A two-component simulation model to teach respiratory mechanics

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Kuebler WM, Mertens M, Pries AR. A two-component simulation model to teach respiratory mechanics. Adv Physiol Educ 31: 218–222, 2007; doi:10.1152/advan.00001.2007.—Interactive learning has been proven instrumental for the understanding of complex systems where the interaction of interdependent components is hard to envision. Due to the mechanical properties and mutual coupling of the lung and thorax, respiratory mechanics represent such a complex system, yet their understanding is essential for the diagnosis, prognosis, and treatment of various respiratory disorders. Here, we present a new mechanical model that allows for the simulation of respiratory pressure and volume changes in different ventilation modes. A bellows reflecting the “lung” is positioned within the inverted glass cylinder of a bell spirometer, which is sealed by a water lock and reflects the “thorax.” A counterweight attached to springs representing the elastic properties of the chest wall lifts the glass cylinder, thus creating negative “pleural” pressure inside the cylinder and inflating the bellows. Lung volume changes as well as pleural and intrapulmonary pressures are monitored during simulations of spontaneous ventilation, forced expiration, and mechanical ventilation, allowing for construction of respiratory pressure-volume curves. The mechanical model allows for simulation of respiratory pressure changes during different ventilation modes. Individual relaxation curves constructed for the lung and thorax reflect the basic physiological characteristics of the respiratory system. In self-assessment, 232 medical students passing the physiology laboratory course rated that interactive teaching at the simulation model increased their understanding of respiratory mechanics by 70% despite extensive prior didactic teaching. Hence, the newly developed simulation model fosters students’ comprehension of complex mechanical interactions and may advance the understanding of respiratory physiology.

Learning has been proven instrumental for the understanding of complex systems where the interaction of interdependent components is hard to envision. Due to the mechanical properties and mutual coupling of the lung and thorax, respiratory mechanics represent such a complex system, yet their understanding is essential for the diagnosis, prognosis, and treatment of various respiratory disorders. Here, we present a new mechanical model that simulates combined pulmonary and chest mechanics during spontaneous breathing and artificial ventilation. The model was designed to 1) visualize dynamic pressure changes during the respiratory cycle, 2) allow for reconstruction of the combined respiratory pressure-volume curve and its two contributing components individually, 3) compare respiratory mechanics in spontaneous and artificial ventilation, and 4) simulate pathophysiological situations, e.g., pneumothorax.

**MATERIALS AND METHODS**

The model is based on Hutchinson’s classical bell spirometer (5), which represents the “thorax” (Figs. 1 and 2). In brief, an inverted Plexiglas cylinder (total volume: 23 liters, 2.8 kg) is placed into a water-filled basin and suspended by a counterweight (3.5 kg) to create an expandable air-sealed compartment. The counterweight is connected to an elastic spring (10 N) allowing for a stable resting position and elastic inspiratory and expiratory movements of the thorax. Excessive “inspiratory” movements that might lift the thorax out of the water seal are prevented by a security line. A cylindrical rubber bellows (art. no. 00029793, Simrit, Weinheim, Germany; extended volume: 30 liters) was installed on a platform within the thorax to simulate the “lung.” A polyethylene tubing (“airways”) connects the bellows with ambient air or a high-volume syringe (model 5530, Hans Rudolph, Kansas City, MO), respectively. Gas volumes delivered to or from the bellows are determined by a volumeter (art. no. A1 2925 S, ACE, Simbach am Inn, Germany).

The thorax and lung are mechanically coupled via the confined “pleural” space between the Plexiglas cylinder and the bellows. Although mechanical coupling between the lung and thorax via air in the pleural space can be expected to be less efficient compared with the fluid-filled pleural space in vivo, the model generates realistic pleural pressures ($P_{\text{pleu}}$) due to the combined tendencies of the thorax to expand and the lung to collapse. An opening at the top of the cylinder is connected via a flexible tubing to a U-tube manometer filled with dyed water to allow visualization of $P_{\text{pleu}}$. Similarly, a manometer is connected via a T-connector to the airway tubing to measure lung pressure ($P_{\text{lun}}$)

By partial inflation of the bellows (via the pump) with parallel withdrawal of equivalent air volumes from the pleural space via a valve in the cylinder and adjustment of the elastic spring mounting, a stable end-expiratory resting position is established with the mechanical characteristics resembling...
those of the respiratory system. Starting from this end-expiratory resting position, the model can be used to perform the following respiratory maneuvers:

1. Spontaneous respiration is effected by adding weights to a platform attached to the counterweight, simulating the activation of inspiratory muscles, which induces inspiration by lifting the thorax and inflating the bellows. Removal of the weights results in “passive” expiration and return of the system to the end-expiratory resting position.

2. Forced expiration is simulated by adding weights on top of the Plexiglas cylinder, thus compressing the thorax and deflating the lung.

3. Mechanical ventilation is simulated by connecting the airway outlet from the lung to a high-volume syringe and subsequent inflation of the lung.

4. External pneumothorax is simulated by opening the valve in the Plexiglas cylinder, thus mechanically uncoupling the thorax from the lung.

From volumetric data and P pleu recordings obtained during spontaneous respiration, forced expiration, and mechanical ventilation, static pressure-volume curves of the respiratory system can be constructed. From P pul and P pleu data obtained during mechanical ventilation, “transpulmonary” pressures were calculated, and pressure-volume curves of the respiratory system and its two individual components, the thorax and lung, were subsequently constructed.

The learning effect of the simulation model was tested in 13 groups of 17–19 medical students during their physiological laboratory course. The students had recently received conventional didactic teaching on respiratory mechanics by lectures and textbooks. First, students were challenged to construct and explain pressure-volume relations during different respiratory maneuvers theoretically. Following a 20-min introduction to the simulation model, students then simulated the respiratory maneuvers and tested their theoretical hypotheses. The observed mechanical characteristics and measured pressure-volume curves were discussed in the group. At the end of the 90-min course, students were asked to complete a questionnaire. On a scale of 1 (worst) to 10 (best), students ranked 1) their comprehension of P pul and P pleu changes during spontaneous and mechanical ventilation, 2) their understanding of respiratory relaxation curves before and after simulation of respiratory mechanics, and 3) the overall usefulness of the model for the understanding of respiratory mechanics.

RESULTS

Spontaneous respiration. Sequential addition of 200-g weights to the counterweight platform results in gradual expansion of the “thoracic” compartment, thereby further reducing P pleu and causing a consecutive passive inflation of the lung. Pressure-volume curves generated from total inflated volumes (from the end-expiratory resting position) and P pleu measured after each weight addition and subsequent system stabilization showed an almost linear pressure-volume relation (Fig. 3, right). Due to the static conditions of the measurement, pressure-volume curves obtained during spontaneous inspiration and expiration are identical and P pul equals atmospheric pressure (data not shown).

Forced expiration. Consecutive addition of 200-g weights on top of the Plexiglas cylinder compresses the thoracic compartment, thus increasing P pleu and simulating expiration of the expiratory reserve volume. Pressure-volume curves constructed for forced expiration again showed an almost linear relation (Fig. 3, left) yet with a flatter slope (150 compared with 199 ml/cmH2O for spontaneous respiration), indicating a lower compliance of the model at lung volumes below the end-expiratory resting position.

Mechanical ventilation and relaxation curves. Lung inflation with defined volumes via the high-volume syringe connected to the outlet mimics mechanical ventilation with positive pressure and allowed for reconstruction of relaxation curves of the respiratory system and its two underlying components, the thorax and lung (Fig. 4). Mechanical lung inflation
increased both $P_{\text{pul}}$ and $P_{\text{pleu}}$, whereas mechanical deflation to lung volumes below the end-expiratory resting position reduced both pressures. The constructed relaxation curves reproduced basic principles of respiratory mechanics: the steep pressure-volume curve constructed for $P_{\text{pleu}}$ reflected the thorax with its tendency to expand toward its resting position. This thoracic resting position, which is defined by the $y$-axis intercept of the thoracic pressure-volume curve ($P_{\text{pleu}} = 0 \text{ cmH}_2\text{O}$), is only reached at a higher volume compared with the end-expiratory resting position of the combined respiratory system. The pressure-volume curve constructed for $P_{\text{pul}} - P_{\text{pleu}}$ reflects the lung with its tendency to collapse with $P_{\text{pul}} - P_{\text{pleu}}$ approaching zero. Similar to the in vivo situation, lung compliance decreases with higher inflation volumes due to elastic expansion limits of the rubber bellows. The pressure-volume relation constructed for $P_{\text{pul}}$ resembles the sigmoidal curve characteristic for the mechanics of the total respiratory system.

Notably, pressure-volume relations of the simulation model diverge from biological relaxation curves in several aspects. First, neither the thoracic resting position nor the flattening of the thoracic pressure-volume curve at lung volumes approaching the residual volume are reflected in the model. These limitations are largely attributable to the fact that pressure changes in the gas-filled pleural compartment are attenuated compared with the fluid-filled pleural space in vivo. As a direct consequence of this, the pressure-volume curves of lung and respiratory system do not intersect, and the compliance of the respiratory system does not decline during mechanical lung deflation. Second, due to its inherent weight, the rubber bellows requires a considerable opening pressure at the end-expiratory resting position, thus shifting apart the thoracic and pulmonary pressure-volume curves on the pressure axis.

Pneumothorax. Following mechanical uncoupling of the lung and thorax by opening the valve in the Plexiglas cylinder, the lung collapses. Active respiratory movements evoked by weights added to either the counterweight platform or Plexiglas cylinder no longer affect lung volume; only mechanical ven-

![Fig. 2. Photograph of the simulation model. Bar = 10 cm.](image)

![Fig. 3. Pressure-volume curve from the simulation model representing the relationship between $P_{\text{pul}}$ and lung volume changes during spontaneous ventilation (right) and forced expiration (left). Lung volume is expressed as the change (in ml) from the end-expiratory resting position.](image)

![Fig. 4. Relaxation curves from the mechanically ventilated simulation model. Relationships constructed from the transpulmonary pressure ($P_{\text{pul}} - P_{\text{pleu}}$) represent the relaxation curve of the lung; relationships constructed from $P_{\text{pleu}}$ represent the relaxation curve of the thorax; and relationships constructed from $P_{\text{pul}}$ represent the relaxation curve of the combined respiratory system. Lung volume is expressed as the change (in ml) from the end-expiratory resting position.](image)
tillation can inflate the lung, in which case \( P_{\text{pul}} \) is positive, whereas \( P_{\text{pleu}} \) remains zero.

**Student evaluation.** Medical students (\( n = 232 \)) with prior conventional theoretical instruction studied respiratory mechanics with the new simulation model. Questionnaires regarding their comprehension of changes in \( P_{\text{pul}} \) and \( P_{\text{pleu}} \) during the respiratory maneuvers and respiratory relaxation curves revealed a considerable subjective increase in physiological understanding (Fig. 5). On a scale of 1–10, students ranked the overall usefulness of the model as 6.8 ± 1.7 (mean ± SE). In their individual comments, students stressed that “this is a good model to finally understand compliance curves and many pathophysiological states” and that “without this model I would never have understood the different characteristics of compliance curves.”

**DISCUSSION**

Here, we present a new mechanical model to improve and facilitate students’ understanding of respiratory mechanics. This teaching tool allows for simulation of spontaneous respiration, forced expiration, and mechanical ventilation and for construction of respiratory relaxation curves and mimics pathophysiological situations such as external pneumothorax.

Respiratory mechanics and pressure-volume relations describe basic principles of ventilation. Hence, their understanding is a fundamental component of physiological training for medical students. A thorough analysis of pressure-volume curves can allow the clinician to diagnose lung disease, customize ventilator settings, follow the course of a disease, and make prognoses (4). Static pressure-volume curves can be used, for example, to optimize airway pressures in ventilated patients with acute respiratory distress syndrome (ARDS) and to identify ARDS patients who will profit from positive end-expiratory pressure ventilation (6, 9). Compliance and elastic values derived from both thorax and lung relaxation curves are often critical for the diagnosis of restrictive respiratory diseases (10) and allow for the important differentiation between the relative contributions of altered lung and chest wall dynamics (3).

Didactic teaching and private studies from textbooks seem to provide only a limited understanding of the complex interaction of lung and chest mechanics. Physiologists have recognized this problem and constructed models to simulate respiratory mechanics of single components of the respiratory apparatus. Weissenberg and Lavy (11) immersed an inflatable balloon in a water-filled measuring cylinder to simulate lung mechanics during inflation and deflation. From the obtained data, pressure-volume curves were plotted and could be used for on-line experiment simulation in silico. DiCarlo and colleagues (2) simulated the characteristic relaxation curves by either substituting a balloon for the lung or a tennis ball for the chest. In probably the most elaborate simulation model so far, Chase and co-workers (1) simulated lung mechanics by mechanically ventilating six rubber bellows that were connected in parallel. The pressures and compliance of each bellows were varied individually by superimposed adjustable weights, thus allowing for simulation of compartmentalized lung disease of different severity. All of these reported models simulate pressure-volume curves for either the isolated lung or isolated thorax but do not take into account the mutual interaction of both components.

To our knowledge, the mechanical model presented here is the first one allowing for simulation and interactive teaching of interdependent lung and chest mechanics. The model is entirely constructed from basic mechanical and inexpensive modules. It is ideally suited for frequent interactive teaching, since it is robust, easy to handle, and does not require elaborate maintenance or additional costs for consumables. Pressure-volume curves derived from the simulation model share the basic characteristics of biological relaxation curves. Notable differences are largely attributable to the higher opening pressure of the bellows compared with the lung and the larger compressibility of the gas-filled pleural space in the model compared with the fluid-filled pleural compartment in vivo. These characteristics might be further improved by adding counteracting mechanical components. Resulting differences to the biological relaxation curves need to be addressed with the students, and a thorough discussion of the underlying constraints will further sharpen their overall understanding of the interrelated respiratory mechanics.

By mechanically uncoupling the lung from the thorax, external pneumothorax was simulated. Potentially, the model could also be used to mimic additional pathophysiological scenarios such as restrictive or obstructive lung disease, e.g., by increasing the elastance of the rubber bellows by additional parallel rubber bands or by limiting airflow by application of a setscrew to the aeriferous inflow tube.

The acceptance and subjective learning benefit of this new interactive approach is reflected by its positive evaluation in a student questionnaire. The better understanding of respiratory mechanics may raise interest and insights into lung (patho-)physiology and ultimately help to improve respiratory patient care.
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