fully constricted and the tidal volume was very sensitive to valve adjustments in this area, while the commercial APL valve has a much wider control range. We may determine the sensitivity in terms of mL of tidal volume per degree of rotation (mL/degree) if we keep comparing the sensitivity of the valve in its linear range of adjustments:

- **Diaphragm valve.** The linear range is shown to be between 60 and 82.5 % of the valve closure. Taking into account the number of turns between full opening and full closing of the valve (2.5 turns = 900degree), the linear range takes place on 202.5 degrees. The tidal volume ranges from 275 mL to 0 mL in this value range. Consequently, our sensitivity is 275/202.5 = 1.36 mL/degree.
- **Needle valve.** The linear range is shown to be between 80 and 100% of valve closure, i.e. on 1300 degrees when taking the number of turns of the valve into account. In this

[1] Valve is also used to refer to flow restrictors

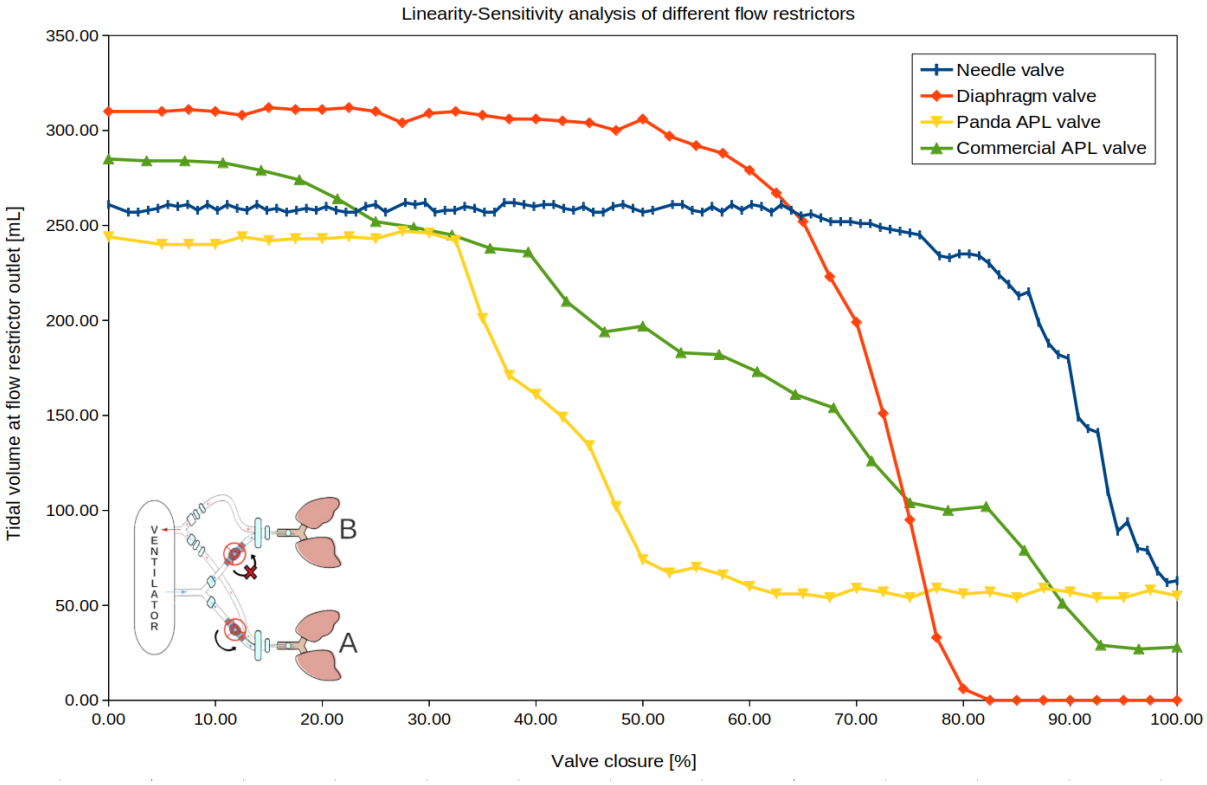


Figure 4.2: Linearity and sensitivity analysis of several flow restrictors while performing differential multi-ventilation. The curves for each valve were generated by measuring the tidal volume supplied to patient A by gradually closing, with the smallest increment possible, the valve of his circuit while the valve of patient B remained fully open. The experiment was performed in pressure-controlled ventilation.

range, the tidal volume ranges from 235 mL to 60 mL. Sensitivity is then 175/1296 = 0.14 mL/degree.

- **Panda APL valve.** The linear range is shown to be between 32.5 and 52.5% of valve closure, i.e. on 340 degrees when taking the number of turns of the valve into account. In this range, the tidal volume ranges from 240 mL to 67 mL. Sensitivity is then 0.5 mL/degree.

- **Commercial APL valve.** The linear range is shown to be between 40 and 92.5% of valve closure, i.e. on 900 degrees when taking the number of turns of the valve into account. In this range, the tidal volume ranges from 230 mL to 30 mL. Sensitivity is then 0.22 mL/degree.

As an example, if we are in the linear range of adjustment, we can state that rotating both valves a 1/8th turn (45 degrees) results in a tidal volume change of around 60 mL for the diaphragm valve, 6mL for the Needle valve, 22,5 mL for the Panda APL valve, and 10 mL for the commercial APL valve.

Secondly, as already mentioned, the curves obtained during this experiment do not all have a linear behaviour. The Needle and Diaphragm tested valves (control at almost full constriction) have a behaviour similar to what is called 'quick opening flow characteristic' in the literature [48]. Flow characteristics, which can be divided into a linear, equal percentage, and quick opening, describe the relationship between flow and valve opening (Figure 4.3a). The flow characteristic of a valve is dictated by how the valve's flow section changes with valve position,

and therefore by the physical shape of the plug and seat arrangement (i.e. valve trim), as represented in figure (Figure 4.3b). A flow restrictor's linear or non-linear tendency is therefore determined by the geometry of the valve trim.

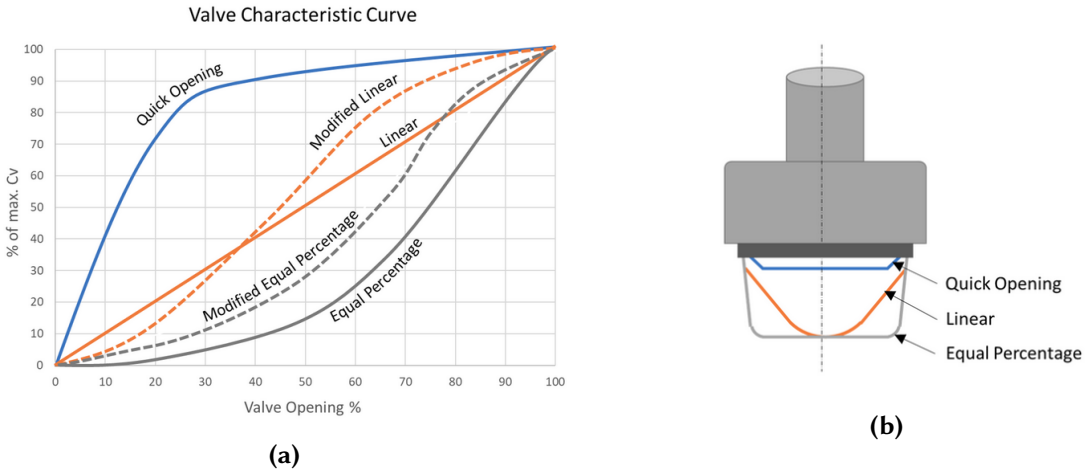


Figure 4.3: Valve flow characteristic curve (a) and different valve trim shapes (b) resulting in these curves

Diaphragm valves are well-known to have a flow characteristic close to the quick-opening characteristic [48] which confirms the results obtained in Figure 4.2. For the tested needle valve, however, the trim doesn't look similar to the quick opening trim shape. The reason why it behaves as such can be explained by the fact that this valve's progressive restriction is done with a 45-degree angle of inclination toward the air passage, whereas it is 0 or 90 degrees for the majority of the commercial valves, and for which the theory of the trim shape valve is verified.

Finally, the graph in Fig.4.2 highlights two observations:

1. The tidal volume delivered to the two patients at the beginning of the experiment differs depending on the valves;
2. Tidal volume at 100% valve closure differs depending on the valves;

The first observation can potentially be explained by the presence of leakage, which would therefore induce a loss of volume during inspiration. However, although this is plausible for the Panda APL valve and the Needle valve, this difference could not be explained for the medical valve, which did not show any leakage. The second observation is described hereafter.

Leakage and sealing at full closure

The objective of this study is to evaluate the leakage and sealing at full closure of the five adjustable valves used to balance tidal volumes.

Sealing at full closure was evaluated based on the graph provided in Fig. 4.2. It corresponds to the tidal volume delivered to patient A at 100% of valve closure. The results are the following:

1. Diaphragm valve : 0 mL;
2. Needle valve: about 60 mL;

3. Panda APL valve: about 55 mL;
4. Commercial APL valve: about 30 mL;
5. Panda Flow restrictor: about 230 mL.

Values for Needle valve can be explained by either (1) the designer purposefully choosing not to fully close the valve for safety purposes or (2) 3D printing flaws. In the case of APL valves, this can be explained by a too low spring constant, which is justified for its use as a PEEP valve (during mechanical ventilation, the PEEP is rarely set for values above 20cmH2O). Panda flow restrictor doesn't constrict the flow. Design errors or printing errors are the explanations given for this problem. What's more, these values could change from one ventilation parameter set to another. Overall, the higher the pressure delivered by the ventilator, the lower the sealing at full closure.

Leaks have been identified for both the Panda PEEP valve and the Needle valve. While leakage was relatively low for the Panda PEEP valve, it was quite high for the needle valve, despite the use of an O-ring.

Egonomics

The results of the ergonomics study are shown in Figure 4.1.

Flow restrictors	Size of the adjusting element	Ease of turning	Easy to change O-rings (if any)
Diaphragm valve	5	3	5
Needle valve	3	2	2
Panda APL valve	4	4	5
Commercial APL valve	2	3	5
Panda valve	4	3	5

Table 4.1: Evaluation of ergonomics of the different flow restrictors

Reusability

Valves that cannot be autoclaved will not be considered reusable.

For 3D printed valves, one has to look at the properties of PLA. In a study, Alexander kutschera [49] states that PLA gets very brittle rather than heat-deformed in an autoclave. And so the autoclave can only be carried out a small number of times.

The commercial APL valve, being specially designed for commercial use, is reusable.

The diaphragm valve, however, cannot be subjected to autoclave because of the diaphragm.

2.2 Comparison of a flow restrictor

Figure 4.4 provide a Spider web-like chart that represents the score of each valve, from 1(Poor) to 5(Good) for each characteristic. From this graph, anyone can select the valve that best suits the objectives of their differential multi-ventilation setup.

In this work, with the weights of each characteristic defined in the chapter material and method, the weighted average for each flow restrictor was computed.

The best suitable flow restrictor for differential multi-ventilationis then first the commercial APL valve, followed by the diaphragm valve.

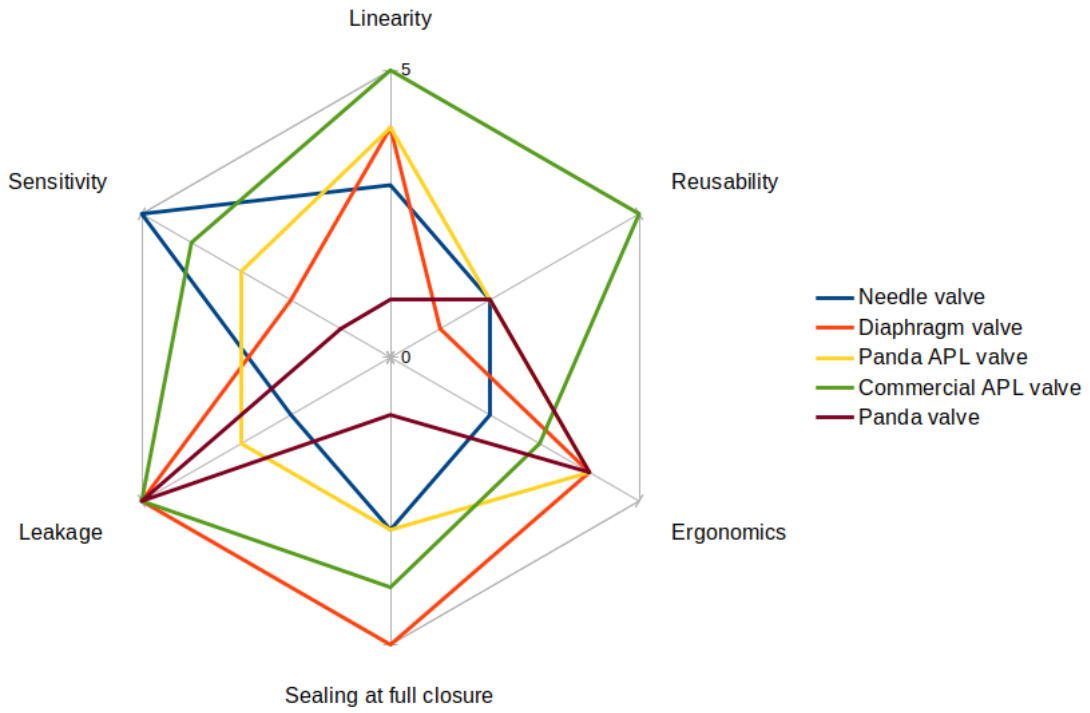


Figure 4.4: Spider web-like chart for selection of a flow restrictor for the differential multi-ventilation

3 Analysis of tidal volume distribution when sharing a ventilator between two patients

In this section, we are interested in how tidal volumes are distributed between test lungs when they share the same ventilator. The case without added flow restrictors, which is equivalent to studying a dual patient ventilation system, and the case with added flow restrictors, which is equivalent to a differential ventilation system were analysed.

An experiment devoted to the first simulation of clinical feasibility is also carried out. It aims to answer practical questions that come to mind when thinking about in-vivo testing.

3.1 Case without added flow restrictor for volume and pressure-controlled ventilation modes

3.1.1 Effect of ventilator parameter settings on tidal volume distribution

Previous research on dual ventilation ([50, 51]) have shown that the tidal volume (V_T) splits equally between patient during dual ventilation as long as they are "matched", which means that they have the same lung characteristics (in terms of compliance of the respiratory system).

However, the majority of these studies are performed for one or two different ventilator settings, which does not rule out certain circumstances in which the tidal volume would not divide equally.

The goal of this study is to determine whether, in the case of dual ventilation, the following ventilator settings have an impact on the distribution of tidal volume between patients:

1. Ventilator's tidal volume setting, in the case of volume-controlled ventilation (VCV);
2. Ventilator's positive inspiratory pressure (PIP) setting, in the case of pressure-controlled ventilation (PCV).

Effect of the tidal volume ventilator setting

Two test lungs of equal compliance are ventilated in VCV through the means of the dual ventilation circuit while varying delivered tidal volume.

Results are shown in Figure 4.5. It illustrates how the tidal volume ventilator setting affects the distribution of tidal volume in the two test lungs.

The graphical visualisation seems to indicate that the volume is divided almost equally independently of the ventilator tidal volume setting. The difference in percentage between delivered tidal volume in the two test lungs was calculated as the absolute difference of these two values divided by the mean of these two values. The data show that there is a difference that ranges from 0.4 % (for a ventilator tidal volume setting of 1100 mL) to 4.8 % (for a ventilator tidal volume setting of 600mL, which corresponds to a difference in volume delivered between the two test lungs of 12 mL).

Analysis of this graph also brings attention to another phenomenon: the sum of the two individual test lung's volumes is always smaller than the volume set on the ventilator.

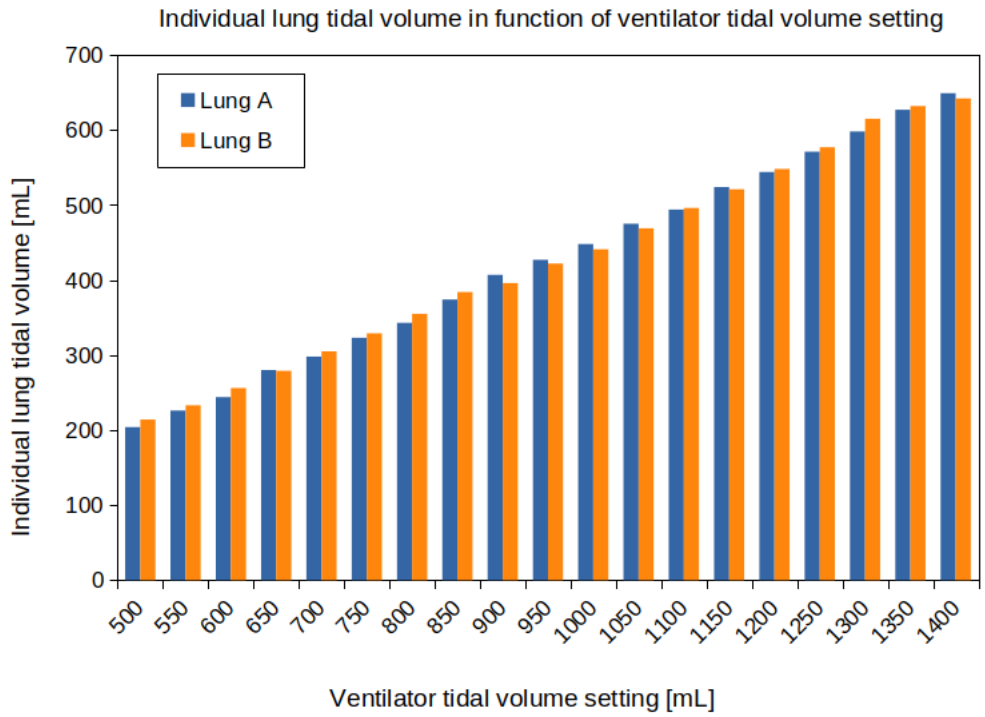


Figure 4.5: Impact of the ventilator tidal volume setting on the delivered tidal volume in two test lungs with the same compliance during dual ventilation. The tidal volume delivered to each patient is recorded using a flow sensor while discretely adjusting the tidal volume setting on the ventilator, starting at 500 ml and rising to 1400 ml.

By plotting the difference between the sum of individual lung tidal volume and the ventilator set tidal volume in function of the ventilator's set tidal volume (Figure 4.6), we see that the difference varies between 82mL and 11mL. Moreover, the average is about 100mL. 4.6.

A possible explanation for the variation of this difference throughout the experiments might simply be the inaccuracy of the ventilator. Indeed, The ventilator's tidal volume setting frequently differs from the volume that is delivered. As an example, the ventilator indicates to transmit 572 mL for a volume value set at 600 mL and 872mL for a setting at 900 mL. However, it is important to bear this in mind when analysing the further results.

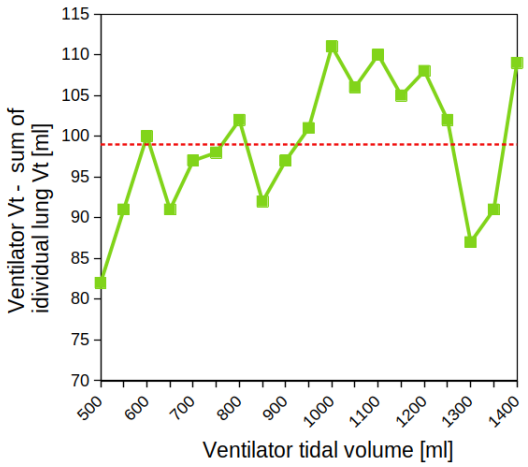


Figure 4.6: Difference between the sum of individual lung tidal volume and the ventilator set tidal volume in function of the ventilator set tidal volume. The average is shown by the red line.

Effect of the PIP ventilator setting

Two test lungs of equal compliance are ventilated in PCV through the means of the dual ventilation circuit while varying PIP parameters on the ventilator.

Results are shown in Figure 4.7. It illustrates how the ventilator PIP setting affects the distribution of tidal volume in the two test lungs.

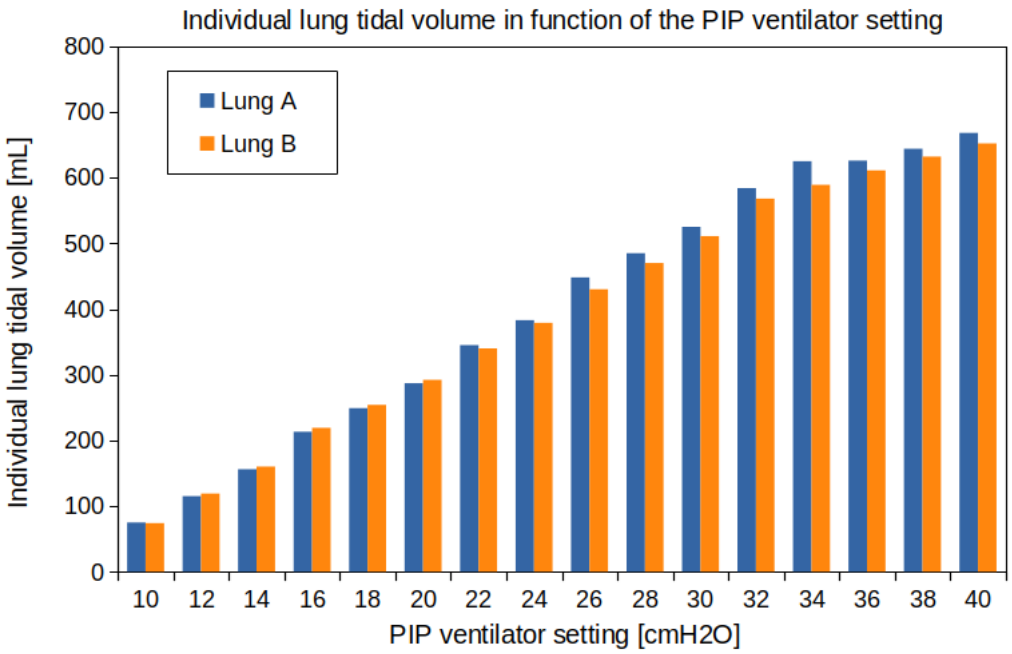


Figure 4.7: Impact of the ventilator PIP setting on the delivered tidal volume in two test lungs same compliance during dual ventilation. The tidal volume delivered to each patient is recorded using a flow sensor while discretely adjusting the tidal volume setting on the ventilator, starting at 10 cmH2O and rising to 40 cmH2O.

Again, the graphical visualisation seems to indicate that the volume is divided almost equally for each value of the ventilator PIP setting. Particularly, data show that there is a difference that ranges from 1 % (for a ventilator PIP setting of 24 cmH2O) to 5.9 % (for a ventilator PIP setting of 34 cmH2O, which corresponds to a difference in volume delivered between the two test lungs of 36 mL).

Both tests trying to assess the effect of ventilator parameters setting on tidal volume distribution enable confirmation that, regardless of the ventilation parameters chosen, the tidal volume is distributed equally (with a maximum variation of about 6%) between two patients with the same pulmonary compliance who share a ventilator in a dual ventilation configuration (i.e., without additional flow restrictor or other modules).

This outcome, which is not unexpected, can be explained by representing the equivalent electrical circuit of the dual ventilation circuit (Figure 4.8).

Each respiratory system may be seen as a flow resistance (airway/tubing resistance) coupled in series with compliance (lung property). These two resistance-compliance circuits are coupled parallel to the ventilator. The flow delivered to each patient and, therefore, their V_T, are determined by the relative flow impedance of their resistance-compliance circuit; the higher the impedance, the lower the flow and VT [52]. The impedance of the resistance is equal to itself, whereas the impedance of the compliance is inversely proportional to its value. Assuming the same resistance in both circuits (equal tubing length and diameter, and no obstructions) and identical compliance (same test lungs used), the impedance and therefore the tidal volume of each circuit is the same. It should be noted that in this analogy with an electrical circuit,

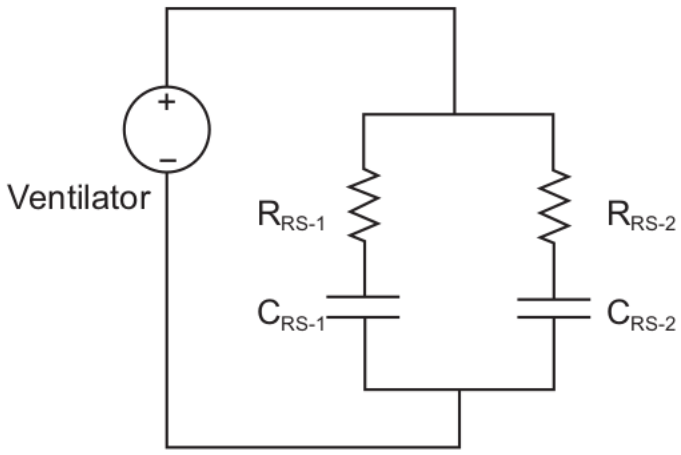


Figure 4.8: Equivalent electrical circuit of the dual ventilation circuit. From([52])

Experiment	1			2			3		
Use case	Balanced compliance			Unbalanced compliance			Unbalanced compliance (severe)		
Patient Medical diagnostic	A Normal	B Normal	Δ	A Normal	B ARDS-Mild	Δ	A Normal	B ARDS-Severe	Δ
C_L (mL/cmH2O)	32	32		32	29		32	27	
V_T (mL) $\pm\sigma^2$	249± 3.9	256±2.8	3%	297±3.3	199±2.6	41%	328±1.9	160±2.2	69%
PIP (cmH2O) $\pm\sigma^2$	20.75 ±0.03	20.72±0.05		22.84±0.25	22.60±0.15		24.13±0.39	24.38±0.47	
PEEP (cmH2O) $\pm\sigma^2$	5.79±0	5.79±0	0%	5.79±0	5.79±0	0%	5.87±0	5.87±0	0%
P_{plat} (cmH2O) $\pm\sigma^2$	17.90±0.02	17.90±0.02	0%	19.66±0.19	19.75±0.22	0%	20.54±0.16	20.54±0.14	0%

Table 4.2: Results of the 3 experiments performed for VC mode with Ventilator setting 1

the pressure difference is comparable to the voltage ($\Delta P \leftrightarrow U$) and flow is comparable to the current ($Q \leftrightarrow I$).

3.1.2 Effect of compliance on tidal volume distribution

The goal of this study was to investigate how the distribution of tidal volume and pressure between two test lungs were affected by compliance differences and how this dependence relates to the modes of ventilation during dual ventilation (i.e. two patients sharing the same ventilator through a ventilation circuit without added flow restrictor or other modules).

The three studied experiments cover different combinations of test lung compliances: balanced compliance, unbalanced compliance and severely unbalanced compliance.

Tables 4.2, 4.3, 4.4 and 4.5 provide the mean and standard deviation (σ^2) of all respiratory cycles that were recorded during each experiment for each of the four tests. In addition, graphical visualisation of the individual lung tidal volume distribution is provided in the figures. 4.10 and 4.9.

We first notice that the delivered tidal volume per connected patient will be a result of their lung compliance, for both pressure-controlled ventilation (PCV) and volume-controlled ventilation (VCV).

For both VCV and PCV, when the compliance of the two simulated test lungs is identical, there is no important difference in outcome variables (up to 3%). This is in accordance with the results of the tests in the section 3.1.1.

For VCV, unbalanced compliance produces a difference in V_T of about 41% for ventilator setting 1 and 27% for ventilator setting 4, and lower compliance resulted in lower V_T. Increasing that difference in compliance further increase differences in the distribution of V_T: this results in a

Experiment	1			2			3		
Use case	Balanced compliance			Unbalanced compliance			Unbalanced compliance (severe)		
Patient Medical diagnostic	A Normal	B Normal	Δ	A Normal	B ARDS-Mild	Δ	A Normal	B ARDS-Severe	Δ
C_L (mL/cmH2O)	32	32		32	29		32	27	
V_T (mL) $\pm\sigma^2$	322±2.6	311±4	3%	377±3.8	286±3.4	27%	410±4.5	252±3.4	48%
PIP (cmH2O) $\pm\sigma^2$	31.52± 0.01	31.50±0.01		34.34±0.35	24.40±0.31		34.95±0.36	35.55±0.33	
PEEP (cmH2O) $\pm\sigma^2$	14.52±0	14.52±0		14.5±0	14.5±0		14.52±0	14.52±0	
P_{plat} (cmH2O) $\pm\sigma^2$	23.9±0.08	23.9 ±0.09		24.50±0.06	26.01±0.12		26.55±0.05	26.57±0.05	

Table 4.3: Results of the 3 experiments performed for VC mode with Ventilator setting 4

Experiment	1			2			3		
Use case	Balanced compliance			Unbalanced compliance			Unbalanced compliance (severe)		
Patient Medical diagnostic	A Normal	B Normal	Δ	A Normal	B ARDS-Mild	Δ	A Normal	B ARDS-Severe	Δ
C_L (mL/cmH2O)	32	32		32	29		32	27	
V_T (mL) $\pm\sigma^2$	306±2.8	302±2.7	1%	301±2.9	221±3.8	30%	307±2.3	165±3.1	60%
PIP (cmH2O) $\pm\sigma^2$	20.20±0.06	20.19±0.07	<1%	20.27±0.09	20.49±0.08		20.13±0.03	20.63±0.03	
PEEP (cmH2O) $\pm\sigma^2$	5.80±0	5.76±0 %	0%	5.72±0	5.72±0		5.70±0	5.70±0	

Table 4.4: Results of the 3 experiments performed for PC mode with Ventilator setting 1

Experiment	1			2			3		
Use case	Balanced compliance			Unbalanced compliance			Unbalanced compliance (severe)		
Patient Medical diagnostic	A Normal	B Normal	Δ	A Normal	B ARDS-Mild	Δ	A Normal	B ARDS-Severe	Δ
C_L (mL/cmH2O)	32	32		32	29		32	27	
V_T (mL)	548±1.9	544±2.4	<1%	538±2.6	394±2.2	31%	535±1.4	282±2.8	62%
PIP (cmH2O)	30.24±0.18	30.32±0.17		29.72±0.13	29.75±0.22		20.72±0.19	20.64±0.17	
PEEP (cmH2O)	10.1±0	10.1±0		10.14±0	10.1±0		10.13±0	10.13±0	

Table 4.5: Results of the 3 experiments performed for PC mode with Ventilator setting 5

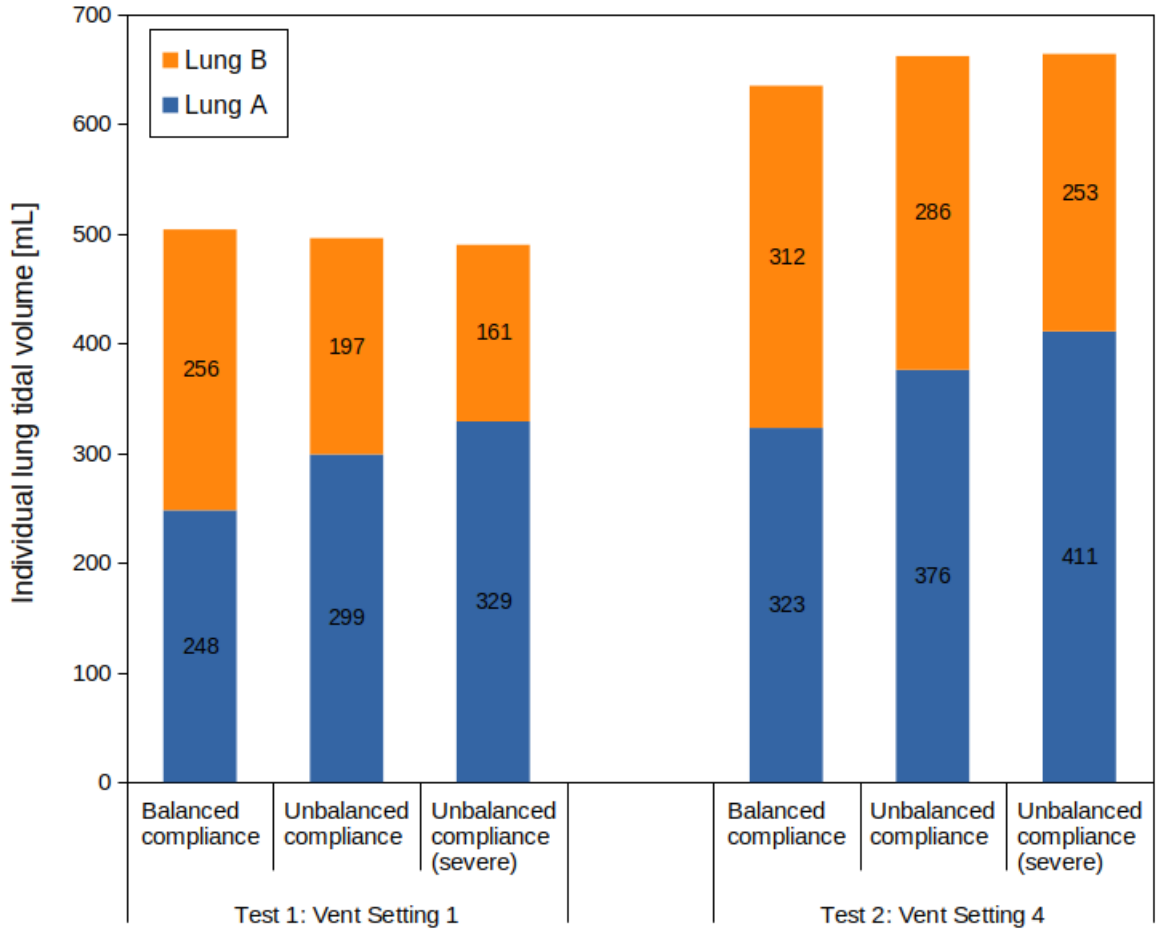


Figure 4.9: Individual lung tidal volume as a function of individual lung compliance for ventilator setting 1 and 4 in VC mode

difference of up to 69% for vent setting 1 and 48% for ventilator setting 4. However, for both setting 1 and 4, the sum of the individual V_T is relatively constant over the 3 experiments. This means that patient A, whose compliance is unchanged over the three experiments, is impacted by the compliance of lung B. More generally, in the case of VCV, the total tidal volume delivered is constant (working principle of the ventilator) and divided in ratios between the two ventilated lungs.

For PCV, unbalanced compliance produces a difference in V_T of about 30% for ventilator setting 1 and 31% for ventilator setting 5, and lower compliance resulted in lower V_T. Increasing that difference in compliance further increase differences in the distribution of V_T: this results in a difference of up to 60% for ventilator setting 1 and 4% for ventilator setting 5.
Figure 4.10 highlights the fact that for PCV, asymmetrical compliance has an impact on the total tidal volume delivered by the ventilator. A decreased compliance of test lung B results in a lower V_T for test lung B, while V_T for lung A stays unchanged and therefore in a lower total tidal volume delivered by the ventilator. The reason for this behaviour can be explained by the analogy with the electrical circuit in Figure 4.8: as compliance of lung B decrease, the associated impedance (inversely proportional to compliance) increase. As a result, the total resistance-compliance impedance of circuit B also increases, resulting in a lower delivered V_T to test lung B.

For your information, the curves generated by the sensors, which served as a basis for the

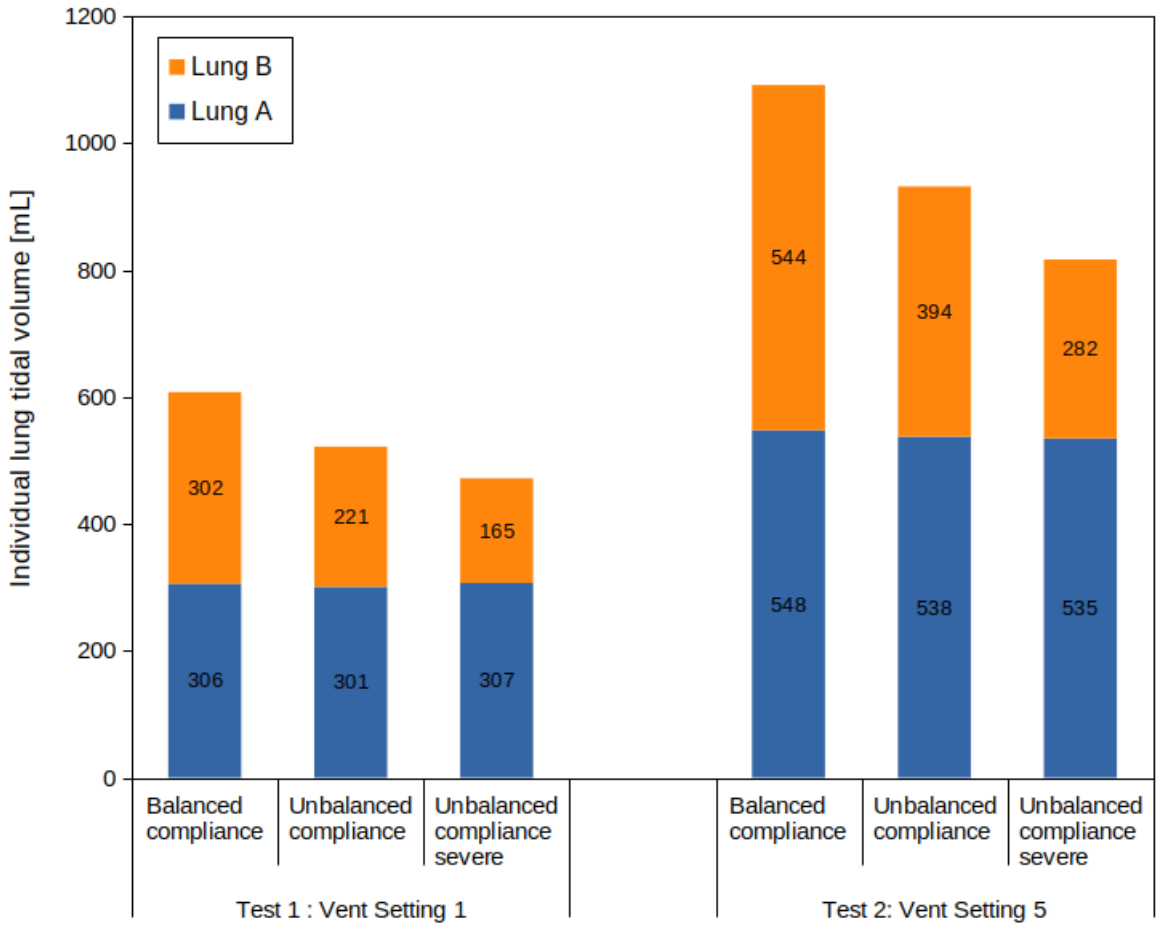


Figure 4.10: Individual lung tidal volume as a function of individual lung compliance for ventilator setting 1 and 5 in PC mode

analysis of the values during this study, are provided in the Appendix C,1.

It is also noticeable that the measured pressures are the same for both patients, for each experiment of each test. This is quite logical since both the inhalation and exhalation circuits of the patients are connected and there is no one-way valve that would allow obtaining a different pressure.

3.2 Case with added flow restrictor for volume and pressure-controlled ventilation modes

According to the state of the art, using flow restrictors in a differential multi-ventilation system theoretically allows for the individualisation of the tidal volumes delivered to two patients sharing the same ventilator.

The purpose of the following tests is to assess the accuracy of this theory. For this purpose, this section is divided into two main parts:

1. Assess the effect of a flow restrictor on tidal volume distribution for both volume and pressure-controlled ventilation modes;
2. Investigate whether it is possible to individually adjust tidal volumes delivered to two patients having different compliance, thanks to the knowledge gained from the first test.

An experiment devoted to the first simulation of clinical feasibility is also carried out. It aims to answer the following question: if the differential multi ventilation differential was indeed feasible in theory, would it be possible to use it in reality? This is the subject of the third section.

3.2.1 Effect of a restriction on the flow distribution

The influence of adding a flow restrictor on the distribution of tidal volume between two test lungs ventilated by the means of a differential multi-ventilation circuit is studied in this experiment.

Since a restriction increases the resistance to flow, studying the influence of a flow restrictor in this setup is like studying the influence of an increase in airway resistance.

Results ares shown in Figures 4.11, 4.12 and 4.13.

At the start of the test, the predetermined ventilation settings provide both test lungs with the same tidal volume. Then as the flow control valve from test lung A was progressively closed from a fully open position, the tidal volume delivered to it began to decrease in a nonlinear way. The behaviour of test lung B, however, depends on the selected ventilation mode.
In VCV, the decrease in tidal volume delivered to test lung A results in an increase in tidal volume to test lung B (see Fig. 4.11). At full closure, the entire volume delivered by the ventilator is delivered to test lung B. In PVC, while the tidal volume of test lung A decreases, the tidal volume delivered to test lung B stays unchanged. Specifically, a difference in 4% between the maximum and the minimum of tidal volume delivered to lung B was observed.

The effect of increasing resistance on the distribution of volumes between the two lungs is therefore similar to the effects produced by decreasing compliance in both cases of ventilation (VCV and PCV). Again, this can be explained by the analogy with the electrical circuit shown in Figure 4.8. Increasing the restriction of one patient's ventilation circuit increases the resistance of that circuit. Since the impedance of a resistor is equal to its value, increasing the resistance increases the impedance of the associated circuit. And the higher the impedance, the less flow, and thus volume, passes through the circuit.

The analysis of the pressures resulting from the change in the resistance of the ventilation circuit of test lung A is done in Figure B.1b. Results are not unexpected. For VCV, the pressure of test lung A decrease as the resistance of its circuit increase. Therefore, since the increase

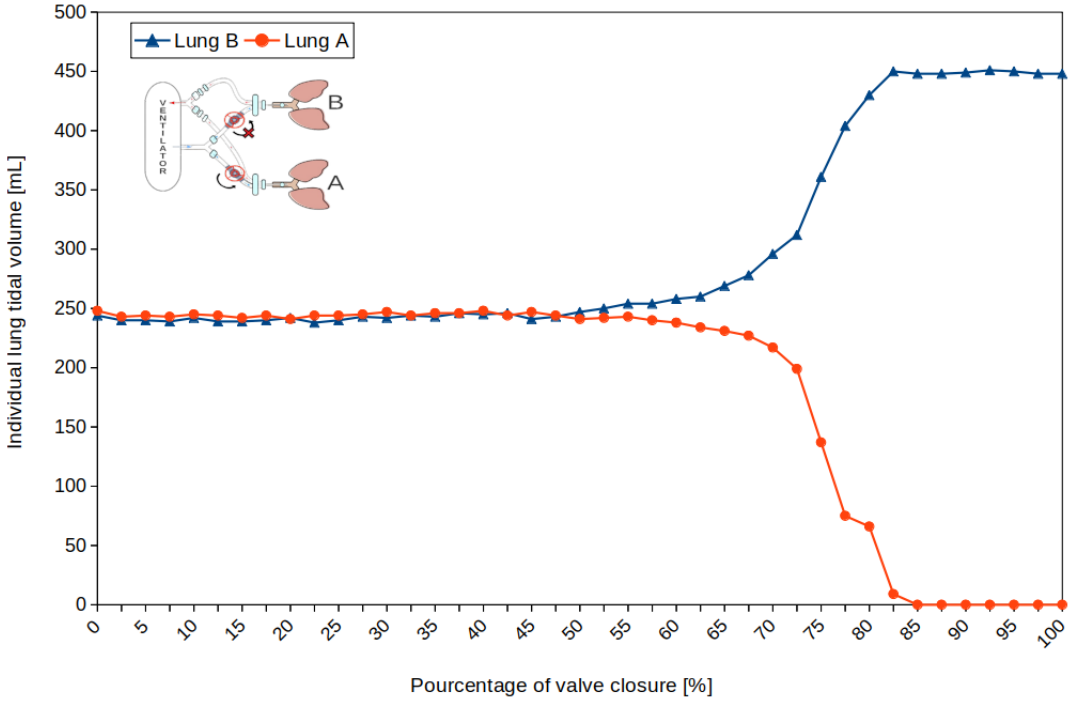


Figure 4.11: Individual Tidal volume of test lung A and B as a function of valve closure of test lung A for ventilation in a volume-controlled mode. Curves were generated by gradually closing patient A valve by increments of one 16th of a turn while the valve of patient B remained fully open. Both lungs have equal compliance of 32 mL/cmH2O. Ventilation settings 1 were used.

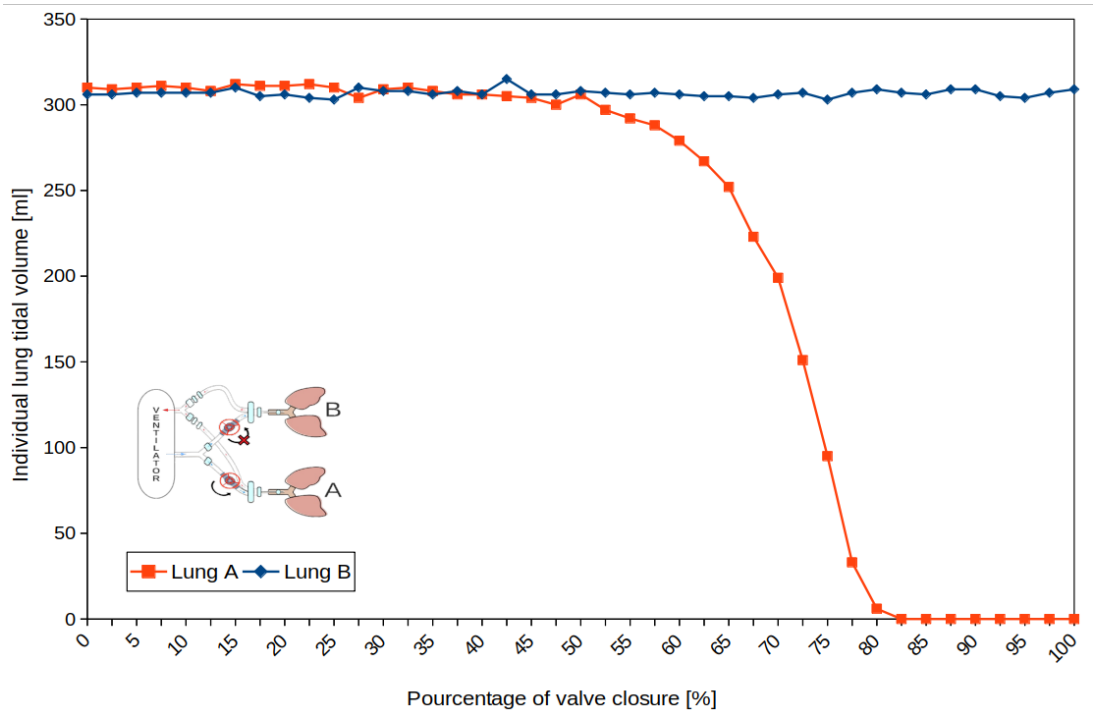


Figure 4.12: Individual Tidal volume of test lung A and B as a function of valve closure of test lung A for ventilation in a pressure-controlled mode. Curves were generated by gradually closing patient A valve by increments of one 16th of a turn while the valve of patient B remained fully open. Both lungs have equal compliance of 32 mL/cmH2O. Ventilation settings 1 were used.

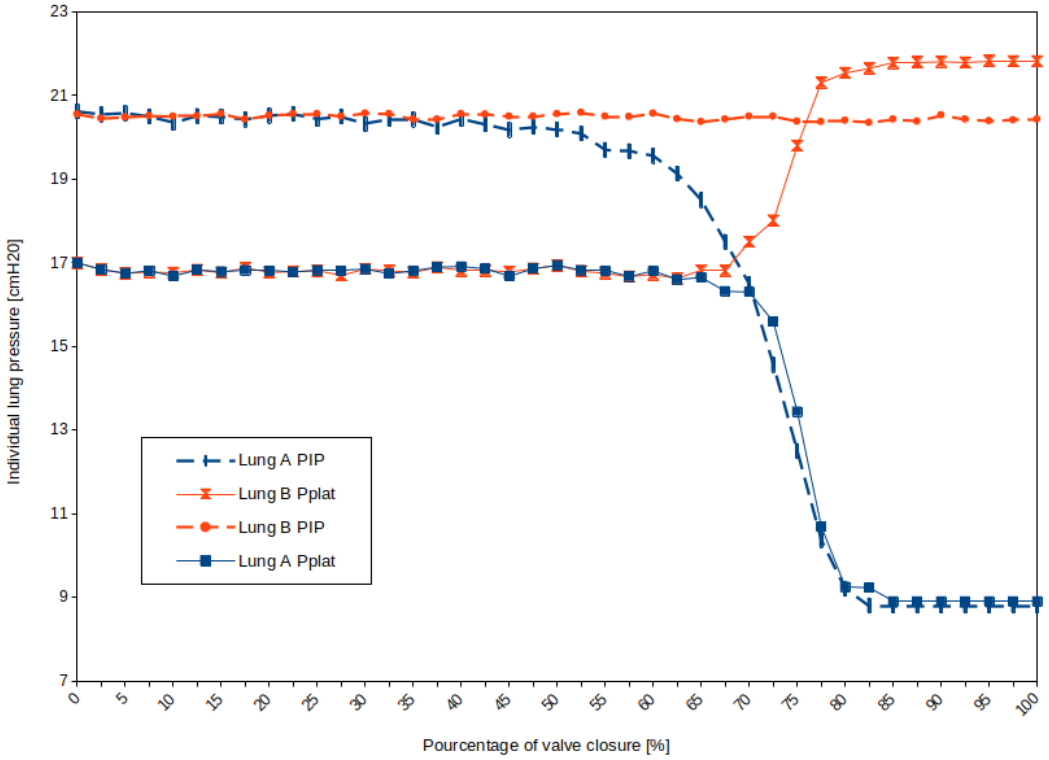


Figure 4.13: Individual pressure in test lung A and B as a function of valve closure of test lung A for ventilation in a pressure-controlled mode (PIP) and ventilation in a volume-controlled mode (P_plat). Curves were generated by gradually closing patient A valve by increments of one 16th of a turn while the valve of patient B remained fully open. Both lungs have equal compliance of 32 mL/cmH2O. Ventilation settings 1 were used for both VC and PC ventilation modes.

in resistance of circuit A leads to an increase in the tidal volume of circuit B, the pressure in circuit B increases. At the total closure of the flow restrictor, the Plateau pressure of circuit B increase by about 25%. For PCV, the pressure of circuit B doesn't change, while the pressure of circuit A decrease with restriction increase. This appends because less volume has been able to flow by the restriction during the ventilator set time, the pressure behind the flow restrictor will therefore be lower than what you would obtain without restriction.

The same experiment was also performed under controlled pressure for two lungs of different compliances. The results obtained are similar to the case where the compliance is identical: the volume delivered to the test lung whose resistance is adjusted with the flow restrictor decreases with increasing resistance (i.e. increasing restriction). The tidal volume of the other test lung, however, does not vary. The graphs are provided in the Appendix C 2.

3.2.2 Individualization of the tidal volume with different lungs compliance

This test aims to analyse the possibility of individualising ventilation for two patients sharing the same ventilator, but whose needs or mechanical properties may vary during mechanical ventilation.

Results are shown in Figure 4.14 for pressure-controlled ventilation and 4.15 for volume controlled ventilation.

Again, it can be seen that in the case of pressure-controlled ventilation, the tidal volume delivered to patient B is constant throughout the experiment, whereas the tidal volume of

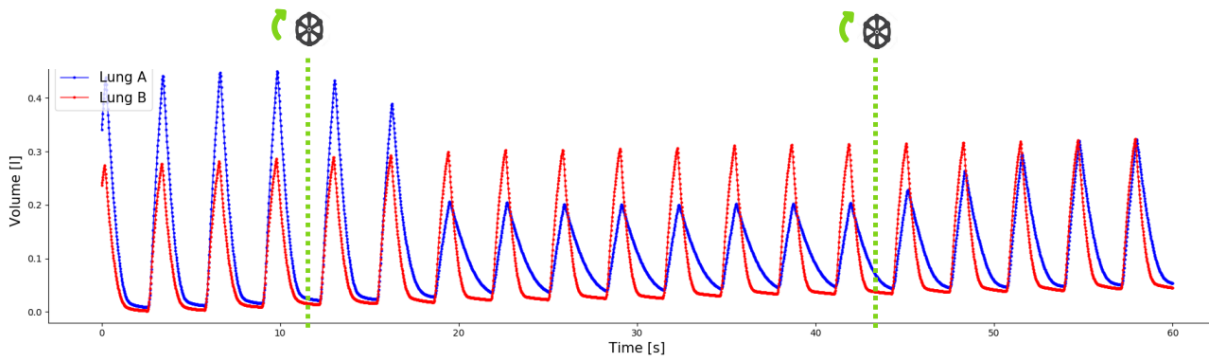


Figure 4.14: Individualisation of Tidal volume of test lungs A and B over time in a pressure-controlled ventilation mode. Test lung A has a compliance C_L=32mL/cmH2O and test lung B has compliance of C_L=27mL/cmH2O. By making titration of flow restrictor A, different values for the tidal volume of test lung A can be achieved without impacting the tidal volume of test lung B.

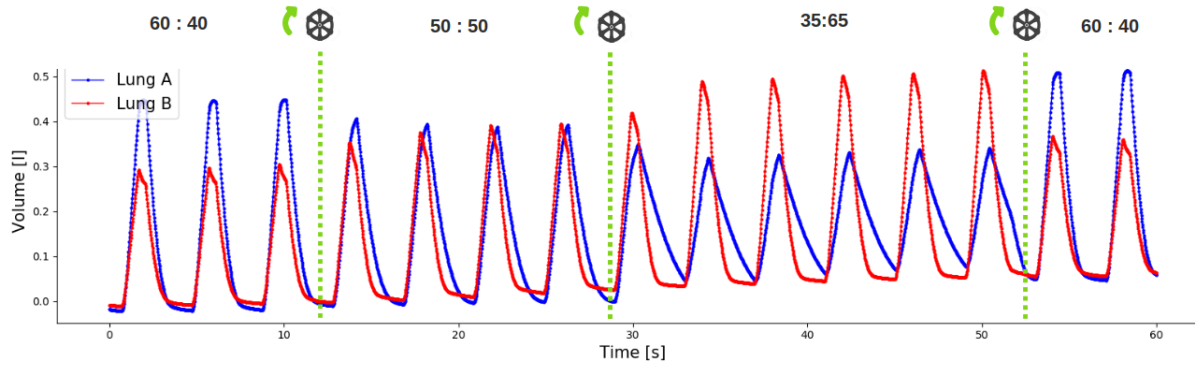


Figure 4.15: Individualisation of Tidal volume of test lungs A and B over time in a pressure-controlled ventilation mode. Test lung A has a compliance C_L=32mL/cmH2O and test lung B has compliance of C_L=27mL/cmH2O. By making titration of flow restrictor A, different spanning ratios for the tidal volume of test lungs A and B can be achieved. Tested spanning ratios (Lung A: Lung B) are 60:40, 50:50, and 35:65.

patient A is adjusted twice by adjusting the valves. This shows that patient A can be ventilated over several tidal volume ranges.

In the case of volume-controlled ventilation, the tidal volume delivered to patient A is initially greater than that delivered to patient B due to his greater compliance. Then, the flow restrictor of patient A is adjusted. Each adjustment of valve A induces an increase in the tidal volume delivered to patient B.

By adjusting the flow restrictor gradually, it was possible to achieve tidal volume ratios of 60:40, 50:50, 35:65 and again 60:40.

However, the resulting pressure analysis shows (see Appendix C.9), an increase in pressure in patient B of approximately 40%.

As information, respiratory curves from both tests are available in Appendix C.8 and C.9.

3.3 First simulation to asses the clinical feasibility of using a differential multi-ventilation system to ventilate two patients using the same ventilator

The objective of the following simulation is to test the ability to connect and disconnect one patient from the differential multiventilation circuit and measure the effects on the other patient.

The experience was done in pressure-controlled ventilation. Fig.4.16 show that a total obstruction of one circuit in pressure-controlled ventilation is possible, and induces no change in the tidal volume delivered to the other patient. It allows therefore to connect or disconnect a patient from the differential multi-ventilation circuit.

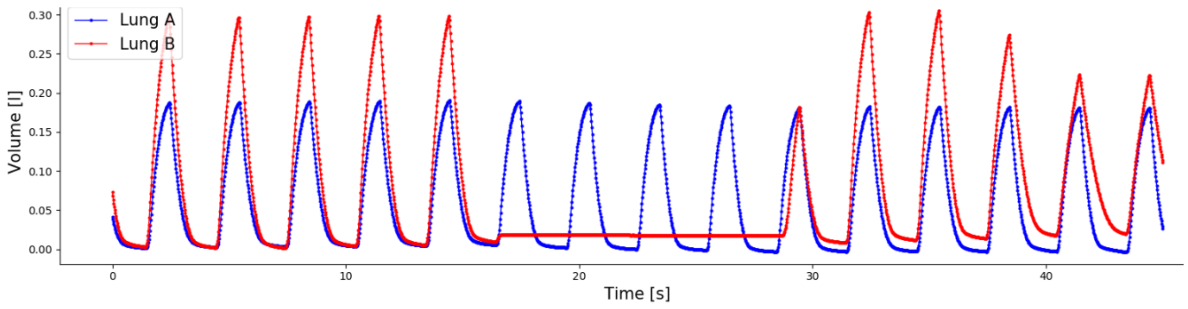


Figure 4.16: Simulation of disconnection of a patient on a differential multi-ventilation circuit and influence on the other patient. Flow restrictor of test lung B is closed to allow its disconnection while variation on the tidal volume delivered to the test lung A is measured.

Chapter 5

Discussion

Ventilator sharing between two is a solution to rapidly increase the capacity of existing ventilators. This technique, however, entails multiple and significant patient safety concerns raised in a Joint Statement. Among these concerns, the impossibility of individualising the tidal volumes delivered to patients was mentioned. Through the use of a differential multi-ventilation system, it is however theoretically possible to address this concern. The aim of this work was therefore to analyse the actual feasibility of individualising tidal volume to provide answers as to the medical validation of this way of ventilation sharing.

Setting up a benchtest

The first part of this work was dedicated to the setting up of a test bench, which served for the performing of all tests during this work. This test bench consists of a ventilator, tubing, test lungs, pressure and volume sensors and their monitoring screen, as well as the various other modules required for a differential multi-ventilation setup.

The present bench test has limitations. First, it is based on a ventilator from the 2000s that is no longer up to date. It still does the job, but hospitals are now using more recent ventilators, which are more accurate and more efficient.

Secondly, the sensors implemented have been calibrated on the basis of the ventilator, so that they provide the same respiratory curves as the one provided by it. However, the repeatability of the sensors has not been tested. The data may therefore vary from one measurement to another. Trust has been placed in the sensors by the simple fact that they have been designed for a medical application.

Finally, to obtain different compliance values for the test lungs in order to simulate the ventilation of patients with different lung mechanical properties, several rubber bands were added around the lungs. This method has also been performed in another study [?]. Results show a linear trend of the compliance, which is the expected result of a test lung compliance. . However, no study was found to prove that this method is reliable. The application of an additional force on the test lung could influence its dynamic behaviour and thus distort the dynamic of the fluids during breathing. Further study is needed to determine this.

Study of different flow restrictors for differential multiventilation.

Thanks to the various characterisation carried out throughout this part of the work, a flow restrictor was selected for differential multi-ventilation from the 5 initially selected.

The characterisation of flow restrictor linearity and sensitivity was the most relevant experiment in this section. The curves obtained show a general less linear behaviour of a diaphragm or needle valve compared to a generally more linear applied pressure regulating valve (APL). Similar trends have also been observed by Tormod Martinson and Christian Tronstad, from Oslo University Hospital, Norway for the same type of valves. This non-linear trend is further explained in the literature to be a "quick opening" valve characteristic curve. Diaphragm valves are well-known to fit that kind of curve. The sensitivity calculated on the linear part of the curve will depend mainly on the number of turns between the fully open and fully closed positions. In general, the greater the number of turns, the more accurately the valve can be set. However, if this number of turns is large, adding a visual cue of the valve's closure seems essential to any design. In order to select a valve, a higher weight has been given to sensitivity than to linearity but this is debatable as in general, linear flow restrictors are easier to operate.

With regard to tightness at full closure, a study [44] advises the use of design limited constriction to avoid the valve shutting off the flow if not careful during its adjustment. However, in this work, we encourage the use of valves that allow to shut down a flow at full closure. This is necessary to be able to disconnect a patient from a differential multi-ventilation circuit. Furthermore, an error in the adjustment of the valve is unlikely, since to use such a ventilation device, the nursing staff will have to be well trained.

Results from leakage, egonomics and reusability are less relevant. These tests are mainly based on personal judgements rather than on a solid scientific approach. For example, it would have been interesting to develop a replicable approach for each valve in order to test its ergonomics (e.g. measuring the closing force or measuring the size of the closing element for example). The final choice of valve is, therefore, to be taken with a grain of salt. The possibility that someone will reach a different conclusion regarding ergonomics, tightness and reusability characterisation exists.

More precisely, the performed study of different flow restrictors for differential multiventilation has several limits. At first, the tests have not been carried out enough times to have any relevant statistical data on the results obtained. Secondly, the selection of the best valve is only made on 6 criteria. It would be interesting to enlarge this list to be more objective. An interesting property to analyse would be for example the feasibility of the flow restrictor in surgical-grade stainless steel, for those wishing to develop a differential multi-ventilation setup completely in this type of steel. The stability of the valve over time is also a characteristic that could be measured.

In conclusion, due to its similarity with other tests in the literature and its close link with the characteristic curve valve, it can be said that the results obtained for the linearity and sensitivity analysis are relevant, despite the lack of statistical data. The final choice of the best valve, however, is not relevant.

Tidal volume distribution when sharing a ventilator between two patients

The third contribution of my work was devoted to the analysis of the distribution of tidal volumes and in particular to the analysis of the possibility of individualisation of tidal volumes for two patients in differential multi-ventilation.

The results from Fig.4.5 and 4.7indicate that when the compliance is identical for both patients ventilated, the volume delivered by the ventilator is distributed quite equally, regardless of the tidal volume or pressure delivered by the ventilator, with a maximal difference between the two patients recorded of 6%. This volume distribution, however, becomes unequal as soon as a difference in mechanical properties (i.e. compliance and resistance) between the two patients occurs, and was large already when the stiffer lung had a 15 % lower compliance than the other lung, see Fig.4.10, 4.9. More specifically, a lung with decreasing compliance or increasing resistance will also have a decreasing tidal volume. In volume-controlled ventilation, lower compliance or higher resistance in one lung directly increases the volume delivered to the other lung. In pressure-controlled ventilation, however, tidal volume and pressure delivered to a patient solely depend on their own lung compliance and resistance. It allows therefore to keep the ventilation to one lung constant while changing compliance or resistance of the other lung. These results are largely in agreement with conclusion of several studies ([4], [40], [41]).

Additional simulations, Fig.4.15, 4.14, demonstrate that any of the two ventilation modes can achieve the individualisation of tidal volume goals. The method of achieving this, however, varies from one ventilation mode to another.

To individualise the tidal volume for pressure-controlled ventilation (PVC), the ventilator parameters must first be adjusted to achieve the desired ventilation on the patient with the lowest compliance. Then the flow restrictor of the other patient is adjusted to provide the desired tidal volume. In the case of volume-controlled ventilation (VCV), the tidal volume on the ventilator should be set as the sum of the target volumes of the two patients and the adjustment of the flow restrictor will change the volume delivered to both patients.

However, although individualisation is possible, the accuracy with which this is done can only be done within the sensitivity of the flow restrictor used. For a patient with ARDS, a study showed that an increase of 1mL/kg in initial tidal volume was associated with 23% increase in ICU mortality risk [53]. Therefore, the use of valves with linear behaviour and small sensitivity per degree of rotation is to be preferred.

Also, from the above-mentioned results, it is suggested that PCV is preferable when performing differential multi-ventilation. It allows for keeping the ventilation constant for one patient while the compliance or resistance of the other patient was changed. CVV is not advised since a rapid rise in one patient's flow impedance would result in a quick increase in the other patient's tidal volume.

The first clinical simulations carried out in the course of this work, see Fig4.16, are promising. Connecting and disconnecting a patient is typically the kind of scenario one would encounter in clinical use. The tests prove that all these manoeuvres of connecting and disconnecting a patient have no influence on the ventilation of the other patient in the case of pressure-controlled ventilation. This test also shows the importance of using a valve with a perfect seal, even if other ways could be used, such as clamping the tubes during the change of the patient, but this would damage the tubing.

In view of the results obtained above, and the similarities with other studies previously carried

out, it is legitimate to assert that the individualisation of tidal volumes during the *in-vitro* differential multi-ventilation is feasible. However, the overall interest of this work is to ensure that the individualisation of tidal volumes in a differential multi-ventilation configuration can be reliably implemented and used in clinical critical care settings to address ventilator shortages. To this end, further steps need to be taken to address the current limitations prior to clinical implementation.

Firstly, in the used set-up, some modules, as well as some connectors, are 3D printed causing slight leaks which may influence the results. Secondly, the data analysis is not done numerically and has to be done at the end of the experiment only. Instantaneous monitoring of patients is therefore not currently feasible. Third, only one ventilator was used for testing. However, the test should be run on a variety of ventilators due to variations in performance characteristics of ICU ventilators. Fourth, 3 combinations of lung compliance, with low value, have been tested. More combinations with larger differences need to be studied. However, the different ventilation settings that have been used throughout the tests have been chosen to cover a large range without any clinical basis behind it. A Standard set of practical, evidence-based parameters would have been more relevant. Fifth, most of the time, only tidal volume monitoring was done, whereas flow curves can bring additional information on the treated data. Finally, the setup has not been tested with active humidification or heat and moisture exchangers.

To summarise, the advances made from this work are shown in Table5.1.

Concern	**Dual Patient Ventilation**	**Differential Multi-ventilation (Theoretical perspective)**
Individualize ventilation to each patient		
-$PEEP$	Shared between patients	Individualized to each patient
-V_T	Shared between patients	Individualized to each patient
-CO_2 measurements	Shared between patients	Individualized to each patient
-$I:E$, RR, FIO_2	Shared between patients	Shared between patients
Rebalancing of the airflow when one patient improves or deteriorates (C, R)	Patients must be rematched as they improve or deteriorate because it is impossible to individually adjust each patient	Flow restrictor adjustments allow desired tidal volume for each patient even if patients improve/deteriorate
Measure of pulmonary mechanics	Shared between patients	Can be computed for each patient
Potential cross- contamination	Great risk	Respiratory filters and one way valves greatly reduce the risk
Patient monitoring	Additional respiratory monitors and heightened clinical vigilance required	Additional respiratory monitors and heightened clinical vigilance required
Safely removing a patient from the circuit	Patient cannot be quickly removed from circuit	Individual patient can be quickly removed from the circuit by flow restrictor adjustments.

Table 5.1: Main concern of shared ventilation. A comparison between dual patient ventilation and differential multiventilation Contributions made through this work are highlighted in blue

Chapter 6

Conclusion

The first contribution of this work was to set up a test bench. More specifically, it was necessary to set up a differential multi-ventilation set-up that served as the basis for all the tests performed in this work. This set-up included a reanimation ventilator, linear test lungs, valves, medical filters, splitters, one-way valves, and flow and pressure sensors. These flow and pressure sensors were implemented using an Arduino. To monitor the volume, pressure and flow curves in each limb of the device, a dynamic interface using MatPlotLib was implemented. The first experiment on this set-up allowed to determine the compliance of three different test lungs, which were used to simulate patients with different compliance.

The second contribution of this work was to determine the flow restrictor that will be used in the set-up to conduct the tests. The literature provides a range of different flow restrictors. Amongst these, five valves were chosen, and each valve was subjected to a set of tests to characterize them. The results obtained are similar to those found in the literature. A comparison between these valves was then made to select a single type of valve, which will be used in the multi-ventilation differential set-up.

The last contribution is the one that answers the research question. It consisted in determining the possibility of individualising the tidal volumes for two patients undergoing differential multi-ventilation in volume and pressure-controlled ventilation. The ultimate goal was to medically validate the differential multi-ventilation. Initial tests with a simple ventilation sharing device were used to determine the impact of a change in compliance on the distribution of volumes between the lungs, then the influence of resistance on this volume, and finally the influence of the ventilation mode on the distribution of these volumes. Tests on the multi-ventilation differential circuit were then carried out to individualise the volume delivered to each patient. This was done in pressure ventilation and volume control. A final test was then carried out in pressure control to determine if it was possible to disconnect one patient from the circuit without impacting the other. The results showed that it was indeed possible.

Even if some limitations are to take into account in the interpretation of the results, like the low amount of combinations of compliance simulations, it has been shown that it is possible to individualise the tidal volume for two patients for pressure and volume-controlled ventilation. However, the pressure-controlled ventilation mode is much safer than the volume-controlled ventilation mode. These results allow us to solve three of the concerns related to shared ventilation. In addition, results show that some clinical scenarios would be difficult to perform in volume-controlled ventilation. However, these analyses do not allow to conclude on the medical validity of differential multi-ventilation. Improving the identified limitations and individual validation of the other concerns of sharing ventilation is crucial before any in vivo testing.

Appendix A

Supplement to state-of-the-art

1 The laws governing behaviour of gases

The laws governing behaviour of gases

Pressure: Based on the reference pressure, the following terminology is used:

- Absolute pressure: measured in relation to vacuum
- Relative pressure: measured in relation to atmospheric pressure
- Differential pressure = measurement of the difference between 2 pressures

The unit of pressure in the International System of Units (SI) is the pascal (Pa), defined as a force of one Newton per square meter ($1Pa = 1N/m^2$). However, depending on the discipline concerned, other units of measurement are sometimes used to measure the pressure:

- Bar [bar] : 1 bar = 10^5 Pa
- Standard atmosphere [atm] : 1 atm = $1,013x10^5$ Pa
- Torr [$mmHg$] : 1 $mmHg$ = $133,32$ Pa
- Centimeters of water [cmH_2O] : 1 cmH_2O = $98,06$ Pa
- Pound per square inch [psi] : 1 psi = 6895 Pa

The atmospheric pressure is the absolute pressure exerted by the weight of the air, at sea level at $0°C$ has a mean value of 101325 Pa = 1 atm = 760 $mmHg$.

Ideal gas law:

$$PV = nRT$$

Where: R = universal gas constant $= 8,314J \cdot K^{-1} \cdot mol^{-1}$, n= number of moles of gas, T = temperature in degrees Kelvin.
From the ideal gas law derives other laws:

- ***Boyle-Mariotte's law :*** at a constant temperature the same amount of gas will decrease in volume with an increase in pressure. With other words, the product

of gas pressure and volume is a constant value in during a isothermal processe at constant mass :

$$p_1 V_1 = p_2 V_2$$

Where 1 and 2 refers to state 1 and 2.

- ***Charle's law :*** at constant pressure, the same amount of gas will increase in volume with an increase in temperature

Flow: For a flow to occur, there must be a pressure difference (Δp). A gaz, like air, always move from higher to lower pressure. During breathing, the movements of the thorax create alternately high and low pressures in the lungs, which explains how gas exchanges can take place (see section 2.2.2).

The flow rate (Q) is directly proportional to the pressure difference : $Q \propto \Delta p$. The greater the pressure difference, the greater the flow rate.

2 Gas exchange in human body

2.1 Gas laws principles (2) and air composition

Understanding the basic principles of gases and their behaviour is essential to fully understand the processes of gas exchange in the lung. In addition to the several law seen before (section 2.2.1), other gas laws help to describe the behavior of gases. Basically:

Gas laws principles (2)

Partial pressure
In a mixture of gases, each constituent gas has a partial pressure which is the notional pressure of that constituent gas as if it alone occupied the entire volume of the original mixture at the same temperature.

Dalton's Law
This empirical law was observed by John Dalton in 1801 and is related to the ideal gas laws. It's definition is the following one : "For a on reacting gaseous mixture, the total pressure of the mixture is the summation of all partial pressures gases that make up its content". Therefore, if a mixture is composed of 3 different gases, men can write :

$$p_{tot} = p_{gase_1} + p_{gase_2} + p_{gase_3} \tag{A.1}$$

Flow
We already know that for a flow to occur, there must be a pressure gradient. However the displacement according to a pressure difference also applies to a single gas, due then to a partial pressure difference. The diffusion of gases within the alveoli follows this mechanism

Air composition

Table A.1 gives a typical composition of dry air [54]. However, atmospheric air is rarely totally dry but is made up of a mixture of dry air and water vapor and is called moist air.

Components	Molar Fraction (% in volume)
Nitrogen (N_2)	78.08
Oxygen (O_2)	20.95
Argon (Ar)	0.93
Carbon dioxide (CO_2)	0.03
Other gases	0.01

Table A.1: Chemical composition of dry air.

Atmospheric air is therefore a mixture of gases, each of wich independently contributes to its total pressure. The pressure of each individual gas is know as partial pressure. Therefore, according to Dalton's law:

$$p_{atm} = p_{N_2} + p_{O_2} + p_{Ar} + p_{CO_2} + p_{Othergases} + p_{H_20} \tag{A.2}$$

Where p_{H_20} is the partial pressure of water, which depends on the temperature. Values can be found in tables.

2.1.1 Process of gas exchange

As mention before, at each generation of the bronchial tree, the total surface of airways increase. According to the principle of mass conservation, gas flow velocity decreases as the area increases. The average velocity of gas through the trachea during a normal breath is roughly 0.7 m/sec, whereas it is just 0.001 mm/sec at the alveolar surface [14]. This is substantially slower than the rate of O2 and CO2 diffusion. Diffusion rather than convective movement, is therefore responsible for gas exchanges in the alveoli.

During this exchange, oxygen diffuses passively across the aveolar capillary membrane from the alveolar gas phase into plasma and red cells, where it binds to haemoglobin. Carbon dioxide diffuses from blood to alveoli in the reverse manner.. The driving force behind the diffusion is the partial pressure difference.

The quantity that may diffuse through the membranes for a given time follow Fick's law of diffusion : [55], [11]:

$$\text{Diffusion rate} \propto \frac{\text{Surface} \times \text{Pressure difference} \times \text{Diffusion coefficient of the gas}}{\text{Thickness of the membrane}}$$

Figure A.1 illustrate the gas exchanges between alveoli and blood for both O2 and CO2 while highlighting the variation of partial pressure throughout those exchange.

Note that the partial pressures of O2 and CO2 within the alveoli are not the same as in the atmosphere. This is mainly due to the fonctional residual capacity (see section **??**). With other word, atmospheric air is brought into the lungs through inhalation, but the lungs are not completely emptied and replaced with outside air with each cycle of breathing. Only a small portion of air in the alveoli is refreshed with each breath. This make the air composition in the alveoli significantly different from that of the inhaled air.

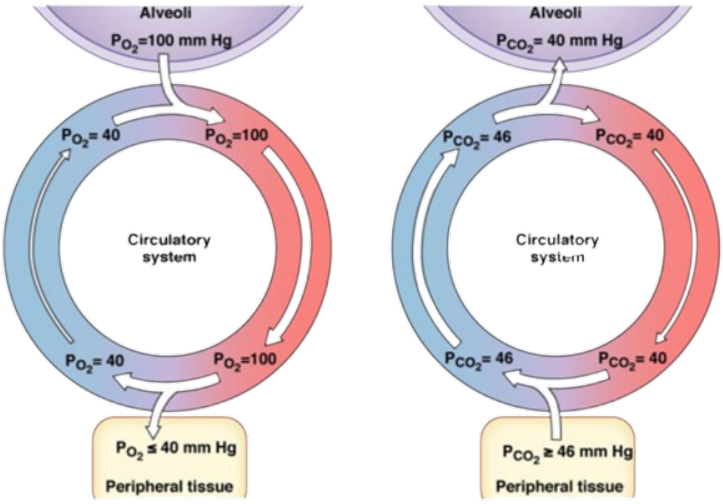


Figure A.1: Gas exchange between alveoli and blood

3 List of flow restrictors

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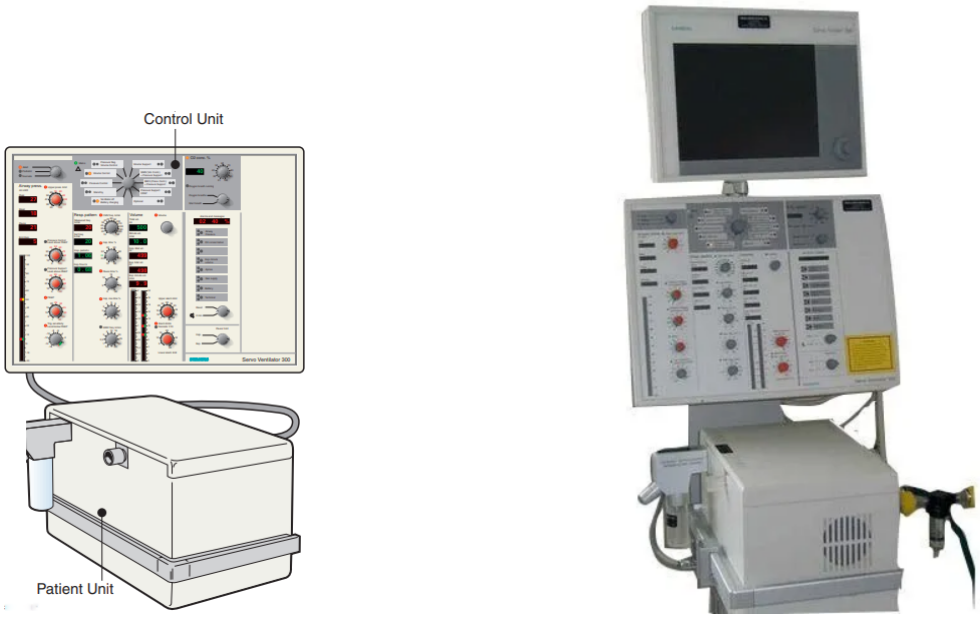
Appendix B

Supplement to materials and methods

1 Test hardware

Hereafter some illustration of the test hardware that was used for testing. See Figures B.1, B.2, B.3, B.4, B.5, B.6,B.7 and B.8



(a) (b)

Figure B.1: Siemens servo ventilator 300A

2 Datasheet of the massflow sensor

3 Flow restrictor leakage

Figure B.9 show how hose clamps and rubber adapter were used for sealing leaks of the Needle valve.

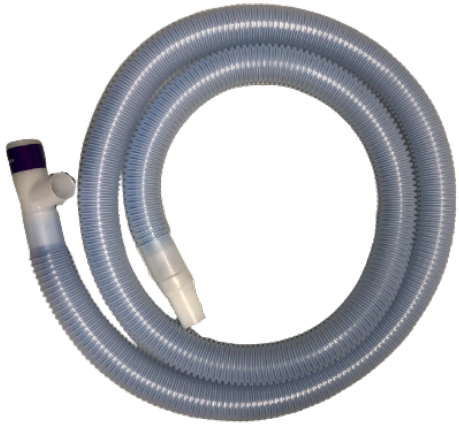


Figure B.2: Double lumen corrugated tubing, 22-mm outer diameter

Figure B.3: GIBECK Iso-Gard Filter S

4 Head loss calculation

The total head loss inside a pipe is the sum of

1. *Major losses*: it is the friction between the walls of the pipe and the fluid.
2. *Minor losses*: it is the additional losses due to entries and exits, fittings and valves.

Major losses

The calculation of major losses is computed using Darcy-Weisbach equation:

$$\Delta h = f \frac{L}{D} \frac{V^2}{2g} \tag{B.1}$$

- Δh : pressure loss in m
- f : Darcy friction factor
- L : Pipe length in m
- D : hydraulic diameter in m
- V : fluid flow average velocity in m/s
- g : standard gravity = 9,81 m/s^2

For a laminar flow , we have : $f = \frac{64}{Re}$ where Re is Raynold number.
For a turbulent flow, f is calculated using Colebrook's empirical formula :

$$\frac{1}{\sqrt{f}} = -2log\left(\frac{\epsilon/D}{3.7} + \frac{2.51}{Re\sqrt{f}}\right) \tag{B.2}$$

A graphical method also exists to calculate the coefficient f, which is the Moody chart.

Minor losses

The calculation of minor losses is computed using the following equation:

$$\Delta h = K \frac{V^2}{2g} \tag{B.3}$$

where K is loss coefficient must be determined for each situation, see Figure B.10.

Figure B.4: AR Adult-Pediatric Electrostatic Filter HME

Figure B.5: Maquet Test lung 190, 1L

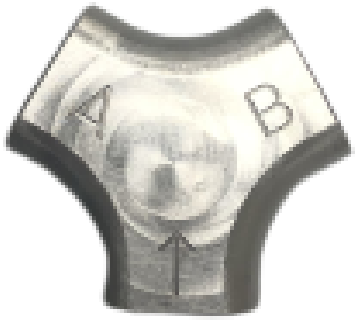


Figure B.6: Y Splitter from MultiVentY

Figure B.7: Sisto diaphragm valve used as flow restrictor

Figure B.8: One way valves

Figure B.9: Sealing leaks for Needle valve

Table 5-3 Representative Loss Coefficients for Various Transitions and Fittings

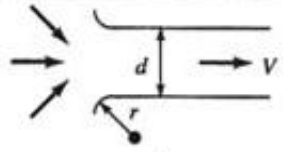
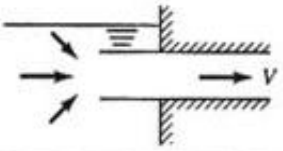
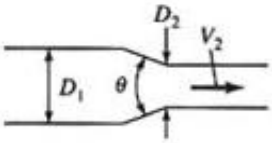
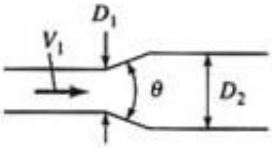
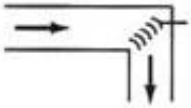
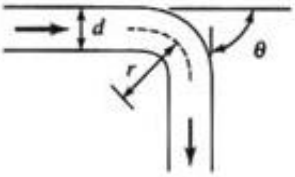
Description	Sketch	Additional Data	K		Source
Pipe entrance $h_L = K_e V^2/2g$		r/d 0.0 0.1 >0.2	K_e 0.50 0.12 0.03		(7)
			1.00		
Contraction $h_L = K_C V_2^2/2g$		D_2/D_1 0.0 0.20 0.40 0.60 0.80 0.90	K_C $\theta = 60°$ 0.08 0.08 0.07 0.06 0.05 0.04	K_C $\theta = 180°$ 0.50 0.49 0.42 0.32 0.18 0.10	(7)
Expansion $h_L = K_E V_1^2/2g$		D_1/D_2 0.0 0.20 0.40 0.60 0.80	K_E $\theta = 10°$ 0.13 0.11 0.06 0.03	K_E $\theta = 180°$ 1.00 0.92 0.72 0.42 0.16	(7)
90° miter bend		Without vanes	$K_b = 1.1$		(42)
		With vanes	$K_b = 0.2$		(42)
Smooth bend		r/d 1 2 4 6	K_b $\theta = 45°$ 0.10 0.09 0.10 0.12	K_b $\theta = 90°$ 0.35 0.19 0.16 0.21	(14) (22) and (30)
Threaded pipe fittings	Globe valve - wide open Angle valve - wide open Gate valve - wide open Gate valve - half open Return bend Tee 90° elbow 45° elbow		$K_v = 10.0$ $K_v = 5.0$ $K_v = 0.2$ $K_v = 5.6$ $K_b = 2.2$ $K_t = 1.8$ $K_b = 0.9$ $K_b = 0.4$		

Figure B.10: Loss coefficent for minor losses calculation

Appendix C

Supplement to the results

1 Effect of compliance on tidal volume distribution

Hereafter, the respiratory curves generated for the experiment "effect of compliance on tidal volume distribution", for VCV with setting 1 (C.1,C.2 and C.3) and PCV with setting 1 (C.4,C.5 and C.6).

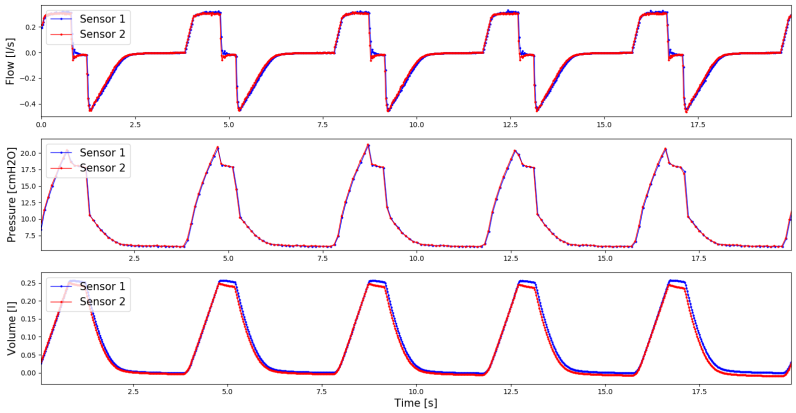


Figure C.1: Effect of compliance study. Monitoring curves for VCV with setting 1, for balanced compliance

2 Effect of a flow restrictor on tidal volume distribtion

Hereafter, the curves of individual tidal volume of test lung A and B as function of valve closure of test lung A for ventilation in a volume controlled mode, for different values of compliance. Fig C.7.

3 Individualisation of the tidal volume with different lungs compliance

Hereafter, the curves generated for the experiment "individualisation of tidal volume with different lungs compliance", for PCV (C.8) and VCV (C.8).

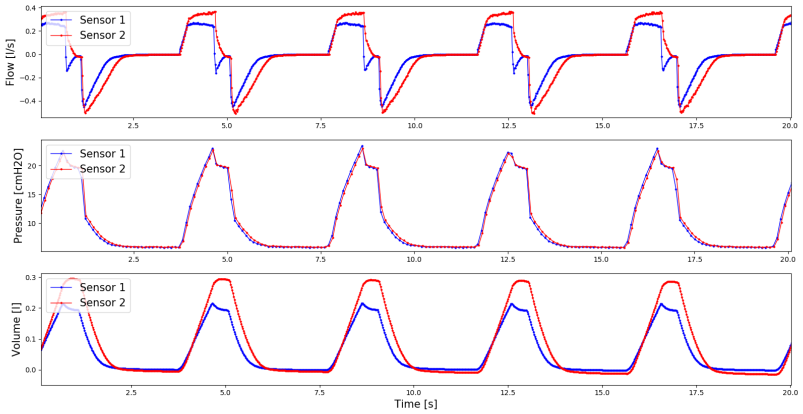


Figure C.2: Effect of compliance study. Monitoring curves for VCV with setting 1, for unbalanced compliance

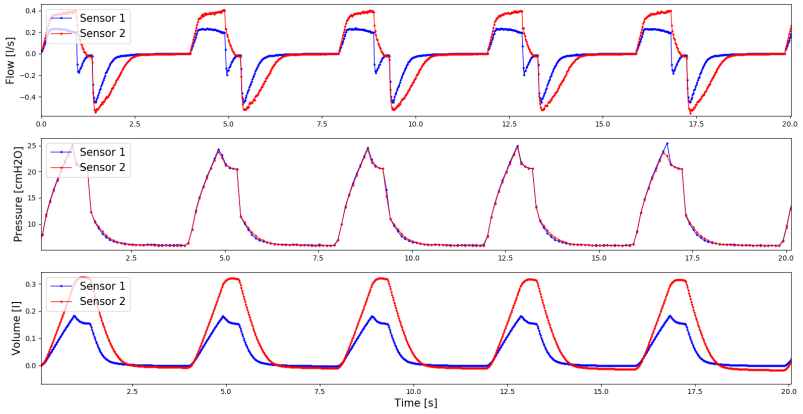


Figure C.3: Effect of compliance study. Monitoring curves for VCV with setting 1, for unbalanced compliance (severe)

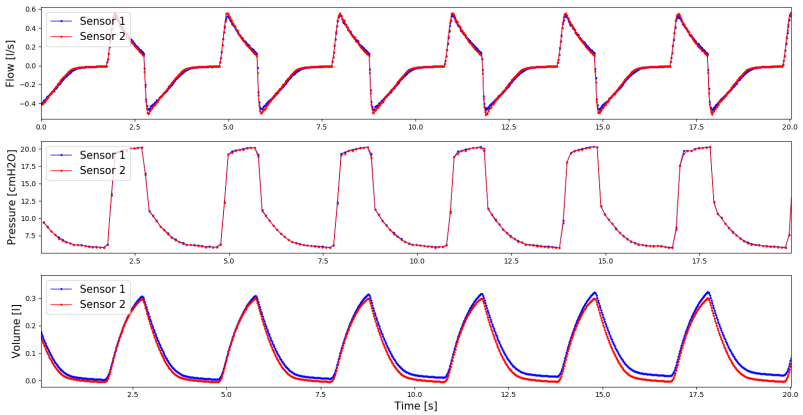


Figure C.4: Effect of compliance study. Monitoring curves for PCV with setting 1, for balanced compliance

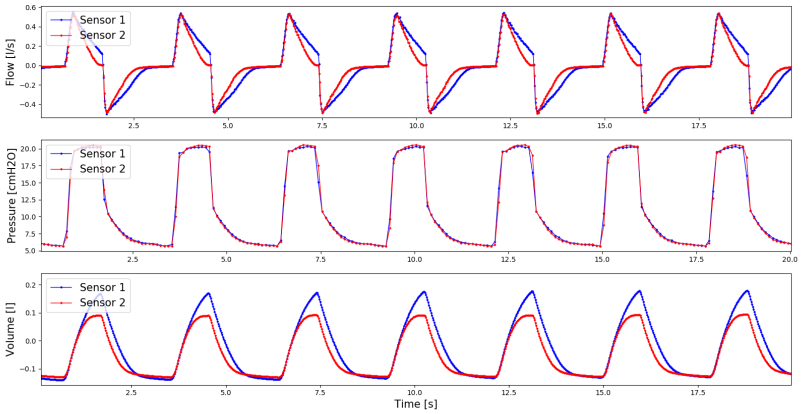


Figure C.5: Effect of compliance study. Monitoring curves for PCV with setting 1, for unbalanced compliance

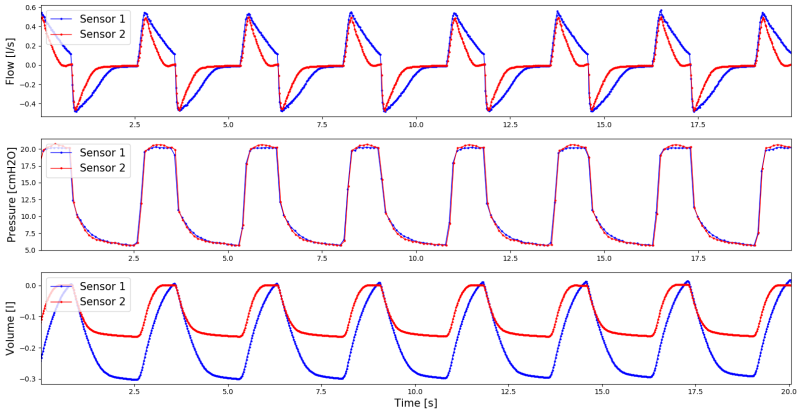


Figure C.6: Effect of compliance study. Monitoring curves for PCV with setting 1, for unbalanced compliance (severe)

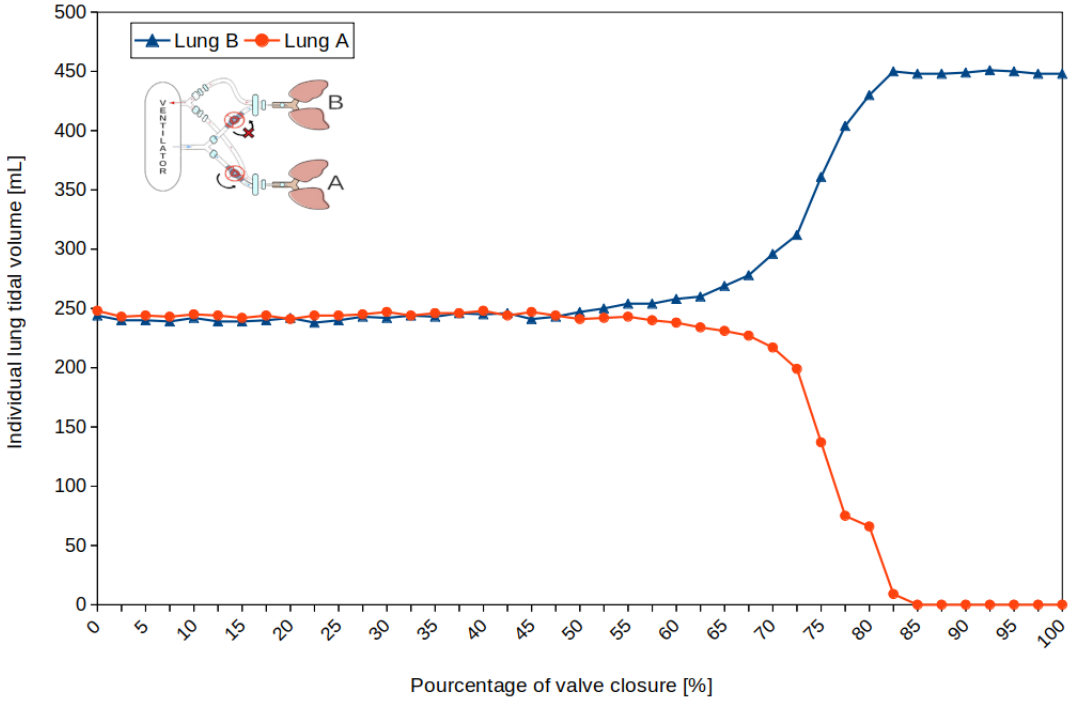


Figure C.7: Individual Tidal volume of test lung A and B as function of valve closure of test lung A for ventilation in a volume controlled mode. Curves were generated by gradually closing patient A valve by increments of one 16th of a turn while the valve of patient B remained fully open. Lung A as a compliance of C_L =32mL/cmH2O and Lung B as a compliance of C_L= 27mL/cmH2O. Ventilation settings 1 were used.

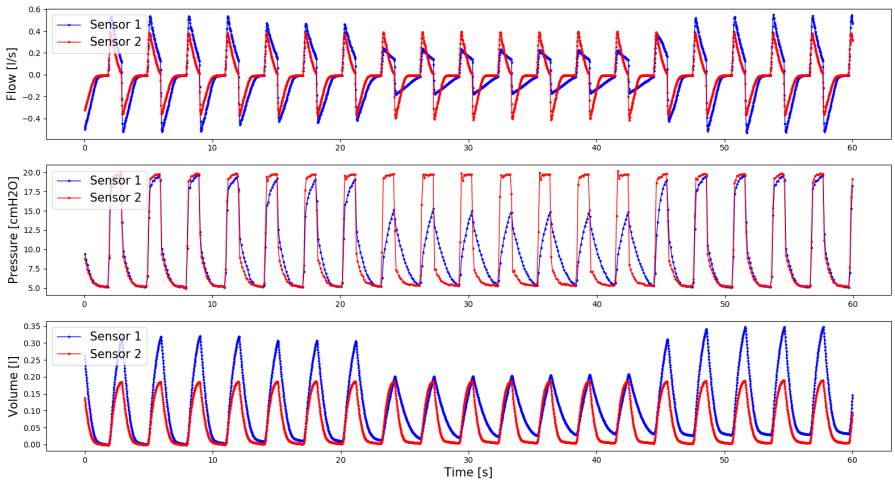


Figure C.8: Individualisation study. Monitoring curves for PCV with setting 1, for unbalanced compliance

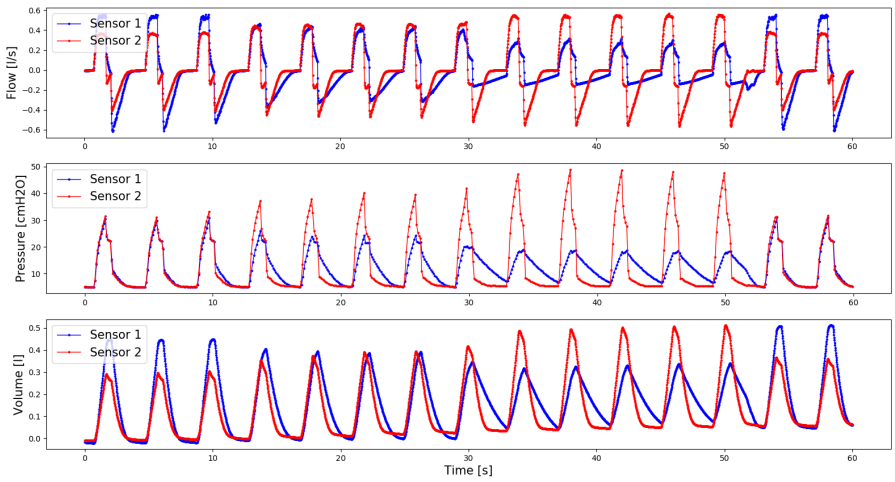


Figure C.9: Individualisation study. Monitoring curves for VCV with setting 1, for unbalanced compliance

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