



Modeling the Lung from a Rheological Perspective

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Introduction

Mechanical ventilation is often interpreted through simplified measures such as airway resistance, compliance, elastance, tidal volume, and driving pressure. These variables remain clinically useful, but they describe only part of lung behavior. The lung is not a purely elastic structure that responds instantaneously and reversibly to every applied force; it is a viscoelastic organ whose response combines elastic recoil, viscous dissipation, and time-dependent memory that vary with flow, frequency, regional structure, and prior mechanical loading.

Rheology: the study of how materials deform and flow provides the framework for linking ventilator settings to tissue behavior over time. Applied to the lung, it explains why the same pressure or tidal volume can produce different effects when delivered at different rates, why airway pressure falls during an inspiratory pause, why inspiratory and expiratory pressure–volume curves do not overlap, and why repeated mechanical loading can shift tissue from reversible deformation toward structural injury.

The central implication is that ventilator-induced lung injury depends not only on how much mechanical load is applied, but also on how quickly it is delivered, where it is concentrated, and how often vulnerable regions are exposed to it. A rheological perspective therefore complements conventional respiratory mechanics by linking ventilator settings to stress distribution, tissue memory, and energy dissipation rather than treating pressure and volume as sufficient descriptors on their own.

This perspective frames lung mechanics as a continuum from bedside variables to tissue-level material behavior: conventional measures such as compliance, elastance, driving pressure, and tidal volume remain useful, but their interpretation improves when they are connected to stress distribution, deformation history, and energy dissipation over time.

This chapter first defines the lung as a viscoelastic material, then connects stress, strain, strain rate, shear, creep, stress relaxation, and hysteresis to ventilator-induced lung injury. It then compares classical integer-order models with fractional-order models and explains why the latter may better represent heterogeneous, history dependent lung tissue.

The Lung as a Viscoelastic Material

An ideal elastic material deforms in proportion to the applied stress and returns completely to its original shape when the stress is removed. An ideal viscous fluid, by contrast, continues to deform under load, and the stress required to produce deformation depends on the rate of deformation. The lung combines both behaviors.

During inspiration, part of the applied mechanical energy is stored reversibly as lung tissue stretches. Another part is dissipated through airway and tissue resistance, interstitial-fluid movement, and structural rearrangement within the extracellular matrix. Elastic behavior is associated mainly with elastin, collagen, and alveolar surface forces, whereas viscous behavior reflects airway resistance, parenchymal resistance, fiber interactions, and fluid movement.

This combination means that lung deformation is time dependent. The pressure required to inflate the lung depends not only on its current volume, but also on the speed of inflation and on the preceding pattern of loading. Viscoelasticity is generally more evident in the lung parenchyma than in the chest wall.

Elasticity and Elastance

Elasticity is the ability of lung tissue to store mechanical energy and return toward its previous configuration after the load is removed. Elastin contributes to recoil over a broad range of deformation, while collagen becomes increasingly important at larger volumes and helps limit excessive distension. Surfactants also contribute by modifying alveolar surface tension.

In clinical respiratory mechanics, the stiffness of the respiratory system is expressed as elastance, the inverse of compliance. A stiff lung has high elastance and low compliance. Although the spring is a useful conceptual model for elasticity, the lung does not behave as a single ideal spring because its stiffness changes with volume, recruitment, disease, and mechanical history.

Viscosity and Resistance

Viscosity describes resistance to deformation that depends on the rate of deformation. In the respiratory system, this behavior is reflected in airway resistance and in the intrinsic viscous properties of the lung tissue. Viscous forces dissipate mechanical energy rather than returning it completely during expiration.

The distinction between elasticity and viscosity is clinically important. A slow inflation and a rapid inflation may reach the same final volume but generate different stresses. Rapid flow increases the rate at which tissue is deformed and may increase the mechanical load imposed on vulnerable regions.

Plasticity, Resilience, and Anisotropy

Viscoelastic behavior is reversible only within limits. Plasticity refers to permanent deformation after stress exceeds the tissue's elastic range. In the lung, this may appear as residual deformation, irreversible reorganization of the extracellular matrix, or remodeling associated with diseases such as fibrosis.

Resilience is the maximum amount of energy that tissue can store elastically without permanent structural damage. Once the applied energy exceeds this tolerance, microstructural injury, inflammation, and progression of ventilator-induced lung injury become more likely.

The lung is also anisotropic, meaning that its mechanical response varies with the direction of applied force. This results from the arrangement of collagen and elastin fibers and differs among lung regions. Anisotropy becomes more pronounced at high inflation volumes, when regional differences in tissue structure and load distribution are amplified.

These material properties establish the foundation for the rest of the chapter: ventilation does not simply move air into an elastic container but loads a heterogeneous tissue whose response depends on deformation, rate, direction, and prior history.

Stress, Strain, and Strain Rate

Rheological analysis uses three closely related concepts: stress, strain, and strain rate.

Stress is the internal force per unit area within a material. In the lung, tissue stress is most closely related to transpulmonary pressure, the pressure difference between the alveoli and pleural space. Driving pressure is often used clinically as an approximation when transpulmonary pressure is not available, but it is not equivalent because it reflects the respiratory system as a whole and may be influenced by chest-wall mechanics as well as lung mechanics.

Strain is the relative deformation produced by stress. In the lung, it describes how much alveolar tissue expands relative to its resting volume rather than the absolute size of the delivered breath. At the whole-lung level, a useful approximation is the ratio of tidal volume to end-expiratory aerated lung volume. The same tidal volume can therefore produce modest strain in a normally aerated lung but much higher strain in a lung with extensive collapse or consolidation.

Strain is also heterogeneous. Open units may expand beside collapsed or poorly aerated units. The resulting regional concentration of deformation means that global airway measurements can underestimate the stress experienced by local tissue. Even modest tissue stretching may activate inflammatory pathways and generate reactive oxygen species, particularly when it occurs repeatedly.

Strain rate is the speed at which deformation occurs. In ventilation, it is related to flow relative to end-expiratory aerated lung volume. High inspiratory flow produces a high strain rate and gives tissue less time to adapt. Consequently, two breaths with the same tidal volume may not have the same biological effect if one is delivered rapidly and the other gradually.

Shear stress acts parallel to a tissue surface and occurs when adjacent layers move relative to one another. It is especially relevant in heterogeneous lungs, where open alveoli lie next to collapsed units. During inspiration, the open regions expand while the collapsed regions resist deformation. This mismatch concentrates tangential forces at their interface.

Repeated collapse and reopening magnify these forces. The resulting mechanical injury, known as atelectrauma, can disrupt alveolar membranes and promote local inflammation. The risk rises when end-expiratory lung volume is insufficient to prevent repetitive opening and closing. Thus, positive end-expiratory pressure must be considered not only in terms of airway pressure but also in relation to regional stability and tissue stress.

Together, stress, strain, strain rate, and shear explain where mechanical injury begins. The next question is how those forces evolve over time, which requires examining creep, stress relaxation, and hysteresis.

Time-Dependent Viscoelastic Phenomena

Creep

Creep is the progressive increase in deformation that occurs when a constant load is maintained. If lung tissue is exposed to a sustained pressure, its volume or strain may continue to increase over time even though the pressure does not change.

Creep may contribute to gradual recruitment of slowly opening lung units, but it can also promote progressive overdistension during prolonged exposure to a fixed pressure. This dual effect is particularly relevant in acute respiratory distress syndrome (ARDS), where the extracellular matrix is altered and mechanical properties are distributed unevenly across the lung.

Stress relaxation

Stress relaxation is the opposite time-dependent response. When tissue is held at a constant deformation, the force or pressure required to maintain that deformation decreases over time. The fall reflects redistribution of stress and microstructural reorganization within the viscoelastic tissue.

Clinically, stress relaxation can be observed during an inspiratory hold. Airway pressure falls in two stages: an initial rapid decrease, largely related to the dissipation of resistive pressure, followed by a slower decrease associated with tissue relaxation. The difference between the initial and final pressures during the pause provides information about the dynamic behavior of the respiratory system. In inflamed or fibrotic tissue, relaxation may be reduced, so pressure remains elevated for longer.

Classical models often describe relaxation with an exponential function. Experimental work, however, suggests that lung tissue may show power-law behavior across multiple time scales. This finding is one reason more advanced models are being developed.

Hysteresis

Hysteresis is the separation between the inspiratory and expiratory branches of the pressure–volume curve. At a given volume, inspiratory pressure is higher than expiratory pressure, creating a loop whose area represents mechanical energy dissipated during inflation and deflation rather than returned elastically to the respiratory system.

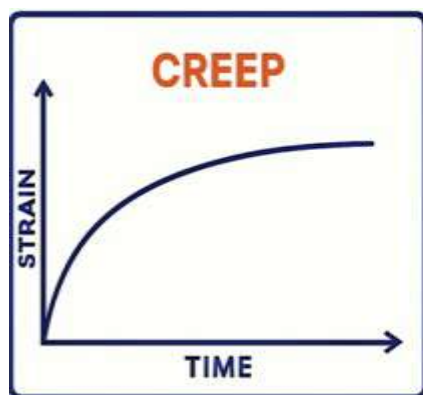
Hysteresis reflects the combined effects of tissue viscosity, surface tension, alveolar recruitment and derecruitment, and the structural organization of the extracellular matrix. In ARDS, the size and shape of the loop may provide information about recruitability, regional heterogeneity, and energy dissipation. A larger hysteresis area may indicate greater recruitable volume, although it should not be interpreted as a direct measure of a single mechanism because several processes contribute to the loop.

The dissipated energy is not recovered during expiration. Repeated energy dissipation may contribute to biotrauma and atelectrauma, particularly when high pressures, large strains, or rapid flow are combined.

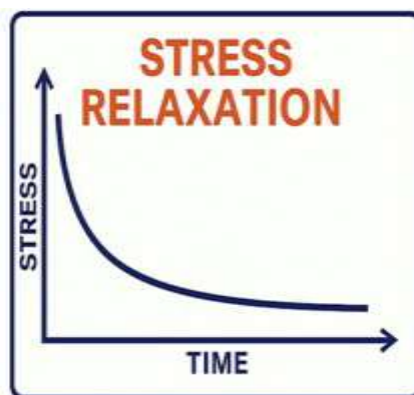
The next sections evaluate how classical rheological formulations represent the phenomena described above. Spring–dashpot models are useful because they translate elasticity, viscosity, creep, stress relaxation, and hysteresis into simple mechanical elements, but they also reveal where simplified representations begin to fail.

Table 1: Terms, definitions and their clinical correlation

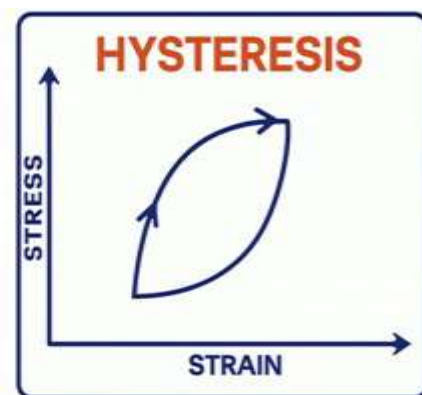
Engineering material term	Rheological definition	Clinical/Physiological correlation	Application in lung-protective mechanical ventilation (LPMV)
Stress (σ)	Internal tension or force per unit area	Transpulmonary pressure (PTP)	Limit to prevent barotrauma/volutrauma
Strain (ϵ)	Relative volumetric deformation	Tidal volume (VT) normalized to EELV (functional lung)	Goal: maintain low VT (6 ml/kg ideal body weight)
Strain Rate (ϵ')	Rate of change in deformation (deformation/time)	Flow ($\dot{V}$)	Control flow patterns (e.g., square vs. decelerating) to manage dynamic stress
Shear Stress (τ)	Tangential stress or force parallel to a surface	Local forces at tissue interfaces (e.g., stress hotspots)	Minimize by optimizing PEEP to avoid cyclic opening/closing (atelectrauma)
Creep	Increase in deformation under constant stress	Delayed alveolar recruitment	Use prolonged inspiratory pauses to optimize EELV
Stress Relaxation	Decrease in stress under constant deformation	Drop in plateau airway pressure during inspiratory pause	Determination of dynamic tissue viscosity and response time
Hysteresis	Deviation of the stress-strain curve during loading and unloading	Difference between inspiratory and expiratory P/V	Indicator of recruitability and energy dissipation



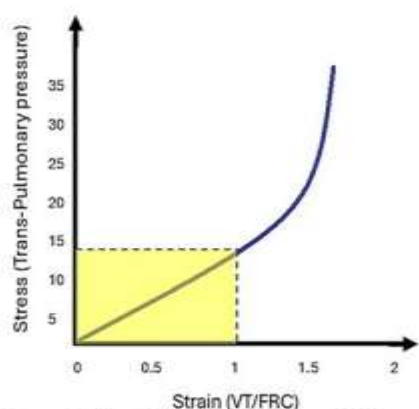
Creep. Progressive deformation (strain) over time under a constant load



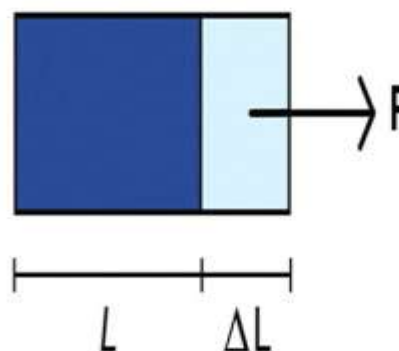
Stress relaxation. The force (stress) required over time to maintain a constant deformation (strain) progressively decreases



Hysteresis. At the same volume, inspiratory pressure exceeds expiratory pressure, resulting in the formation of a loop in the P-V curve, the area of which represents energy dissipated by the tissue



Stress – Strain relationship. The yellow shaded area represents the safe zone of ventilation



Stress and strain. F : Force = Stress = PTP, L : Length = Volume). Strain = $\Delta L/L$.

Figure 1: Creep, Stress relaxation, Hysteresis, Stress-Strain relationship

Classical Integer-Order Models

A rheological model expresses the relationship between stress and strain over time. The simplest models use two idealized elements. A spring represents elasticity: it responds immediately, stores energy, and returns toward its original length when the load is removed. A dashpot represents viscosity: its resistance depends on the rate of deformation and the energy supplied to it is dissipated. These elements are governed by ordinary, or integer-order, differential equations.

Integer-order models usually assume small deformations, linear behavior, and one or a few characteristic relaxation times. They remain valuable because they make the main mechanical components visible and can approximate selected clinical or experimental observations. Their limitations become important when lung behavior is heterogeneous, nonlinear, or spread across many time scales, because a small set of exponential responses cannot fully represent the continuous mechanical memory of injured parenchyma.

Maxwell model

The Maxwell model places a spring and dashpot in series. It is useful for representing stress relaxation: when deformation is held constant, stress decays over time. It also captures the memory of prior loading. Its main weakness is its prediction of unbounded, linear creep under constant stress. That behavior is not a realistic description of intact lung parenchyma over prolonged ventilation, so the model is best used as a conceptual or limited short-term approximation.

Kelvin–Voigt model

The Kelvin–Voigt model places a spring and dashpot in parallel. It is suited to delayed deformation, or creep, because the dashpot slows the spring's response. It represents resistance to rapid deformation and approaches elastic behavior during slow loading. It cannot, however, reproduce stress relaxation after a sudden imposed deformation. This makes it incomplete for interpreting inspiratory holds and other maneuvers in which pressure falls while volume is held constant.

Zener model

The Zener model, also called the standard linear solid, combines two springs with a dashpot. It is the simplest classical model that can represent both creep and stress relaxation, including partial recovery

after unloading. In this arrangement, the parallel spring provides an immediate elastic response, while the spring–dashpot pathway represents delayed tissue reorganization.

In a physiological interpretation, the springs may be associated broadly with elastin and collagen, while the dashpot represents tissue rearrangement, interstitial-fluid movement, and other rate-dependent processes. The model can also reproduce hysteresis and is useful for normal or mildly altered parenchyma. Its principal limitation is that it still describes relaxation with a small number of exponential time constants and may predict a long-term response that differs from the finite, power-law relaxation observed experimentally.

Anti-Zener model

The Anti-Zener model uses two dashpots and a spring and has a predominantly fluid-like response with temporary elastic behavior. It is not a good representation of healthy lung parenchyma, which normally retains elastic recoil. It may nevertheless be useful for pathological materials in which viscous behavior becomes prominent, such as airway secretions, interstitial edema, or tissues undergoing progressive structural alteration.

Burgers model

The Burgers model combines Maxwell and Kelvin–Voigt behavior in a four-element configuration. It can represent both relaxation and creep, including delayed deformation and terminal flow. Because it contains multiple characteristic times, it is more flexible than the two-element models and can be useful in detailed research analyses.

Its additional parameters also create practical problems. The model is more difficult to identify from clinical data and may overfit measurements. Most importantly, it still predicts exponential rather than power-law relaxation. A larger network of elements can improve the fit, but this increases complexity without fully solving the underlying problem.

Generalized Maxwell model

The generalized Maxwell, or Wiechert, model places multiple Maxwell units in parallel. Each unit has a different relaxation time, allowing the model to approximate the broad range of responses produced by different populations of alveoli or acinar regions. It is especially relevant to ARDS, in which collapsed, aerated, edematous, and overdistended regions coexist.

This approach improves accuracy but requires many parameters. It approximates a continuous distribution of relaxation times with a discrete set of elements. The result may be mathematically useful while remaining difficult to interpret physiologically or apply at the bedside.

Why Integer-Order Models Are Not Enough

Classical spring–dashpot models are useful first approximations, but they become insufficient when the lung is treated as a heterogeneous, history-dependent biological material. Their shared limitation is not simply mathematical complexity; it is that they compress distributed tissue behavior into one or a few discrete relaxation times. This simplification becomes especially problematic in ARDS, where normal, collapsed, edematous, fibrotic, and overdistended regions can coexist within the same lung.

Experimental observations indicate that lung stress relaxation is better described by a rapid initial fall followed by prolonged, slow relaxation. This is a power-law response rather than a simple exponential response. In schematic form, stress may be expressed as:

$$\text{Stress}(t) \propto t^{-\beta}$$

where t is elapsed time and β describes the rate of decay.

An exponential model reduces the influence of past deformation relatively quickly. A power-law model preserves the influence of prior loading over a longer range of time scales. This distinction matters during repeated ventilation because the mechanical state of the tissue is shaped not only by the immediately preceding breath, but also by the accumulated history of deformation, relaxation, recruitment, and derecruitment.

Integer-order models can approximate a power law only by combining many discrete elements. That strategy increases the number of parameters, complicates calibration, and weakens the physiological meaning of each parameter. The modeling problem is therefore not simply to add more parameters, but to represent a distributed material response without losing interpretability. Fractional-order modeling addresses this problem more compactly by treating tissue memory as continuous rather than as a collection of isolated relaxation elements.

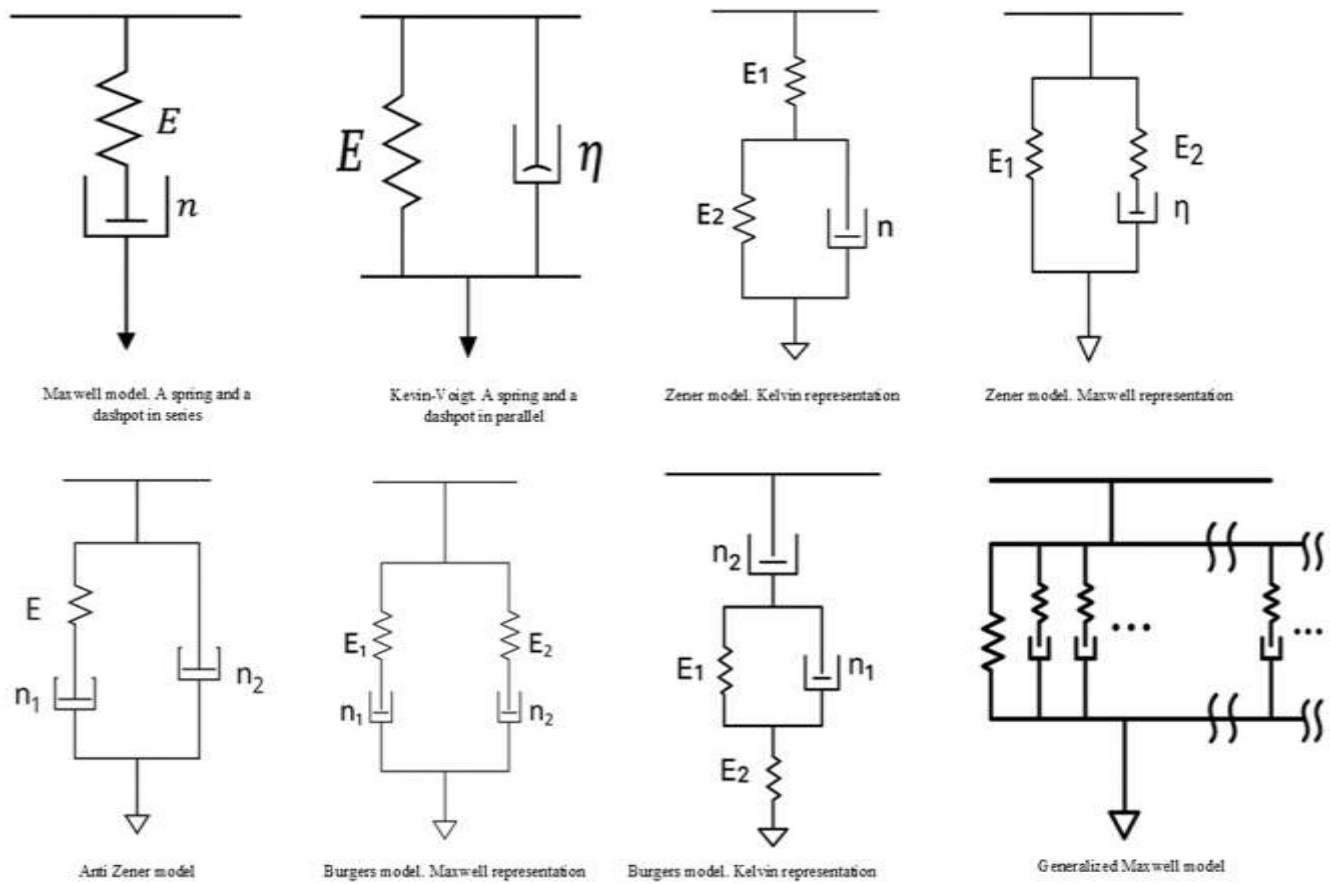


Figure 2: Classical Integer-Order Models

Fractional-Order Viscoelastic Models

Fractional calculus extends ordinary differentiation and integration to non-integer orders. In viscoelasticity, a fractional derivative is non-local: the current response depends on the history of deformation rather than only its instantaneous rate. This mathematical memory is well suited to a lung whose response reflects a broad, continuous distribution of relaxation times.

The spring-pot

The central element of fractional rheology is the spring-pot, also called the Scott–Blair element. It continuously interpolates between a spring and a dashpot through a fractional order, usually denoted by α .

- When $\alpha = 0$, the element behaves as a pure spring.
- When $\alpha = 1$, it behaves as a pure dashpot.
- When $0 < \alpha < 1$, it combines elastic energy storage with viscous energy dissipation and produces power-law behavior.

The fractional order is more than a curve-fitting parameter. It can summarize how closely the tissue behaves like an ideal solid or an ideal fluid and may reflect the degree of structural heterogeneity. Values in an intermediate range are consistent with tissue that is predominantly elastic but has a substantial viscous component. The exact value must be established and interpreted through validated measurements rather than treated as a universal clinical threshold.

Basic fractional models

The fractional **Maxwell model** replaces the classical dashpot with a spring-pot in series with a spring. It avoids the classical model's prediction of indefinitely linear creep and introduces a damped, history-dependent response.

The fractional **Kelvin–Voigt** model places a spring-pot in parallel with a spring. It improves the description of delayed creep but, like its classical counterpart, does not reproduce stress relaxation adequately and does not provide an instantaneous strain response to an abrupt stress.

Composite Fractional Models

The fractional **Zener model**, or fractional standard linear solid, combines a spring with a fractional Kelvin–Voigt or fractional Maxwell branch. It can represent an immediate elastic response, delayed relaxation, and a non-zero long-term equilibrium response in a compact structure.

The fractional **Burgers model** combines fractional Maxwell and fractional Kelvin–Voigt components. It can describe primary creep, prolonged creep, stress relaxation, and both immediate and delayed recovery. Its ability to generate power-law responses makes it a high-fidelity model for biological tissues, although the number of parameters and the difficulty of parameter identification remain important limitations.

The fractional **Anti-Zener model** combines a fractional Maxwell branch with a Scott–Blair element in parallel. It is particularly relevant when viscous flow and gradual yielding dominate, as may occur in severely injured lung tissue. It can represent linear creep, residual fractional elasticity, and power-law relaxation. Because of its fluid-like behavior, it is less appropriate for healthy lung tissue than for advanced pathological states.

Fractional models are not automatically superior in every application. They require careful fitting, validation against independent data, and clinically interpretable parameter estimation. Their value is strongest when they help bridge tissue-scale mechanics with clinical interpretation, explaining why the same tidal volume, driving pressure, or inspiratory flow can have different effects in lungs with different degrees of heterogeneity and mechanical memory.

Table 2: Comparative summary of fractional viscoelastic models for lung tissue mechanics

Model name	Configuration	Key features	Advantages	Limitations	Typical applications
Fractional Maxwell Model	Spring and spring-pot (fractional dashpot) in series	Captures power-law stress relaxation; fractional order α between 0 and 1	Accurate for long-term viscoelastic behavior; fewer parameters	Complex parameter estimation; limited physical interpretation	Modeling lung tissue viscoelasticity; polymer rheology
Fractional Kelvin-Voigt Model	Spring and spring-pot in parallel	Describes delayed creep; fractional order enhances flexibility	Good for short-term creep; simple structure	Less accurate for stress relaxation; parameter coupling	Soft tissue mechanics; damping analysis
Fractional Zener Model	Combination of spring, spring-pot, and dashpot (series/parallel)	Captures both creep and relaxation; intermediate complexity	Versatile; fits wide range of viscoelastic responses	Higher computational cost; parameter identifiability issues	Biomechanics; composite material modeling
Fractional Burger Model	Fractional Maxwell in series with Fractional Kelvin-Voigt	Describes complete creep and recovery cycles; power-law fitting	Superior fidelity for biological tissues; continuous relaxation spectrum	Most complex; requires advanced parameter estimation	Full viscoelastic characterization ; respiratory system
Fractional Anti-Zener Model	Scott-Blair dashpot in parallel with Fractional Maxwell	Models linear creep and fractional elasticity; ideal for pathological lungs	Quantifies energy dissipation; predicts power-law relaxation; memory effect	Advanced; best for severe pathology; less used in healthy tissue	ARDS, fibrosis, asthma; ventilator-induced injury

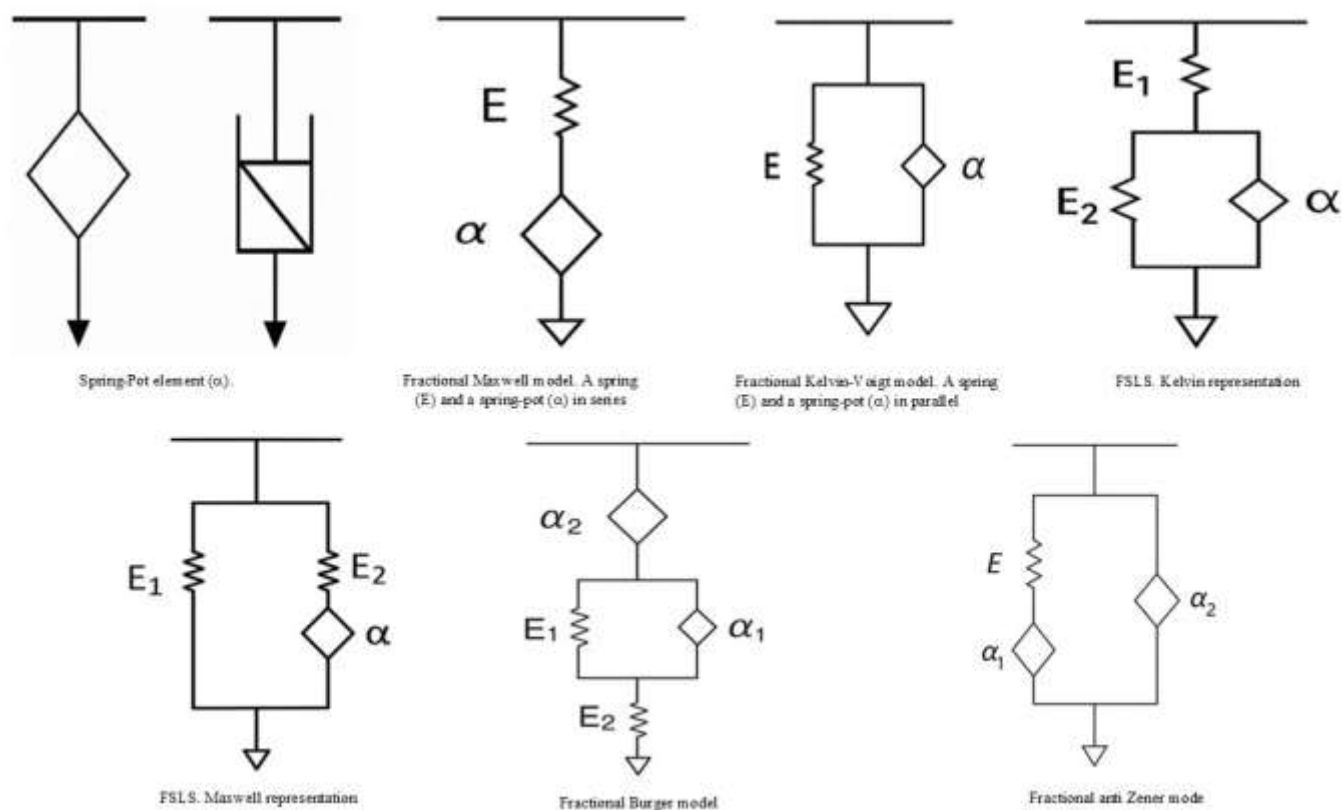


Figure 3: Fractional-Order Viscoelastic Models. FSLs: fractional standard linear solid

Clinical Relationship With Ventilator-Induced Lung Injury

A rheological interpretation unifies several mechanisms of ventilator-induced lung injury. High stress can produce structural damage. Excessive strain can cause overdistension. Repetitive collapse and reopening can cause atelectrauma. Rapid energy delivery increases strain rate and may intensify tissue stress even when total tidal volume is unchanged. The energy dissipated through hysteresis contributes to ergotrauma, or injury related to excessive mechanical energy transfer.

The key variables remain interconnected:

- **Stress** is related clinically to transpulmonary pressure and, imperfectly, to driving pressure.
- **Strain** is related to tidal volume relative to end-expiratory aerated lung volume.
- **Strain rate** is related to inspiratory flow and inspiratory time.
- **Shear stress** increases at interfaces between neighboring open and collapsed regions.
- **Creep** describes delayed deformation under sustained load.
- **Stress relaxation** appears as a pressure decline during constant-volume holding.
- **Hysteresis** reflects energy dissipation during the respiratory cycle.

Fractional models may eventually support real-time estimation of viscoelastic properties, but their clinical role would depend on whether their parameters can be measured reliably and interpreted consistently at the bedside. In principle, changes in the fractional order or related parameters could indicate alterations in tissue heterogeneity, recruitment potential, or disease severity. Such estimates might help individualize positive end-expiratory pressure, adjust flow and inspiratory time, and identify conditions in which a nominally safe tidal volume produces excessive regional strain.

These applications remain promising rather than established. A model-based parameter should not replace direct clinical assessment, and no single rheological index can determine the safest ventilator settings without considering gas exchange, hemodynamics, chest-wall mechanics, recruitability, and overdistension.

The practical objective is to keep stress, strain, strain rate, shear, and energy dissipation within a recoverable range. Tidal volume should be interpreted relative to end-expiratory aerated lung volume. Driving pressure and transpulmonary pressure should be considered when possible. Flow should be limited when it generates unnecessary strain rate. Positive end-expiratory pressure should be selected with attention to both recruitment and overdistension. Inspiratory holds and pressure–volume analysis can

provide useful information about resistance and relaxation, but their interpretation must account for airway resistance and tissue heterogeneity.

Overall, the models form a progression from interpretability to physiological fidelity. Classical models are easier to explain and estimate but compress lung behavior into a few discrete time constants. Fractional models preserve a compact structure while representing power-law relaxation and longer memory, making them better suited to heterogeneous injured tissue, provided their parameters can be validated and interpreted clinically.

Conclusion

The lung is a dynamic, heterogeneous, and history-dependent viscoelastic organ. Elasticity allows energy storage and recoil; viscosity produces rate-dependent resistance and energy dissipation. Their interaction creates creep, stress relaxation, and hysteresis, which cannot be fully represented by a model with fixed resistance and elastance.

Integer-order spring–dashpot models remain useful teaching tools and first approximations. Maxwell and Kelvin–Voigt models isolate relaxation and creep, the Zener model combines both behaviors, the Anti-Zener model represents more fluid-like pathological materials, and Burgers and generalized Maxwell models extend the range of phenomena that can be described. Across this family of models, the central limitation is the reliance on a small number of discrete relaxation times.

Fractional-order models address this weakness by using non-integer derivatives and the spring-pot element. They represent memory, power-law behavior, and a continuous distribution of relaxation times with comparatively compact formulations. They therefore offer a promising framework for describing healthy and injured lung tissue across broad time and frequency domains.

The move from classical to fractional models is therefore not merely mathematical. It reflects a shift from viewing the lung as a simple mechanical structure to viewing it as a multiscale biological material whose response depends on architecture, regional heterogeneity, loading history, and energy dissipation. The clinical value of this approach will ultimately be measured by whether it improves individualized ventilation and reduces irreversible injury.

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