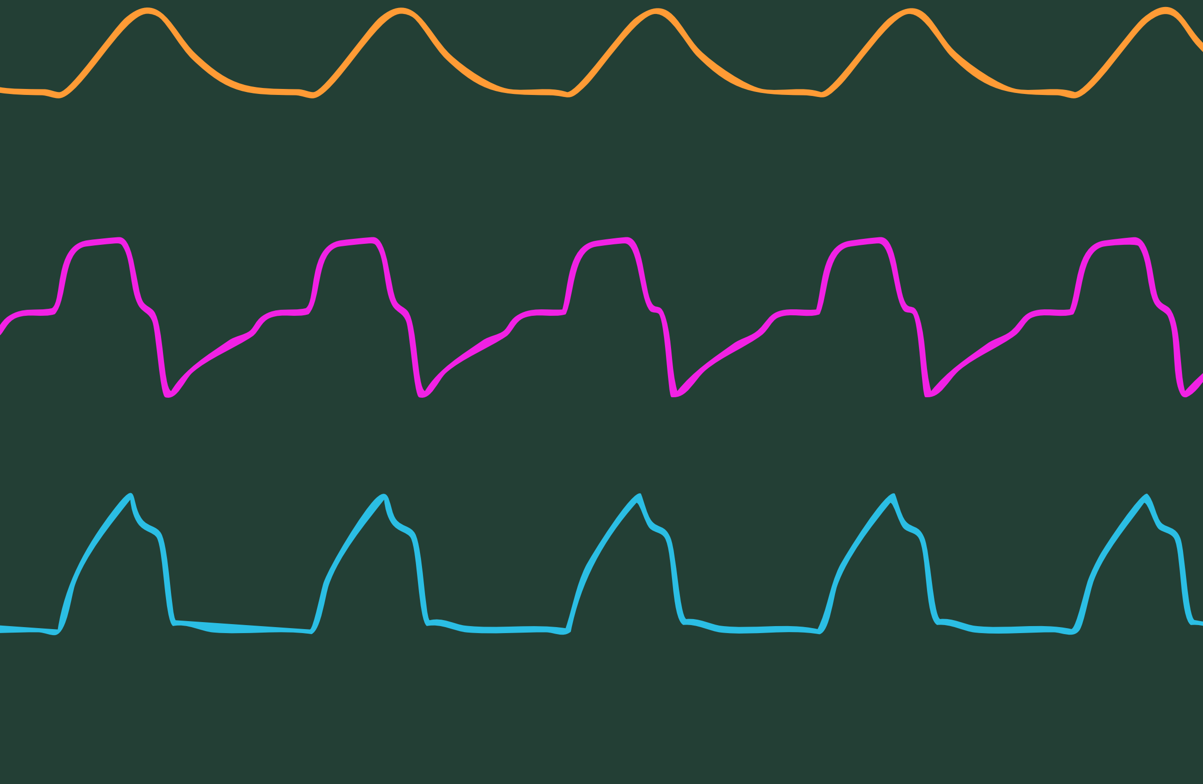
INTRODUCTION TO RESPIRATORY MECHANICS



Matías Madorno

INTRODUCTION TO RESPIRATORY MECHANICS

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FluxMed Academy

A space for academic information regarding respiratory physiology and mechanical ventilation is set up. It is called FluxMed Academy. This book is the first material that will be available in this space.

You can access it on the www.fisbol.com website or by scanning this QR code:

The structure and material that this space will have has not yet been defined. We invite the community of mechanical ventilation and respiratory physiology enthusiasts to send us (academy@fisbol.com) their proposals, ideas, comments, complaints and any opinion they have on how we can make this space add to the community.

Preface

This book presents respiratory physiology in an accessible way with a quantitative analysis, under the idea that understanding the relationship between the different variables and components helps to have a better notion of what is happening with a patient who is on mechanical ventilation.

It is useful to think that the behavior of respiratory physiology is according to simplified physical models. This helps to understand and predict what is happening with the patient. Even though these models are big simplifications they are very useful as an initial approach to understanding the interaction of the different components.

In this book we will see how from a few basic components we can assemble different models of the respiratory system, and with them analyze the behavior. Starting with simpler models and then making them more complex to explain some more advanced situations. The idea of this book is that the reader leaves with a robust understanding of the simplest models that allow explaining many everyday scenarios.

The decision of how to ventilate a patient is much more complex than what is presented in this book and should not be taken as the only source to decide the patient's mechanical ventilation strategy.

> Throughout the text the reader will find these sections in a different style where analogies are made, or the topic is seen from another angle to support understanding. An everyday example helps to understand the logic behind it. We are used to performing mathematical operations daily when we buy things. The same mathematics is what is needed to understand respiratory physiology. These analogies are marked in another color so that whoever uses it can look for it or skip it as they prefer.
>
> The reader is invited to do the calculations while maintaining the units of the different variables. Through the calculations the units should cancel out, and the result should be in the units of the resulting magnitude. A quick example, if a flow of 15 Liters/minute is sustained for 2 minutes the total volume is: 15 Liters/minutes * 2 minutes = 30 Liters. This helps to better understand what is being done and is a way to detect errors. Doing the math with the units, the result must be in the unit they expect from the result.
>
> Units of measurement are your friends.

Chapter 1.

Introduction to Quantitative Respiratory Physiology

Basics

In order to have the tools to be able to accurately analyze what is happening, and instead of saying a pressure difference that makes a volume move, being able to say how much pressure and how much volume, the narrative should be more precise: the values, relations and proportions have to add up.

Let's start by reviewing the variables that are studied in respiratory physiology associated with mechanical ventilation.

Pressure

Pressure is a force applied to an area. To measure it, the total force applied divided by the area where it is applied could be used.

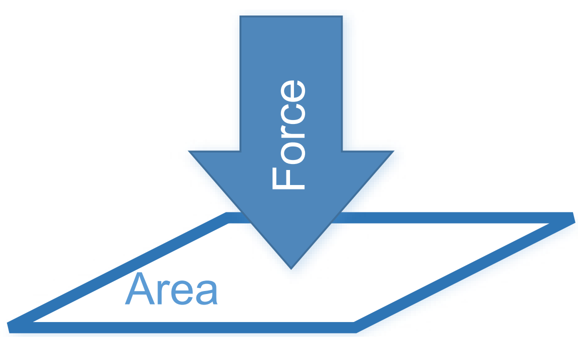

Figure 1. A pressure is defined as a force applied over an area

The molecules of a gas exert pressure on the walls of the container that contains it. In mechanical ventilation, pressure is usually measured in centimeters of water (cmH_2O).

A standing person will exert a force on the floor proportional to his weight. The pressure he generates on the floor will depend on the area on which he rests. If the surface area is smaller, the pressure will be greater. If this person stands on a soft ground with a shoe with studs, as the area is small, the pressure will be greater, and he will dig into the ground. If he changes the shoes for one with a larger surface, such as the snowshoes, with the same weight the surface will be larger making the pressure lower and thus he will not sink into the ground. Another example is when we force ourselves to nail a pushpin into a cork wall. The force that the finger makes against the pushpin is the same as the force that the pushpin does against the wall. As the surface that is in contact with the finger is much larger than the surface that is in contact with the wall, the pressure on the finger is much lower than on the wall.

Volume

Volume is the space occupied by a body. In respiratory physiology we will mainly measure the volume of air (or of the different mixtures of gases breathed). Volume is usually measured in liters (L) or milliliters (mL), where 1L equals 1000mL.

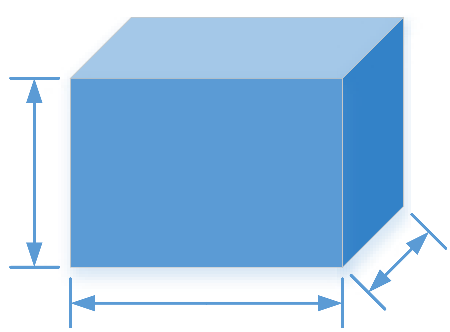

Figure 2. Volume is the space occupied by a body.

Some volumes are commonly referenced so they have a specific name. For example, the volume inspired and expected in a breath is the tidal volume.

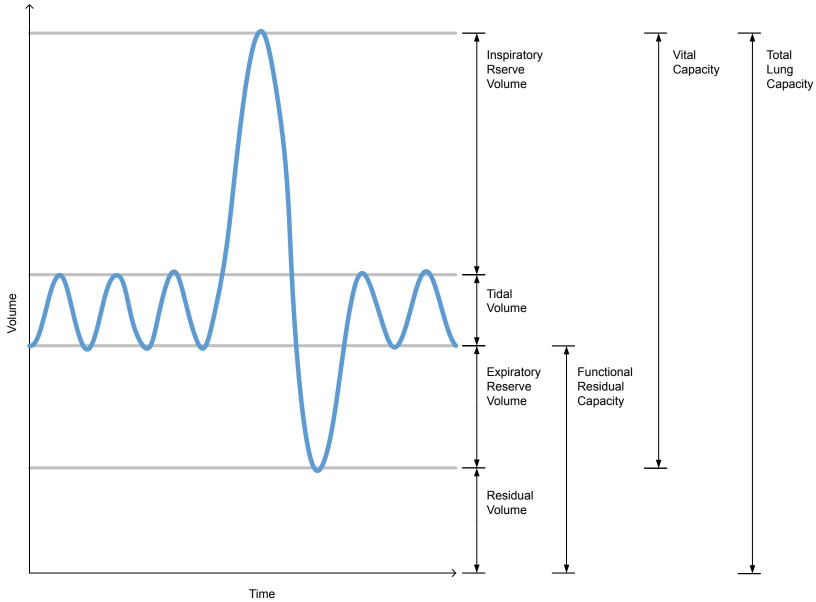

Figure 3. Lung volumes and capacities seen on a volume-time tracing during a respiratory function test.

Flow

Flow is the movement of the volume of air per unit of time; it is the speed at which the volume of air moves. In mechanical ventilation, flow is usually expressed in in liters per minute (L/min) or sometimes in liters per second (L/sec). Flow is considered positive when circulating to the patient and negative when exhaling. In spirometry it is the other way around, expiration is positive and inspiration negative.

It is worth noting the difference between flow and the linear velocity of a gas. Flow is how many liters of fluid travel per unit of time, whereas linear velocity is the distance a molecule travels per unit of time. The flow is equal to the linear velocity multiplied by the area of the tube section.

Flow and volume are intimately related. They are linked by time. Volume is the integral of flow and flow is the derivative of volume.

$$F(t) = \frac{dV(t)}{dt}$$

$$V(t) = \int F(t)dt$$

The information contained in a flow signal is the same as in the volume signal, since from one the other can be obtained. This establishes that if the flow or volume curve is defined, the shape of the other is also defined.

Both curves are usually plotted because when interpreting that information visually it can be much easier to see in one or the other curve.

> Flow is a velocity and volume is distance. This is analogous to a car moving at a certain speed and distance it travels. Speed and position are related by time. If you know the position you were at each moment, it is possible to calculate the speed you had between each point. Or, if we know how fast you travel and for how long you keep that speed, you can calculate the distance you traveled.
>
> Another way to look at the relationship between flow and volume is with the bank account. If we know all the movements (flow) we can add them up and have the amount of money in the account (volume). Or if we have the total amount of money in the account (volume) at any given moment, by seeing how it changes, we can know the movements (flow).

Resistance

Resistance is the force that opposes the circulation of flow. It is the pressure difference necessary to generate one flow unit.

In respiratory physiology, resistance is usually measured in the centimeters of water required to generate one liter per second ($cmH_2O/L/sec$) or the centimeters of water to generate one liter per minute ($cmH_2O/L/min$). Resistance is the ratio of the pressure difference to the flow.

If two compartments that have different pressures are connected with a tube, it is intuitive to see that the fluid will circulate through the tube. Going from the compartment with higher pressure to the compartment with lower pressure. The flow will be larger, the greater the pressure difference is between the two compartments.

The speed at which the fluid moves from one container to the other depends on the resistance of the tube that connects them. The higher the resistance, the lower the flow that circulates for a given pressure difference. This relationship is explained by Ohm's law.

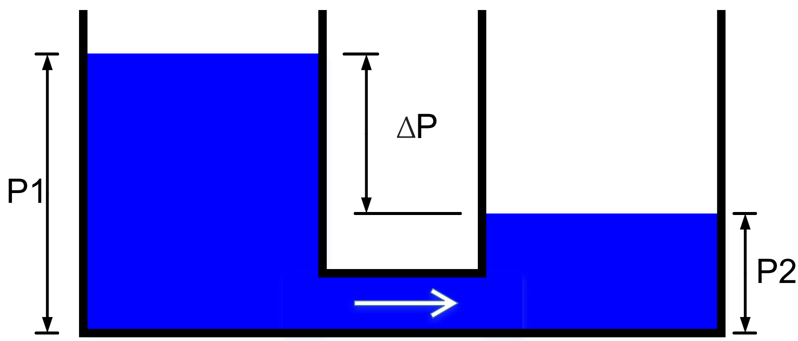

Figure 4. If two containers that are connected by a tube have different levels of pressure, there will be a flow of through the tube.

Ohm's law establishes the relationship between flow and the pressure necessary for it to circulate.

$$P_{res} = \Delta P = F.R$$

where Pres is the resistive pressure, F is the flow and R is the resistance.

Resistive pressure is the pressure difference between the two ends of the resistance. In the example of the containers, it will be the difference in pressure between the two containers.

Resistance is the relationship between flow and resistive pressure. This is how much you must pay, with pressure, for a flow unit to circulate. Resistive pressure is the total value of pressure that must be paid for the total flow that is passing. Resistance is the price, it is the pressure for one unit of flow.

In the same way that the price of an apple is how much we pay for each apple, the total spent on apples is equal to the amount multiplied by the price. What we pay for apples has the same form as Ohm's law:

Total amount to pay = Number of apples . price of an apple

If the apples cost $3 each and we buy 2 apples:

Total amount to pay = 2 apples . 3($/apple) = 6 $

This is totally equivalent to having a resistance of 3 cmH_2O for each Liter/second and we want a flow of 2 Liters/second.

Resistive Pressure = Flow . Restence

Resistive Pressure=2 L/s . 3(cmH_2O/L/s) =6 cmH_2O

Elastance or compliance

A spring is a one-dimensional elastic component (it moves along a line), and when stretched on the spring line, it generates a force to return to its state of equilibrium. This force will be proportional to how much it is stretched with respect to its point of equilibrium. The more you stretch it, the bigger the force.

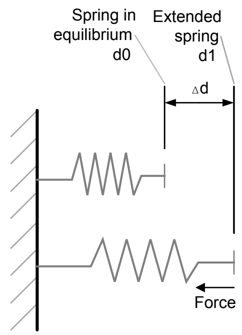

Figure 5. Force of a spring proportional to the distance that separates from the point of equilibrium.

In a three-dimensional elastic body, when a volume is introduced, this increase in volume will stretch the walls of the body and generate an increase in pressure within the body. The greater the volume, the greater the distention of the walls and the pressure generated inside.

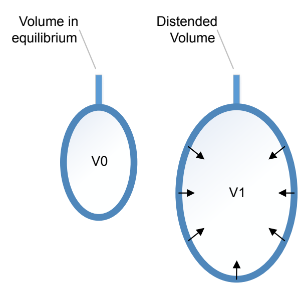

Figure 6. By increasing the volume in an elastic structure, the force made by the walls generates the elastic recoil pressure.

This shows that in an elastic body there is a relationship between the change in volume and the change in pressure. In respiratory physiology, this relationship is called elastance or compliance.

Elastance is how much the pressure of the elastic body increases per unit volume. The elastance is usually measured by how many centimeters of water increases if the body has its volume increased by one liter (cmH_2O/L). Compliance is the volume that increases per unit pressure, usually measured in milliliters of volume increase for each centimeter of water (ml/cmH_2O). For example, a compliance of 50 ml/cmH_2O means that if the pressure increases by 5 cmH_2O, the volume will increase to 250 ml.

$$C = \frac{\Delta V}{\Delta P}$$

$$E = \frac{\Delta P}{\Delta V}$$

The relationship between compliance and elastance is inverse:

$$C = \frac{1}{E}$$

$$E = \frac{1}{C}$$

For example, to convert a compliance of 50 ml/cmH_2O to elastance you should calculate E = 1/ (50 ml/cmH_2O) that is, 1 cmH_2O / 50 ml = 0.02 cmH_2O/ml. Converting milliliters to liters gives 20 cmH_2O/L

The colloquial use of the word elastic is associated with something that is easy to stretch. In physics (and physiology) an elastic body is a body that when deformed stores energy that is applied to it when it is deformed, and will return that energy when it regains its original shape. The elastic property is associated with the point that it is energy that is recovered. This makes an elastic body that stores a lot of energy with little deformation very elastic and something that with little energy deforms a lot to have a low elastance. This definition goes against the colloquial use of the word elastic. A hard elastic element is more elastic than a soft one.

Hooke's law establishes the relationship between the force generated by a spring in relation to the distance it is stretched from the point of equilibrium:

$$F =- k\Delta d$$

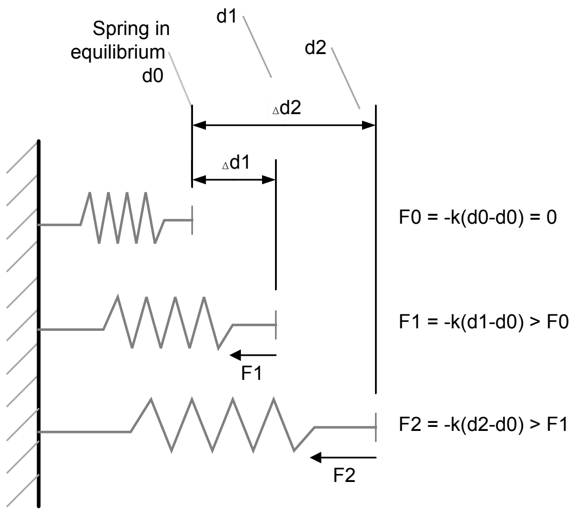

Figure 7. Example of Hooke's Law. A spring that is stretched further away form its equilibrium point will generate a bigger force.

Where F is the force, Δd is the distance that is stretched from the equilibrium point, and k is the elastic constant of the spring.

When instead of applying to a spring the force is applied to an elastic body that varies its volume, Hooke's Law takes the following form:

$$P_{elast} = E.V$$

where Pelast is the elastic recoil pressure, V is the volume variation, and E is the elastance of the body.

Knowing that compliance is the inverse of elastance, the pressure generated by the elastic component can be expressed in relation to compliance rather than elastance:

$$P_{elast} = \frac{V}{C}$$

Elastance and compliance are equivalent terms, explaining the relationship between elastic recoil pressure and the volume of the body. In the clinical field, we often talk about compliance and when discussing physiological models, elastance is often used.

We can see elastance as the price to pay in pressure to stretch the elastic body by one unit of volume. If we buy peaches, the price of peaches is the elastance, the volume added is the quantity of peaches and the elastic recoil pressure is the total amount of money paid to buy the peaches. If they are more expensive, we will pay more for the same number of peaches. In the same way, if a system has a higher elastance, more pressure will be needed to achieve the same volume. Or, if we lower the volume (we buy fewer peaches), then it will require less pressure.

Total amount to pay=8 peaches . 3($/peach) = 24$

Compliance is the inverse, it's another way of showing how much something costs. If the price is expressed as 4 peaches for each dollar. Now instead of multiplying by quantity (as we did with elastance) we must divide.

Total amoun to pay=8 (peaches/4 peaches/$) = 2$

The respiratory system.

Simple model.

One of the simplest models we can use to analyze the behavior of the respiratory system is a two-element model. It is a model made of a resistive component that is in the entrance of an elastic component. Where the resistive component will require a pressure difference for a flow to circulate and the elastic component will increase the pressure with volume that flows into it.

The interior of the elastic component represents the alveolar pressure, the pressure at the end of the resistive component is the airway pressure. The volume of the elastic element will be the volume of the lung; the variation of this volume will be the tidal volume (Vt) which is given by the flow of air in and out of the patient.

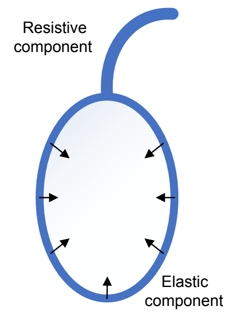

Figure 8. Two-element respiratory system model

This elastic body has its equilibrium point with a volume, the Functional Residual Capacity (FRC).

Analysis of a respiratory cycle.

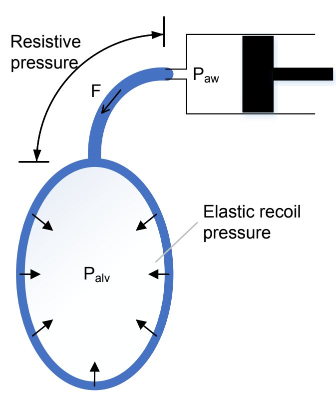

Figure 9. Resistive pressure and elastic pressure in the two-element model. Resistive pressure is the difference between airway pressure and alveolar press ure. The elastic recoil pressure is generated by the volume that distends the elastic component.

Starting from a system that is in equilibrium. When the ventilator triggers, increasing the pressure to a new level, it generates a pressure difference between the airway and the alveolus, this pressure gradient will generate a flow that will be limited by resistance. This flow, as time passes by, will turn into volume, increasing the lung volume.

As the volume increases, the elastic structures of the respiratory system are distended, and this causes the alveolar pressure to increase. If the alveolar pressure increases and the airway pressure is constant, the pressure difference between both sides of the resistive component is reduced, leading to a decrease in flow. This process leads to an exponential process of variation in flow, volume and alveolar pressure. If the new pressure level is maintained long enough, a new equilibrium situation will be reached.

When the ventilator cycles, it reduces the airway pressure. This means that the airway pressure is now lower than the alveolar pressure. There is pressure difference between both sides of the resistive component. A flow is generated that now circulates from the patient to the ventilator. During the beginning of the exhalation, the pressure difference is maximum, generating a maximum expiratory flow. As the flow is negative, the lung volume is reduced, that relaxes the elastic forces, reducing the alveolar pressure, which in turn reduces the pressure difference along the resistive component and reduces the flow. Generating an exponential curve in the flow and expiratory volume.

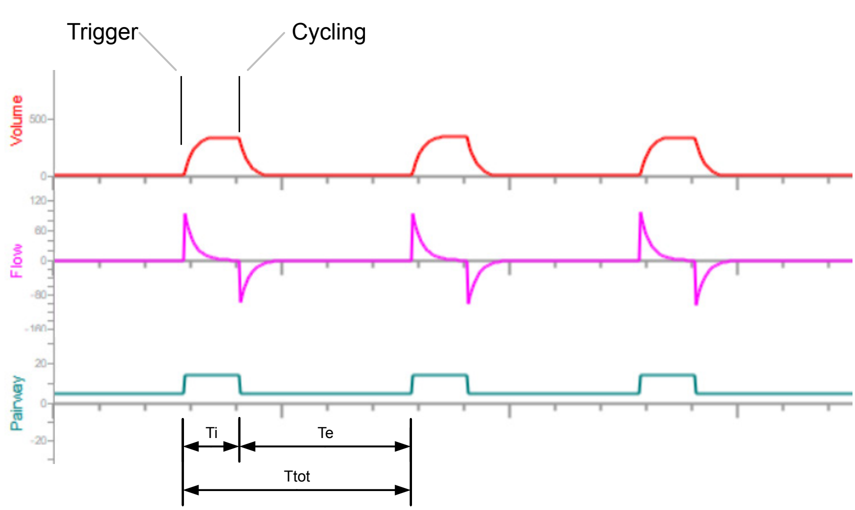

Figure 10. Inspiratory and expiratory phase. Trigger is the start of the pressurization and cycling is the switch from inspiration to expiration. This image was created based on FluxMed simulator using a two-element linear model.

In this model, the pressure generated by the ventilator is used to distend the elastic structures of the lung (elastic recoil pressure Pelast = E*V) and to overcome the resistive component (resistive pressure Pres=R*F). The variation in airway pressure is equal to the sum of the resistive pressure and the elastic recoil pressure.

$$\Delta P_{aw} = P_{res} + P_{elast}$$

$$\Delta P_{aw} = R.F + E.V$$

where R is the resistance of the respiratory system and E the elastance of the respiratory system.

Since the variation in pressure is over the positive end-expiratory pressure (PEEP). The equation can be completed as:

$$P_{aw} = R.F + E.V + PEEP$$

This model, although very simple, is extremely useful to explain and predict the behavior of respiratory mechanics. It is the model that is normally used as a first approximation when evaluating a patient.

Ventilator control variable

Given these relationships, what does the ventilator generate? Does it generate a flow? Pressure? A volume? Think before you go any further.

These are tricky questions since flow, volume and pressure are variables that are linked. If the ventilator generates flow, for the flow to circulate it must generate a pressure and if it generates a pressure, there will be flow, and the flow is the variation in volume, meaning that if there is flow there is volume. The trick is that if you generate one of the variables, you also generate the others.

The thing is that what the ventilator does is control the value of one variable. The value of the other will be a consequence of what is defined by physiology. The ventilator can determine one of the variables and the others will be a consequence of the patient's mechanics (the patient's resistance and elastance).

As mentioned above, flow is the derivative with respect to the time of the volume. In physics, the derivative of a variable with respect to time is usually expressed by placing a dot over the variable. This gives another way to write the above equation is as follows:

$$P_{aw} = R.\dot{V} + E.V$$

It is easier to see that flow and volume are the same variable and are related to pressure (Paw) by the resistance R and elastance E which are properties of the patient.

We can only define the value of a variable, the pressure or the volume. On the one hand, if we define the pressure values over time, the volume will be a consequence of the patient's mechanical properties (resistive and elastic component). On the other hand, if we define the volume values (we also define flow), the airway pressure values will be a consequence of the patient's mechanics.

This means that the ventilator can control only one, pressure or volume.

> We can go again to the example of buying apples. If there is a given price for the apples, I have two options. I can decide how many apples to buy, and the total amount to pay will be consequence of that decision. The other alternative is that I can decide how much I will spend in total, and then the number of apples I buy will be the consequence. If the price is already fixed, I can't decide how much to spend and the number of apples to buy.

Broadly speaking, ventilators have two ventilatory modes: by pressure if it controls pressure and by volume if it controls flow (which is the same as controlling volume). When the ventilator controls pressure, the pressure curve tells us about the ventilator, and in the flow and volume curves will be the patient's information (the elastic and resistive component). In the same way, when the ventilator controls the volume (or flow), the patient's information is on the pressure curve. There are more advanced modes where some clarifications should be made, but so far, we are looking at the basics.

Timing of a Respiratory Cycle

When the ventilator starts pressurizing the ventilation, the trigger can be due to time or because it detects a patient's effort. The trigger marks the beginning of pressurization. At the end of the inspiratory pressurization, the ventilator can perform an inspiratory pause forcing flow to be cero. Cycling is when the ventilator goes into the expiratory phase. In which it sets that the airway pressure to the value of PEEP (Positive End Expiratory Pressure). Cycling can be time-based or associated with the patient's mechanics.

The ventilatory mode is what defines the control variable, the trigger, pressurization and cycling criteria.

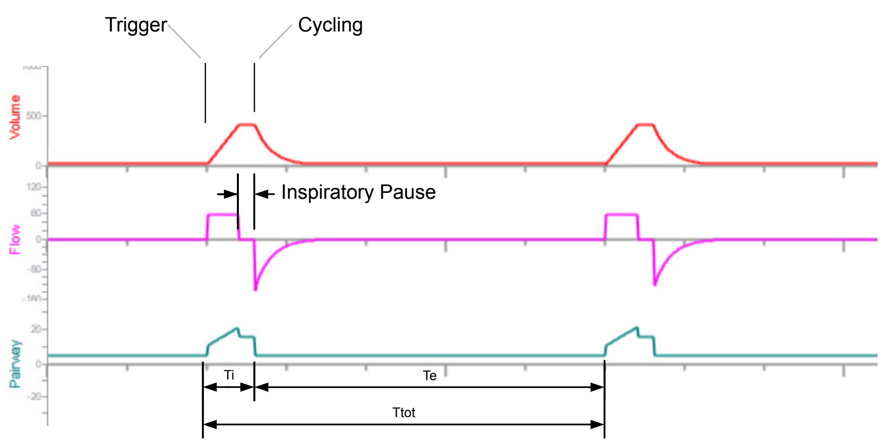

Figure 11. Inspiratory and expiratory phase. Trigger is the start of the pressurization and cycling is the switch from inspiration to expiration. The inspiratory pause is a zero flow period between the insufflation and the expiration This image was created based on FluxMed simulator using a two-element linear model.

The respiratory rate (RR) will specify the total time (Ttot) of the respiratory cycle. Rate is defined as the number of breaths per minute. For example, if the frequency is 10 breaths per minute, each cycle will last 6 seconds.

$$Ttot = \frac{60\frac{seconds}{minutes}}{10\frac{breaths}{minute}} = 6\frac{seconds}{breaths}$$

Inspiratory time is part of the ventilator configuration, depending on the way it is defined by each ventilator's brand and model. In some ventilators it is defined directly - there is an inspiratory time setting button. But in other it is indirect, for example, with the I:E relationship. The I:E ratio is the ratio of time you are breathing in and the one you are exhaling.

An I:E ratio of 1:3 means that expiration lasts three times longer than inspiration: a quarter of the time is used for inspiration in and three quarters of the time to exhale. Following the previous example, if the total time is 6 seconds, with an I:E ratio of 1:3 inspiration will last 1.5 seconds and expiration 4.5 seconds.

The objective of this text is to understand the relationship between the variables. It's important to know your tools and know how your ventilator works.

Pressure Control Ventilation (PCV)

In pressure-controlled modes, the ventilator will provide whatever flow is necessary to maintain the pressure that is set as a target. Therefore, the flow (and volume) will depend on the mechanical characteristics of the patient's respiratory system. When setting up the ventilator a pressure level for the inspiration, a pressure level for the expiration (PEEP), inspiratory time and respiratory rate are established. These variables may be defined indirectly.

During inspiration, the ventilator's control system's target will be to apply the pressure established as inspiratory pressure. During expiration, the goal will be to apply PEEP.

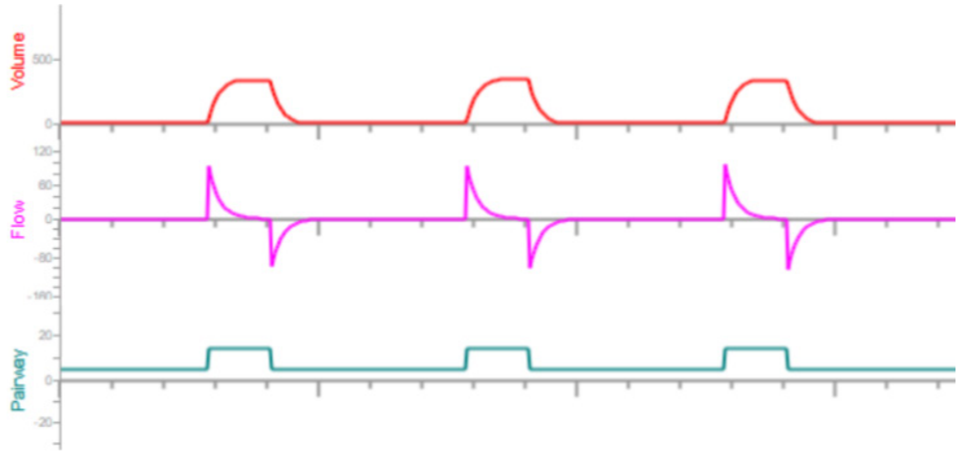

Figure 12. Pressure Controlled Ventilation. During the entire cycle the ventilator makes its best effort to keep the pressure at the configured value. This image was created based on FluxMed simulator using a two-element linear model.

If the patient is ventilated with a pressure of 8 cmH_2O during the inspiratory phase and 0 cmH_2O PEEP (it is rare for PEEP to be zero, but it is chosen to keep the example simple), with a frequency of 10 breaths per minute and an inspiratory time of 1.5 seconds, and starting from the equilibrium point at 0 cmH_2O in the airway, there is no flow in the system, so the alveolar pressure is 0 cmH_2O. When the ventilator triggers the inspiratory pressurization, it takes the airway pressure to the inspiratory pressure of 8 cmH_2O in this example. Having a higher pressure in the patient's airway than in the alveolus generates an inspiratory flow. The flow value will depend on the resistance of the respiratory system, but it will be a positive flow as it is air going into the patient.

If there is a positive flow, over time this flow becomes a volume that enters the respiratory system. This volume distends the elastic structures of the respiratory system, generating an elastic recoil pressure. This is an increase in alveolar pressure. By increasing the alveolar pressure, the pressure difference between the airway and the alveolus is reduced, and this reduces the flow. This leads to the volume increasing more slowly, the alveolar pressure will increase more slowly reducing the flow even more in one cycle. When the elastic recoil pressure is equal to the pressure applied by the ventilator, the alveolar pressure is equal to that of the airway. If the pressure level is maintained for long enough, the flow becomes equal to zero.

The flow will depend on the patient's resistance, and the total added volume (if enough time is provided) will depend on the patient's compliance. Flow and volume will have an exponential curve. The total time it takes for the flow to reach zero depends only on the patient's time-constant. The time-constant is calculated as:

$$\tau = R.C$$

Where τ is the time-constant, R the patient's Resistance, and C the patient's Compliance. It is reasonable to consider that 3 time-constants are necessary for flow to become zero.

A similar process occurs when the ventilator cycles, it decreases the pressure to the level of PEEP configured. As the alveolar pressure is greater than the pressure of the airway, an expiratory airflow is generated. This flow will depend on the pressure difference and resistance. As the volume of the respiratory system decreases, the distention of the elastic structures is reduced, decreasing the alveolar pressure. This causes the flow to decrease as the volume is reduced, reaching zero flow when the alveolar pressure becomes equal to the airway pressure.

If the ventilatory mode is pressure-controlled, the volume will depend on the patient's mechanics. If the patient's mechanics (resistance or elastance) change, the speed and amount of volume that enters and leaves the patient will change. When we ventilate the patient where pressure is controlled, we must be aware of the volume that is delivered.

In the expiration the same exponential process is present and will take 3 time-constants to become zero. In inspiration it is not necessary that flow reaches zero but it is very important than, in expiration, flow does reach zero.

Volume Control Ventilation (VCV)

With the advancement of technology, ventilators integrated more advanced control systems into their equipment. This allowed the flow signal to be used as a control variable, and therefore to control the volume provided.

To control the flow signal, the ventilator will generate the pressure needed to maintain the desired flow. Therefore, the pressure in the airway will depend on the mechanical characteristics of the patient.

The ventilator operator must set up what the flow signal will be like during inspiration. Depending on the manufacturer of the ventilator, how this signal is defined may vary. It can be defined for example with the volume to be delivered, the flow to which it is to be delivered and the duration of the pause.
Another possibility is with the volume to be delivered, the respiratory rate and the ratio of the duration of inspiration and expiration (I:E ratio).

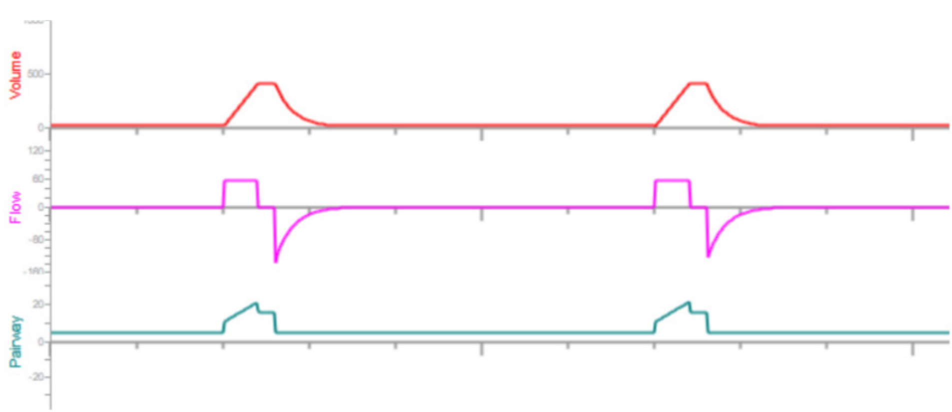

Figure 13. Volume Controlled Ventilation. The shape of the inspiratory flow is defined by the volume and insufflation time, the ventilator will apply whatever pressure is required to generate the target flow. During expiration the ventilator will control pressure as it will make airway pressure equal to PEEP. This image is based on the FluxMed simulator using a two-element linear model.

When the ventilator starts the pressurization, it applies the pressure necessary for the inspiratory flow to be equal to the programmed one. Flow will become volume as time goes by, and thus, increase the volume within the respiratory system that distends the structures - increasing the alveolar pressure. If the alveolar pressure increases, the flow will tend to decrease, so the ventilator must increase the pressure in the airway to sustain the desired flow. If an inspiratory pause is configured, once the programmed volume is delivered the ventilator will generate the inspiratory pause making the flow zero, it will not deliver volume or allow the volume to come out.

The ventilator cycles and switches to control pressure, setting PEEP as the objective goal. As the alveolar pressure is greater than that of the airway, an expiratory flow is generated in the same way as in the PCV mode. During expiration, ventilators always controlled pressure. The control algorithm aims to maintain the PEEP level, this is a pressure control method.

In volume-controlled ventilation, during inspiration the ventilator controls flow (volume), so the pressure waveform will be consequence of the patient's mechanics. During expiration, the ventilator controls pressure so the flow (and volume) curve will be consequence of the patient's mechanics.

PEEP and intrinsic PEEP (PEEPi or AutoPEEP)

When the system is at rest, in equilibrium, without the patient exerting force and without the ventilator applying pressure, the system has a volume. This volume is called functional residual capacity (FRC). When an external PEEP is applied, the end-expiratory point has a volume greater than the FRC. At the end of expiration, if the flow becomes zero, the alveolar pressure becomes equal to PEEP. This implies that the elastic recoil pressure of the respiratory system is equal to PEEP, that is, because of the increased volume compared to FRC.

In the previous example, a PEEP of 0 cmH_2O was used, so that at the end of expiration, if the flow becomes zero, the lung volume is equal to the functional residual capacity (FRC) of the patient. In many cases, it may be desirable to have a positive PEEP. As the airway pressure is greater than the atmospheric pressure, the equilibrium point will be with an increased lung volume with respect to the FRC. The volume at end of expiration is an interesting factor to keep in mind. It is sometimes referenced as the *End Expiratory Lung Volume (EELV)*.

This becomes easier to understand if we consider that when airway pressure increases from 0 cmH_2O to a given PEEP level, this change produces a pressure variation similar to the one occurring during inspiration in a pressure-controlled mode. The volume entering the patient corresponds to the difference between FRC and EELV. Thus, the higher the PEEP, the greater the lung volume at end-expiration.(1–3).

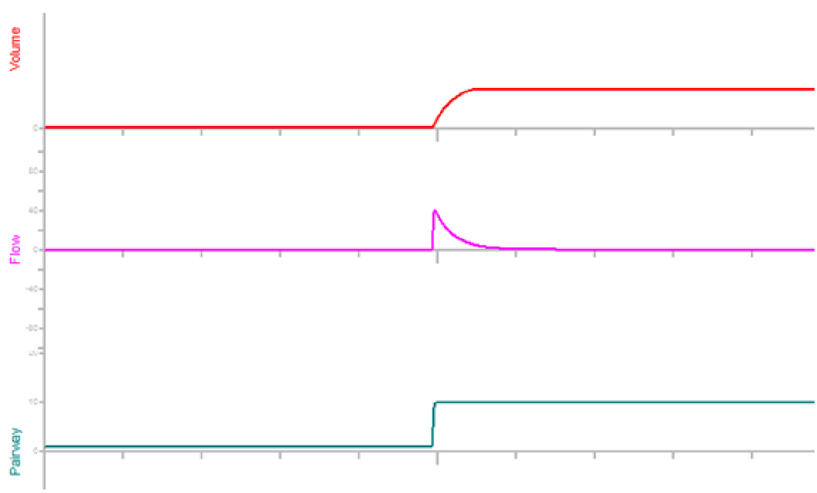

Figure 14. The variation of PEEP is a change in pressure, this generates a variation in volume. This image was created based on FluxMed simulator using a two-element linear model.

Making a change in end-expiratory pressure will affect the volume of the respiratory system at the end-expiration. As volume is measured from the end of expiration, there is no easy way to measure the volume at the end of expiration.

The equation of motion when written without the term PEEP the volume V is measured from the equilibrium point from the relaxed system.

$$P_{aw} = R.F + E.V$$

As normally the volume is measured from the end of the previous expiration, it is a different volume. An easy way to correct the difference is to add the PEEP term to the equation of motion.

$$P_{aw} = R.F + E.V + PEEP$$

Adding PEEP is a way to change the reference point from where volume is measured. When the PEEP term is present volume is measured from the end of expiration.

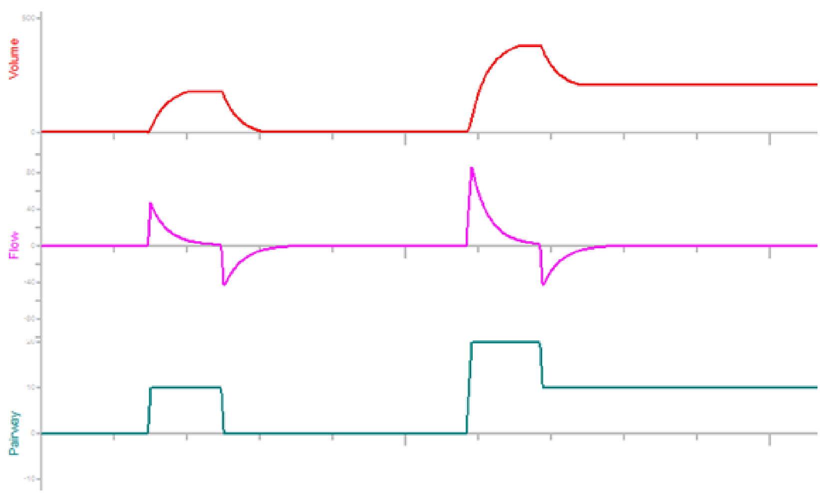

Figure 15. A change of PEEP will generate additional volume in the elastic component. In the breath on the right the inspiratory volume is bigger than the expiratory volume. This image was created based on FluxMed simulator using a two-element linear model.

If the flow becomes zero, it is because the pressure of the two compartments (airway and alveolus) is equal. If they were different, there would be flow. If at expiration, the ventilator triggers and increases the pressure before the flow reaches zero, the alveolar pressure is higher than PEEP, as the flow is going from the alveolus to the airway. This is evidence that some volume is trapped. When an expiratory pause is generated, the ventilator, instead of pressurizing the next breath, closes the valves and forces a zero flow. As flow is zero, airway pressure will equal with alveolar pressure. If the pressure in the expiratory pause increases, it is because there is an intrinsic PEEP (PEEPi) or AutoPEEP. It is important to keep this phenomenon in mind, because if the flow does not reach zero in expiration, the respiratory system has more pressure than what is being seen in PEEP.

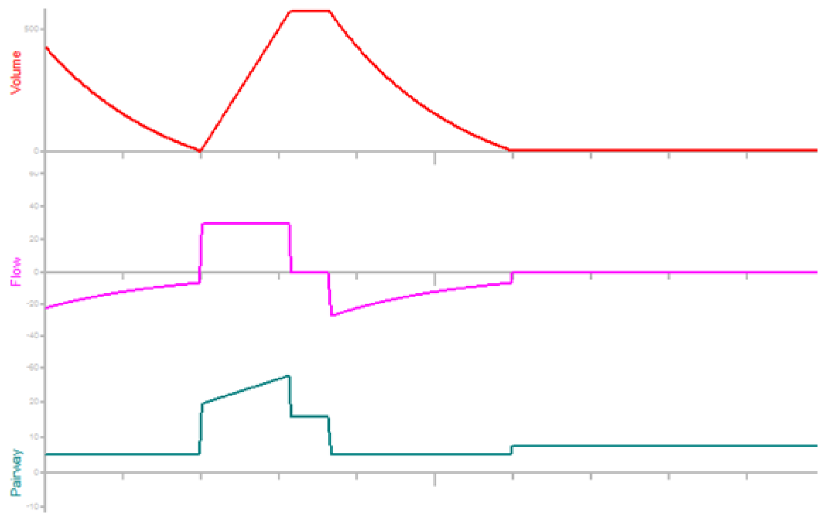

Figure 16. If expiratory flow does not reach zero. There is more pressure in the alveolus than on the airway (the air is flowing out). This is AutoPEEP or intrinsic PEEP (PEEPi). During an expiratory pause, it can be measured as the increases of pressure over PEEP. This image was created based on FluxMed simulator using a two-element linear model.

PCV and VCV from the equation of motion.

Analyzing these controlled modes in detail from the linear two-element model, we know that the total pressure applied by the ventilator is equal to the elastic recoil pressure (Pel = E*V) plus the resistive pressure (Pres = R*F) plus the PEEP.

$$P_{aw} = P_{res} + P_{elast} + PEEP = R.F + E.V + PEEP$$

This equation explains both pressure and volume-controlled ventilation very well.

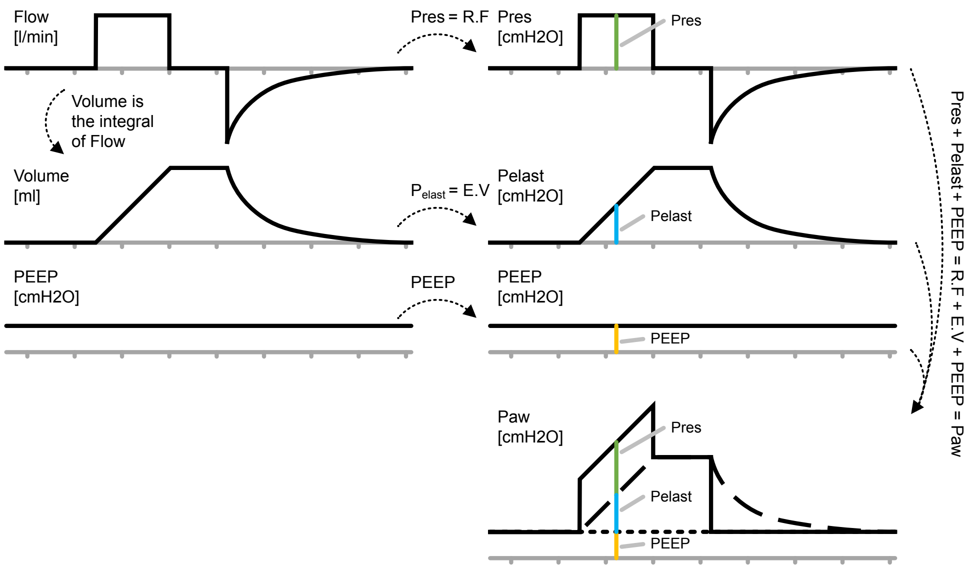

Figure 17. Flow times Resistance (R) is the Resistive Pressure (Pres) Volume is the integral of Flow. Flow times Elastance (E) is the Elastic Recoil Pressure (Pelast). PEEP is pressure. The sum of the Resistive Pressure, the Elastic Recoil Pressure and PEEP is equal to the Airway Pressure (Paw). The morphology of the Airway Pressure waveform can be explained from the equation of motion.

The three terms of the equation of motion are: one associated with flow passing through resistance, another related to the volume which distends structures during insufflation and the starting pressure (PEEP).

Figure 17 shows the flow curves, volume curves and PEEP level on the left. Multiplying the flow curve (on the top left) by the patient's resistance will generate the resistive pressure waveform (top right). This is the pressure required for the flow to circulate through the resistive component. Multiplying the volume curve (in the second row on the left) by the patient's elastance, yields the elastic recoil pressure curve (second row to the right). This is the pressure that must be generated for that volume to enter the elastic component. PEEP (in the third row) is already pressure, so it can be placed on the right column.

If we add the three pressures waveforms from the right: resistive pressure, elastic pressure and PEEP, we get the curve from the bottom right. This is the airway pressure. These curves are a graphical representation of the equation of motion.

To perform these calculations in this way, it is necessary to know the patient's resistance and elastance (or compliance). We don't know the patient's resistance and elastance (or compliance), but we can do the calculation in reverse and by analyzing the airway pressure curve, which we do know, it is possible to calculate the patient's resistance and elastance (or compliance).

Measurement of respiratory mechanics parameters.

Flow and volume are related by time. Since the volume curve is the integral of the flow, and we cannot change the passage of time, when defining flow or volume the other is defined.

From the equation

$$P_{aw} = R.F + E.V + PEEP$$

It's the same as

$$P_{aw} = R.F + \frac{V}{C} + PEEP$$

The mechanical characteristics of the patient are given by the resistance R and the compliance C (or elastance E). Since PEEP is set, airway pressure and volume remain to be defined. As airway pressure and volume are related by the equation, only one can be chosen.

If the ventilator controls the volume curve by defining how it wants the volume to be delivered to the patient, the pressure curve will be defined by the patient's mechanics. This gives us the guideline that if the patient is ventilated by volume in the pressure waveform will have the information of the patient's R and C.

This is equivalent when the ventilator controls the pressure curve. Then the volume (and flow) waveform will be defined by the patient's mechanics. In pressure-controlled ventilation, in the volume curve (and the flow curve) it will have information of the patient's R and C.

If we look at the resistive pressure behavior in VCV, we can see that during inspiration the resistive pressure is a rectangle. The height of the rectangle is the resistive pressure necessary to make the inspiratory flow go through the patient's resistance.

In the airway pressure we can see that the resistive pressure can be measured as the difference between the Peak Inspiratory Pressure (PIP) and the Plateau Pressure (Pplat). Ohm's law relates resistive pressure to resistance and flow:

$$P_{res} = R.F$$

As we know the resistive pressure and the flow, it is possible to calculate the resistance of the patient's airway:

$$R = \frac{P_{res}}{F} = \frac{PIP - Pplat}{F}$$

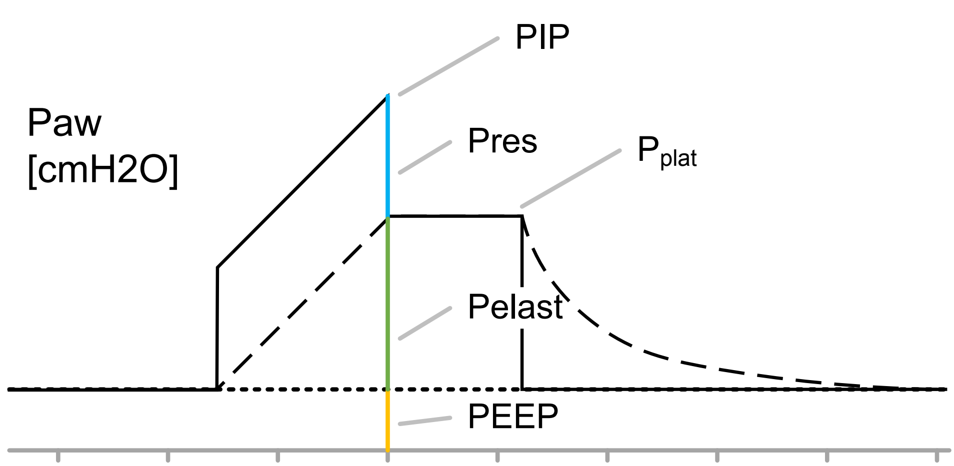

Figure 18. Resistive pressure is the difference between the Peak Inspiratory Pressure (PIP) and the Plateau Pressure (Pplat).

Spontaneous breathing.

The respiratory muscles act on the elastic body, changing the volume and therefore the pressure within the respiratory system. The inspiratory muscles decrease the pressure while the expiratory muscles increase it.

In a person breathing without any assistance, without the ventilator, and starting from the equilibrium point where the alveolar pressure is equal to the atmospheric pressure, since the two pressures are equal, the flow is zero. The inspiratory muscles generate a force that throws the elastic body out of equilibrium, increasing the volume, which leads to a drop in alveolar pressure. As the alveolar pressure is lower than the atmospheric pressure, an inspiratory air flow is produced. Given the pressure difference, the magnitude of the flow will depend on the resistance of the airway.

By relaxing the inspiratory muscles, the elastic force compresses the air inside and increases the alveolar pressure. As the alveolar pressure is greater than the atmospheric pressure, an expiratory air flow is generated.

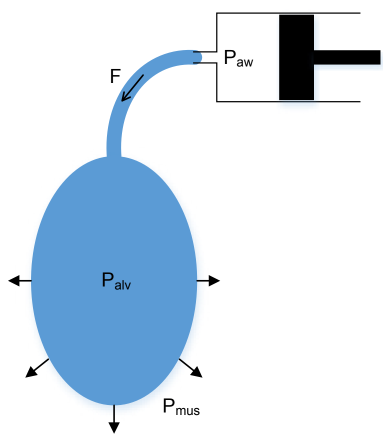

Figure 20. In a two element model the pressure generated by the respiratory muscles reduces the alveolar pressure but it can't be measured.

In spontaneous breathing we usually breathe using only inspiratory muscles. During the inhalation with the muscles, we deliver energy to remove the elastic component from equilibrium and to generate the inspiratory flow through the resistive component. The energy stored in the elastic component is released during expiration and is used to generate the expiratory flow through the resistance.

We can use the expiratory muscles to increase flow and reduce expiratory time. Either because we have an increased ventilatory demand and we need to increase the flow and reduce the expiration time. Or, simply, to blow out a candle on a birthday cake.

The most basic measurement of respiratory monitoring is performed only in the airway, measuring airway pressure and flow. By integrating flow, volume is obtained. In the case of a patient breathing spontaneously, the pressure in the airway will always be atmospheric and therefore the pressure is zero.

Since it is not possible to measure the pressure inside the model, in this model we cannot quantify the pressure produced by the respiratory muscles during spontaneous ventilation. In the equation we have been using so far if airway pressure is always zero flow and volume must be zero. Our model has a limitation.

There is a technique used to assess the strength of the respiratory muscles that involves occluding the airway and asking the subject to perform an inspiratory or expiratory effort. Since there is no flow, there is no pressure difference between the alveoli and the airway, thus allowing the pressure generated by the respiratory muscles to be observed in the airway pressure.

Chapter 2.

Lung and chest wall model

Model of the 3-element respiratory system.

We know from our anatomical knowledge that the respiratory system is made up of the chest wall and the lung, where the lung only has elastic behavior while the chest wall has both elastic and active (the respiratory muscles) behavior.

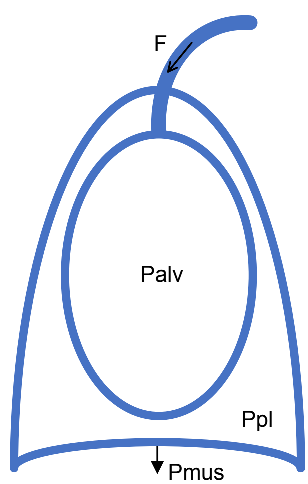

Figure 21. Three-element respiratory system model. A resistive component and two elastic components. All the resistance of the respiratory system is in the resistive component; an elastic component representing the lung is inside the chest wall that has an elastic behavior and can generate muscular pressure.

A model of a respiratory system formed by an airway with a resistive component, a lung with an elastic component that is inside a chest wall that has an elastic component and an active component, allows quantifying the pressure generated by the respiratory muscles and separating the pressure applied to the lung and that applied to the chest wall(10,11).

It should be noted that the lung parenchyma is a much more fragile tissue than the chest wall. It is in the interest of keeping the pressure applied to the lung limited to avoid injury to the lung. The chest wall can withstand greater pressure without being a risk to the patient.

Developing the equation of motion of this model, we have:

$$P_{aw} + P_{mus} = R.F + E_{lung}.V + E_{cw}.V$$

Where R is the resistance of the airway, F the flow, Elung is the elastance of the lung, Ecw is the elastance of the chest wall, V is the variation in volume and Pmus is the magnitude of the pressure generated by the muscles in the chest wall.

The relation between lung and chest wall elastance with respiratory system elastance is defined by the fact that the volume distends both structures at the same time. The lung is obviously within the chest wall.

$$E_{lung}.V + E_{cw}.V = \left(E_{lung} + E_{cw}\right).V = E_{rs}.V$$

The elastance of the respiratory system is equal to the sum of the elastance of the lung and the elastance of the chest wall. Compliance is the inverse of elastance.

$$C_{lung} = \frac{1}{E_{lung}}$$

$$C_{cw} = \frac{1}{E_{cw}}$$

In controlled ventilation, Pmus is zero and the only source of pressure is the ventilator. In the same way that with two elements we can see what the elastic component of the lung and the chest wall have a similar behavior.

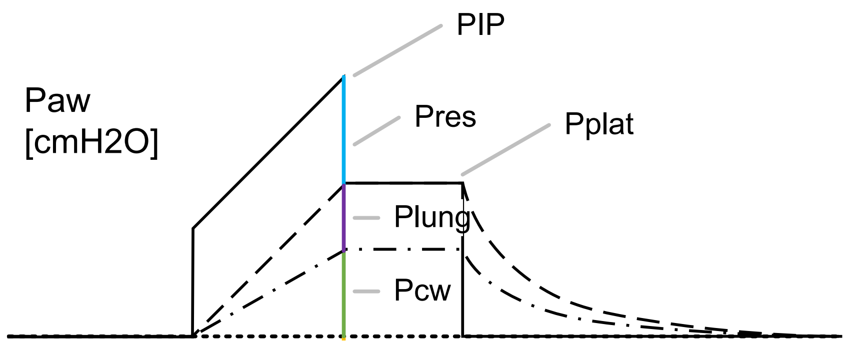

Figure 22. The three-element model can explain the independent elastic behavior of the patient's lung and chest wall. The total elastic pressure is the same.

In this model, the pressure outside the lung and inside the chest wall is the pleural pressure. Since pressures are measured with respect to atmospheric pressure, which is the pressure on the surface of the body, pleural pressure will be the pressure of the chest wall. The chest wall is formed of an elastic component and, at the same time, an active component, which is the respiratory muscles. The pressure resulting from the sum of the elastic recoil of the chest wall and that generated by the muscles will be the pressure of the pleural space. As the chest wall tends to open and the lung tends to collapse, a point of equilibrium is reached.

The pressure difference between the pleura and the alveolus is the pressure that drops through the lung parenchyma, the transpulmonary pressure (Plung). As the lung parenchyma has only one elastic component, there is no active component in the lung. Transpulmonary pressure is the result of the elastic recoil pressure of the lung.

It is not possible to measure alveolar pressure directly but knowing that alveolar pressure is equal to airway pressure when the flow is zero. Transpulmonary pressure can be measured as the difference between airway pressure and pleural pressure.

$$P_{lung} = P_{aw} - P_{pleura}$$

This is true only when the resistive pressure is zero, which is true when the flow is zero.

Knowing the volume variation and the pressure variation of the transpulmonary pressure can calculate the compliance (or elastance) of the lung.

$$C_{lung} = \frac{\Delta V}{\Delta P_{lung}} = \frac{VT}{P_{lung_{exp}} - P_{lung_{insp}}}$$

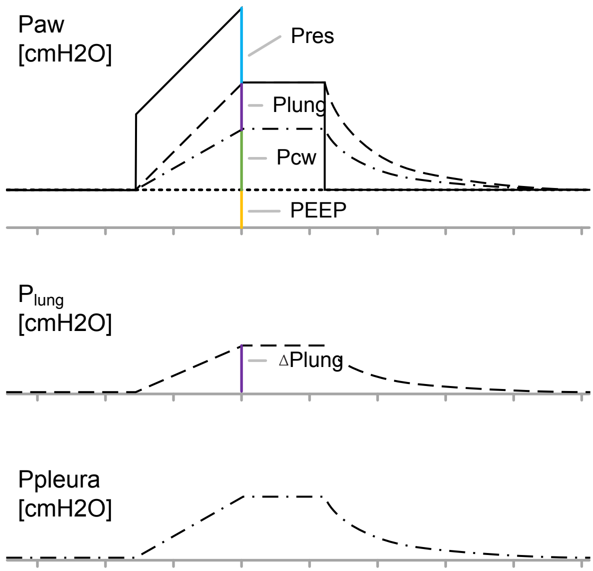

Figure 23. By measuring Pleural Pressure (esophageal pressure) it is possible to assess the transpulmonary pressure variation, and the chest wall pressure variation.

As the chest wall has an elastic component and an active component, to measure the compliance of the chest wall it is necessary that the muscles are not generating any pressure. If this does not happen, it is not possible to discriminate whether the pressure is a consequence of the elastic component of muscle activity. If the patient is not making any effort, the pressure variation will be a consequence of the elastic component.

$$C_{cw} = \frac{\Delta V}{\Delta P_{pleura}} = \frac{VT}{P_{pleura_{espi}} - P_{pleura_{inspi}}}$$

It is not possible to measure the elastic component of the chest wall when the patient is generating some muscular pressure. However, there are methods to estimate the value of the compliance of the chest wall from the subject's anthropometric data.

Taking chest wall compliance as 4% of inspiratory vital capacity for each cmH_2O. Inspiratory vital capacity is calculated from age (a in years), height (h in cm), and sex as follows(12,13):

Adults (at > 18)		
	Male:	$IVC = 6.10h - 0.028a - 4.65$
	Female:	$IVC = 4.66h - 0.026a - 3.28$
Teenagers (at < 18 years old and h > 150 cm)		
	Male:	$IVC = 8.4h - 9.9$
	Female:	$IVC = 5.0h - 4.5$
Children: (at < 18 years old and h < 150 cm)		
	Male:	$IVC = 5.70h - 5.26$
	Female:	$IVC = 5.50h - 5.39$

Esophageal pressure

The pleura is the space between the lung and the chest wall. The lungs are not attached to the chest wall, but because there is no fluid or air in the pleural space, the lung and the chest wall are next to each other. On a lung ultrasound, you can see how the lung parenchyma slides with respect to the chest wall.

Directly measuring pressure in the pleural space is not trivial. On the one hand, if the pleura is perforated and air enters the pleura, it will generate a pneumothorax, which is not recommended for the patient. On the other hand, if a chest tube or a measuring balloon is introduced, it must be considered that the presence of a foreign body into this narrow space could affect the measurement.
Part of the esophagus is near the pleural space. In 1949, Buytendijk was the first to show that esophageal pressure could be used as a surrogate for pleural pressure. Followed by Dornhorst and Leathart et al. and then by Cherniak et al.(14–16).

There is no doubt that variations in esophageal pressure reflect variations in pleural pressure. Although there are factors that can affect the absolute value, it is possible to use esophageal pressure as a surrogate for pleural pressure(17–21).

Esophageal pressure can be measured with an air- or fluid-filled catheter or with a pressure transducer placed directly into the esophagus. The most frequent is the use of a probe with an air balloon. Depending on the properties of the balloon it will be inflated with different volumes, it is important that you use the value suggested by the manufacturer of the probe you are using, as an error in inflation can induce an error in the measured value.

Placement of the esophageal pressure probe

To generate a quality esophageal pressure record, it is important to place it in the correct location and inflate it with the proper volume. The procedure is similar to the procedure of placing a feeding tube.

The objective is to place the balloon in the middle or lower third of the esophagus and inflate it so that it has an adequate transmission of information from the pleural space to the probe and to the device where the measurement is made (22).

The steps to correctly place the catheter are:

1. Prepare the materials: Esophageal catheter, 5 ml syringe, three-way stopcock, extension and measuring device. FluxMed pressure probes include the three-way stopcock and extension that are suitable for information transmission.

2. Insert the catheter and connect it to the monitor. The positioning in the middle or lower third is usually between 33 and 40cm from the nostril.

    a. One possibility is to place the probe in the stomach and remove it in steps of 2 or 3 cm verifying its position.

    b. Positioning is easier if the patient is in a semi-seated position.

    c. Pay attention to whether the patient coughs or if there are leaks outside the endotracheal tube because this may indicate that the probe is in the trachea.

    d. Connect the probe to the three-way stopcock and the extension and the extension to the equipment.

e. FluxMed probes have a Y-piece so that you can measure without removing the stylet.

3. Balloon probe insufflation

a. Connect the syringe to the three-way stopcock and remove the air from the probe. Pull the syringe until you feel some resistance because no more air is coming out.

b. Disconnect the syringe and place 5ml in it. Connect the syringe to the three-way stopcock and inflate the 5ml into the probe, then remove 3.5ml so that the probe is left with 1.5ml of air. These values are for the FluxMed BA-A-008 probe. It is important that you follow the instructions of the manufacturer of the catheter you use.

4. Check for heart oscillations in the esophageal pressure tracing.

a. If the values of the esophageal pressure tracing are equal to those of the airway when the flow is zero. The probe is likely placed in the airway. Deflate the catheter, remove it, and put it back in.

5. Confirmation of the position of the probe using the occlusion test.

a. Make an expiratory pause on the ventilator.

b. In a patient who does not make an inspiratory effort, perform a bilateral compression on the patient's chest.

c. In a patient who makes inspiratory efforts, wait for the patient to make an inspiratory effort.

d. Measure the variation in esophageal pressure ΔPes and airway pressure variation ΔPaw. Verify that the ratio ΔPes/ΔPaw is between 0.8 and 1.2. During an occlusion (zero flow) the airway pressure variation must be equal to the esophageal pressure variation. That is why when the ratio ΔPes/ΔPaw is close to 1 we can validate the

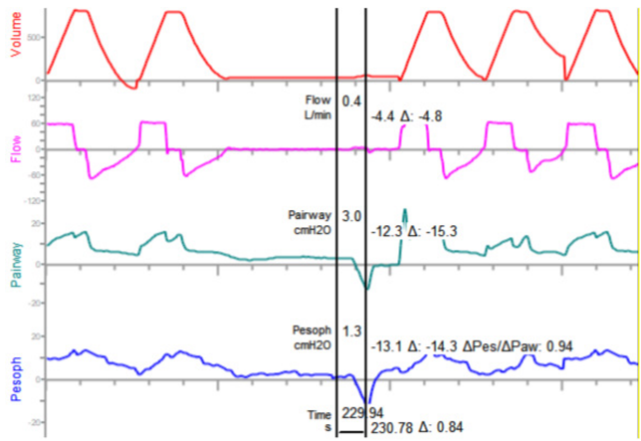

Figure 24. Patient's inspiration during an Occlusion Test. Airway pressure drop is -15.3 cmH2O and esophageal pressure drop is -14.3 cmH2O. ΔPes/ ΔPaw ratio is 0.94 showing that the esophageal pressure is placed and working correctly. This image was created based on FluxMed monitor acquisition software FluxView.

e. If the value is out of range, check the location and/or inflated until the value is correct.

f. If the probe was initially placed in the stomach, the patient's inspiratory efforts will generate a positive deflection of pressure in the probe. When the probe is slowly removed, the pressure increases as it passes through the esophageal sphincter and then falls when it reaches the esophagus.

g. An alternative to check the location of the probe is by looking at radiopaque marks on an X-ray.

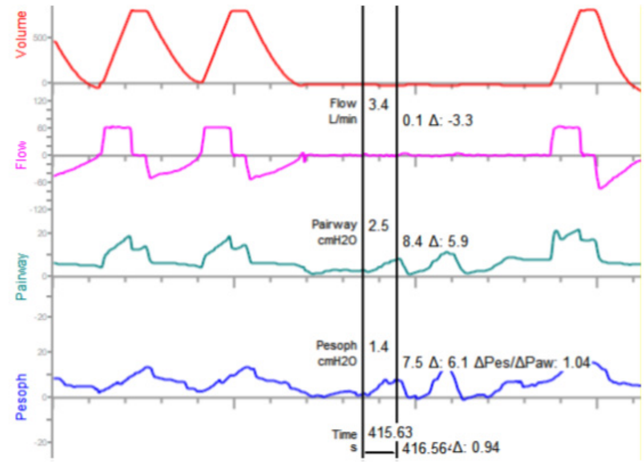

Figure 25. Chest compression during an Occlusion Test. Airway pressure increases by 5.9 cmH2O and esophageal pressure increases by 6.1 cmH2O. ΔPes/ΔPaw ratio is 1.04 showing that the esophageal pressure is placed and working correctly. This image was created based on FluxMed monitor acquisition software FluxView.

6. Secure the probe to prevent it from moving. Take note of the distance at which the catheter was placed for reference in case it moves.

    a. Periodically repeat the test to verify the accuracy of the measurement and, if necessary, deflate and reinflate the catheter or accommodate its location.

    b. The probe without the stylet is more comfortable for the patient and has a slight improvement in the quality of information transmission.

Technical aspects of an esophageal manometry probe

An esophageal manometry probe is a measurement tool. It is important that the same tool does not modify the value to be measured. Understanding the technical aspects of manometry catheters allows us to understand how the measurement can be affected. The catheter is an instrument that transmits pressure from the measurement site (the esophagus) to the equipment.

At the measuring site it has multiple perforations to take the pressure to be transmitted. It must be inflated so that the pressure can be transmitted from the inside of the balloon to the inside of the tubing. The volume range that allows adequate pressure transmission is the working volume of the catheter. On the one hand, if you have too little air, there will not be proper transmission of pressure from the outside of the probe to the inside. On the other hand, if you inflate too much, the walls of the balloon will be tightened, generating pressure instead of transmitting pressure from outside.

The catheter is evaluated based on three characteristics. The working volume, which is the range of inflation volumes in which the probe correctly transmits the external pressure of the balloon. It is intended to be very wide so that an error in inflation has a low error in the measurement. Hysteresis, which is the pressure difference generated by the catheter when it is inflated and deflated. At a given volume, the pressure generated by the catheter must be the same regardless of how the volume is reached by inflating or deflating. The dynamic response, which is the evaluation of the catheter's ability to follow a change in pressure. The goal is for it to be faster than the physiological changes of pressure to make sure it can follow them.

To measure the working volume, the probe is placed in a way that the balloon is in the air without touching anything. With a three-way stopcock it is connected to a syringe and a pressure gauge. The pressure gauge can be the same monitor placing a cursor that allows the value to be measured at any given time.

With the syringe, all the air is removed from the probe, for example, reaching a pressure of -10 cmH_2O. This is the pressure that corresponds to a zero volume of inflation volume. The pressure is then measured at different inflation volumes. For example, if volume is added in steps of 0.5 ml. Wait for the measurement to stabilize between the insufflation and the measurement. Take as many steps as necessary to reach a pressure of +20 cmH_2O.

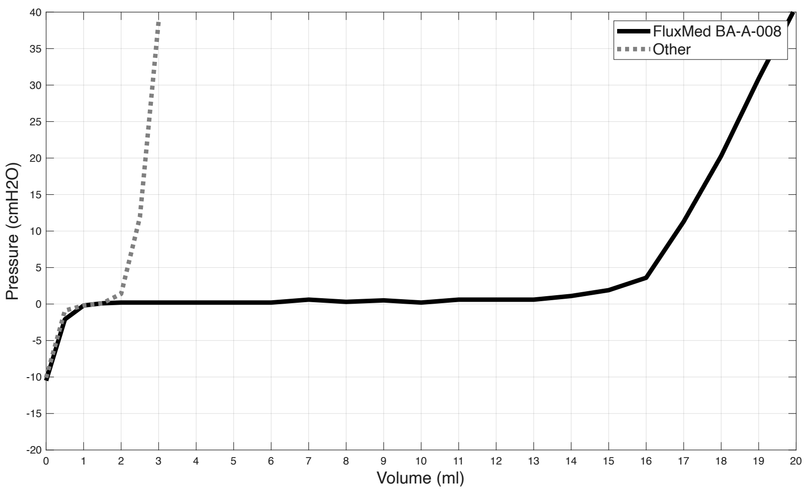

Figure 26. Pressure Volume curve of two pressure probes. Considering zero volume the volume necessary to generate a -10 cmH2O pressure volume is added in steps of 0.5 or 1 ml until a pressure of 40 cmH2O is reached. The working volume is the volume at which the pressure is equal to the pressure outside the prob (cero in this case).

This volume pressure graph shows the volumes at which the probe reflects the pressure that is outside the probe. If the balloon is inflated in a range where the pressure is equal to zero on the graph, it will reflect the pressure on the outside of the probe. If a volume that is below the minimum or greater than the maximum is used, the walls of the balloon will change the measured value. This test can be repeated by placing the catheter in a pressurized container at different pressure levels, the static response of the catheter at different pressures could be seen.

The materials that make up the catheter may have a viscoelastic response. This causes the balloons to generate different pressures at the same volume if the volume is reached by inflating or deflating. This can be seen as the hysteresis of the catheter when it is inflated and deflated. It is desirable that the catheter does not have hysteresis so that there are no more sources of error in the measurement.

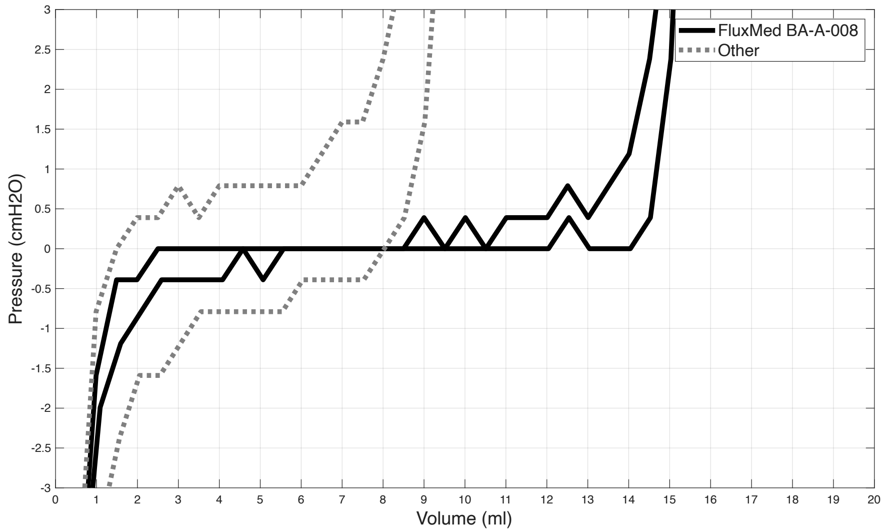

Figure 27. Pressure Volume curve of two pressure probes for hysteresis analysis. The inflation and deflation curves should overlap. At any given inflation value the difference between inflation and deflation will be the error introduced in the measurement. The error will depend on the way how the filling volume was reached.

Another important technical aspect is the dynamic response, i.e. the speed at which pressure changes are transmitted. A slow catheter will take longer to reflect a change in pressure, while a fast catheter will show it immediately. Slow or fast is related to the application. In the case of esophageal pressure, Milic-Emili showed that it must be able to measure a bandwidth of up to 15 Hz (24). The bandwidth of the probe can be observed by the time it takes to reflect 95% of the pressure change. To meet the 15 Hz bandwidth, a probe must reflect 95% of the change in at least 32 milliseconds.

A simple way to measure this is by placing the esophageal probe inside a balloon, inflating it, and popping it. When popped, the measured pressure will drop immediately, the time it takes for the esophageal probe to reflect this change will show the dynamic response of the probe.

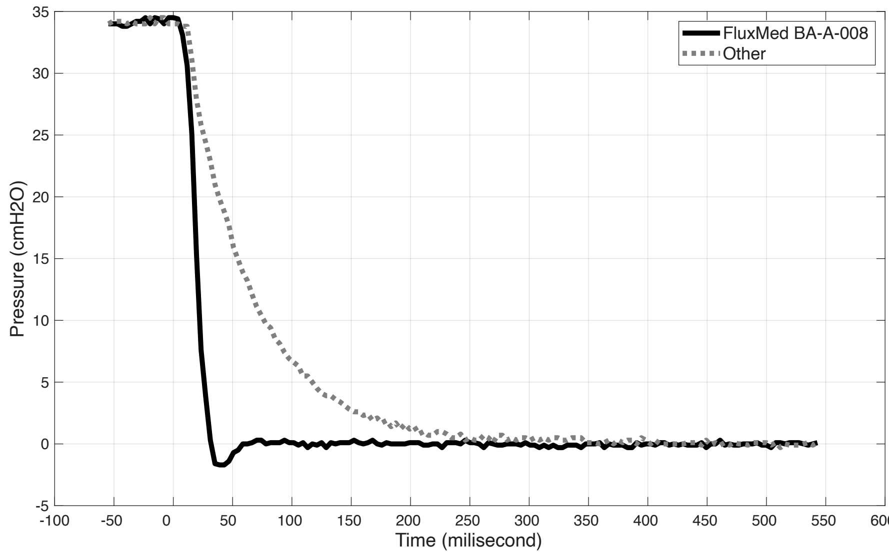

Figure 28. Dynamic response of two different probes. At time cero pressure changed to zero. The tracings show the time it takes to each probe to reach the new value.

The clinical impact of the dynamic response can be observed when pressure changes are assessed. For example, in a P0.1, if a catheter takes more than 100 milliseconds to reflect the change, the measurement will be underestimating the value. If you are looking to measure esophageal pressure swing or evaluate asynchronies, a slow dynamic response probe could affect the measurements.

It should be noted that these aspects are considering the catheter alone. When used in a patient, the probe is placed in the esophagus to measure the pressure of the pleural space. Therefore, it must be considered to minimize the effect that the esophagus adds to the measurement. Because the esophagus is naturally collapsed, when the probe is inflated inside it, volumes near the lower limit of the catheter's working range are used.

Chapter 3.

Ventilation Modes and Patient Effort

There are different degrees of control that the patient may have on the breathing pattern in a ventilator. Different ventilation modes will allow the patient to determine more aspects of the breathing pattern. The patient's effort will have different consequences depending on the variable the ventilator is controlling

Triggering

As part of the ventilator setting, the respiratory rate is determined, this defines the total time between cycles. The ventilator measures the time from the last pressurization and when the total time is reached, it triggers the next pressurization. This is a time triggered breath. If the ventilator detects that the patient is making an inspiratory effort, it can start pressurization before the total time is reached. By controlling the triggering, the patient can increase the configured respiratory rate.

The pressurization triggered by the patient is usually by two methods: by pressure or by flow. In pressure triggering, the ventilator analyzes the pressure curve during expiration, and if the pressure falls below PEEP by a value greater than a threshold, then the ventilator triggers the pressurization. The threshold is configurable so that healthcare professionals can adjust the triggering sensitivity.

In flow triggering, the ventilator analyzes the flow during expiration. If it detects that the patient generates a positive flow that is above a threshold, it triggers the pressurization. The flow threshold is configurable by the healthcare professional to determine how sensitive they want it to be.

Pressure or flow trigger is a configuration completely independent of the ventilatory mode.

Patient effort in a pressure-controlled mode

When the ventilator controls the pressure, the patient's information is seen in the flow and volume waveforms. If the patient makes an inspiratory effort, then the flow and volume waveforms will be affected.

If we think about it from the equation of motion:

$$P_{aw} + P_{mus} = R.F + E.V$$

If airway pressure is defined by the ventilator, when there is muscle pressure the left side of the equation is larger. Airway pressure plus muscular pressure adds up. The elastic and resistive components are the same. To keep the equality, the flow and volume must increase. In pressure modes, the patient's effort increases the flow and volume.

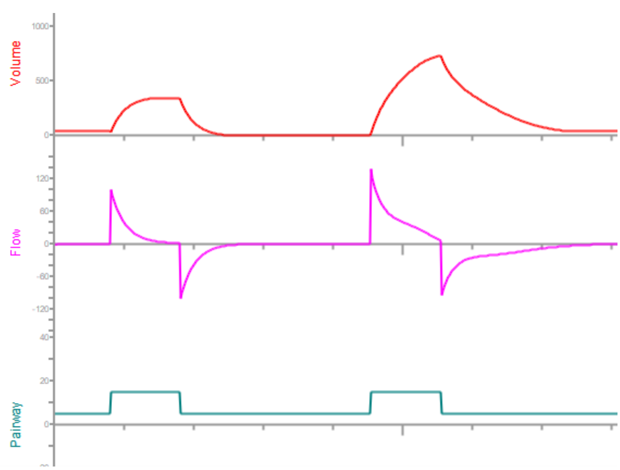

Figure 29. These are two PCV breaths. The one on the left is completely passive, the ventilator the only source of pressure. The one on the right is the patient is making some inspiratory effort. It has a higher flow and volume because of the additional muscular pressure. This image was created based on FluxMed simulator using a three-element linear model.

The ventilator's control algorithm has a pressure target. It will deliver the flow that is necessary to reach that pressure. If something else is generating pressure it will not change the pressure the ventilator is generating.

Patient effort in a volume-controlled mode

During inhalation, the ventilator controls the volume through the way in which it is delivered (the flow). We will see the patient's information in the airway pressure waveform. In a volume-controlled mode, the flow and volume curve should not change as they are controlled by the ventilator. If the patient's effort adds muscular pressure, the airway pressure tracing will be affected.

From the equation of motion:

$$P_{aw} + P_{mus} = R.F + E.V$$

Flow and volume are controlled by the ventilator; the elastic and resistive component of the patient are the same so the right side of the equation must remain the same. On the left side of the equation, if the muscular pressure increases, the airway pressure must reduce to add up to the same value.

The muscular pressure term can be passed to the right of the equal sign. Showing how

$$P_{aw} = R.F + E.V - P_{mus}$$

In a volume-controlled mode, when the patient makes an inspiratory effort, muscle pressure increases. If we look at the equation, resistance, elastance, flow and volume are the same, so when muscle pressure increases, airway pressure decreases. In a volume-controlled mode, when the patient adds an effort, the ventilator reduces the assistance.

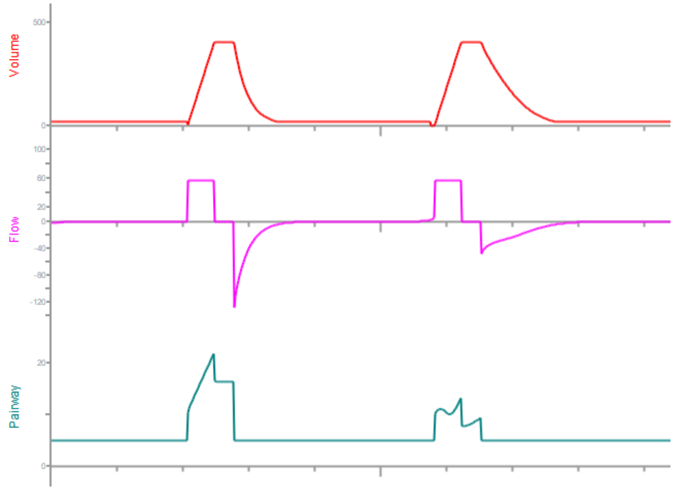

Figure 30. These are two VCV breaths. The one on the left is completely passive, the ventilator the only source of pressure. The one on the right is the patient is making some inspiratory effort. It has the flow and volume waveforms are the equal during the inspiration. On the one on the right the pressure is reduced, as the patient is providing some of the pressure that is needed to achieve the inspiratory flow. During expiration the ventilator controls pressure (PEEP), it can be seen how the relaxation of the inspiratory muscles reduces the peak expiratory flow. This image was created based on FluxMed simulator using a three-element linear model.

The ventilator's control algorithm when it controls volume, it has a flow target. It will generate the airway pressure that is necessary for the air to move at the intended speed. As your speed target if someone else (the patient) starts generating pressure it is easier for the ventilator to get to target speed. The ventilator will reach the target flow with less pressure.

Pressure support

So far, the patient could only control the trigger. The level and duration of assistance will be given by the configuration of the ventilator. In VCV, a fixed volume will be delivered and in PCV it will be pressurized at a level and for a given time that is decided by the person that configures the ventilator.

In a pressure support mode, it is a pressure-controlled mode where the patient can vary the duration of inspiration. The patient has a way of controlling the cycling. In pressure support, the ventilator triggers when it senses the patient's effort and cycles when the inspiratory flow tracing drops to certain level.

Let's assume that we have a PEEP of 5 cmH_2O, a pressure support of 10cmH_2O over PEEP, and the cycling off criteria is set at 25% of the peak inspiratory flow. When the ventilator detects the patient's effort, it will go from the PEEP pressure of 5cmH2O to a value of 15cmH_2O (10cmH_2O of pressure support plus 5 cmH_2O of PEEP). This will produce an inspiratory flow by the increase of airway pressure and the patient's muscular pressure. Let's suppose that the peak inspiratory flow is 120 liters/min. If the patient's effort is short, the muscular pressure will be short, the flow will quickly fall. When flow drops to a value of 25% of the peak inspiratory flow (in this case: 40 liters/min), the ventilator will cycle and lower the pressure to 5cmH_2O of PEEP. If the inspiratory effort lasts longer, the muscular pressure will take longer to drop, and the flow will take longer to reach the threshold and therefore will delay the cycling, making longer the inspiration.

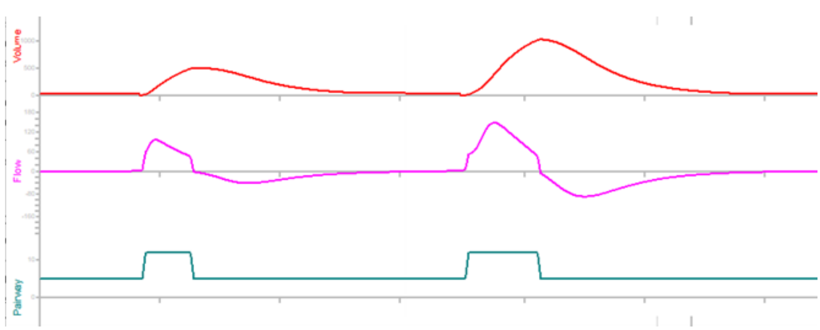

Figure 31. These are two breaths on Pressure Support. The one on the left has a smaller effort. The one on the right is the patient is making a smaller inspiratory effort than on the breath on the right. The bigger patient effort causes the inspiratory flow to take longer to decrease up to 25% of the peak inspiratory flow. Makin the inspiratory time longer. This image was created based on FluxMed simulator using a three-element linear model.

In pressure support, the patient can have some control over the cycling. In pressure support, the patient can regulate the respiratory rate and the duration of the inspiratory time. The level of pressure applied over PEEP, continues to be a factor defined by the healthcare professionals that configures the ventilator.

The cycling off threshold is the percentage of the peak inspiratory flow the flow must drop to, to cycle the breath. This means that, if everything else is the same, with a lower cycling off value the inspiration is longer and with a higher value the inspiration is shorter.

Proportional Modes

There are two modes in which the patient can control the triggering, cycling, and magnitude of assistance. They are called proportional modes. The pressure generated is proportional to the patient's effort.

PAV+, which stands for Proportional Assist Ventilation, is a proportional mode in Puritan Bennett ventilators. In this mode, the ventilator makes short inspiratory pauses that it uses to measure the patient's compliance and resistance. Using the equation of movement, it calculates muscular pressure that the patient is generating, on real time, and that allows it to generate a pressure that is proportional to the muscular pressure.

NAVA stands for Neurally Adjusted Ventilatory Assist, is a proportional mode of Getinge ventilators. It uses a probe that is placed in the patient's esophagus with 10 electrodes that measure the electrical activity of the diaphragm. Which allows it to measure the magnitude of the total electrical activity of the diaphragm. You can configure how much pressure to apply in proportion to this electrical activity (it is set as cmH2O/μV). Allowing the ventilator to generate pressure proportional to the patient's electrical activity of the diaphragm (Edi).

Know your tools

Ventilators are complex devices. Mode settings, tools and maneuvers vary significantly between different manufacturers. Each ventilator has different interfaces and controls.

Many of the variables and parameters that are configured are related to one another, so by selecting one value there are others that are also defined. For example, the inspiratory time and the I:E. In some ventilators, one is defined and in others the other.

The suggestion I have for you is that you should know your tools. Connect a test lung and play around with the settings to see how the tool you use every day to save lives responds.

From here on out

I hope that this text has been of interest to you and has contributed to the understanding of the physiology of the ventilated patient. I invite you to make your own measurements and question the assumptions until you reach your own conclusions.

The models proposed here are simplifications of reality. The respiratory system has 300 million alveolar units and here we explain them with a handful of linear parameters. The strength of these models is that they can provide a proper description with great simplification. It is in the understanding of the model to know when it ceases to be useful.

About the Author

Matías Madorno has been designing medical technology for almost a quarter of a century and has been developing technology associated with respiration monitoring for more than 20 years. He is a Full Professor at Buenos Aires Institute of Technology (Instituto Tecnológico de Buenos Aires, ITBA) where he has more than 20 years of experience in teaching on the technical side. He has extensive teaching experience in clinical aspects, and he is a professor responsible for a course in the anesthesia residency of the Buenos Aires Association of Anesthesia Analgesia and Reanimation (Asociación de Anestesia, Analgesia y Reanimación de Buenos Aires, AAARBA) and has collaborated in several courses and congresses on mechanical ventilation for health professionals.

He studied computer engineering, completed a postgraduate degree in design and maintenance of medical equipment and a doctorate in engineering at ITBA. His doctoral thesis was related to monitoring ventilated patients.

He is the creator and main responsible for the design and development of the FluxMed monitoring systems. He has a moderate scientific activity with a few dozen scientific publications.

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Understanding mechanical ventilation begins with understanding the physiology behind it.

This book offers a clear and accessible introduction to respiratory mechanics, built on simple but powerful quantitative models that explain how pressure, flow, and volume interact in the ventilated patient.

Designed for respiratory therapists, critical care physicians, anesthesiologists, and trainees, it guides the reader through the fundamental principles that govern the behavior of the respiratory system: resistance, compliance, elastance, time constants, transpulmonary pressure, and patient–ventilator interaction.

Beginning with the simplest linear models and progressively incorporating more complex elements such as chest wall mechanics, intrinsic PEEP, and patient effort, the text provides a robust conceptual framework to interpret waveforms and anticipate responses, and a tool to make sound clinical decisions at the bedside.

Rather than offering recipes, this book cultivates understanding. Through intuitive explanations, analogies, and quantitative reasoning, it equips clinicians to analyze what is happening breath by breath, and why.

A practical and physiologically grounded introduction for anyone who wants to think more clearly about mechanical ventilation.