Fundamentals of Mechanical Ventilation

A short course in the theory and application of mechanical ventilators

Robert L. Chatburn

Fundamentals of

Mechanical Ventilation

This is a unique book, written from the perspective of how ventilators work. Unlike other texts on the subject that focus on clinical applications, this book shows you how to think about ventilators, when to use various modes, and how to know if they are doing what you expect. It does not say much about how to use ventilators in various clinical situations or how to liberate patients from the machine. Mechanical ventilation is still more of an art than a science. This book focuses on how to master the instrument. Once you have done this, you will be able to make the best use of other books and actual clinical experience.

FEATURES

- Defines jargon
- Written at three levels to support (1) basic understanding, (2) comprehensive understanding, and (3) subject mastery.
- Covers ventilator design and how to understand and select modes.
- Comprehensive section on graphic displays: waveforms and loops.
- Accurate waveform illustrations based on mathematical models.
- Review questions throughout text.
- Self-assessment questions at the ends of chapters, with answers.

ABOUT THE AUTHOR

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Preface

Find a better way to educate students than the current books offer.
If you can't improve on what's available, what's the point?

Earl Babbbie
Chapman University

This book is about how ventilators work. It shows you how to think about ventilators, when to use various modes, and how to know if they are doing what you expect. This book does not say much about how to use ventilators in various clinical situations or how to liberate patients from the machine. Mechanical ventilation is still more of an art than a science. This book leads you to expertise with the theory and tools of that art. Once you have done this, you will be able to make the best use of other books and actual clinical experience.

There are 18 books devoted to mechanical ventilation on my bookshelf. They are all well written by noted experts in the field. Some are commonly used in colleges while others have fallen into obscurity. Yet, in my opinion, they all have the same limitation; they devote only a small fraction of their pages to how ventilators actually work. Most of their emphasis is on how ventilators are used to support various disease states, the physiological effects of mechanical ventilation, weaning, and adjuncts like artificial airways and humidifiers. This book is different.

The reason I made this book different may be clarified by analogy. Suppose you wanted to learn how to play the guitar. You go to the library, but all you can find are books that give you a few pages describing what different guitars look like and all the fancy names and features their manufacturers have made up. There may be a little information about how many strings they have and even what notes and chords can be played. Unfortunately, many of the books use words with apparently conflicting or obscure meanings. There is no consistency and no music theory. They all devote most of their content to a wide variety of song scores, assuming the few pages of introduction to the instrument will allow you to play them. How well do you think you would learn to play the guitar from these books? If you have ever actually tried it, you would see the difficulty. That approach works for a simple instrument like a harmonica, but it does not work well for a complex device like a mechanical ventilator. In a similar fashion, we don't let our teenagers drive cars after simply pointing out the controls on the dashboard; they have to sit through weeks of theory before ever getting behind the wheel. You can kill or injure somebody with a ventilator just as fast as you can with a car.

Certainly there is a great need for understanding the physiological effect of mechanical ventilation. But most authors seem to put the cart before the horse. In this book, I have tried to present the underlying concepts of mechanical ventilation from the perspective of the ventilator. All terminology has been clearly defined in a way that develops a consistent theoretical framework for understanding how ventilators are designed to operate. There is one chapter devoted to how to use ventilators, but it is written from the perspective of what the ventilator can do and how you should think about the options rather than from what clinical problem the patient may have. There is also a chapter devoted to monitoring the ventilator-patient interface through waveform analysis, a key feature on modern ventilators. In short, this book will teach you how to think about ventilators themselves. It teaches you to how to master the instrument. That way you are better prepared to orchestrate patient care. Only after thoroughly understanding what ventilators do will you be in a position to appreciate your own clinical experience and that of other expert authors.

The unique approach of this book makes it valuable not only to health care workers but to those individuals who must communicate with clinicians. This includes everyone from the design engineer to the marketing executive to the sales force and clinical specialists. Indeed, since manufacturers provide most of the education on mechanical ventilation, the most benefit may come from advancing their employees' level of understanding.

How to Use This Book

This book may be read on a variety of levels depending on your educational needs and your professional background. Look at the different approaches to reading and see what is most appropriate for you.

Basic Familiarity: This level is appropriate for people not directly responsible for managing ventilators in an intensive care environment. This may include healthcare personnel such as nurses, patients on home care ventilators, or those not directly involved at the bedside such as administrators or ventilator sales personnel. Study the first two chapters and the section on alarms in Chapter 3. Skim the others for areas of interest, paying attention to the figures in Chapter 5.

Comprehensive Understanding: Respiratory care students should achieve this level along with physicians and nurses who are responsible for ventilator settings. Some sales personnel may wish to understand ventilators at this level in order to converse easily with those who buy and use them. Study all the chapters, but skip the "Extra for Experts" sections. Pay attention to the "Key Idea" paragraphs and the definitions in the Glossary. Make sure you understand Chapter 5.

Subject Mastery: This level is desirable for anyone who is in a position to teach mechanical ventilation and particularly for those who are involved with research on the subject. All material in the book should be understood, including the "Extra for Experts" sections. A person at this level should be able answer all the questions and derive all the equations used throughout.

Of course, these levels are only suggestions and you will undoubtedly modify them for your own use.

Acknowledgement

The central ideas of this text came from two seminal papers I published in Respiratory Care, the official scientific journal of the American Association for Respiratory Care. The first was published in 1991, and introduced a new classification system for mechanical ventilators (Respir Care 1991:36(10):1123-1155). It was republished the next year as a part of the Journal's Consensus Conference on the Essentials of Mechanical Ventilators (Respir Care 1992:37(9):1009-1025). In the years that followed, those papers became the basis for book chapters on ventilator design in every major respiratory care textbook including:

- Tobin MJ. *Principles and Practice of Mechanical Ventilation*, 1994. McGraw-Hill.
- Branson RD, Hess DR, Chatburn RL. *Respiratory Care Equipment*, 1st and 2nd editions, 1995 & 1999. Lippincott.
- White GC. Equipment for Respiratory Care 2nd edition, 1996, Delmar.
- Hess DR, Kacmarek RM. *Essentials of Mechanical Ventilation*, 1996. McGraw-Hill.
- Pilbeam SP. Mechanical Ventilation. Physiological and Clinical Applications, 3rd edition, 1998. Mosby.
- Scanlan CL, Wilkins RL, Stoller JK. *Egan's Fundamentals of Respiratory Care* 7th edition, 1999. Mosby.
- Branson RD. MacIntyre NR. *Mechanical Ventilation*, 2001. WB Saunders.
- Hess DR, MacIntyre NR, Mishoe SC, Galvin WF, Adams WB, Saposnick AB. *Respiratory Care. Principles & Practice*, 2002. Saunders.

In 2001, my coauthor, Dr. Frank Primiano Jr., and I introduced a new system for classifying modes of ventilation, tying in with the principles established in the earlier publications (Respir Care 2001; 46(6):604-621). That paper received the Dr. Allen DeVilbiss Technology Paper Award for best paper of the year.

1. INTRODUCTION TO VENTILATION

During breathing, a volume of air is inhaled through the airways (mouth and/or nose, pharynx, larynx, trachea, and bronchial tree) into millions of tiny gas exchange sacs (the alveoli) deep within the lungs. There it mixes with the carbon dioxide-rich gas coming from the blood. It is then exhaled back through the same airways to the atmosphere. Normally this cyclic pattern repeats at a breathing rate, or frequency, of about 12 breaths a minute (breaths/min) when we are at rest (a higher resting rate for infants and children). The breathing rate increases when we exercise or become excited.[1]

Gas exchange is the function of the lungs that is required to supply oxygen to the blood for distribution to the cells of the body, and to remove carbon dioxide from the blood that the blood has collected from the cells of the body. Gas exchange in the lungs occurs only in the smallest airways and the alveoli. It does not take place in the airways (conducting airways) that carry the gas from the atmosphere to these terminal regions. The size (volume) of these conducting airways is called the **anatomical dead space** because it does not participate directly in gas exchange between the gas space in the lungs and the blood. Gas is carried through the conducting airways by a process called "convection". Gas is exchanged between the pulmonary gas space and the blood by a process called "diffusion".

Key Idea

One of the major factors determining whether breathing is producing enough gas exchange to keep a person alive is the ventilation the breathing is producing. Ventilation (usually referred to as **minute ventilation**) is expressed as the volume of gas entering, or leaving, the lungs in a given amount of time. It can be calculated by multiplying the volume of gas, either inhaled or exhaled during a breath (called the **tidal volume**), times the breathing rate (eg, 0.5 Liters x 12 breaths/min = 6 L/min). The level of ventilation can be monitored by measuring the amount of carbon dioxide in the blood. For a given level of carbon dioxide produced by the body, the amount in the blood is inversely proportional to the level of ventilation:

$$\text{carbon dioxide in blood} \propto \frac{\text{carbon dioxide produced by metabolism}}{\text{minute ventilation}}$$

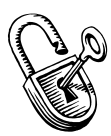
Therefore, if we were to develop a machine to help a person breathe, or to take over his or her breathing altogether, it would have to be able to produce a tidal volume and a breathing rate which, when multiplied together, produce enough ventilation, but not too much ventilation, to supply the gas exchange needs of the body. During normal breathing the body selects a combination of a tidal volume that is large enough to clear the dead space and add

[1] This section is adapded from: Primiano FP Jr, Chatburn RL. What is a ventilator? Part I. www.VentWorld.com;2001.

fresh gas to the alveoli, and a breathing rate that assures the correct amount of ventilation is produced. However, as it turns out, it is possible, using specialized equipment, to keep a person alive with breathing rates that range from zero (steady flow into and out of the lungs) up to frequencies in the 100's of breaths per minute. Over this frequency range, convection and diffusion take part to a greater or lesser extent in distributing the inhaled gas within the lungs. As the frequency is increased, the tidal volume that produces the required ventilation gets smaller and smaller.

There are two sets of forces that can cause the lungs and chest wall to expand: the forces produced when the muscles of respiration (diaphragm, inspiratory intercostal, and accessory muscles) contract, and the force produced by the difference between the pressure at the airway opening (mouth and nose) and the pressure on the outer surface of the chest wall. Normally, the respiratory muscles do the work needed to expand the chest wall, decreasing the pressure on the outside of the lungs so that they expand, which in turn enlarges the air space within the lungs, and draws air into the lungs. The difference between the pressure at the airway opening and the pressure on the chest wall surface does not play a role in this activity under normal circumstances. This is because both of these locations are exposed to the same pressure (atmospheric), so this difference is zero. However, when the respiratory muscles are unable to do the work required for ventilation, either or both of these two pressures can be manipulated to produce breathing movements, using a mechanical ventilator.

It is not difficult to visualize that, if the pressure at the airway opening (ie, the mouth and nose or artificial airway opening) of an individual were increased while the pressure surrounding the rest of the person's body remained at atmospheric, the person's chest would expand as air is literally forced into the lungs. Likewise, if the pressure on the person's body surface were lowered as the pressure at the person's open mouth and nose remained at atmospheric, then again the pressure at the mouth would be greater than that on the body surface and air would be forced into the lungs.



Key Idea

> Thus, we have two approaches that can be used to mechanically ventilate the lungs: apply positive pressure (relative to atmospheric) to the airway opening - devices that do this are called **positive pressure ventilators**; or, apply negative pressure (relative to atmospheric) to the body surface (at least the rib cage and abdomen) - such devices are called **negative pressure ventilators**.

Sometimes positive airway pressure is applied to a patient’s airway opening without the intent to ventilate but merely to maintain a normal lung volume. Originally, devices were designed to present resistance to expiratory flow, and hence provide positive pressure throughout expiration. The pressure at end expiration was called positive end expiratory pressure or **PEEP**. The problem with these early devices was that the patient had to inhale with enough force to drop the airway pressure through the PEEP level to below atmospheric pressure before inspiratory flow would begin. This often increased the work of breathing to intolerable levels. Newer devices were designed to avoid this problem. The key was to design the device so that the patient could inspire by dropping the pressure just below the PEEP level, rather than all the way to atmospheric pressure. As a result, the pressure in the patient’s lungs remained positive (ie, above atmospheric) throughout the breathing cycle. Thus, the new procedure was called continuous positive airway pressure or **CPAP**. Almost all current ventilators provide CPAP rather than PEEP. There are also devices that just produce CPAP for patients that are breathing without a ventilator.

As time passed, people forgot the historic reasons for the distinction between PEEP and CPAP. The original PEEP therapy is now called "positive airway pressure", PAP, and is used to help patients (who are not connected to mechanical ventilators) to mobilize airway secretions and reverse atelectasis. Currently, the term PEEP is applied to the continuous positive airway pressure provided during assisted ventilation by a mechanical ventilator. Assisted ventilation means simply that the ventilator helps the patient with the timing and/or work of inspiration. The term CPAP is usually applied to continuous positive airway pressure provided while the patient breathes unassisted, such as for infants with respiratory distress syndrome after extubation or adults with sleep apnea.

It is important to remember that CPAP and PEEP themselves are not forms of assisted ventilation, in the sense that they do not supply any of the work of breathing. They may, however, make it easier for the patient to breathe by lowering airway resistance or increasing lung compliance.

Self Assessment Questions

Definitions

Explain the meaning of the following terms:

- Anatomical dead space
- Minute ventilation
- Tidal volume
- PEEP
- CPAP

True or False

1. Gas exchange is the function of the lungs that is required to supply oxygen to the blood for distribution to the cells of the body, and to remove carbon dioxide from the blood that the blood has collected from the cells of the body.
2. Gas exchange occurs in the all the conducting airways and the alveoli.
3. Minute ventilation is calculated as the product of tidal volume and breathing rate.
4. The unit of measurement for minute ventilation is liters.
5. It is possible to keep a person alive with breathing rates that range from zero (steady flow into and out of the lungs) up to frequencies in the 100's of breaths per minute.

Multiple Choice

1. The forces that expand the lungs and chest wall during inspiration are:

a. The forces produced when the muscles of respiration (diaphragm, inspiratory intercostal, and accessory muscles) contract.

b. Positive end expiratory pressure (PEEP).

c. The force produced by the difference between the pressure at the airway opening (mouth and nose) and the pressure on the outer surface of the chest wall.

d. Both a and c.

2. In order to generate an inspiration, the following condition must be present:

   a. Lung pressure must be higher than pressure at the airway opening.

   b. Airway pressure must be higher than body surface pressure.

   c. Body surface pressure must be higher than airway pressure.

   d. Pleural pressure must be lower than body surface pressure.

3. In order to generate an expiration, the following condition must be present:

   a. Lung pressure must be higher pressure at the airway opening.

   b. Pressure at the airway opening must be higher than body surface pressure.

   c. Body surface pressure must be higher than pressure at the airway opening.

   d. Body surface pressure must be lower than lung pressure.

Key Ideas

1. What two variables determine whether breathing is producing enough gas exchange to keep a person alive.

2. Explain how the level of ventilation can be monitored by measuring carbon dioxide in the blood. Why not just measure tidal volume and frequency?

3. Describe the difference between positive pressure ventilators and negative pressure ventilators.

2. INTRODUCTION TO MECHANICAL VENTILATORS

A **mechanical ventilator** is an automatic machine designed to provide all or part of the work the body must produce to move gas into and out of the lungs. The act of moving air into and out of the lungs is called breathing, or, more formally, ventilation.

The simplest mechanical device we could devise to assist a person's breathing would be a hand-driven, syringe-type pump that is fitted to the person's mouth and nose using a mask. A variation of this is the self-inflating, elastic resuscitation bag. Both of these require one-way valve arrangements to cause air to flow from the device into the lungs when the device is compressed, and out from the lungs to the atmosphere as the device is expanded. These arrangements are not automatic, requiring an operator to supply the energy to push the gas into the lungs through the mouth and nose. Thus, such devices are not considered mechanical ventilators.

Automating the ventilator so that continual operator intervention is not needed for safe, desired operation requires:

- a stable attachment (interface) of the device to the patient,
- a source of energy to drive the device,
- a control system to regulate the timing and size of breaths, and
- a means of monitoring the performance of the device and the condition of the patient.

Types of Ventilators

We will consider two classes of ventilators here. First are those that produce breathing patterns that mimic the way we normally breathe (ie, at rates our bodies produce during our usual living activities: 12 - 25 breaths/min for children and adults; 30 - 40 breaths/min for infants). These are called **conventional ventilators** and their maximum rate is 150 breaths/minute.[1] Second are those that produce breathing patterns at frequencies much higher than we would or could voluntarily produce for breathing - called **high frequency ventilators**. These ventilators can produce rates up to 15 Hz (900 breaths/minute).

Conventional Ventilators

The vast majority of ventilators used in the world provide conventional ventilation. This employs breathing patterns that approximate those produced by a normal spontaneously breathing person. Tidal volumes are large enough to clear the anatomical dead space during inspiration and the breathing rates are in the range of normal rates. Gas transport in the airways is dominated by convective flow and mixing in the alveoli occurs by *"**Health Devices** has repeatedly stressed the need for users to understand the operation and features of ventilators, regardless of whether they will be used to ventilate neonatal/pediatric or adult patients. The fact that*

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[1] This is a limit imposed by the Food and Drug Administration on manufacturers.

ventilators are such an established technology by no means guarantees that these issues are clearly understood…we continue to receive reports of hospital staff misusing ventilators because they're unaware of the devices' particular operational considerations."

ECRI Health Devices July 2002, Volume 31, Number 7

3. HOW VENTILATORS WORK

If you want to understand how ventilators work, and not just how to turn the knobs, it is essential to have some knowledge of basic mechanics. We begin by recognizing that a ventilator is simply a machine designed to transmit applied energy in a predetermined manner to perform useful work. Ventilators are powered with energy in the form of either electricity or compressed gas. That energy is transmitted (by the ventilator's drive mechanism) in a predetermined manner (by the control circuit) to assist or replace the patient's muscular effort in performing the work of breathing (the desired output). Thus, to understand ventilators we must first understand their four mechanical characteristics:

1) Input power
2) Power conversion and transmission
3) Control system
4) Output (pressure, volume, and flow waveforms)

We can expand this simple outline to add as much detail about a given ventilator as desired. A much more detailed description of ventilator design characteristics can be found in books on respiratory care equipment.[2]

Input Power

The power source for a ventilator is what generates the force to inflate the patient's lungs. It may be either electrical energy (Energy = Volts × Amperes × Time) or compressed gas (Energy = Pressure × Volume). An electrically powered ventilator uses AC (alternating current) voltage from an electrical line outlet. In addition to powering the ventilator, this AC voltage may be reduced and converted to direct current (DC). This DC source can then be used to power delicate electronic control circuits. Some ventilators have rechargeable batteries to be used as a back-up source of power if AC current is not available.

A pneumatically powered ventilator uses compressed gas. This is the power source for most modern intensive care ventilators. Ventilators powered by compressed gas usually have internal pressure reducing valves so that the normal operating pressure is lower than the source pressure. This allows uninterrupted operation from hospital piped gas sources, which are usually regulated to 50 p.s.i. (pounds per square inch) but are subject to periodic fluctuations.

[2] Branson RD, Hess DR, Chatburn RL. Respiratory Care Equipment, 2nd Ed. Philadelphia: Lippencott Williams & Wilkins, 1999. ISBN 0-7817-1200-9

Power Transmission and Conversion

The power transmission and conversion system consists of the drive and output control mechanisms. The drive mechanism generates the actual force needed to deliver gas to the patient under pressure. The output control consists of one or more valves that regulate gas flow to and from the patient.

The ventilator's drive mechanism converts the input power to useful work. The type of drive mechanism determines the characteristic flow, and pressure patterns the ventilator produces. Drive mechanisms can be either: (1) a direct application of compressed gas through a pressure reducing valve, or (2) an indirect application using an electric motor or compressor.

The output control valve regulates the flow of gas to and from the patient. It may be a simple on/off exhalation. An example would be the typical infant ventilator. The valve in the exhalation manifold closes to provide a periodic pressure waveform that rises to a preset limit during inspiration (ie, forcing gas into the lungs) then opens to allow pressure to fall to another preset limit during exhalation (ie, allowing gas to escape from the lungs). Alternatively, there can be a set of output control valves that shape the output waveform. An example would be the Hamilton Galileo ventilator. This ventilator uses an exhalation manifold valve that closes to force gas into the lungs or opens to allow exhalation. There is also a flow control valve that shapes the inspiratory flow waveform once the exhalation manifold closes.

Control System

The Basic Model of Breathing (Equation of Motion)

We use models of breathing mechanics to provide a foundation for understanding how ventilators work. These models simplify and illustrate the relations among variables of interest. Specifically, we are interested in the pressure needed to drive gas into the airway and inflate the lungs.

The physical model of breathing mechanics most commonly used is a rigid flow conducting tube connected to an elastic compartment as shown in Figure 3-1. This is a simplification of the actual biological respiratory system from the viewpoint of pressure, volume, and flow.

The mathematical model that relates pressure, volume, and flow during ventilation is known as the **equation of motion** for the respiratory system:

muscle pressure + ventilator pressure = (elastance x *volume) + (resistance* x *flow)*

This equation is sometimes expressed in terms of compliance instead of elastance.

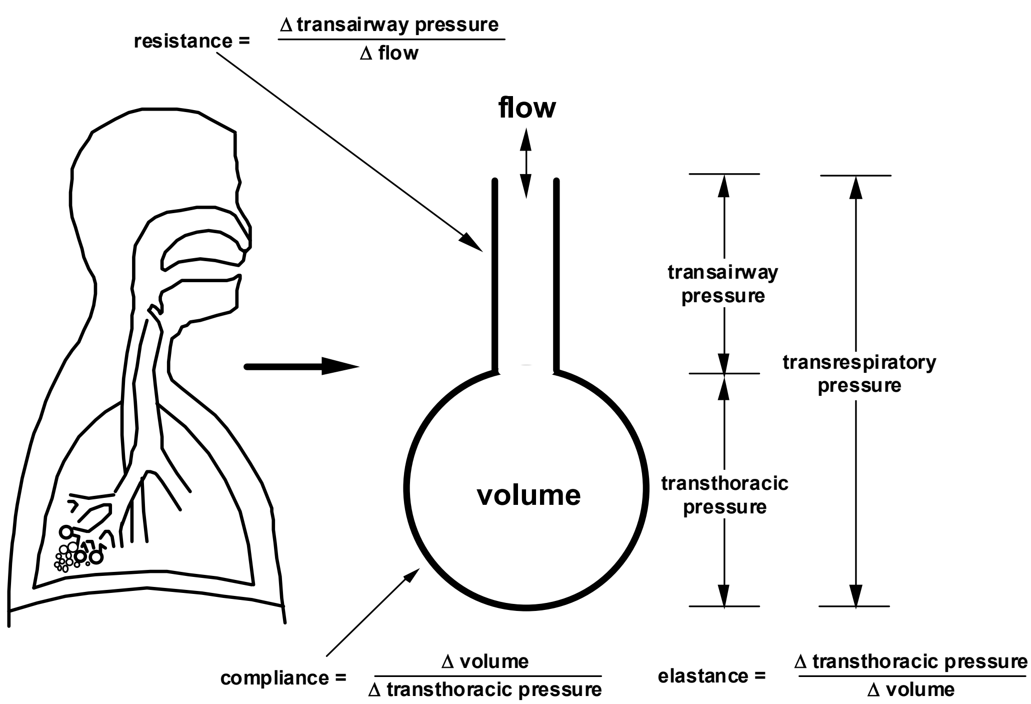
$$\textit{muscle pressure} + \textit{ventilator pressure} = (\textit{volume} / \textit{compliance}) + (\textit{resistance} \times \textit{flow})$$

Pressure, volume and flow are variable functions of time, all measured relative to their end expiratory values. Under normal conditions, these values are: muscle pressure = 0, ventilator pressure = 0, volume = functional residual capacity, flow = 0. During mechanical ventilation, these values are: muscle pressure = 0, ventilator pressure = PEEP, volume = end expiratory volume, flow = 0. Elastance and resistance are constants.

When airway pressure rises above baseline (ie, when ventilator pressure increases), inspiration is assisted. The pressure driving inspiration is called **transrespiratory system pressure**. It is defined as the pressure at the airway opening (mouth, endotracheal tube or tracheostomy tube) minus the pressure at the body surface. Transrespiratory system pressure has two components, **transairway pressure** (defined as airway opening pressure minus lung pressure) and **transthoracic pressure** (defined as lung pressure minus body surface pressure). We may occasionally use the term **transpulmonary pressure**, defined as airway opening pressure minus pleural pressure.

Figure 3-1. Models of the ventilatory system. P = pressure. Note that compliance = 1/elastance. Note that intertance is ignored in this model, as it is usually insignificant.

Key Idea



Equation of Motion for the Respiratory System

$$P_{vent} + P_{muscles} = \text{elastance} \times \text{volume} + \text{resistance} \times \text{flow}$$

Muscle pressure is the imaginary (ie, unmeasurable) transrespiratory system pressure generated by the ventilatory muscles to expand the thoracic cage and lungs. Ventilator pressure is transrespiratory system pressure generated by the ventilator. The combined muscle and ventilator pressures cause gas to flow into the lungs.

Elastance (elastance = Δpressure/Δvolume) together with **resistance** (resistance = Δpressure/Δflow) contribute to the load against which the muscles and ventilator do work (note that load has the units of pressure, so the left side of the equation equals the right side). So the equation of motion may also be expressed as:

muscle pressure + ventilator pressure = elastic load + resistive load

Elastic load is the pressure required to deliver the tidal volume (elastance times tidal volume) and **resistive load** is the pressure required to deliver the flow (resistance times flow). Note: it is sometimes more convenient to speak of compliance instead of elastance. **Compliance** is defined as Δvolume/Δpressure and is equal to 1/elastance.

Key Idea

From the equation, we see that if ventilator pressure is zero, the muscles provide all the work of breathing. This is normal, unassisted breathing. Note that if the patient is connected to a ventilator and the ventilator provides exactly the flow demanded by the patient's inspiratory effort, the airway pressure will not rise above baseline (ie, $P_{vent} = 0$ throughout inspiration). If the ventilator does not provide enough flow to meet the demand, airway pressure will fall below baseline. On the other hand, if the ventilator provides more flow than is demanded by the patient, then airway pressure will rise above baseline and inspiration is said to be "assisted". If both the muscle pressure and the ventilator pressure are non-zero, the patient provides some of the work and the ventilator provides some work. This is called **partial ventilatory support**. If the muscle pressure is zero, the ventilator must provide all the work of breathing. This is called **total ventilatory support**.

Review and Consider

1. The equation of motion for the respiratory system can be traced to Newton's Third Law of Motion: Every action has an equal and opposite reaction. In fact, the equation of motion is sometimes called a "force balance" equation. Why? (*Hint: what is the unit of measurement that results from multiplying elastance by resistance or multiplying resistance by flow?*)
2. Rewrite the equation of motion in using only transrespiratory pressure, thransthroacic pressure and transairway pressure.
3. Write the equation of motion for unassisted spontaneous inspiration and for assisted ventilation of a paralyzed patient.
4. Write the equation of motion for passive expiration.
5. If lung elastance increases, what happens to lung compliance?

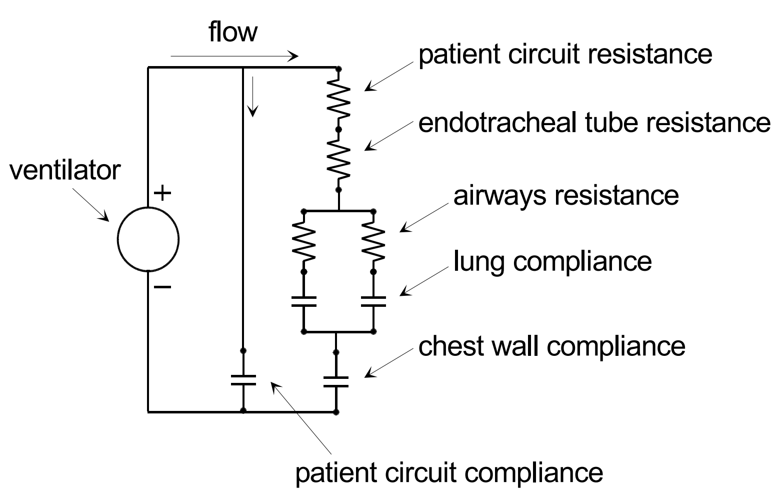
6. Use the equation of motion to show what happens to airway pressure if airway resistance decreases during mechanical ventilation.

Extra for Experts

The model shown in Figure 3-1 is really an oversimplification of the actual respiratory system. For example, it lumps together chest wall and lung compliance as well as lumping together the compliances of the two lungs. In addition, it lumps together the resistances of all the many airways. It also ignores inertance (the constant of proportionality between pressure and the rate of change of flow) because the inertia of the gas, lungs, and chest wall are insignificant at normal frequencies.

For some discussions, it is more useful to have multi-compartment models. To simplify the drawing of such models, we borrow symbols from electrical engineering. Specifically, a resistor in electronics is used to represent airway resistance and a capacitor is used to represent compliance. The ventilator may be represented as a constant voltage source (ie, a pressure controller) as shown in Figure 3-1 or it may be represented as a constant current source (ie, a flow controller). Figure 3-2 shows a multi-compartment model using electrical components.

Figure 3-2. Multi-compartment model of the respiratory system connected to a ventilator using electronic analogs. Note that the right and left lungs are modeled as separate series connections of a resistance and compliance. However, the two lungs are connected in parallel. The patient circuit resistance is in series with the endotracheal tube. The patient circuit compliance is in parallel with the respiratory system. The chest wall compliance is in series with that of the lungs. The function of the exhalation manifold can be shown by adding a switch that alternately connects the patient and patient circuit to the positive pole of the ventilator (inspiration) or to ground (the negative pole, for expiration). Note that inertance, modeled as an electrical inductor, is ignored in this model as it is usually negligible.



"The full advantage of the newly available ventilator modes and features will be realized only if your facility is willing and able to dedicate time to an extensive in-house training program Often, clinicians overlook the new, and perhaps more optimal, modes because they are more familiar with the older modes."

ECRI Health Devices July 2002, Volume 31, Number 7

4. HOW TO USE MODES OF VENTILATION

The clinical use of the many available modes of ventilation is a much debated topic. A full coverage of all the issues is beyond the scope of this presentation. However, we will review the clinical application of the major breathing patterns shown in Table 3-2. This should provide the student with a solid understanding of the basic approaches to ventilatory support and a framework for learning more through actual experience and further reading.

Volume Control vs Pressure Control

Figure 4-1 illustrates the important variables for *volume control* modes. It shows that the primary variable we wish to control is the patient's minute ventilation. Minute ventilation is usually adjusted by means of a set tidal volume and frequency (rate), but some ventilators require you set minute volume directly. In addition, some ventilators allow you to set rate directly while on others rate is an indirect result of the inspiratory and expiratory time settings. Tidal volume is a function of the set inspiratory flow and the set inspiratory time. Inspiratory time is affected by the set frequency and, if adjustable, the set I:E ratio. The mathematical relations among all these variables are given in Table 4-1.

Extra for Experts

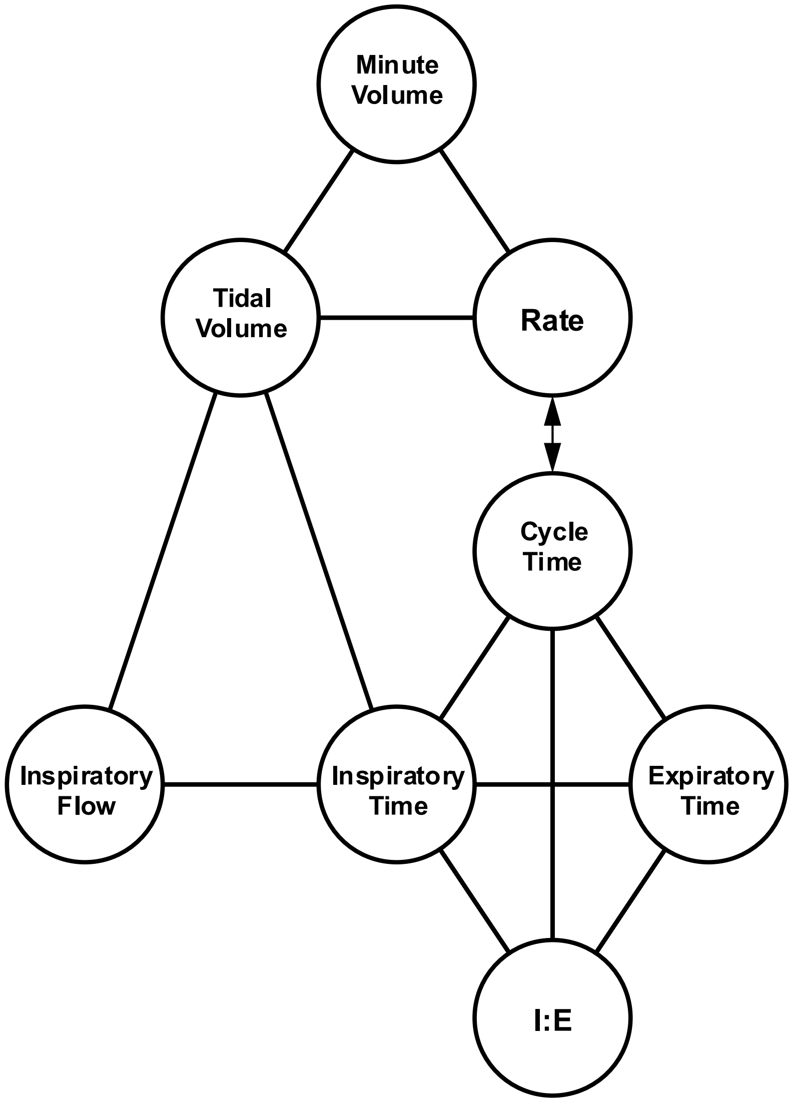
The equations for volume and flow were derived by solving the equation of motion using calculus. The equation of motion itself is not an algebraic equation, as it may appear, but rather a linear differential equation with constant coefficients (elastance and resistance). Its more technical form is

$$p(t) = E \cdot v(t) + R \cdot \frac{dv}{dt}$$

During volume control, if flow is constant (k) then the solution for $dv/dt = k$ is $v(t) = kt$. This says that volume as a function of time equals the constant flow times time. Thus, tidal volume is simply the constant (or average) flow times inspiratory time $\left(V_T = \dot{V} \times T_I\right)$ as shown in Table 4-1.

During pressure control, pressure, as a function of time is assumed constant. Thus $p(t)$ in the equation of motion becomes k and the equation is then solved to give the exponential equations for volume and flow shown in Table 4-1.

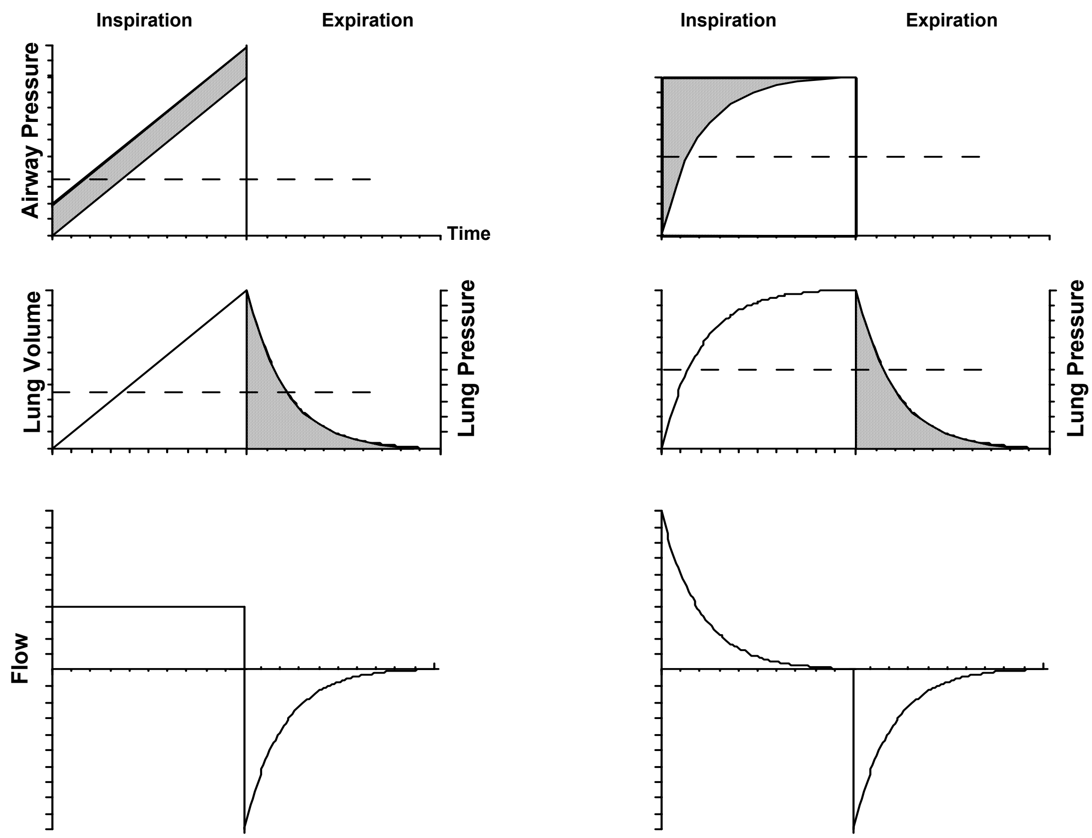
Figure 4-1. Influence diagram showing the relation among the key variables during volume controlled mechanical ventilation.



This diagram represents the most fundamental ideas of mechanical ventilation. Without a complete understanding of the variables and how they are related, you will not be able to understand how to manage even the simplest mode of ventilation.

Figure 4-4. Comparison of volume control using a constant inspiratory flow (left) with pressure control using a constant inspiratory pressure (right). Shaded areas show pressure due to resistance. Unshaded areas show pressure due to compliance. The dashed line shows mean airway pressure. Note that lung volume and lung pressure have the same waveform shape.

Key Idea



Continuous Mandatory Ventilation (CMV)

Continuous mandatory ventilation (sometimes referred to as the "Assist/Control" mode) is intended to provide full ventilatory support. All breaths are mandatory. They are delivered by the ventilator at a preset volume or pressure, breath rate and inspiratory time. When people use the term "Assist/Control" they usually mean volume controlled CMV with "assist" referring to the possibility of patient triggering and "control" meaning that the patient becomes apneic, breaths will be machine triggered.

The ventilator will deliver a patient triggered breath if the patient has spontaneous inspiratory efforts, so it is important to set an appropriate trigger level. The ventilator delivers time triggered breaths if patient efforts are absent. The ventilator may **autotrigger** (ie, repeatedly trigger itself when the trigger level is set too sensitive). As a result, hyperventilation, air trapping and patient anxiety often ensue. However, if the trigger level is not sensitive enough, the ventilator will not respond to the patient's inspiratory efforts, which results in an increased work of breathing. In the case where spontaneous triggering is counterproductive (eg, the patient tends to hyperventilate), sedation or paralysis may be required or another breathing pattern may be tried.

Volume Control

Indications

Theoretically, volume control (with a constant inspiratory flow) results in a more even distribution of ventilation (compared to pressure control) among lung units with different time constants where the units have equal resistances but unequal compliances (eg, ARDS).[1] Volume controlled CMV is indicated when it is necessary to maintain precise regulation of minute ventilation or a blood gas parameter such as $PaCO_2$ in patients who have minimal respiratory drive such that synchrony with the ventilator is not a problem.

During VC-CMV, changes in the patient's lung mechanics result in changes in airway pressure. A reduction in lung compliance and or an increase in resistance will cause higher peak airway pressures. Care should also be taken to avoid setting a flow setting that fails to meet patient needs or exceeds their demand (if the patient is making inspiratory efforts). If the flow exceeds patient demands, inspiration may be prematurely shortened. An insufficient flow rate could result in an increase in the work of breathing and a concomitant increase in oxygen consumption. This would be visualized on a waveform monitor as the airway pressure dipping below baseline. To date, only one ventilator has software capable of avoiding this problem. The Flow Augment feature of the Bear 1000 will increase the flow above the set value if the patient's inspiratory effort causes airway pressure to go below baseline.

Example

Perhaps the most common application of this mode of ventilation is to facilitate therapeutic hyperventilation in the patient with traumatic brain injury. Patients are often sedated to reduce oxygen consumption, ventilator asynchrony, and to minimize the patient's response to noxious stimuli. VC-CMV can achieve precise regulation of $PaCO_2$ and support efforts to alleviate intracranial hypertension and reduce the likelihood of secondary complications from cerebral ischemia.

[1] Respir Care 1994;(39)10:979-988.

ACHTUNG!

Das Machine ist nict fur derfingerpoken und mittengrabben. Ist easy schnappen der Springerwerk, blowenfusen und corkenpoppen mit spitzensparken. Ist nict fur gwrwerken by das Dummkopfen. Das rubbernecken Sightseeren keepen hands in das pockets. Relaxen und watch das Blinkenlights.

Old German folk saying

5. HOW TO READ VENTILATOR GRAPHIC DISPLAYS

Rapid Interpretation of Graphic Displays

When learning to read ECGs or radiographic displays, it is useful to look for key features in a specific order. We can apply that technique to waveform displays. The following procedure for routine inspection of ventilator graphics will guide you to look at general features first and then focus on details and potential problems.

1. *Check the overall quality of the display*. Make sure the pressure, volume and flow scales are set correctly and that no portions of the waveforms are cut off. The time scale should be set appropriately; use a fast sweep speed (shorter time scale) if you want to see details of individual breaths and slow sweep (longer time scale) if you want to look at trends. Check to see if leaks (eg, around uncuffed tubes or through chest tubes) are distorting the volume and flow waveforms or preventing an effective determination of static pressure from an inspiratory hold.
2. *Identify the mode of ventilation.* First, try to distinguish between mandatory and spontaneous breaths to determine the breath sequence (eg, CMV vs IMV). Then determine if mandatory breaths are volume controlled or pressure controlled. Of course you could just look at the ventilator to see what mode is set. However, there can be an infinite number of waveform displays for any given mode depending on what the patient is doing, and the whole point is to assess the ventilator-patient interaction. Besides, if the ventilator is not set properly, it may not be delivering the expected mode. For example, if the trigger sensitivity is not set correctly, the patient may be in IMV instead of SIMV or the pressure support setting may not actually be active.
3. *Check for signs of asynchrony.* From step 2 you should have determined if the patient is making spontaneous breathing efforts. If so, make sure the trigger sensitivity is correctly set. Waveform displays are good for checking how hard the patient must work to trigger a breath and loop displays are good for checking the synchrony between spontaneous and mandatory breaths (particularly useful during infant ventilation). For volume controlled breaths, see if the inspiratory flow is high enough. For pressure controlled breaths, check for proper setting of pressure rise time and cycling threshold (if the ventilator allows such settings). For both types of breaths, determine if there is noticeable gas trapping.
4. *Check for optimal settings and therapeutic response.* This should include an inspection of pressure-volume loops to detect signs of over-distention, indicating a need to reduce tidal volume (this is only reliable for volume controlled breaths). When using an

inspiratory hold, you can look at the distance between peak and plateau pressures as an indication of changes in airway resistance (eg, tube kinking or response to bronchodilator). Changes in plateau pressure indicate changes in compliance and may be used to set optimal PEEP levels. Plateau pressure should be below 35 cmH_2O to lessen the risk of lung damage from too large a tidal volume.

You will learn exactly how to look at graphic displays and implement the four step procedure in the rest of this chapter.

Waveform Display Basics

To understand heart physiology we study ECGs and blood pressure waveforms. In the same way, to understand ventilator-patient interaction, we must examine ventilator output waveforms. Ventilator graphics are usually presented in one of two ways. The most common is to plot control variables (pressure, volume, and flow) on the vertical axis and time on the horizontal axis (see Figure 5-1). This type of graphic is often called a **waveform display** (sometimes called a scalar display). The other type of graph plots one control variable against another (such as flow on the vertical axis and volume on the horizontal axis). This is referred to as a **loop display**. We will begin our discussion of output graphics with a detailed description of waveform displays.

Key Idea

It is important to remember that pressure, volume, and flow are all variables measured relative to their baseline or end-expiratory values. Also, convention dictates that positive flow values (above the horizontal axis) correspond to inspiration, and negative flow values (below the horizontal axis) correspond to expiration. The vertical axes are in units of the measured variables (usually cmH_2O for pressure, liters or mL for volume, and L/minute or L/second for flow). The horizontal axis of these graphs is time, usually in seconds.

Because pressure, volume, and flow are all related by the equation of motion, and because waveform displays are graphs of the variables in the equation of motion, it follows that the shapes of the waveforms are related. To understand this, let's take a closer look at the output waveforms first shown in Chapter 2 (Figure 2-1, redrawn in Figure 5-1 below).

Volume Controlled Ventilation

Keep in mind that the equation of motion represents a physical model composed of a rigid flow-conducting tube connected to an elastic chamber (much like a straw connected to a balloon, as shown in Figure 3-1). This physical model represents the airways and the lungs/chest wall. In Figure 5-1, waveforms are plotted for volume controlled ventilation, with constant inspiratory flow, through the airways alone (A), the lungs alone (B), and the two connected (C).

In Figure 5-1A, we see the graph for a model with resistance (the airways) only. A constant flow during inspiration produces a rectangular flow waveform if we include the instantaneous rise at the start of inspiration and the fall back to zero at end inspiration. Volume is the integral of flow (in calculus terms, this is the area between the flow curve and the time axis). For a constant flow, lung volume equals the product of flow and time. Thus, for any inspiratory time, tidal volume equals inspiratory time multiplied by the constant flow.

This produces the graph of a straight line with a slope equal to the flow (slope = $\Delta x/\Delta y$, which in this case is change in volume divided by change in time).

Transairway pressure is the product of resistance and flow. Both resistance and flow are constants and the graph of a constant function of time is a straight horizontal line. Another way to look at it is that at each moment, the pressure waveform is just the flow waveform multiplied by the constant resistance, producing the same shape but a different scale. Thus, a rectangular flow waveform produces a rectangular airway pressure waveform. We called this pressure "resistive load".

Figure 5-1. Pressure, volume and flow waveforms for different physical models during volume controlled ventilation. **A** Waveforms for a model with resistance only showing sudden initial rise in pressure at the start of inspiration and then a constant pressure to the end. **B** Waveforms for a model with elastance only showing a constant rise in pressure from baseline to peak inspiratory pressure. **C** Waveforms for a model with resistance and elastance, representing the respiratory system. Pressure rises suddenly at the start of inspiration due to resistance and then increases steadily to peak inspiratory pressure due to elastance.

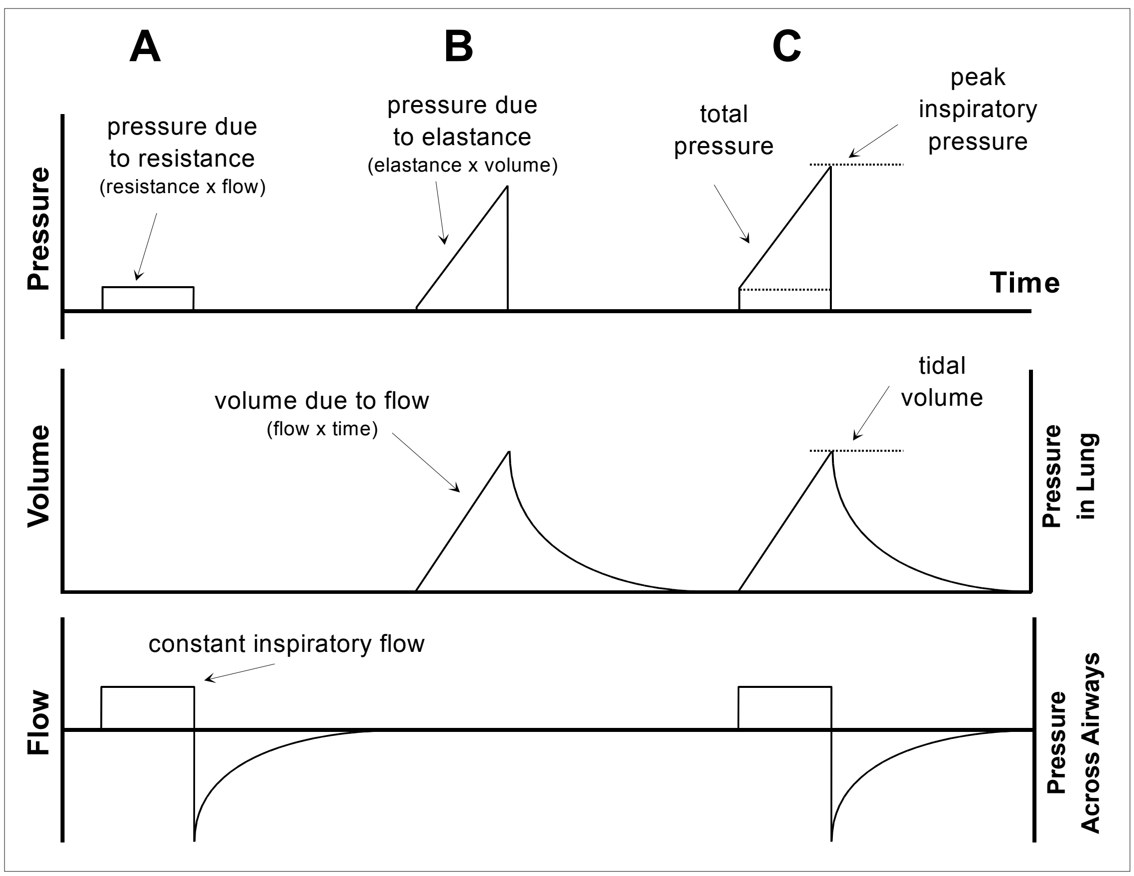


Figure 5-1B shows the results for elastance (the lungs) only. The flow and volume waveforms are of course the same, but the airway pressure waveform is triangular. This is because airway pressure is the product of elastance and volume (a constant and a variable). The result is a graph of a straight line with a slope proportional to the elastance.

Because lung pressure is the product of elastance and volume, lung pressure has the same waveform as volume with just a different vertical scale (with elastance acting like a scaling factor). We called this lung pressure "elastic load" in our earlier discussions.

In Figure 5-1C, we see the waveform for the model of resistance connected in series with elastance (ie, airways and lungs) as it might appear on a ventilator's airway pressure monitor. The flow and volume waveforms are again the same but the airway pressure waveform is the sum of the waveforms in A and B. You can visualize the triangle on top of the rectangle. Another way to look at this is that, for each moment in time, the height of the flow waveform in 5-1C added to the height of the volume waveform in 5-1C equals the height of the pressure waveform in 5-1C.

Predicting the Effects of Changes in Mechanics

Once you understand the above discussion, it will be easy for you to interpret or predict waveform changes associated with changes in respiratory system mechanics. For example, what happens to peak inspiratory pressure if airway resistance increases? Take a minute to figure this out for yourself before you read the answer.

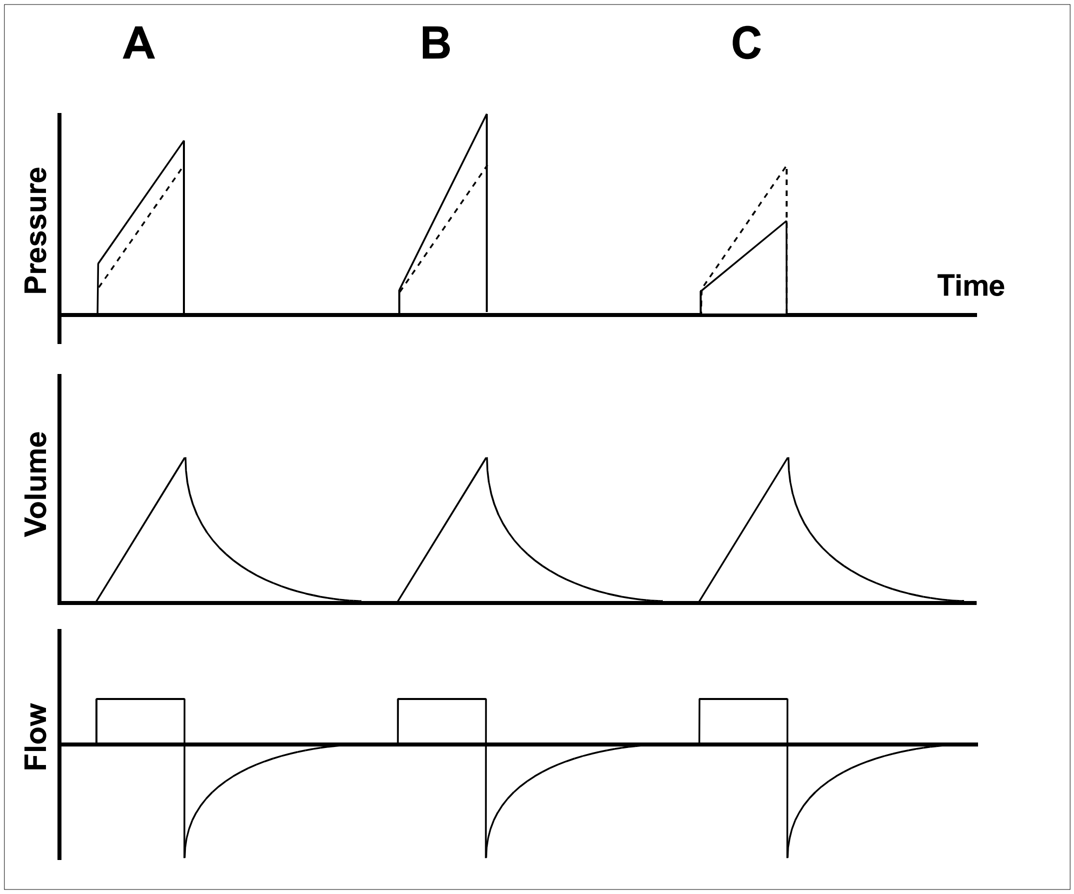
We know that the resistive load is the product of resistance and flow. Thus, if flow stays the same and resistance increases, the resistive load increases and the height of the rectangle in Figure 5-1A would be greater. Thus, the peak pressure in Figure 5-1C would be higher because the triangle sits on a higher rectangle. The same effect would occur if resistance stayed the same and the flow increased.

Here is a little harder question. What happens to peak inspiratory pressure if the patient suddenly gets a massive pneumothorax and one lung collapses? The first thing we have to recognize is that if the remaining lung must accept all the delivered tidal volume, it will take more pressure to expand it. If it takes more pressure to deliver the same volume, then we know that elastance has increased (remember that elastance = Δpressure/Δvolume). In reality, airway resistance may also increase because of the loss of airways, but we will ignore that effect here.

We also know that the elastic load is the product of elastance and volume. Thus, when the patient gets the pneumothorax, the elastic load increases. This is reflected in a higher triangle in Figure 5-1B and a higher peak inspiratory pressure in Figure 5-1C. Also, we said the slope of the inspiratory pressure line is proportional to elastance, so the higher the elastance, the steeper the line.

The results of these analyses are shown in Figure 5-2.

Figure 5-2. Effects of changing respiratory system mechanics on airway pressure during volume controlled ventilation. Dashed line shows original waveform before the change **A** Increased resistance causes an increase in the initial pressure at the start of inspiration and a higher peak inspiratory pressure and higher mean pressure. **B** An increase in elastance (decrease in compliance) causes no change in initial pressure but a higher peak inspiratory pressure and higher mean pressure. **C** A decrease in elastance (increase in compliance) causes no change in initial pressure but a lower peak inspiratory pressure and lower mean pressure.

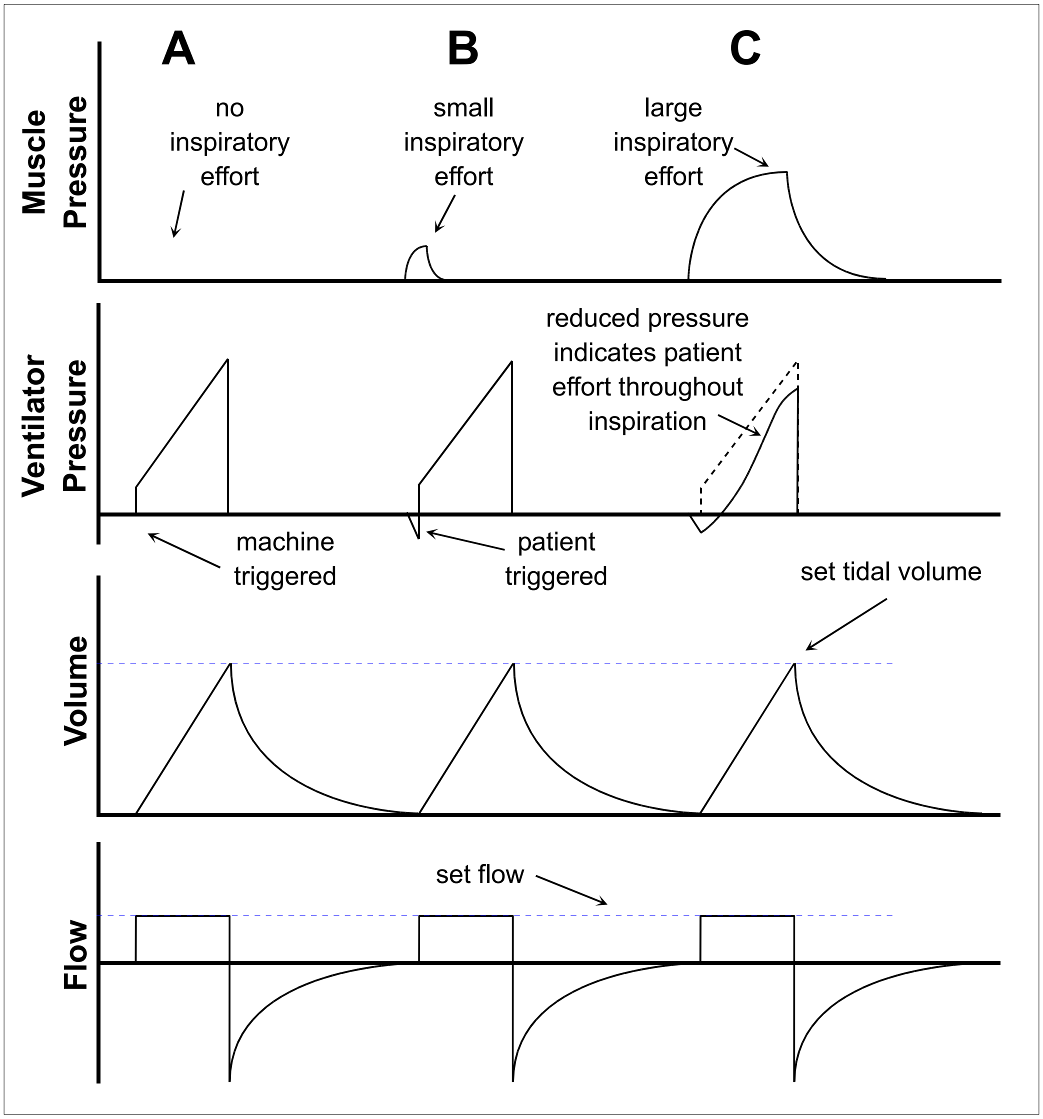


While we are on the subject of mean airway pressure, what happens if ventilatory frequency is increased? From Table 4-1 again we see that of the four variables that determine mean airway pressure, frequency is not mentioned. But from Figures 4-1, 4-3 and Table 4-1 we see that frequency is related to I:E ratio. Frequency can be increased while I:E is held constant (by decreasing both T_I and T_E) in which case mean airway pressure is unchanged. Alternatively, if frequency is altered by changing I:E, then mean airway changes in the same direction as I:E. For example, increasing frequency by decreasing expiratory time increases the I:E ratio and increases mean airway pressure.

The analyses above were for volume controlled ventilation with a constant inspiratory flow. Similar analyses can be applied to volume controlled ventilation with other flow waveforms and to pressure controlled ventilation. Take another look at Figure 4-4. Perhaps now you can better appreciate the use of shading and the dashed lines indicating mean airway pressure.

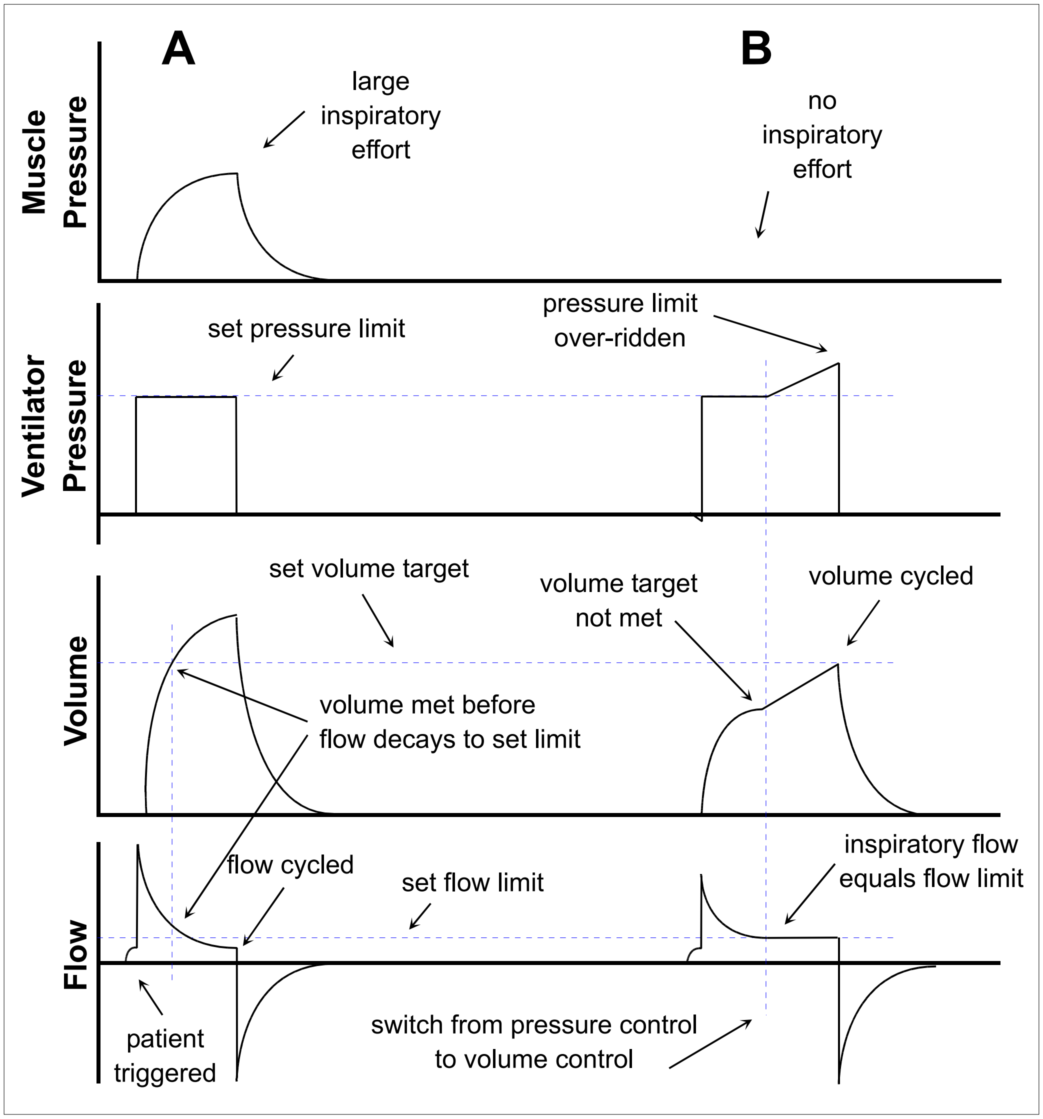
Continuous Mandatory Ventilation (CMV)

VC-CMV



Breaths during volume controlled continuous mandatory ventilation can be machine triggered (A) or patient triggered (B). Although VC-CMV is often chosen to rest the patient, patient effort does not necessarily stop. The area of the pressure curve *under* the baseline is proportional to the work the patient does on the ventilator (imposed work). The area under the pressure curve *above* the baseline is the work the ventilator does on the patient. The difference in the areas between passive inspiration (A) and active inspiration (C) is proportional to the work the patient does during inspiration.

DC-CMV (dual control within breaths: pressure control to volume control)

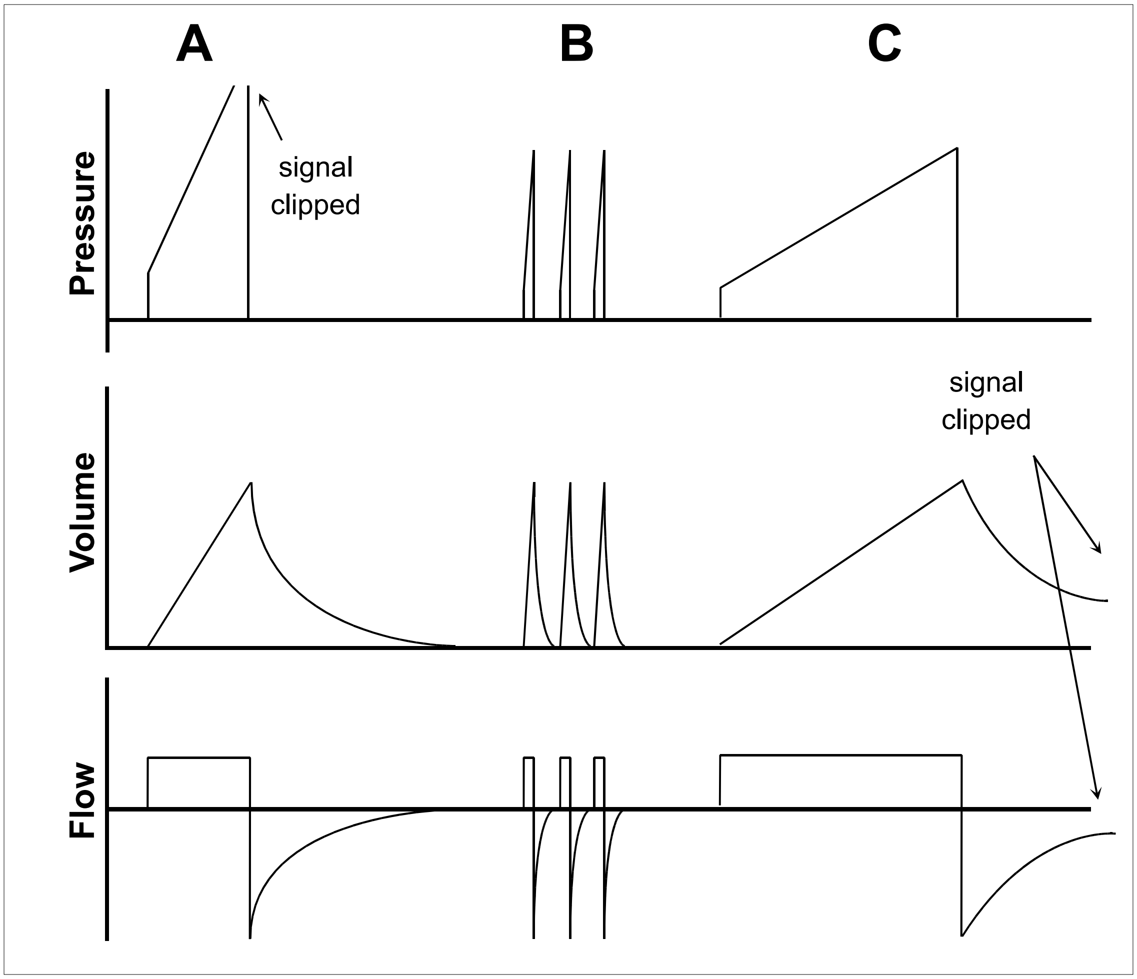


In (A) the patient's inspiratory effort is assisted by a Pressure Support type breath. The tidal volume target is met before flow decays to the set flow limit, and the breath is flow cycled. If the patient effort decreases enough that tidal volume is not met by the time flow decays to the set flow limit, then the control variable switches from pressure to volume (B). The breath proceeds at the set flow limit and is volume cycled when the target tidal volume is met. Notice that when this happens, airway pressure rises above the set pressure limit. Examples of this mode are Volume Assured Pressure Support on the Bird 8400 and Tbird ventilators and Pressure Augmentation on the Bear 1000 ventilator.

How to Detect problems

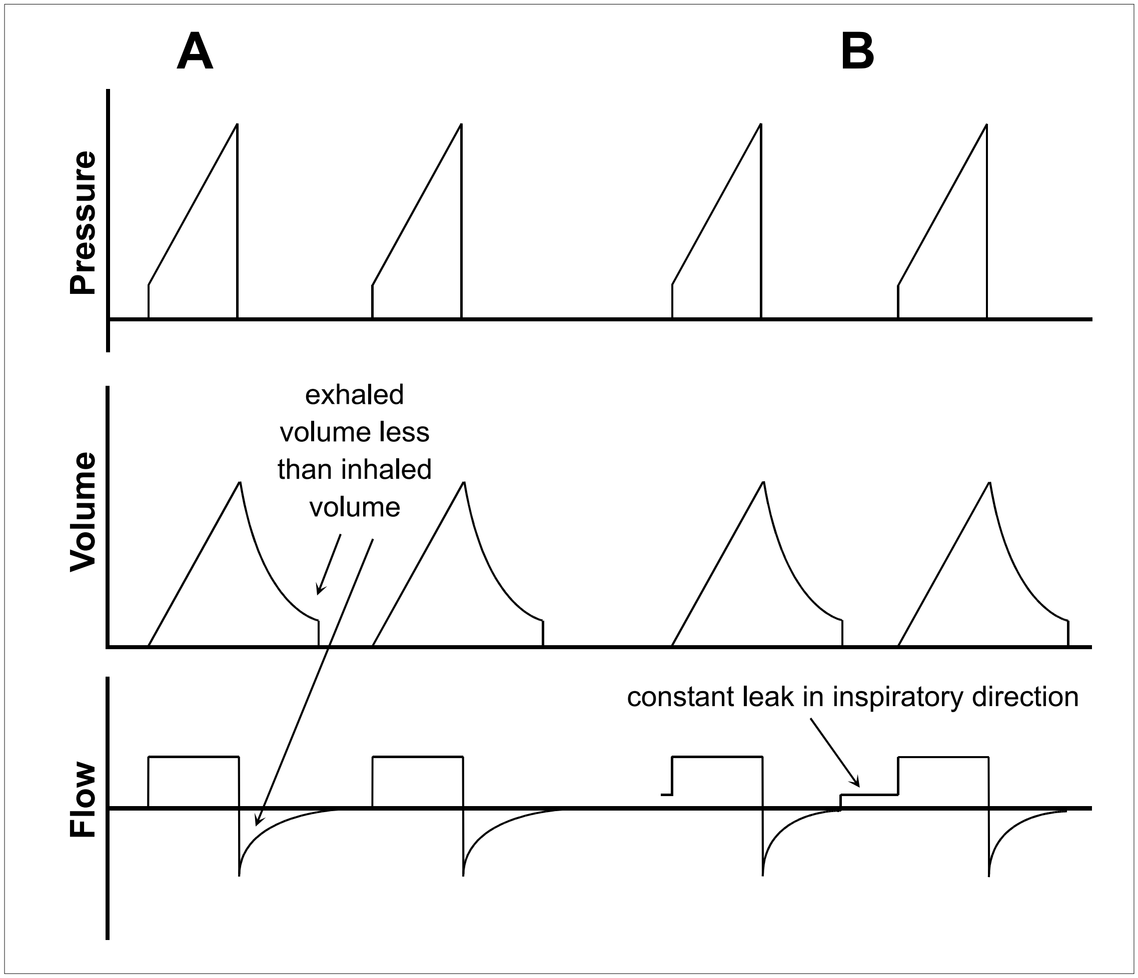
Assessing Overall Quality

Appropriate Scales



(A) The pressure scale is too small and the waveform is cut-off or "clipped". You cannot see the peak inspiratory pressure. (B) The sweep speed is too slow. This makes the time scale too long and you lose detail in the waveforms. (C) The sweep speed is too fast. The time scale is too short, the expiratory waveforms are clipped and you cannot see a whole breath. These problems are avoided with ventilator monitors that automatically adjust scales. You may run into these problems with stand alone monitors or if you are doing research using strip chart recorders.

Leak



(A) Intermittent leak, as through a chest tube or around an uncuffed endotracheal tube. Volume leaks out mostly during inspiration when pressure is elevated and the airways are dilated. The volume waveform abruptly returns to zero because the integrator resets when the software thinks expiration has ended (ie, flow crosses zero). However, it is clear that the height of the inspiratory portion of the volume curve is bigger than the expiratory portion. Flow returns to zero but the area under the flow curve (which is proportional to volume) is smaller for expiration than inspiration. These are both signs that the exhaled volume is less than the inhaled volume. (B) Constant leak. Notice the same signs of leak as in (A) but the leak is present throughout the breath. This is indicated by the expiratory flow curve that remains above zero, indicating a constant positive (inspiratory direction) flow. Large leaks in the patient circuit or through a chest tube may do this. The CPAP level may be reduced by a large leak if the ventilator cannot compensate.

Loop Displays

Loop displays plot one control variable against another as compared to waveform displays that plot individual control variables against time. This allows a rapid assessment of the patient's respiratory system compliance and resistance. Recall that compliance is the change in volume for a given change in pressure. Graphically, that is equivalent to the slope of a curve with pressure on the horizontal axis and volume on the vertical axis. Thus, plotting the pressure-volume curves for inspiration and expiration yield information about compliance. Resistance is assessed from a plot of volume versus flow. As with the waveform displays, we will discuss idealized loops first and then show how real loops look. Then we will review briefly the ways loops may be interpreted.

Pressure-Volume Loop

Pressure-volume loops are commonly used both in clinical situations and for research purposes.

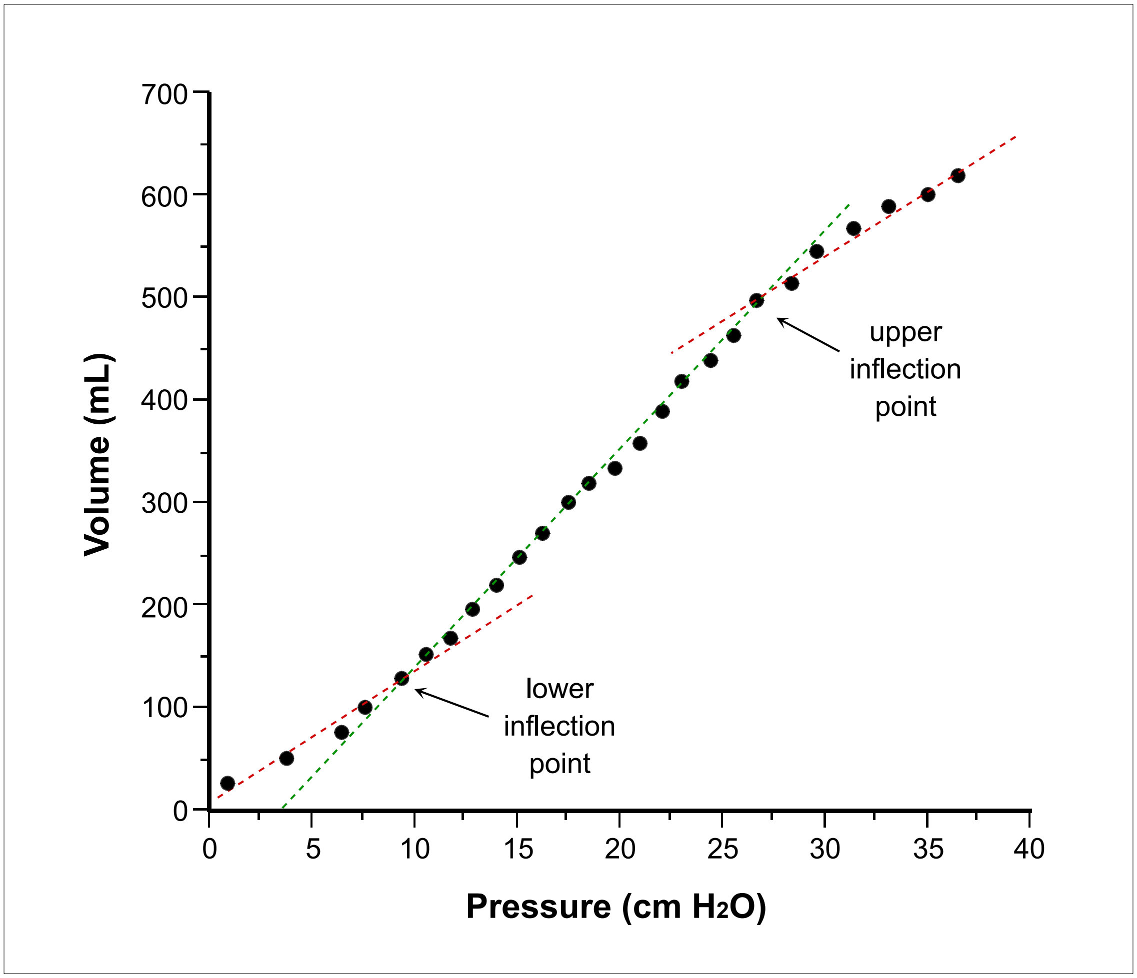
Key Idea

There are two distinct types of pressure-volume loop and it is very important not to confuse them. They are the *static pressure-volume loop* and the *dynamic pressure-volume loop*. The static loop has been used by physiologists to describe lung characteristics for decades. It is produced by injecting an isolated set of lungs or the intact respiratory system with known amounts of gas and recording the associated pressures. In practice, a large, calibrated "super syringe" is used to deliver incremental volumes of 50 to 100 mL from functional residual volume up to total lung capacity (TLC). Static airway pressure is measured following a 1 to 2 second pause after each volume increment. Stepwise inflation is continued up to a predetermined maximum pressure (usually about 35-40 cm H_2O, which normally corresponds to TLC). The same measurements are made during stepwise deflation. Corrections for temperature, humidity, compressible volume and pulmonary gas exchange are required for accurate volume measurement.

The purpose of the pauses between volume increments is to remove any effects of resistance on the pressure measurements. That way, the pressure volume curve can be used to assess respiratory system compliance. This is a cumbersome clinical procedure because it requires that the patient be paralyzed, disconnected from the ventilator and stable enough to last through the whole process. Unfortunately, the patients you most want the data from are often so sick they cannot tolerate the procedure. An alternative technique is to inflate the lungs steadily with a very low constant flow (1-2 L/min). With the slow flow, resistance effects are negligible. This not only speeds up the procedure somewhat but also makes it possible for the ventilator to do it automatically. To date, only one ventilator has this capability (Hamilton Galileo).

The dynamic pressure-volume curve is what you see on ventilator monitors during normal mechanical ventilation. It is simply a plot of airway pressure on the horizontal axis versus volume on the vertical axis at normal inspiratory and expiratory flows.

Static Pressure-Volume Loop (compliance curves)



Static pressure-volume curve of a patient with ARDS during inspiration. The dotted lines represent areas of approximately linear compliance. The slope of a line is the compliance:

$$slope = \Delta\ volume/\Delta\ pressure = compliance$$

Key Idea

> Maximum compliance is found between the two inflection points. This is where tidal ventilation should occur to minimize lung damage. An inflection point is a transition from one linear compliance to another. Compliance is less below the lower inflection point due to alveolar collapse. Compliance above the upper inflection point is also less due alveolar over-distention. There is no universally accepted technique for determining the inflection points and it is often done visually. Also, inflection points are not always evident on any particular patient.

Flow-Volume Loop

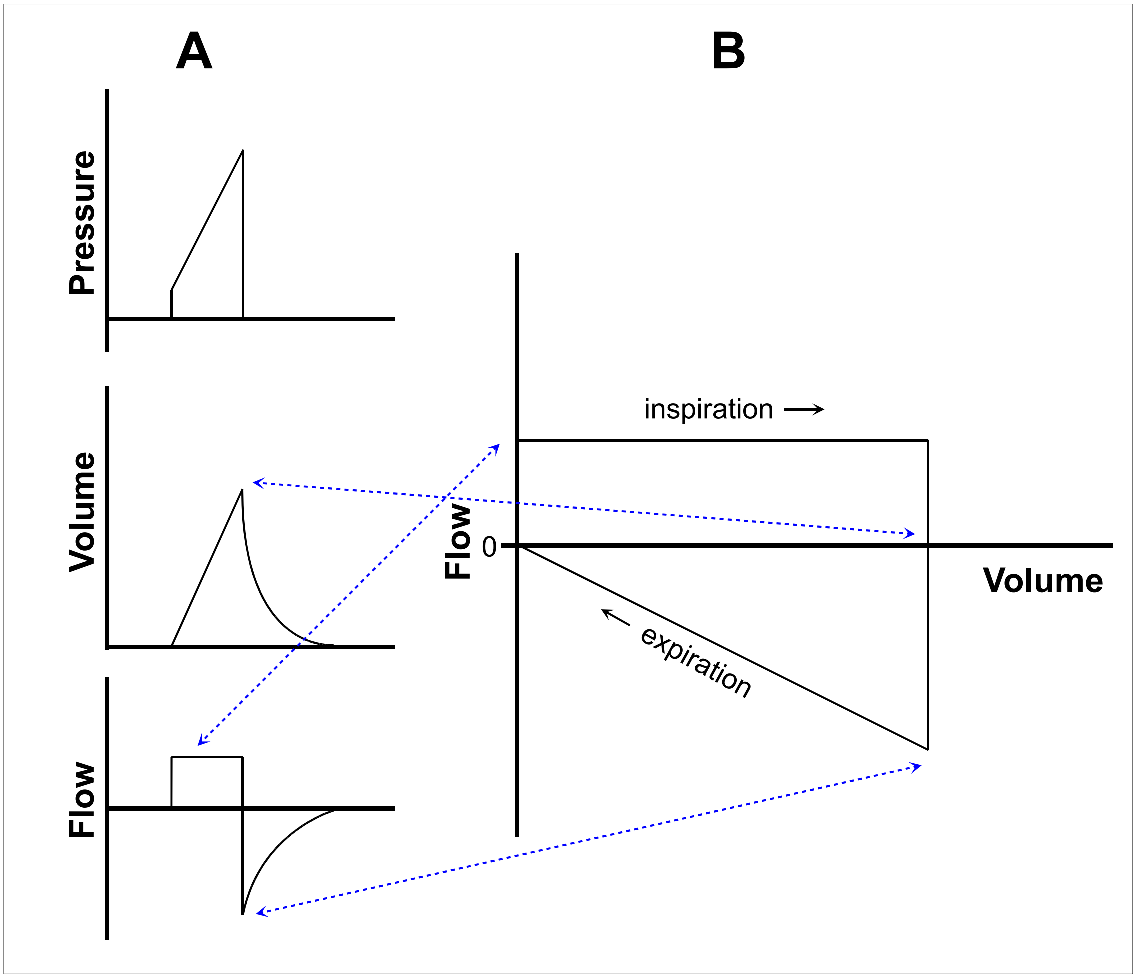
Flow volume loops have traditionally been used in the pulmonary function lab to diagnose lung disease. Flow-volume loops generated during mechanical ventilation are different in three major respects from those obtained in the pulmonary function lab:

1. Flow volume loops obtained during mechanical ventilation are normally passive, compared to the forced vital capacity maneuver required of patients in the lab.
2. The shape of the inspiratory portion of the loop during mechanical ventilation is determined by the flow waveform generated by the ventilator.
3. The expiratory portion of the loop does reflect changes in airway resistance but does not necessarily indicate maximal flow limitation as in pulmonary function studies.

One practical problem with flow-volume loop displays is that there is no consensus on how the axes should be oriented. Sometimes flow is shown on the vertical scale and sometimes on the horizontal scale. Sometimes inspiration is shown as a positive (upward) flow and sometimes it is downward. For consistency with the previous graphs, the flow-volume loops in this book will show flow on the vertical scale with inspiration in the positive, upward direction.

Flow-volume loops can be used to detect autoPEEP and leaks like waveforms can, but they are most useful in the evaluation of bronchodilator response. As you will see below, examination of the expiratory portion of the flow-volume loop clearly shows a change in the shape of the curve as resistance changes. Another practical use of flow-volume waveforms is to access asynchrony during pressure controlled ventilation of neonates. If the infant breathes spontaneously, and out of phase with mandatory breaths, tidal volume delivery will be highly variable. This problem can be minimized by proper setting of the sensitivity or if SIMV is not available, setting the ventilatory frequency to hopefully entrain the patients spontaneous rate (ie, to get the patient to breathe at the same rate or some even multiple of the ventilator rate). Synchrony between the patient and ventilator and the resultant decrease in tidal volume variability is easily seen with flow-volume loops that overlap for several breaths. Low tidal volume variability is desirable not only to stabilize gas exchange but also to minimize the risk of intracranial bleeds.

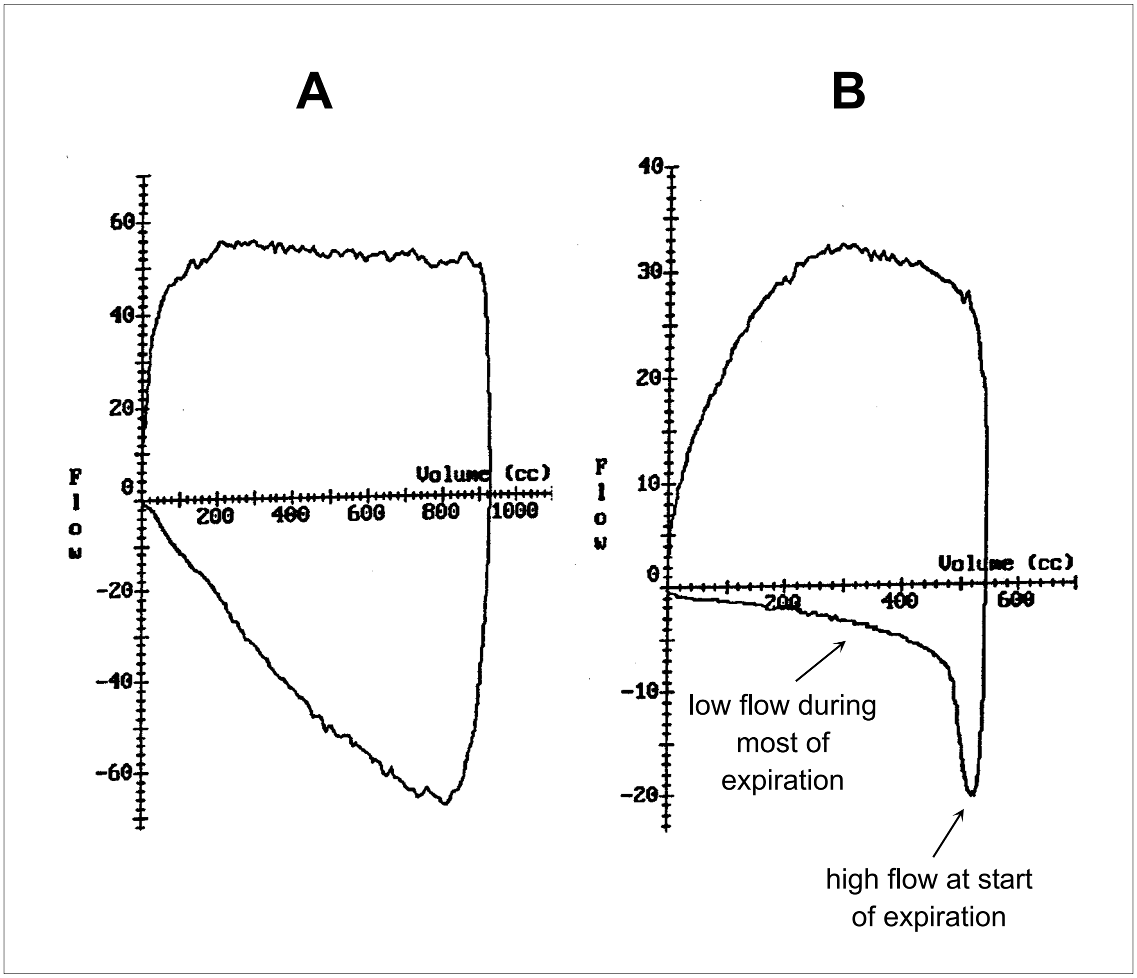
Volume Control (idealized)



Idealized waveforms (A) for volume controlled ventilation with corresponding idealized flow-volume loop (B). Note that the loop is drawn clockwise, with inspiration going to the right and down and expiration to the left and up. The dotted arrows show the correspondence between the waveform display and the loop display for the initial flow rise and tidal volume. The shape of the inspiratory portion of the flow-volume loop is dominated by the constant flow, which hides the change in volume and results in a horizontal line.

During expiration, both volume and flow decrease exponentially with the same time constant (see the discussion about time constants in Chapter 4). When one exponential function is plotted against another and there is a single time constant for the system, the result is a straight line. This can be proven mathematically but you can understand it intuitively by recognizing that if volume and flow are changing at the same rate (ie, the same time constant) then the ratio of flow to volume (the slope of the expiratory portion of the flow-volume curve) must remain constant. A curve with a constant slope is a straight line.

Volume Control (real; normal vs COPD)



Actual flow-volume loop from a ventilator monitor during volume controlled ventilation of a normal patient (A) and one with chronic obstructive lung disease, COPD (B).

Patients with normal lungs have relatively straight expiratory curves while those with marked airway collapse (eg, COPD patients) may show a biphasic curve. If the expiratory portion of the curve is not a straight line, it indicates that the respiratory system really does not behave as if it has only a single time constant. In patients with lung disease, lung units will be affected to different degrees and each will have its own time constant.

The high flow at the start of expiration in B is due to the gas compressed in the patient circuit, the relatively low flow for the rest of expiration comes from the lungs and may represent flow limitation due to collapsed airways.

Calculated Parameters

Most of the ventilators that provide graphic displays also show various calculated parameters. Unfortunately, ventilator operator's manuals do not always define these terms or explain the equations used. The following parameters are the most common.

Mean Airway Pressure

Mean airway pressure is defined as the average pressure at the airway opening over a given time interval. It is defined graphically as the area under the pressure time curve for one breath cycle divided by the total cycle time. If all breaths are identical, then the average pressure for one breath is the average over any period. However, because mandatory breaths may be mixed with spontaneous breaths and because the pressure waveforms for each of these may change on a breath-by-breath basis, mean airway pressure is usually calculated as a moving average over several breath cycles. The ventilator measures airway pressure every few milliseconds for a few breath cycles. Then it sums the measurements and divides by the number of measurements to get the average. The average is "moving" because as the measurement for the next breath come in, they replace those of the first breath in the average. For example, the Hamilton Galileo calculates mean airway pressure as the moving average of 8 breath cycles. In contrast, the Puritan Bennett 840 displays mean airway pressure for each breath.

Figure 5-8 shows two ways to calculate mean airway pressure. The graphical method is based on the mean value theorem from calculus.[1] This theorem says that the mean value of a periodic waveform is the constant value that will give the same area as the waveform over the same period. The area under the waveform on the left in Figure 5-8 can be represented by the number of boxes between the pressure curve and the time axis. Notice that the constant (mean) pressure on the right has the same area.

The numerical method is used by ventilators and is just the arithmetic average of a large number of pressure measurements. The shorter the sampling period, the larger the number of pressure measurements the more closely the waveform is followed and the more accurate the estimate of mean airway pressure. This method may be extended over several breaths to give a moving average that does not change as quickly as a breath-by-breath display.

Mean airway pressure can also be estimated by hand based only on peak inspiratory pressure (PIP), positive end expiratory pressure (PEEP) and the duty cycle:

$$\overline{P}_{AW} = k(PIP - PEEP)(T_I / TCT) + PEEP$$

where k is the waveform constant whose value is 1.0 for a rectangular waveform (pressure control ventilation) and 0.5 for a triangular waveform (volume control ventilation), T_I = inspiratory time and TCT is total cycle time (respiratory period).

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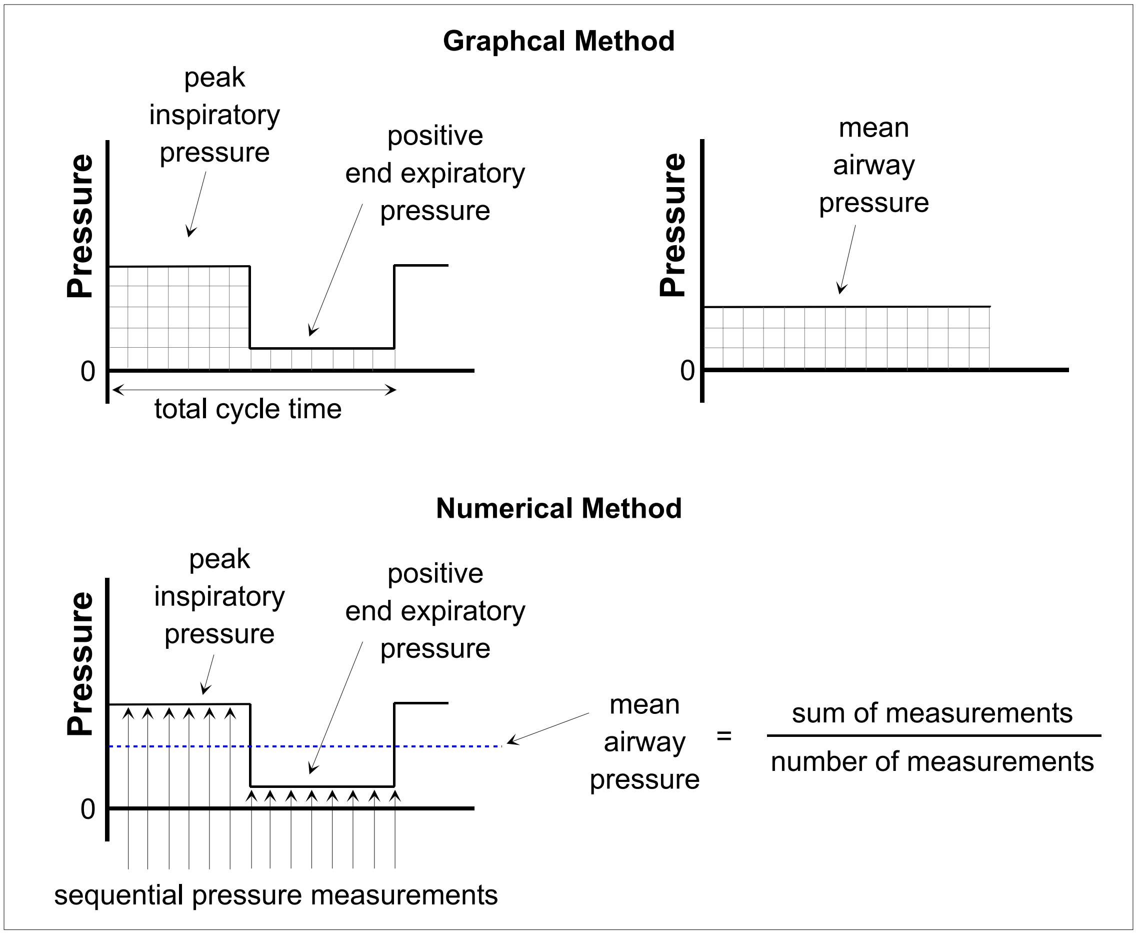
[1] Crit Care Med 1982;10(6):378-383.

Figure 5-8. Two methods of calculating mean airway pressure.



Leak

The leak volume is the difference between the inspired tidal volume and the expired volume. It may be a moving average calculated over several breaths. This value is particularly important in patients with uncuffed endotracheal tubes or with broncho-pleural-cutaneous fistulas.

Calculating Respiratory System Mechanics: Static vs Dynamic

There are two methods used to calculate compliance and resistance at the bedside; a static method and a dynamic method. Both of these methods rely on pressure measurements taken at the airway opening. As we have discussed many times, airway pressure has two components, one due to compliance and volume; the other due to resistance and flow. The technical problem in calculating mechanics is to separate these two components without having to make pressure measurements within the lung (ie, to obtain transairway pressure for resitance and transthoracic pressure for compliance). The solution is to make the measurements in such a way that the pressure measured at the airway opening reflects only the component of interest. The "trick" is to realize that we are measuring a change in pressure not only between two points in space (ie, airway

The static method accomplishes this by taking measurements at the start of inspiration (when flow and volume are zero) and at the end of an inspiratory hold maneuver (when flow is zero and volume is the tidal volume). Because flow is zero at both times, airway pressure is the same as the pressure in the lung. The pressure change between those times is due only to the volume change. Compliance can then be calculated as the volume change (tidal volume) divided by the airway pressure change (plateau pressure minus PEEP).

In a similar fashion, if we take measurements at two times when flow is different but volume is the same, we can calculate resistance. These two times are at the end of inspiratory flow time and at the end of inspiratory hold time. Thus, resistance is simply the pressure change (peak inspiratory pressure minus plateau pressure) divided by the change in flow. The change in flow is the flow at the end of the inspiratory flow time minus the flow at end expiratory time (zero) which simplifies to just the end inspiratory flow.

The dynamic method of calculating respiratory system mechanics was designed for situations when an inspiratory hold is not practical. The basic idea is still the same; compliance is assessed by making measurements of pressure between two points in time when flow is zero (start and end inspiration) and resistance is assessed between two points in time when volumes are equal. However, because we do not have the luxury of a few milliseconds of inspiratory hold to read the pressure values from the ventilator's gauge, we need to analyze pressure, volume, and flow waveforms. Pressure measurements for compliance are read off the graph at times when the flow waveform crosses zero. Pressure measurements for resistance are read off the graph at times when the volumes are equal, usually mid inspiration and mid expiration.

While the dynamic method described above has been used extensively for research, it is not very practical for routing bedside calculations. For that reason, ventilator manufacturers have made use of the microprocessor's capability of making rapid measurements and calculations to simplify things. Because the ventilator is already making continuous pressure, volume and flow measurements for control and alarm purposes, they simply use that data for the additional purpose of calculating resistance and compliance. This is done by fitting the equation of motion to the data using linear regression as described below in the next section. The advantage of this, besides being automated, is that both inspiratory and expiratory parameters can be calculated on a breath-by-breath basis without disturbing the patient's breathing pattern.