Chapter 10: Pulmonary Mechanics

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Introduction

For venous blood to be properly arterialized, the distribution of air and blood within the lung is automatically matched to ensure effective gas exchange across alveolar–capillary membranes. Arterialization comprises a series of interrelated processes that begin with the mechanical performance of the ventilatory apparatus—that is, the lungs and the chest wall, including the rib cage, diaphragm, and abdominal wall. The ventilatory apparatus is critical for replenishing fresh air to the lungs for gas exchange. Although the function of each component of the lung and of the chest bellows can be deranged by injury or disease, the design of the ventilatory apparatus provides for considerable reserve. As a result, mechanical derangements are usually quite severe by the time clinical symptoms appear or arterial blood-gas levels become abnormal.

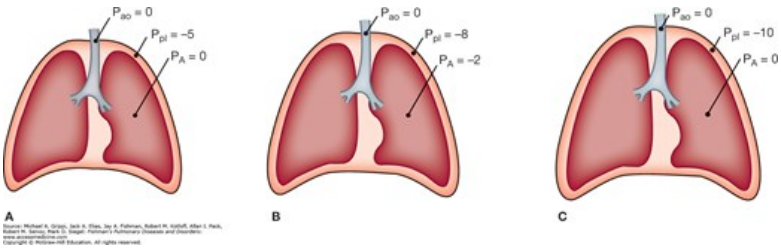
Depending on the nature of the underlying disorder, assessment of the mechanical properties of the ventilatory apparatus provides several different types of information. In some instances, characterization of the mechanical abnormality provides insight into pathogenesis and affords a quantitative measure of severity. In others, once the nature of the mechanical disorder is understood, the mystery surrounding a life-threatening disorder in gas exchange may be dispelled. Finally, certain breathing patterns make sense only if the mechanical performance of the chest bellows is taken into account.

During breathing, the lungs and chest wall operate in unison. The lungs fill the chest cavity so that the visceral pleura are in contact with the parietal pleura of the chest wall. The two pleural surfaces are separated by only a thin liquid film, which provides the bond holding the lungs and chest wall together.

At the end of a normal exhalation when the respiratory muscles are at rest, the ventilatory apparatus is in a state of mechanical equilibrium. The pressure along the entire tracheobronchial tree from the airway opening to the alveoli is equal to atmospheric pressure. The tendency of the lung is to deflate, however, and lung elastic recoil is directed centripetally. This is counterbalanced by the elastic recoil of the chest wall, which is directed centrifugally to favor an increase in volume. These opposing forces generate a subatmospheric pleural pressure of about −5 cmH_2O (Fig. 10-1A). The tendency for the lung to recoil inward and for the chest wall to recoil outward is illustrated by the observation that when the chest is opened at autopsy, the lungs collapse to a nearly airless state and the thorax expands.

Figure 10-1

Respiratory pressures during a breathing cycle. **A.** End expiration. **B.** During inspiration. **C.** End inspiration. P_{pl}, pleural pressure; P_A, pressure in the alveoli; P_{ao}, pressure at the airway opening.



Although it is conventional to consider pleural pressure as a single, mean value that reflects mechanical events within the entire ventilatory apparatus, this is clearly an oversimplification on several accounts: (1) pleural pressure is not directly determinable because normally there is only a potential space between the visceral and parietal pleura; (2) on conceptual grounds, distinctions exist between surface and liquid pleural pressures; (3) pleural

pressures are not uniform over the surface of the lungs, being strongly affected by gravity; and (4) transmission of pleural pressures at the surface to alveoli located at different depths and loci with the lungs depends on the structural interplay among supporting structures in the alveolar walls (interdependence), which resists any inclination of individual alveoli or even a lobule to collapse.[1] Nonetheless, the concept of mean pleural pressure, as generally used in considerations of respiratory system mechanics, has proved to be of great practical value.[2]

The contraction of the muscles of inspiration produces the forces that permit the flow of gas along the tracheobronchial tree and the expansion of the lungs and chest. The movement of air into the lungs requires a pressure difference between the airway opening and the alveoli sufficient to overcome the resistance to airflow of the tracheobronchial tree. Also, a pressure difference across the alveolar walls (between the alveoli and pleural space) must be generated to overcome elastic recoil and inflate the lungs. During spontaneous breathing, the action of the inspiratory muscles causes an increased outward pull on the chest wall.[3] As a result, the pleural pressure becomes more subatmospheric. This pressure change is transmitted to the interior of the lungs, so alveolar pressure also becomes subatmospheric (Fig. 10-1B). In contrast, during artificial ventilation with a positive-pressure ventilator, a supra-atmospheric pressure applied at the inlet to the airways creates the proper pressure gradient between the airway opening and alveoli for airflow.

Expansion of alveoli depends on the achievement of an appropriate distending pressure across alveolar walls. This distending pressure or transpulmonary pressure is the difference between alveolar (P_A) and pleural (P_{pl}) pressures. As shown in Figure 10-1A, the transpulmonary pressure at end expiration ($P_A - P_{pl}$) is 5 cmH_2O. At the end of inspiration (Fig. 10-1C), the lungs contain more air and the distending pressure which also represents the recoil pressure is greater.

The energy used during inspiration to overcome the elastic resistance of the lungs is stored. Expiration occurs when these forces are released. When the inspiratory muscles relax, the recoil of the lungs causes the alveolar pressure to exceed the pressure at the mouth, and air flows out of the lungs. Although expiration during quiet breathing is passive, the expiratory muscles are engaged at high levels of ventilation to assist the movement of air out of the lungs.

Lung Volumes

The lung volumes and capacities (Table 10-1) are also considered elsewhere in this book (see Appendix B). The end-expiratory position of the lungs, functional residual capacity (FRC), is the major reference point for the subdivisions of lung volume. This position is set by the opposing recoil forces of the lung and chest wall when the respiratory muscles are at rest.

Table 10-1

Lung Volumes and Subdivisions

The *functional residual capacity* (FRC) is the volume of air that remains in the lungs at the end of a normal expiration.
The *tidal volume* (TV) is the volume of air that is drawn into the lungs during inspiration from the end-expiratory position (and also leaves the lungs passively during expiration) in the course of quiet breathing.
The *expiratory reserve volume* (ERV) is the maximum volume of air that can be forcibly exhaled after a quiet expiration has been completed (i.e., from the end-expiratory position).
The *residual volume* (RV) is the volume of air that remains in the lungs after a maximal expiratory effort.
The *inspiratory capacity* (IC) is the maximum volume of air that can be inhaled from the end-expiratory position. It consists of two subdivisions: tidal volume and the *inspiratory reserve volume* (IRV).
The *total lung capacity* (TLC) is the total volume of air contained in the lungs at the end of a maximum inspiration.
The *vital capacity* (VC) is the volume of air that is exhaled by a maximum expiration after a maximum inspiration.

Total lung capacity (TLC), the total volume of air contained in the lungs after a maximal inhalation, is determined by the balance between the force-generating capacity of the inspiratory muscles and the opposing elastic recoil forces of the lung and chest wall.[4] Weakness of the muscles of inspiration or increased stiffness of the lung reduces TLC. Loss of retractive forces exerted by the lung, as in emphysema, enlarges TLC.

Residual volume (RV), the volume of air remaining in the lungs after a complete exhalation, is set by the balance between the actions of the expiratory muscle and the recoil forces of the lung, which act to decrease lung volume, and the outward recoil forces of the chest wall, which favor lung expansion. In middle-aged and older individuals, closure of airways at low lung volumes, with air trapping in the lung, is an important determinant of RV.[5]

Static Mechanical Properties of the Respiratory System

To assess the elastic properties of the ventilatory apparatus, it is expedient to evaluate the elastic properties of the lungs and chest separately. Elastic properties are conventionally assessed over a fixed range of volumes during periods of arrested airflow.

Elastic Properties of the Lungs (Pulmonary Compliance)

The change in transpulmonary pressure required to effect a given change in the volume of air in the lungs is a measure of the distensibility, or compliance, of the lungs. Pulmonary compliance is calculated as the ratio of the change in lung volume to the change in transpulmonary pressure—that is,

$$C = \frac{\Delta V_L}{\Delta(P_A - P_{pl})}$$

where

$$C = \text{lung compliance}$$
$$\Delta(P_A - P_{pl}) = \text{change in transpulmonary pressure}$$
$$P_A = \text{alveolar pressure and } P_{pl} = \text{pleural pressure}$$
$$\Delta V_L = \text{change in lung volume}$$

Compliance denotes distensibility, the ease of stretch or inflation. The inverse of compliance (i.e., elastance) refers to the stiffness or the tendency to

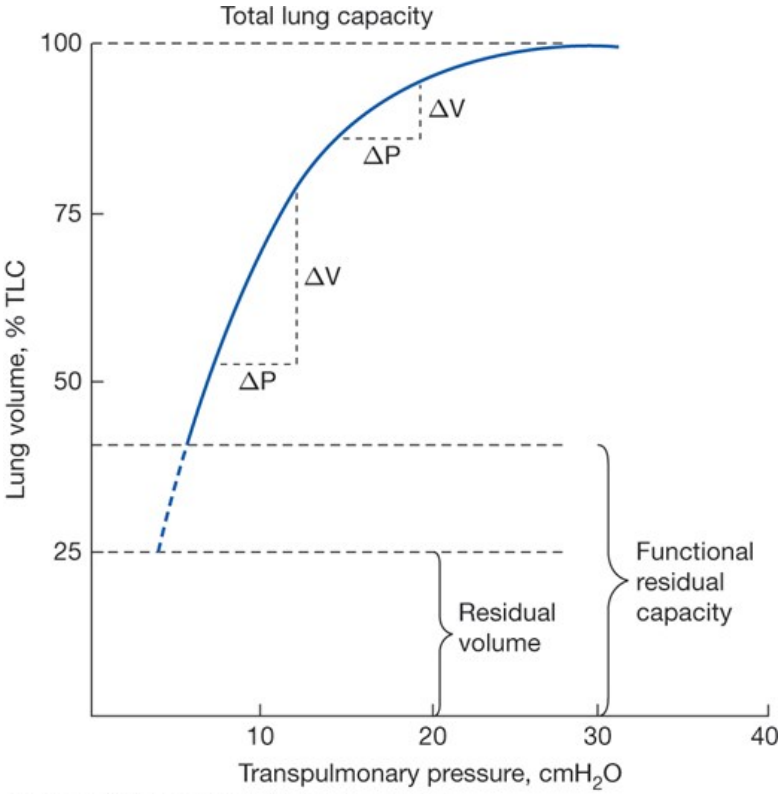
resist distortion and to return to the original configuration when the distorting force is removed.

In practice, pulmonary compliance is determined by relating the changes in transpulmonary pressures to the changes in lung volume during interruptions in the course of an expiration after a maximal inspiration (i.e., starting from TLC).

The pressure–volume characteristics of the lung are nonlinear. As lung volume increases, the elastic elements approach their limits of distensibility, and a given change in transpulmonary pressure produces smaller and smaller increases in lung volume.[6,7] Thus, the compliance of the lung is least at high lung volumes and greatest as RV is approached (Fig. 10-2). Elastic recoil forces favoring collapse of the lung can be demonstrated throughout the range of the vital capacity, even at low lung volumes approaching the RV. If the opposing forces of the chest wall on the lungs are eliminated – for instance, by removing the lungs from the thorax or by opening the chest – the lung collapses to a near-airless state. A minimal volume of air does remain in the lungs because of closure of small airways resulting in the trapping of air in more distal airspaces.

Figure 10-2

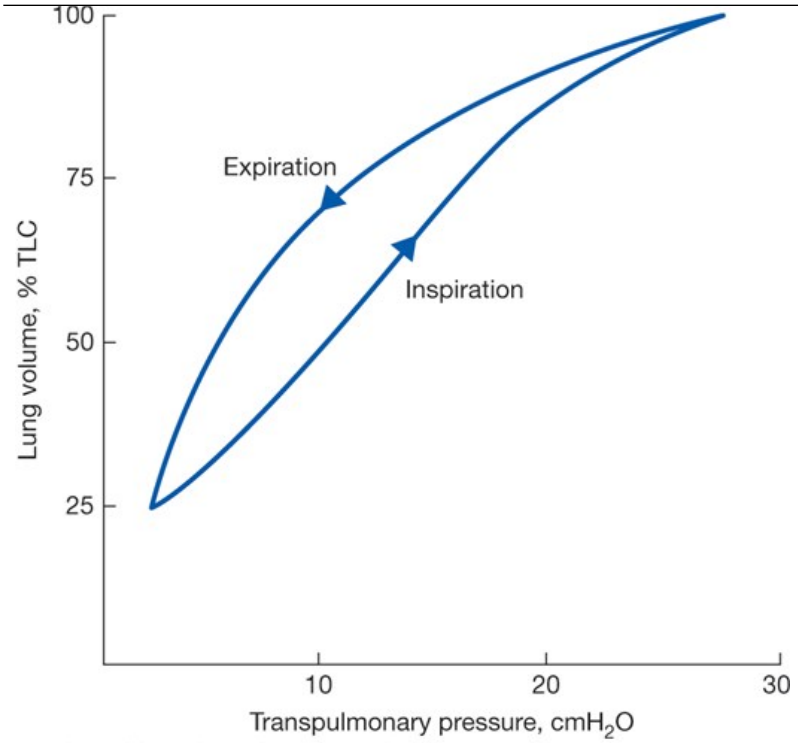
Pressure–volume curve of the lung. The static elastic recoil pressure of the lung is approximately 5 cmH_2O at FRC and 30 cmH_2O at TLC. The compliance of the lung (ΔV/ΔP) is greater at low lung volumes than at high lung volumes.



If static measurements of transpulmonary pressure are made during lung inflation rather than deflation, the pressure–volume curve has a different configuration (Fig. 10-3). This indicates that the elastic recoil of the lung depends not only on the lung volume at which the determination is made but also on the "volume history" of the lung.[8]

Figure 10-3

Pressure–volume curves of the lung during inspiration and expiration.



Hysteresis

Differences in the pathways of the static pressure–volume curve during inspiration (when force is applied) and expiration (when force is withdrawn) are designated as hysteresis, which is a property of all elastic structures. In the lungs, it is due to the surface forces and the properties of the surface material lining the alveolar walls and also to the elastic properties of the tissues. The tissues of the lung are also subject to stress adaptation whereby over time, the pressure required to maintain a given lung volume will decline.[9] An additional factor relates to the closure of small airways at low lung volumes. Once these airways close, the lung units that they serve will not expand during inspiration until a critical opening pressure has been exceeded; only then will the closed units inflate. Recruitment of additional lung units as increasing transpulmonary pressure expands the lungs from low lung volume contributes to the hysteresis of the pressure–volume curve.

The elastic behavior of the lung depends on two factors: the physical properties of the lung tissue, per se, and the surface tension of the film lining of the alveolar walls.

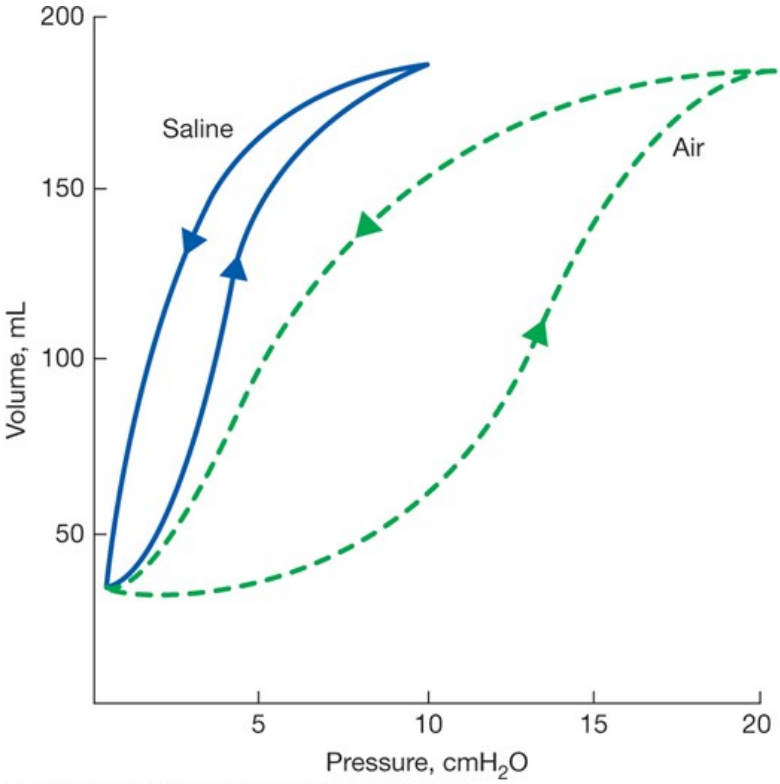
Surface Forces

The interior surfaces of the alveoli are lined by a thin liquid layer of osmophilic material. The surface tension at the air–liquid interface of the alveoli, in addition to the elastic properties of the parenchyma, contributes importantly to the elastic recoil of the lungs and acts to decrease lung compliance.[10] The cohesive forces between the molecules of the liquid lining of the alveoli are stronger than those between the film and alveolar gas, thereby causing the film to shrink to its smallest surface area. The behavior of this surface film has been examined in experimental animals by comparison of pressure–volume relationships of air-filled lungs with those of saline-filled lungs; saline eliminates the liquid–air interface without affecting elastic properties of the tissue. A lung distended with saline requires a lower transpulmonary pressure to maintain a given lung volume than a lung that is inflated with air.[11] Also, hysteresis is less in the saline-filled lung. The greater hysteresis in the air-filled lung is explained by the surface tension of the film lining the alveoli, which is higher during inflation as the film expands than it is during deflation as the film is compressed (Fig. 10-4).

Figure 10-4

Comparison of pressure–volume relationships of air- and saline-filled excised lungs. *Arrows* directed upward indicate inflation; those directed

downward indicate deflation. Since saline eliminates surface forces at the liquid–air interface without affecting tissue elasticity, the difference in pressure between the two curves, at any lung volume, is that required to overcome surface forces. To maintain a small lung volume, a large proportion of the pressure is used to overcome surface forces. In contrast, at high lung volumes a greater fraction of the pressure is used to overcome tissue elasticity.



By considering the alveolus to be a sphere, Laplace's law can be applied. Laplace's law states that the pressure inside a spherical structure—for example, the alveolus—is directly proportional to the tension in the wall and inversely proportional to the radius of curvature:

$$\text{Alveolar pressure} = \frac{2T}{r}$$

where

T = tension (dyn/cm)

r = radius

Abolition of the liquid–air interface by the instillation of saline into the alveolar spaces eliminates surface forces, thereby reducing the transpulmonary pressure required to maintain a given lung volume.

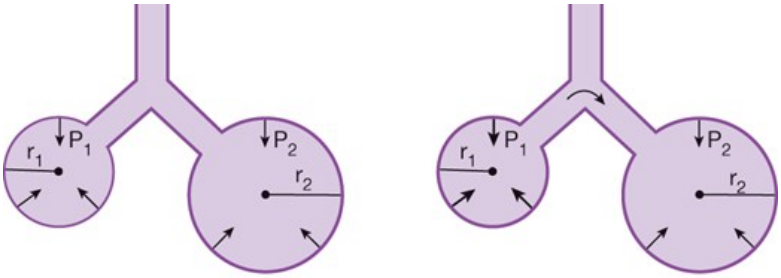
The surface film lining the alveoli of the lung is termed surfactant.[12] The superficial layer of the film facing the alveolar air is made up of surface-active phospholipids, notably dipalmitoyl lecithin. The deeper layer termed the hypophase consists of surface-active phospholipids linked to protein. Surfactant is generated by type II alveolar cells and undergoes a continuous cycle of formation, removal, and replenishment.[13]

Surfactant serves several important functions. The surface tension of surfactant is inherently low and decreases even further at low lung volumes when the surface area of the film is reduced. The minimization of surface forces, particularly at low lung volumes, minimizes the adherence of the walls of distal airways that tend to close at low lung volumes and increases the compliance of the lung and decreases the work required to inflate the lungs during the next breath. The automatic adjustment of surface tension as lung volume changes also promotes stability of alveoli at low lung volumes; if the surface tension were to remain constant instead of changing with lung volume, the transpulmonary pressure required to keep an alveolus open

would increase as the radius of curvature diminished with decreasing lung volume. Therefore, small alveoli would empty into the larger ones with which they communicate, and atelectasis would be a regular occurrence (Fig. 10-5). Surfactant dysfunction as occurs with acute lung injury results in marked increases in surface tension causing stiffening and instability of alveoli and leads to alveolar collapse.

Figure 10-5

The effects of surfactant in maintaining alveolar stability. **A.** Surfactant lowers the tension of the alveolar walls at low lung volumes. Consequently, the transpulmonary pressure (P) of large and small communicating airspaces is the same. $r_1 < r_2$, $T_1 < T_2$, $P_1 \leftarrow P_2$. **B.** Without surfactant, the surface tension remains constant as lung volume changes, and the recoil pressure of small airspaces exceeds that of larger ones. As a result, small alveoli tend to empty into larger ones. $r_1 < r_2$, $T_1 = T_2$, $P_1 > P_2$.



Interdependence and Collateral Ventilation

The low surface tension of surfactant is not the most important determinant of alveolar stability. In reality, the alveoli form a froth rather than individual bubbles.[14] The walls of each alveolus are shared in common with those of adjacent alveoli so that contiguous airspaces attached by their connective tissue framework are tethered to one another and are not free to move independently. The tendency of any one alveolus to collapse is opposed by the traction exerted by the surrounding alveoli. This mechanical interdependence of adjacent airspaces resists the collapse of individual alveoli and serves as a stabilizing influence and ensures uniform inflation.[15] Even when a distal airway is completely obstructed, the alveoli served by the airway can still be ventilated through collateral channels between alveoli (pores of Kohn) and from bronchioles to alveoli (canals of Lambert). This collateral ventilation also prevents alveolar collapse and enhances the uniformity of ventilation, particularly in patients with lung disease.[16]

Physical Properties of Lung Tissue

A number of different tissue components contribute to lung elasticity. The pleura, the intralobular septa, peripheral airway smooth muscle tone, and pulmonary vasomotor tone, as well as the tissues of the alveolar walls, play a role in shaping lung elastic recoil.

The major connective-tissue elements of the alveolar walls are the collagen and elastin fibers.[17] Elastin fibers in the alveolar walls and surrounding the bronchioles and pulmonary capillaries have a low tensile strength but can be stretched to over twice their resting length. Elastin fibers are thought to bear most of the stress in the lung at low volumes. Collagen fibers have high tensile strength but are poorly extensible and probably act to limit expansion at high lung volumes.[18] Like a stretched nylon stocking, expansion of the lungs appears to entail an unfolding and geometric rearrangement of the fibers and only slight elongation of individual fibers.

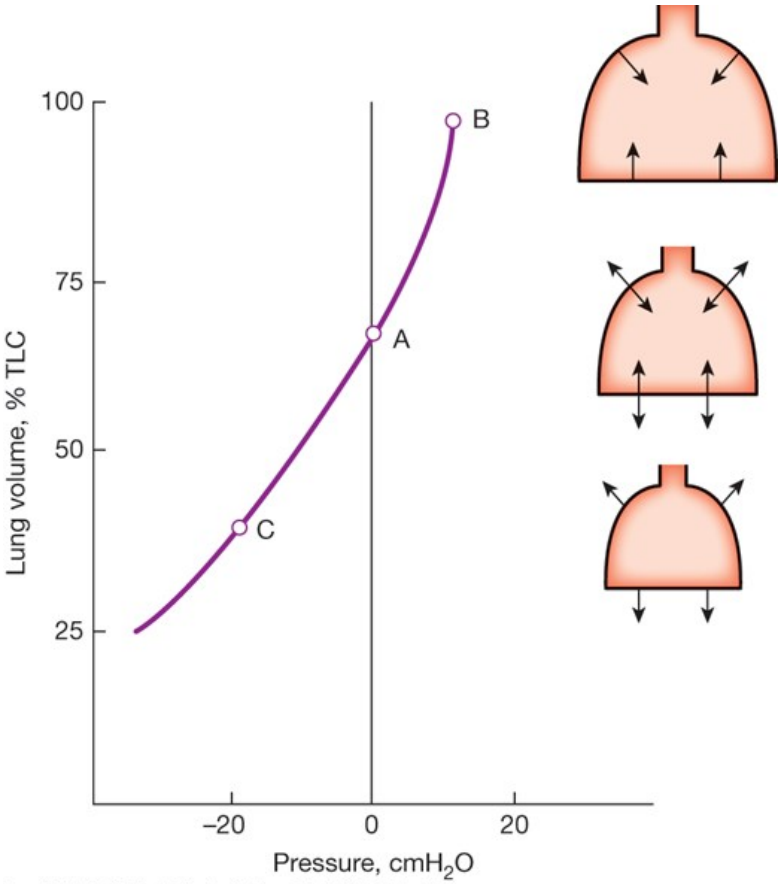
As a result of alterations in the elastin and collagen fibers in the lung, the distensibility of the lungs (measured as compliance) increases with age.[7] This is part of the normal aging process. Pulmonary compliance is also increased by the destruction of alveolar walls and the enlargement of alveolar spaces that characterize pulmonary emphysema. In contrast, the distensibility of the lungs is reduced by pulmonary fibrosis, which stiffens its interstitial tissues.[6]

Elastic Properties of the Thorax

The elastic recoil of the chest wall is such that if it were unopposed by the lungs, the chest would enlarge to approximately 70% of TLC. This position represents its equilibrium or resting position.[19] In this position (when the respiratory muscles are completely relaxed), the pressure difference across the chest wall – that is, the difference between pleural pressure and the pressure at the surface of the chest – is zero. If the chest were forced to enlarge beyond its equilibrium position by an increasingly positive pleural pressure or by the application of subatmospheric pressure at the body surface, it would, like the lung, recoil inward, resisting expansion and favoring return to its equilibrium position. Conversely, at volumes less than 70% of TLC, the recoil of the chest is opposite that of the lung and is directed outward (Fig. 10-6).[20] The chest wall can also be represented as a two-compartment system consisting of the rib cage and the abdomen, and volume changes can be partitioned between the two compartments.[21] Changing from the upright to the supine position at a constant overall lung volume produces a shift in volume from the abdominal to the rib cage compartment. The compliance of the rib cage is similar in the supine and upright positions, but the compliance of the abdominal compartment – particularly at high volumes – is greater in the supine position.[22]

Figure 10-6

Pressure–volume relationships of the isolated chest wall. The direction of the recoil forces across the chest wall is represented by the *arrows.* The equilibrium position of the chest wall at point A, unopposed by the lungs, is approximately 70% of the total lung capacity. In this position, the pressure difference across the chest wall is zero. At larger volumes (B), there is inward recoil of the chest wall; at volumes below the equilibrium position (C), the recoil of the chest wall is directed outward, favoring expansion.



The elastic recoil properties of the chest wall play an important role in determining the subdivisions of lung volume. They may be seriously deranged by disorders affecting the chest wall, such as marked obesity, kyphoscoliosis, and ankylosing spondylitis.

Elastic Properties of the Respiratory System as a Whole

During breathing the lung and the chest wall move together and operate mechanically in series. At any given lung volume the elastic recoil pressure of

the total respiratory system (P_{rs}) can be calculated as the algebraic sum of the pressures exerted by the elastic recoil of the lung (transpulmonary pressure) and the elastic recoil of the chest wall.[23]

Since the elastic recoil of the lung is determined (under static conditions of arrested airflow) as the difference between alveolar pressure (P_A) and pleural pressure (P_{pl}) – that is, $P_A - P_{pl}$ – and the elastic recoil of the chest wall is determined (while the respiratory muscles are completely at rest) as the difference between pleural pressure and the pressure at the external surface of the chest (P_{bs})—that is, $P_{pl} - P_{bs}$, the elastic recoil of the entire respiratory system can be expressed as the sum of the two:

$$P_{rs} = (P_A - P_{pl}) + (P_{pl} - P_{bs}) = P_A - P_{bs}$$

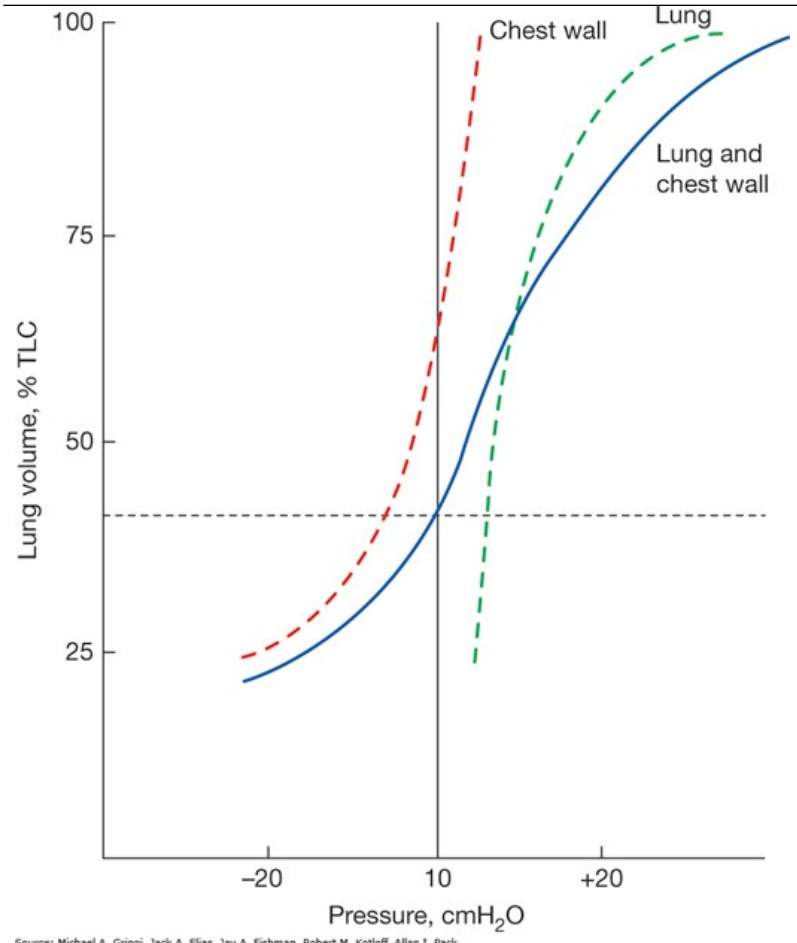
Thus, a measure of the elastic recoil of the respiratory system is supplied by the alveolar pressure, provided that the respiratory muscles are completely at rest and the pressure of the body surface is at atmospheric levels. In the absence of airflow into or out of the lung and when the glottis is open, alveolar pressure corresponds to the pressure at the mouth.

Relaxation Pressure–Volume Curve

The elastic properties of the entire respiratory system can be determined from the relaxation pressure–volume curve (Fig. 10-7). FRC represents the equilibrium position of the lung–chest wall system while the respiratory muscles are relaxed. At this point, the opposing recoils of the lung and chest wall are of equal magnitude, and the recoil pressure of the entire respiratory system is zero. With increases in lung volume above FRC, the recoil pressure of the entire system becomes positive, owing to the combination of an increase in centripetal elastic recoil of the lungs and a decrease in the centrifugal recoil of the chest wall. The net effect favors a decrease in lung volume, and lung volume can be maintained with the airway open to the atmosphere only by the action of the inspiratory muscles. As lung volume exceeds 75% of TLC, the recoil of the chest wall also becomes centripetal and the recoil pressure of the chest wall adds to the inward forces acting to diminish lung volume. TLC represents the lung volume at which the inward passive elastic recoil pressure of the respiratory system reaches the maximum force that can be generated by the inspiratory muscles.

Figure 10-7

Relaxation pressure–volume curves. The lungs and the chest wall function mechanically in series so that the elastic recoil pressures of the total respiratory system, represented by the *solid line*, is the algebraic sum of the separate recoil pressures of the lung and chest wall. At the volume represented by the horizontal *dashed line*, the recoil pressures of the lung and chest wall are equal but in opposite directions. Consequently the net recoil pressure is zero, and the respiratory system is in a position of equilibrium.



At lung volumes below FRC, when the centrifugal recoil of the chest wall exceeds the reduced centripetal recoil of the lungs, the relaxation pressure is negative and this net effect favors an increase in lung volume. Lung volumes below FRC are achieved and maintained by the muscles of expiration.

A switch from the sitting to the supine position decreases FRC because of the effects of gravity. In the upright position, gravity pulls the abdominal contents away from the chest wall. In contrast, in the supine position, the push of the abdominal contents against the diaphragm decreases the centrifugal recoil of the chest wall. The chest wall pressure–volume curve – and, consequently, the pressure–volume curve of the entire respiratory system – is displaced to the right.

Dynamic Mechanical Properties of the Respiratory System

The total nonelastic resistance of the lungs consists of the resistance of the airways to airflow (airway resistance), defined in terms of the driving pressure and the resulting rate of airflow, and the frictional resistance of the lung tissues to displacement during breathing (tissue resistance). Normally, tissue resistance makes up only 10% to 20% of the total pulmonary nonelastic resistance, but in diseases of the pulmonary parenchyma, it may increase considerably.

Airway Resistance

A large fraction of the resistance to airflow is in the upper respiratory tract, including the nose, mouth, pharynx, larynx, and trachea. During nasal breathing, the nose constitutes up to 50% of total airway resistance. During quiet mouth breathing, the mouth, pharynx, larynx, and trachea constitute 20% to 30% of the airway resistance; but they account for up to 50% of the total airway resistance when minute ventilation increases—during vigorous exercise, for example. Most of the remainder of airway resistance is in medium-sized lobar, segmental, and subsegmental bronchi up to about the seventh generation of airways.[24,25] Additional branching distally causes a progressive increase in the number of airways in any generation. While the

caliber of individual airways in daughter branches compared to the parent branch is reduced, the total cross-sectional area of all of the airways in a given generation increases tremendously with successive generations along the tracheobronchial tree. Consequently, in the normal lung, the small peripheral airways, particularly those less than 2 mm in diameter, constitute only about 10% to 20% of the total airway resistance.[26]

Airway Caliber

The airways, like the pulmonary parenchyma, exhibit elasticity and can be compressed or distended. Therefore, the diameter of an airway varies with the transmural pressure applied to that airway—that is, the difference between the pressure within the airway and the pressure surrounding the airway. The pressure surrounding intrathoracic airways approximates pleural pressure, since these airways are tethered to the parenchymal tissue and are exposed to the expansive forces that are active in overcoming the elastic recoil of the lung.[1]

Airway resistance varies inversely with lung volume. As the lung volume increases, the elastic recoil forces of the lung increase; the traction applied to the walls of the intrathoracic airways also increases, widening the airways and decreasing their resistance to airflow. Conversely, at low lung volumes, the transmural airway pressure is lower and airway resistance increases.[27,28] If the elastic recoil of the lung is reduced – by destruction of alveolar walls in pulmonary emphysema, for instance – the transmural airway pressure at any given lung volume decreases correspondingly; the airways are narrower and airway resistance is greater even though there is no disease of the airways per se.

The effects of a change in transmural pressure on airway caliber depend on the compliance of the airways—which, in turn, is determined by their structural support. The trachea, for example, is almost completely surrounded by cartilaginous rings, which tend to prevent complete collapse even when the transmural pressure is negative. The bronchi are less well supported by incomplete cartilaginous rings and plates, whereas the bronchioles lack cartilaginous support. All airways can be stiffened, albeit to different degrees, by contraction of smooth muscle in their walls.

In patients with airway disease, mucosal edema, hypertrophy and hyperplasia of mucus glands, increased elaboration of mucus, and hypertrophy of smooth muscle further compromise airway caliber and increase airway resistance.

Neural pathways and humoral mechanisms are also important in controlling airway smooth muscle tone and regulating airway caliber. Cholinergic parasympathomimetic stimulation originating from the vagus nerve and mediated through the release of acetylcholine causes airway smooth muscle contraction and airway narrowing. Noncholinergic parasympathetic activity may play a role in airway smooth muscle relaxation through the release of vasoactive intestinal peptide and the subsequent production of nitric oxide.[29] Beta-adrenergic receptors in bronchial smooth muscle activated by various sympathomimetic agents promote airway smooth muscle relaxation and airway dilatation.[30]

Pressure–Flow Relationships: Theoretical Considerations

In the lungs, pressure–flow relationships are extremely complicated because the airways consist of a system of irregular branching tubes that are neither rigid nor perfectly circular. For purposes of simplification, pressure–flow relationships in rigid tubes are generally regarded as a model for those in the airways.

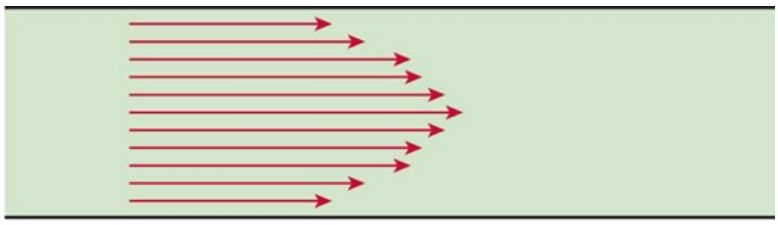
The driving pressure that produces flow of air into and out of the lung must suffice to overcome friction and to accelerate the air. Acceleration in the lungs is of two types: local (i.e., changes in the rate of airflow with time when flow is initiated) and convective (i.e., acceleration of molecules of air over distance while flow is constant). The driving pressure required for convective acceleration is proportional to the gas density and to the square of the flow rate. It is important during expiration because, as air moves downstream from the alveoli toward the airway opening, the total cross-sectional airway diameter decreases; therefore, molecules of air must accelerate through the converging channels even though the overall flow rate remains unchanged. Also, the driving pressure that produces high expiratory flow rates at large lung volume serves for convective acceleration rather than for overcoming friction.[31,32]

The driving pressure required to overcome friction depends on the rate and the pattern of airflow. Two major patterns of airflow warrant special consideration: laminar and turbulent. Laminar flow is characterized by streamlines that parallel the sides of the tube and are capable of sliding over one another. Also, because the streamlines at the center of the tube move faster than those closest to the walls, the flow profile is parabolic (Fig. 10-8). The pressure–flow characteristics of laminar flow depend on the length (l) and the radius (r) of the tube and the viscosity of the gas (η) according to Poiseuille's equation:

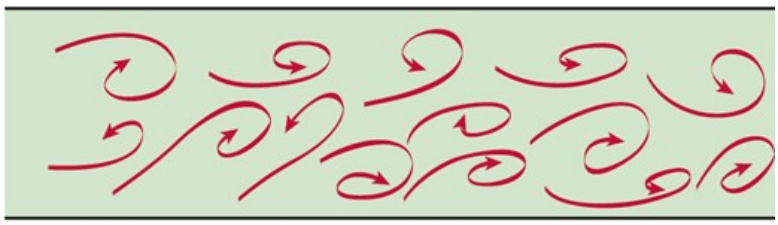
$$\Delta P = \frac{\dot{V} 8 \eta l}{\pi r^4}$$

Figure 10-8

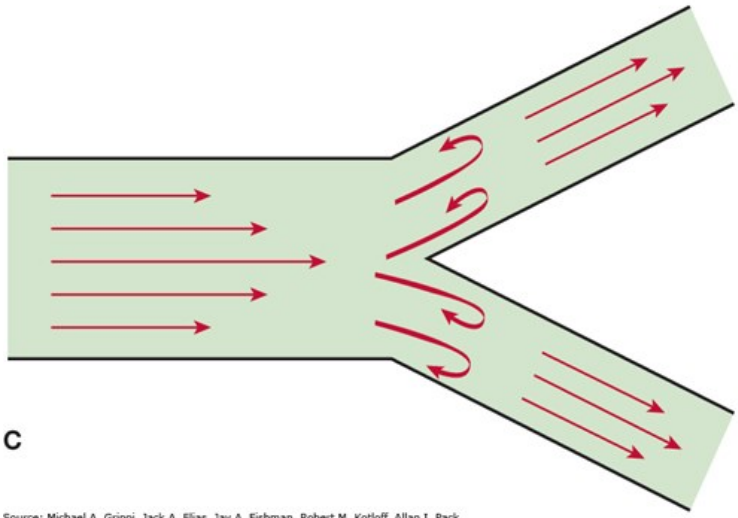
Patterns of airflow. **A.** Laminar flow. **B.** Turbulent flow. **C.** Transition flow that occurs at bifurcations.



A



B



C

where

ΔP = the driving pressure (pressure drop between the beginning and the end of the tube)

$\dot{V}$ = the flow rate that the driving pressure produces

r = the radius of the tube

The critical importance of tube radius in determining the driving pressure for a given flow is apparent in the previously mentioned equation. If the

radius of the tube is halved, the pressure that is required to maintain a given flow rate must be increased 16-fold. Laminar flow patterns occur only in small peripheral airways, where, because of the enormous overall cross-sectional area, flow through the individual airways is exceedingly slow.

Turbulent flow occurs at high flow rates and is characterized by a complete disorganization of streamlines, so that the molecules of gas move laterally, collide with each other, and change velocities. Under these circumstances, pressure–flow relationships change. In contrast to laminar flow, the rate of turbulent airflow is no longer proportional to the driving pressure. Instead, the driving pressure to produce a given rate of airflow is proportional to the square of flow and is dependent on gas density. Turbulent flow occurs regularly in the trachea.

At lower flow rates during expiration – particularly at branches in the tracheobronchial tree, where flow in two separate tubes comes together into a single channel – the parabolic profile of laminar flow becomes blunted, the streamlines separate from the walls of the tube, and minor eddy formation develops. This is referred to as a mixed, or transitional, flow pattern. In a mixed pattern of airflow, the driving pressure for a given flow depends on both the viscosity and the density of the gas.

Whether airflow is laminar or turbulent is predictable from the Reynolds number (Re), a dimensionless number that depends on the average velocity ($\bar{v}$), the density of the gas (ρ), the viscosity of the gas (η), and the diameter of the tube (D), so that

$$\mathrm{Re} = \frac{\bar{v} D \rho}{\eta}$$

In straight, smooth, rigid tubes, turbulence occurs when the Reynolds number exceeds 2000. Therefore, turbulence is most apt to occur when the average velocity is high, gas density is high, gas viscosity is low, and the tube diameter is large. Since most of the resistance to airflow in the normal lung is in large airways, where airflow is largely turbulent and where resistance is density dependent, breathing a mixture of 80% helium and 20% oxygen (a mixture that is 64% less dense than air) reduces the Reynolds number favoring a conversion from turbulent to laminar flow. Consequently airflow increases at a given driving pressure and airway resistance falls.[24]

Calculation of Airflow Resistance

The driving pressure along the tracheobronchial tree – that is, the difference between alveolar pressure and the pressure at the airway opening (mouth) that is required to produce a given rate of airflow into the lungs – provides a measure of the flow resistance of the airways, according to the equation

$$R_{aw} = \frac{P_A - P_{ao}}{\dot{V}}$$

where

$\dot{V}$ = airflow (L/s)

P_A = alveolar pressure (cmH_2O)

P_{ao} = airway-opening pressure (cmH_2O)

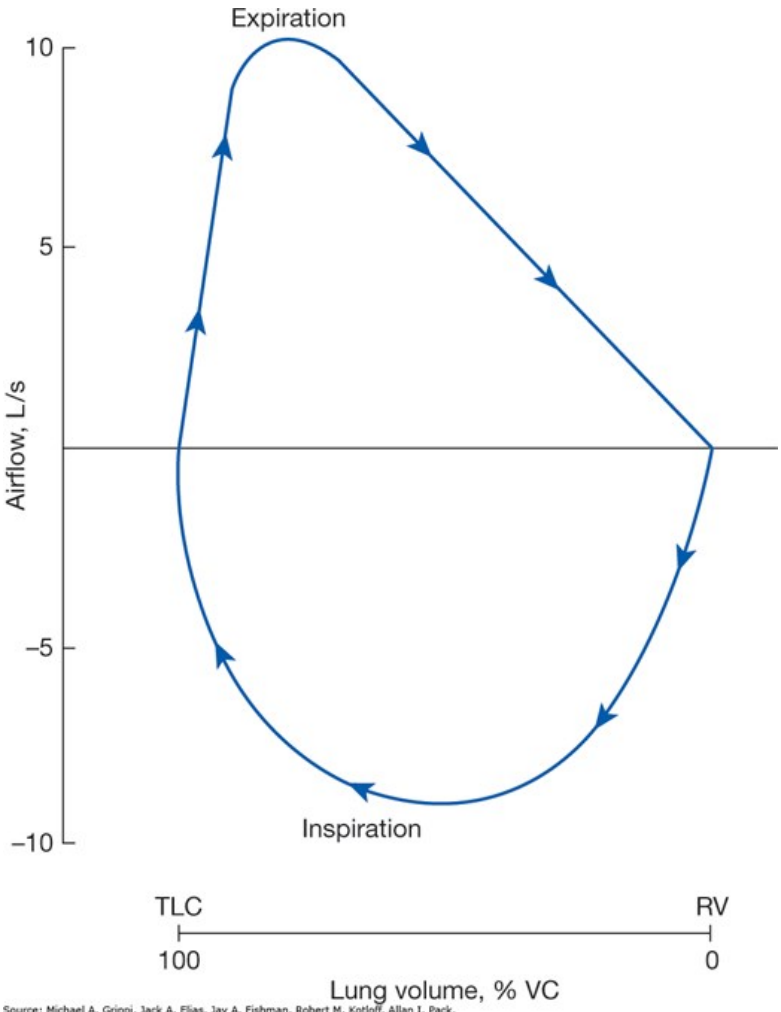
R_{aw} = airway resistance (cmH_2O/L/s)

Flow–Volume Relationships

Considerable insight into the flow-resistive properties of the airways can be obtained from the relationship between airflow and lung volume during maximal expiratory and inspiratory maneuvers.[33] In practice, a person inhales maximally to TLC; then exhales as forcefully, rapidly, and completely as possible to RV; and then returns to TLC by a rapid, forceful inhalation (Fig. 10-9). During the maximal expiration, the rate of airflow peaks at a lung volume that is close to the TLC; as the lung volume decreases and intrathoracic airways narrow, airway resistance increases, and the rate of airflow decreases progressively.

Figure 10-9

Maximal expiratory and inspiratory flow–volume loop.

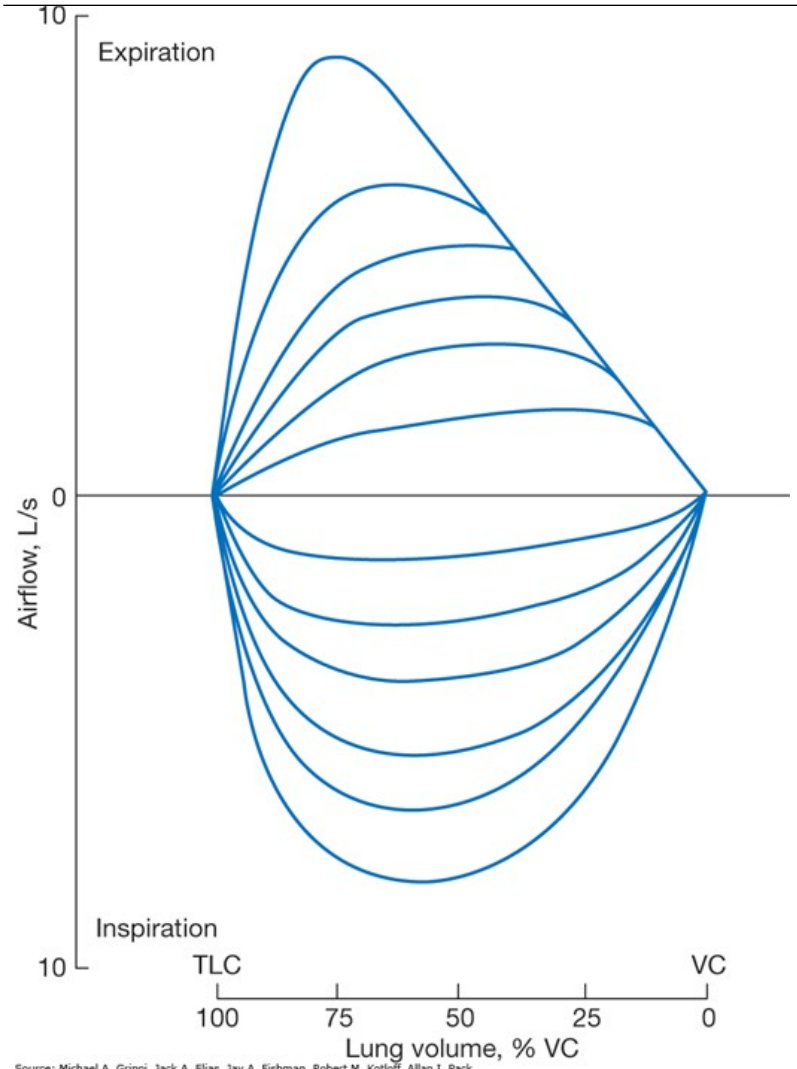


During maximal inspiration, the pattern of airflow is different. Because of the markedly negative pleural pressure and large transmural airway pressure, the bronchi are wide, and their calibers increase further as lung volume increases. Consequently, inspiratory flow becomes high while the lung volume is still low and remains high over much of the vital capacity, even though the force generated by the inspiratory muscles decreases as they shorten.

A family of flow–volume loops is produced by repeating full expiratory and inspiratory maneuvers over the entire range of the vital capacity using different levels of effort (Fig. 10-10). The greater the effort exerted during inspiration, the greater is the rate of airflow over the entire range—that is, from RV to TLC. Similarly, during expiration, the rate of airflow increases progressively with increasing effort at large lung volumes close to TLC. At intermediate and low lung volumes, the rate of expiratory airflow reaches a maximum while the effort expended is only moderate; thereafter, airflow does not increase further despite increasing expiratory efforts.

Figure 10-10

Series of flow–volume loops constructed from complete inspiratory and expiratory maneuvers repeated at different levels of effort.

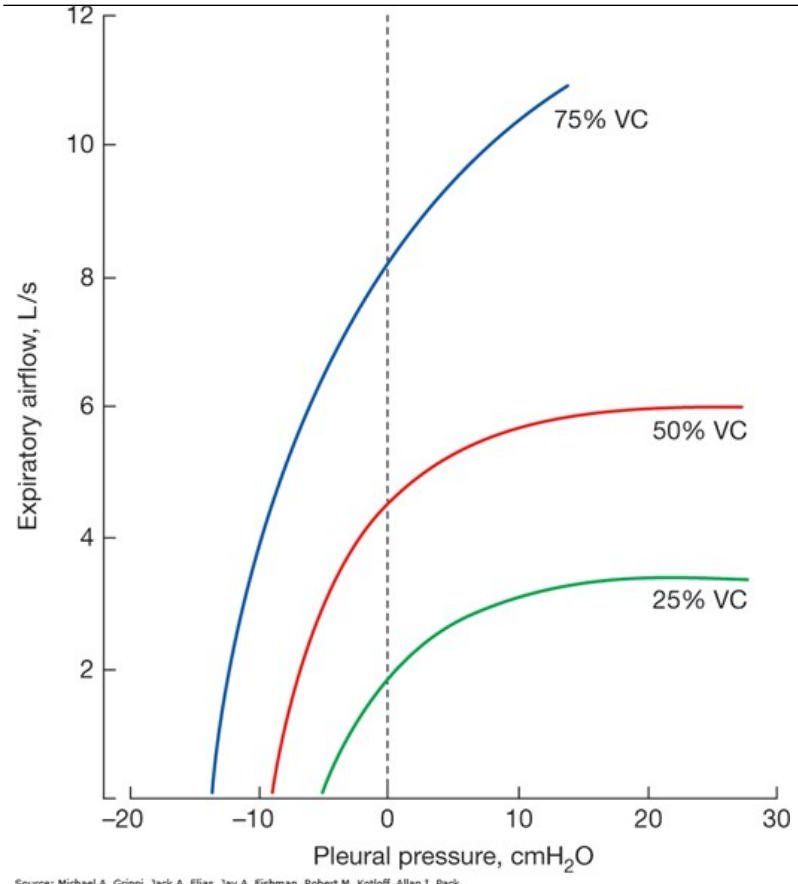


Isovolume Pressure–Flow Curves

Separation of the effects of increasing effort from those of changes in lung volume on the rate of airflow during expiration can be accomplished by using isovolume pressure–flow curves (Fig. 10-11). During repeated expiratory maneuvers performed with varying degrees of effort, simultaneous measurements are made of airflow rate, lung volumes, and pleural pressure. For each lung volume the rate of airflow is plotted against the pleural pressure, as an index of the degree of effort.[34]

Figure 10-11

Isovolumetric pressure–flow curves. At lung volumes greater than 75% of the vital capacity, airflow is effort dependent; that is, airflow increases progressively with increasing effort. At lower lung volumes, airflow is effort independent; that is, airflow becomes fixed at a maximum level and does not increase despite further increases in effort.



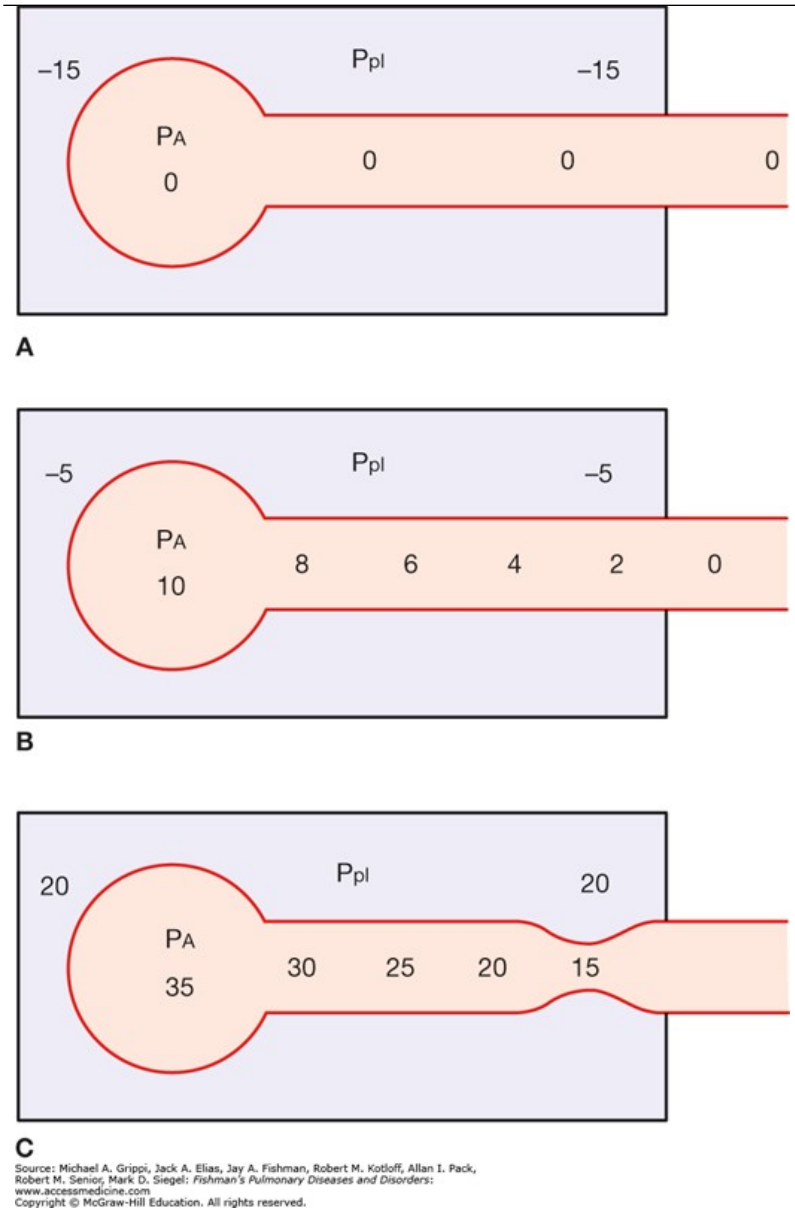
As expiratory effort is increased at any given lung volume, the pleural pressure increases toward, and then exceeds, atmospheric pressure; correspondingly, the rate of airflow increases. At lung volumes above 75% of the vital capacity, airflow increases progressively as pleural pressure increases; it is considered to be effort dependent. In contrast, at lung volumes below 75% of the vital capacity, the rate of airflow levels off as the pleural pressure exceeds atmospheric pressure and becomes fixed at a maximum level. Thereafter, further increases in effort, and in pleural pressure, effect no further increase in the rate of airflow; at these lower lung volumes, airflow is considered to be effort independent. Since the rate of airflow remains constant despite increasing driving pressure, it follows that the resistance to airflow must be increasing in direct proportion to the increase in driving pressure. This increase in resistance is attributed to compression and narrowing of large intrathoracic airways.

Equal Pressure Point Theory: Dynamic Compression of Airways

To illustrate the mechanisms that normally limit airflow during a maximal expiratory maneuver, it is useful to consider a model of the lung where the alveoli are represented by an elastic sac and the intrathoracic airways by a compressible tube, both enclosed within a pleural space (Fig. 10-12).[35]

Figure 10-12

Schema of the distribution of pleural, alveolar, and airway pressures at rest and during expiration, illustrating the equal pressure point concept. **A.** End expiration. **B.** Quiet expiration. **C.** Forced expiration.



At a given lung volume, when there is no airflow (as during breath holding with the glottis open), pleural pressure is subatmospheric, counterbalancing the elastic recoil pressure of the lung. The alveolar pressure (P_A), which is the sum of the recoil pressure of the lung and pleural pressure (P_{pl}), is zero (Fig. 10-12A). Since airflow has ceased, the pressure along the entire airway is also atmospheric.

At the same lung volume during a quiet expiration, pleural pressure is less subatmospheric. Since lung volume and the elastic recoil pressure of the lung are unchanged, alveolar pressure is now positive with respect to atmospheric pressure; airflow occurs. The alveolar pressure is gradually dissipated along the airway in overcoming resistance so that the pressure at the airway opening (P_{ao}) is zero. All along the airway, however, the airway pressure exceeds pleural pressure and the transmural pressure is positive; the airways remain open, and flow continues (Fig. 10-12B).

A forceful expiration raises pleural pressure above atmospheric pressure and further increases alveolar pressure (Fig. 10-12C). Airway pressure again falls progressively from the alveolus toward the airway opening. But at some point along the airway – the equal pressure point – the drop in airway pressure is equal to the recoil pressure of the lung; intraluminal pressure and the pressure surrounding the airways are equal and the same as pleural pressure. Downstream (i.e., toward the airway opening) the transmural pressure is negative, because the intraluminal airway pressure is less than pleural pressure; the airways are subjected to dynamic compression.

The equal pressure point divides the airways into two components arranged in series: an upstream segment, from the alveoli to the equal pressure

point, and a downstream segment, from the equal pressure point to the airway opening. With increasing expiratory effort as the pleural pressure becomes more and more positive with respect to atmospheric pressure, the equal pressure point moves upstream. Once maximum expiratory flow is achieved, the position of the equal pressure point becomes fixed in the region of the lobar or segmental bronchi. Further increase in pleural pressure by increasing expiratory force simply produces more compression of the downstream segment without affecting airflow through the upstream segment.

The driving pressure of the upstream segment – that is, the pressure drop along the airways of that segment – is equal to the elastic recoil of the lung. The maximum rate of airflow during forced expiration (V_{max}) can be expressed in terms of the elastic recoil pressure of the lung (P_l) and the resistance of the upstream segment (R_{us}), as follows:

$$\dot{V}_{max} = \frac{P_L}{R_{us}}$$

Measurements of the rate of airflow during force expiration form the basis of many tests used to assess the flow-resistive properties of the lung. It is evident, however, that the maximum rate of expiratory airflow depends on many factors: The lung volume at which airflow is determined; the force of expiration (particularly at high lung volumes [i.e., above 75% of vital capacity]); the elastic recoil pressure of the lung; the cross-sectional area of large airways; the collapsibility of large intrathoracic airways; and the resistance of small peripheral airways.

Wave Speed Limitation Theory

An alternative explanation for airflow limitation during forced expiration is based on principles of wave speed theory.[36] The wave speed theory proposes that flow is limited by the velocity of propagation of pressure waves along the wall of the tube. The velocity of propagation (v) varies proportionally with the cross-sectional area of the tube (A) and with airway stiffness. At a site where the linear velocity of gas molecules equals the velocity of propagation of pressure waves that is, wave speed, a choke point develops, preventing further increases in flow rate. The flow rate at wave speed is a function of the cross-sectional area of the tube at the choke point (A) and the stiffness of the choke segment (dP/dA), where P is the transmural airway pressure. Where choke points occur in the tracheobronchial tree depends on the lung volume: at large lung volumes, a choke point is situated in the vicinity of the lower trachea; at lower lung volumes, choke points develop more upstream along the bronchial tree. Extension of the neck exerts longitudinal tension and stiffens the trachea, increases wave velocity, and increases maximum expiratory flow rates at large lung volumes.[37]

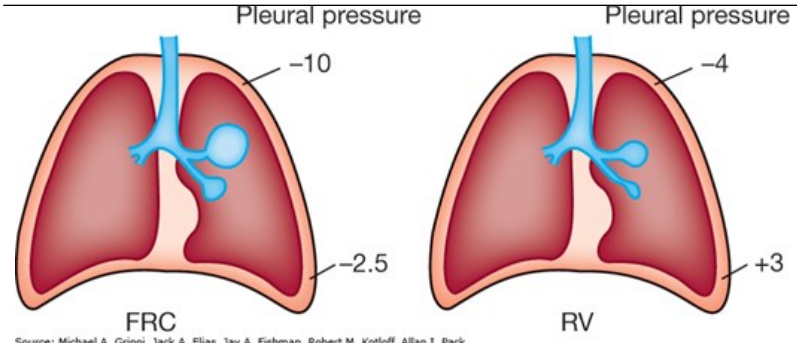
Mechanical Determinants of Regional Ventilation

The lung is not homogeneous, and the mechanical properties of all airways in a given generation and of all alveoli are not the same. This results in important nonuniformities of regional ventilation.

Pleural pressure in the upright person is more subatmospheric at the apex than at the base of the lung, because of the effects of gravity and the weight of the lung.[38] Pleural pressure topography and regional lung expansion are also determined by the shape of the chest wall and by the forces required for the lung to conform to the thoracic cavity shape[9].[39] The rate of increase in pleural pressure from top to bottom is approximately 0.25 cmH_2O per centimeter of vertical distance. Consequently, the transpulmonary pressure – that is, alveolar pressure minus pleural pressure – is greater at the top than at the bottom of the lung. Therefore, at most lung volumes, the alveoli at the lung apexes are larger (more expanded) than those at the lung bases (Fig. 10-13).

Figure 10-13

Pleural pressure gradients in the upright lung at FRC (left) and at RV (right). The effect of the gradient on alveolar volumes is shown schematically for each case.



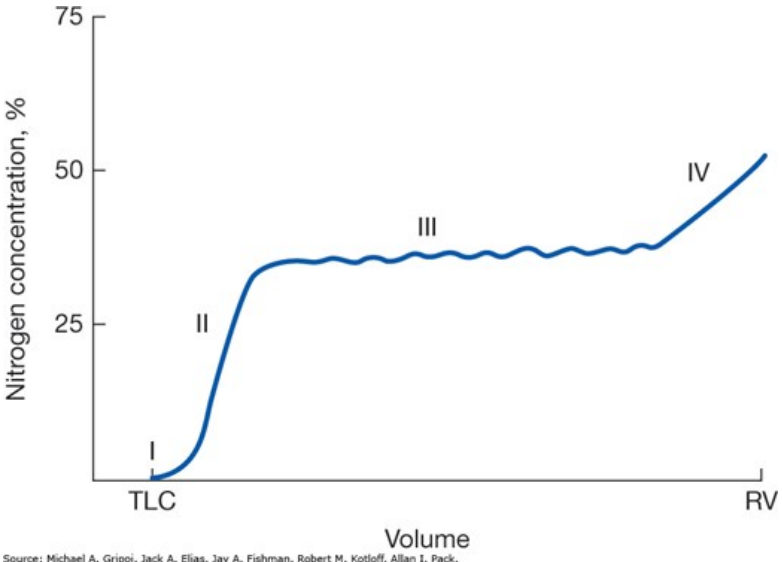
Because of regional variations in lung compliance, ventilation is not uniform, even in the normal lung. With the use of external scanners after the inhalation of a radioactive gas, such as ^{133}Xenon, it has been demonstrated that within the range of normal tidal volume, lung units are better ventilated, and ventilation per alveolus is greater, at the bottom than at the top of the lung.[40] This is because alveoli near the top of the lung are positioned on the upper, flatter part of the pressure–volume curve and are less compliant than alveoli at the lung bases positioned on the lower, steeper portion of their pressure–volume curves.

At low lung volumes (i.e., near the RV), pleural pressure at the bottom of the lung actually exceeds airway pressure and leads to closure of peripheral airways (Fig. 10-13). During a breath taken from RV, air that enters the lungs first is preferentially distributed to the lung apexes.

The distribution of ventilation within the lungs and the volume at which airways at the lung bases begin to close can be assessed by the single-breath N_2 washout test.[41] This test requires a maximum expiration into an N_2 meter after a maximal inspiration of pure O_2 from RV; the changing concentration of nitrogen is plotted against expired lung volume (Fig. 10-14). Because the inspiration starts at the RV, the initial portion of the breath containing dead-space gas, rich in nitrogen, is distributed to alveoli in the upper lung zones. The rest of the breath, which contains only O_2, goes preferentially to lower lung zones. Consequently, the concentration of nitrogen is lower in the alveoli at the lung bases than in the alveoli at the apexes of the lungs.

Figure 10-14

Tracing of expired nitrogen concentration during a slow expiration from TLC to RV after a full inspiration of pure O_2. The four phases are indicated. For further details, see text.



During expiration, the initial portion of the breath consists of O_2 remaining in the large airways; it contains no N_2 (phase I). As alveolar gas containing

N_2 begins to be washed out, the concentration of N_2 in the expired air rises to reach a plateau. The portion of the curve where the concentration of N_2 rises steeply is phase II. The plateau is phase III. Phase III depends on the uniformity of the distribution of ventilation in the lung. If gas enters and leaves alveoli throughout the lung synchronously and equally, phase III is flat. But when the distribution of ventilation is nonuniform, so that gases coming from different alveoli have different N_2 concentration, phase III slopes upward.

At low lung volumes, airways at the lung bases close; only alveoli at the top of the lung continue to empty. Since the concentration of N_2 in the alveoli of upper lung zones is higher than in the alveoli at the lung bases, the slope of the N_2–volume curve increases abruptly, marking the start of phase IV. The volume, above RV, at which phase IV begins is termed the closing volume.[42]

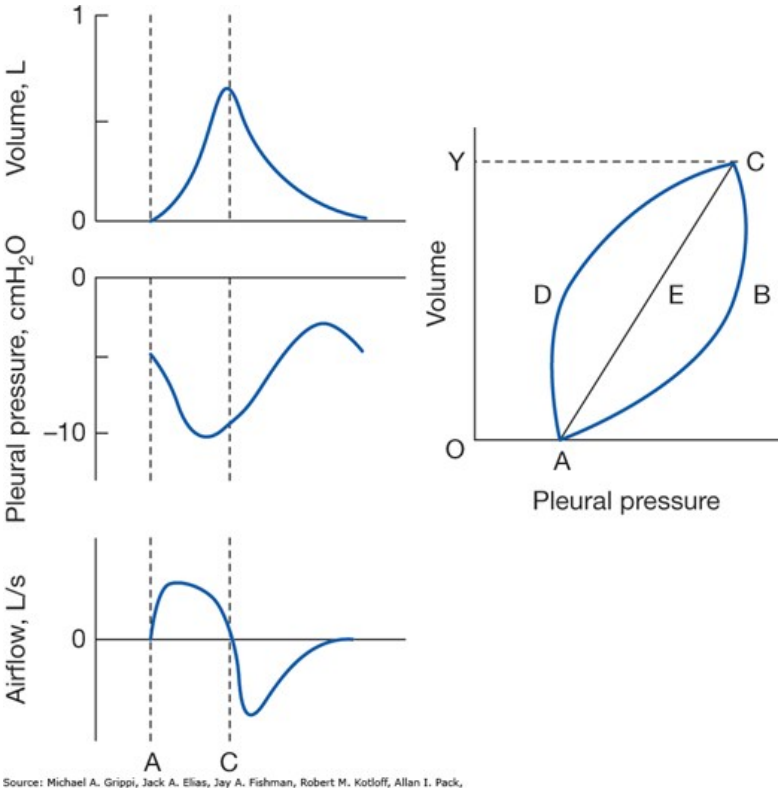
The closing volume increases with advancing age. With diseases of the peripheral airways closing volume may rise to levels above FRC. This results in lung units that are perfused but poorly ventilated leading to reductions in arterial oxygenation.[43,44]

Dynamic Compliance of the Lungs

The relationship between changes in volume and changes in pleural pressure during a normal breathing cycle is shown in Figure 10-15. Airflow momentarily ceases at the end of expiration (A) and at the end of inspiration (C); the change in pleural pressure between these two points reflects the increasing elastic recoil of the lung as the volume of air in the lungs increases. The slope of the line connecting the end-expiratory and end-inspiratory points (AEC in the figure) on a pressure–volume loop provides a measure of the dynamic compliance of the lungs.

Figure 10-15

Individual tracings of tidal volume, pleural pressure, and airflow, taken simultaneously during a single complete breath, are shown on the left. The relationship between volume and pleural pressure is illustrated by the dynamic pressure–volume loop on the right. Dynamic compliance is determined as the slope of the line AEC. The work of breathing during inspiration to overcome the elastic forces of the lung is represented by the area of the trapezoid OAECY, and the work required to overcome nonelastic forces is represented by the area of the loop ABCEA. The loop AECDE represents the work required to overcome airflow resistance during expiration.



In normal persons, dynamic compliance closely approximates inspiratory static lung compliance and remains essentially unchanged even when breathing frequency is increased up to 60 breaths per minute. This indicates that lung units that are parallel with each other normally fill and empty uniformly and synchronously, even when airflow is high and the change in lung volume is rapid. The rate of filling and emptying of a lung unit depends on its time constant—that is, the product of its resistance and compliance. Lung units with high resistance and high compliance take longer to fill and empty more slowly compared to units with low resistance and low compliance. In order for the distribution of ventilation in parallel lung units to be independent of the rate of airflow, the resistance and compliance of these units must be matched so that the time constants of individual units throughout the lungs are approximately the same. The time constants of lung units distal to airways 2 mm in diameter are approximately 0.01 second, and fourfold differences in time constants are necessary to cause dynamic compliance to become frequency dependent.[26]

Patchy narrowing of small peripheral airways produces regional differences in time constants. At low breathing frequencies, when the rate of airflow is low, ventilation is fairly evenly distributed. As the breathing frequency increases, however, ventilation tends to be distributed to areas that offer the least resistance to airflow. Therefore, lung units fed by narrowed airways receive proportionally less ventilation than do areas of the lung where the airways remain normal; the change in pleural pressure required to effect the same change in overall lung volume increases. As a result, the dynamic compliance falls.

Measurements of frequency dependence of dynamic compliance are time-consuming and technically difficult, but this test has proved useful in the diagnosis of obstruction in small peripheral airways when results of other conventional tests of lung mechanics are still within normal limits.[45]

Work of Breathing

During breathing, the respiratory muscles work to overcome the elastic, flow-resistive, and inertial forces of the lungs and chest wall.[46] The elastic work of breathing is done to overcome the elastic recoil of the lungs and chest wall; the resistive work is done in overcoming the resistance of airways and tissues. The mechanical work of breathing can be determined by relating the pressure exerted across the respiratory system to the resulting change in volume, since the product of pressure (P) and volume (V) has the dimension of work, according to the equation

$$\text{work} = \int P \, dV$$

Recordings of pleural pressure and lung volume changes during spontaneous breathing can be used to measure the work of breathing; the work of breathing performed on the lungs can be determined from the area of the dynamic pressure–volume loop (Fig. 10-15) and fractionated into its elastic and resistive components. During inspiration, the work done to overcome the elastic forces of the lung is determined from the area of the trapezoid OAECY (Fig. 10-15). The area of the loop ABCEA is the work in overcoming nonelastic forces during inspiration, and the area of the loop OABCY is the total work of breathing during inspiration.

Expiration during quiet breathing is passive, since the elastic recoil of the lung suffices to overcome the expiratory airflow resistance. Some of the stored elastic energy is also used to overcome inspiratory muscle activity that persists into the expiratory phase of breathing. At high levels of ventilation and when airway resistance is increased, additional mechanical work during expiration is required to overcome nonelastic forces. Under these circumstances, the pleural pressure exceeds atmospheric pressure, and the loop AECDA extends beyond the confines of the trapezoid OAECY (Fig. 10-15).

The work of breathing at any given level of ventilation depends on the pattern of breathing. Large tidal volumes increase the elastic work of breathing, whereas rapid breathing frequencies increase the work against flow-resistive forces. During quiet breathing and during exercise, people tend to adjust tidal volume and breathing frequency to values that minimize the force and the work of breathing.[47] Similar adjustments are also seen in patients with pulmonary disorders. Patients with pulmonary fibrosis, which is characterized by an increased elastic work of breathing, tend to breathe shallowly and rapidly; those with airway obstruction and increased nonelastic work of breathing usually breathe more deeply and slowly.

The work done on the chest wall during breathing is calculated by subtracting the work performed on the lung from the total mechanical work of breathing. The total mechanical work of breathing cannot be readily measured during spontaneous breathing because the respiratory muscles that perform the work also make up part of the resistance offered by the chest wall. But the total mechanical work can be determined during artificial ventilation by using either intermittent positive airway pressure or negative pressure applied to the chest, provided that the respiratory muscles are completely at rest. For this determination, the change in lung volume is related to the pressure difference across the respiratory system—that is, differential pressure between the mouth and the body surface. Disturbances of the chest wall, such as kyphoscoliosis and obesity, increase the work of

breathing severalfold.

Oxygen Cost of Breathing

In order to perform their work, the respiratory muscles require O_2. The O_2 cost of breathing, which reflects the energy requirements of the respiratory muscles, provides an indirect measure of the work of breathing.[48,49] The O_2 cost of breathing is assessed by determining the total O_2 consumption of the body at rest and at an increased level of ventilation produced by voluntary hyperventilation or CO_2 breathing. Provided there are no other factors acting to increase O_2 consumption, the added O_2 uptake is attributed to the increased metabolism of the respiratory muscles.

The O_2 cost of breathing in normal subjects is approximately 1 mL/L of ventilation and constitutes less than 5% of the total O_2 consumption. At high levels of ventilation, however, the O_2 cost of breathing becomes progressively greater. There is a dramatic increase in the O_2 cost of breathing at high levels of ventilation in some diseases of the lung, such as pneumonia, pulmonary fibrosis, and emphysema, and in disorders of the chest wall, such as obesity and kyphoscoliosis. The increase in the energy requirement of the respiratory muscles during increased ventilation, concomitant with a decrease in O_2 supply secondary to arterial hypoxemia, contributes to muscle fatigue, thereby limiting the amount of exertion that these patients can sustain.[3]

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