Design of a Lung Simulator for Teaching Lung Mechanics in Mechanical Ventilation

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Over the last 10 years, noninvasive ventilation has become a treatment option for respiratory insufficiency in pulmonology services. The technique is currently included in pulmonology teaching programs. Physicians and nurses should understand the devices they use and the interaction between the patient and the ventilator in terms of respiratory mechanics, adaptation, and synchronization. We present a readily assembled lung simulator for teaching purposes that is reproducible and interactive. Based on a bag-in-box system, this model allows the concepts of respiratory mechanics in mechanical ventilation to be taught simply and graphically in that it reproduces the patterns of restriction, obstruction, and the presence of leaks. It is possible to demonstrate how each ventilation parameter acts and the mechanical response elicited. It can also readily simulate asynchrony and demonstrate how this problem can be corrected.

Key words: *Noninvasive ventilation. Lung simulator. Patient–ventilator interaction. Asynchrony. Leaks.*

Diseño de un simulador de pulmón para el aprendizaje de la mecánica pulmonar en ventilación mecánica

Desde la última década la ventilación no invasiva se ha incorporado al tratamiento de la insuficiencia respiratoria en los servicios de neumología, y actualmente forma parte del plan de formación de esta especialidad. Médicos y enfermeras deben conocer los equipos con los que trabajan y entender la interacción que se produce entre el paciente y el ventilador en términos de mecánica respiratoria y de adaptación y sincronización. Presentamos un modelo de simulador de pulmón de fácil montaje, reproducible e interactivo, que permite alcanzar estos objetivos. Basado en un sistema de *bag-in-box*, este modelo permite aprender de forma sencilla y gráfica la mayoría de los conceptos de la mecánica respiratoria en ventilación mecánica, pues reproduce patrones de restricción, obstrucción o presencia de fugas. Puede comprobarse cómo actúa cada parámetro del ventilador y la respuesta mecánica que genera, y permite simular numerosas asincronías, así como el modo correspondiente de corregirlas.

Palabras clave: *Ventilación no invasiva. Simulador de pulmón. Interacción paciente-ventilador. Asincronías. Fugas.*

Introduction

Noninvasive mechanical ventilation (NIV) consists of applying positive pressure to the airways in order to increase alveolar ventilation without having to intubate the patient. The interface between the patient and the ventilator is normally a nasal or face mask, which avoids the complications involved in endotracheal intubation or tracheotomy. However, the system is subject to leaks, which can cause asynchronies and lead to failure of this technique.[1] Over the last 10 years, NIV has become a treatment option for respiratory insufficiency in intensive care units, emergency departments, pulmonology services, and in the home.[3-5] In learning this technique, trainee pulmonologists are often faced with a patient and a ventilator without knowing how the ventilator works and how it interacts with the patient. Unlike invasive ventilation, NIV has the added difficulty that the patient is conscious and required to cooperate actively in the process; if the ventilator parameters are not correct, the patient may reject the treatment. In this case, NIV may lead to unnecessary intubation and be worse than the conventional treatment. In order to use this technique correctly, a detailed knowledge of how each parameter works is necessary. It is also necessary to be able to detect, in time, problems with adaptation and the complications that may arise from leaks, and to recognize asynchronies on a monitor and correct them. Before use, practical training is needed to ensure an optimum and effective treatment.[6,7] We present a model for a lung simulator as a useful tool in NIV training.

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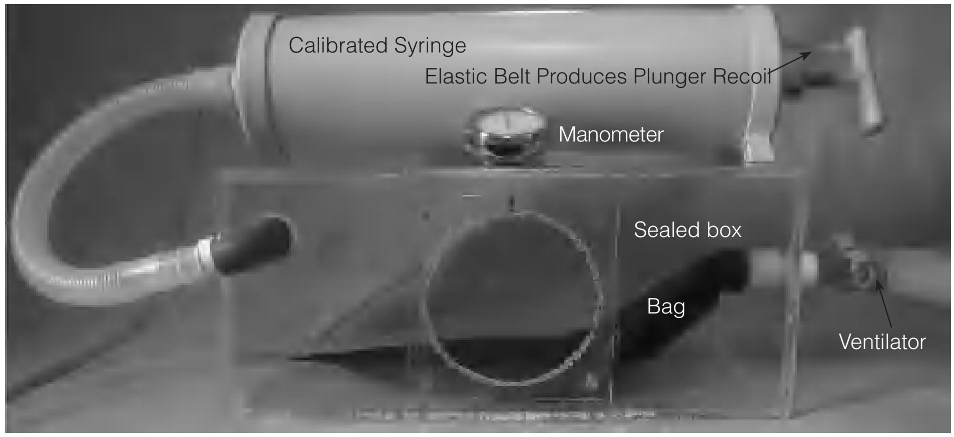

Figure 1. Bag-in-Box Lung Simulator System. The 2-L bag is connected to the ventilator and the 8-L box is connected to a 3-L calibrated syringe with an elastic recoil to depress the plunger.

Description of the Technique

This is a technique for teaching mechanical ventilation, aimed at providing an understanding of asynchronies and the mechanics of ventilation. The technique involves the use of the lung simulator that consist of a closed bag-in-box system (Figure 1), in which the bag is connected to the ventilator and the box is connected to a calibrated syringe with elastic recoil of the plunger. The pressure inside the box—corresponding to the intrapleural pressure—is monitored by means of a manometer. The system allows data on respiratory mechanics to be monitored as the manometer connected to the box records the pressures exerted by the ventilator and the volume administered can be determined from the position of the plunger in the calibrated syringe. This basic design is simple and easy to use and can also be coupled to a computerized system that makes it possible to record the pressure, volume, and flow signals, and present them alongside the signals provided directly by the ventilator.

When the simulator is functioning with a ventilator, at the end of expiration, the bag is at residual functional capacity with no pressure in the system and undergoes a sudden, small change in shape, returning toward its resting position in the absence of any external force. This change in the shape of the bag generates a small change in pressure and flow that, although minimal, is detected by the ventilator as the beginning of an inspiratory impulse that starts the ventilator on a new cycle and makes it possible to keep the system in assisted ventilation mode.

Different accessories can be fitted to simulate a restrictive pattern (increasing the elastic recoil on the plunger), an obstructive pattern (fitting a valve to limit airflow through the expiration tube), or behavior in the event of different sized leaks. The simulations reported here were performed using an Elysée 150 double-tube pressure support ventilator (Saime, Savigny le Temple, France) with a touch screen that provides analysis of flow, pressure, and volume signals in real time. A computer interface makes it possible to view these curves and to analyze them together with inspiratory and expiratory tidal volume, compliance of the system, breathing rate, inspiratory and expiratory flow rate, intrinsic positive end-expiratory pressure (PEEP), peak pressure, etc. All pressure, flow, and volume curves over time shown in the figures in this article come from real data obtained using the procedure described below.

Procedure

When the simulator is functioning with the ventilator, the model can be configured to reproduce, for teaching purposes, the most important characteristics of the mechanical behavior of the respiratory system during NIV. It is possible to simulate basic restrictive and obstructive lung-function patterns and to observe the consequences of ventilation when these patterns are present. Furthermore, the model makes it possible to reproduce the existing responses when the parameters of the ventilator are altered and thus simulate asynchronies that are similar, in practice, to those that occur in real cases of patients under NIV and that can be controlled by adjusting the ventilatory parameters.

Restrictive Model, Obstructive Model, Air Entrapment, and Positive End-Expiratory Pressure

The simulator can reproduce a restrictive pattern by increasing the elastic load on the plunger. By operating the simulator manually, the observer can see that greater effort is required to retract the syringe plunger in order to inflate the bag than under normal conditions, thus reflecting the increased effort required by the patient's respiratory musculature. Similarly, the observer can see that, when the syringe plunger is released, the increased elastic recoil causes the bag to empty quickly and effectively, thus demonstrating that the principal mechanical problem is filling and not emptying. If the ventilator is connected to the simulator under these conditions, it will be seen that high pressures and continuously maintained pressure ramps are required to achieve uniform filling and to regulate the bag (Figure 2A).

An obstructive pattern can be easily reproduced by means of different combinations of reducing the elastic recoil on the syringe plunger and placing an obstruction in the expiratory tube (Figure 2B). By operating the

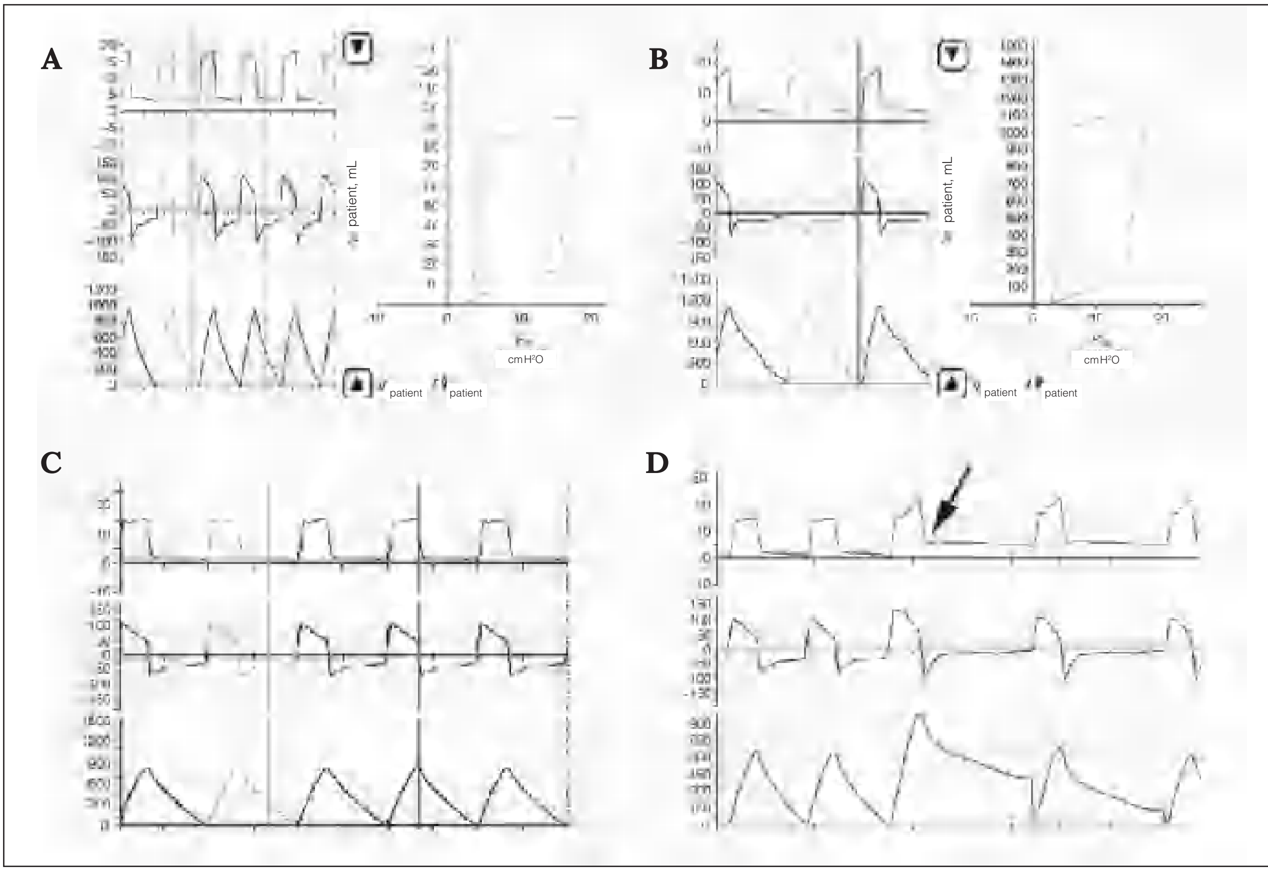

Figure 2. **A: restrictive model. Note the reduced compliance in the pressure-volume curve. B: obstructive model. Note the increased compliance in the pressure-volume curve and the prolonged expiratory time in comparison with the restrictive model. C: positive end-expiratory pressure (PEEP) generated by maintaining rapid auto-cycling with expiratory obstruction. Entrapment of air in the bag is shown in the flow curve and in the pressure curve, which do not return to 0 at the end of expiration. D: auto-PEEP corrected with extrinsic PEEP. Note that applying an external PEEP (arrow) equal to the intrinsic PEEP (shown in the graph by the fact that the end expiratory flow does not reach 0) improves tidal volume and reduces the breathing rate.**

simulator manually, the observer can see that very little effort is required to retract the plunger but that, when it is released, the bag empties far more slowly than under normal conditions. When the simulator is functioning with the ventilator in this configuration, the need for a long emptying time shows that the bag may not completely empty on expiration, leaving a larger residual volume that can be easily measured by means of the calibrated syringe and the manometer on the box. At this point, the manometer will show the end-expiratory pressure that corresponds to the intrinsic PEEP (Figure 2C). It may also be seen how adding an extrinsic PEEP equal to the intrinsic PEEP (Figure 2D) improves the tidal volume and hence the expansion of the bag in the following cycle, and reduces the breathing rate, exactly as occurs in patients under the same conditions.[8] When the simulator is not subject to an obstructive pattern, application of an extrinsic PEEP during ventilation shows how the bag does not empty completely and the manometer registers a positive pressure at the end of expiration.

Simulating Asynchronies

The design makes it possible to work with a single-tube calibrated-leak system or a double-tube system with no leaks. As the main cause of asynchronies during NIV is the existence of leaks, the ability to produce leaks in this model in the inspiratory phase, expiratory phase, or both, increases the usefulness of the simulator from a teaching point of view. When the mechanical ventilator is connected to the simulator, it is possible to observe the behavior of both types of leak and the responses obtained when the different ventilatory parameters, such as inspiratory and expiratory thresholds, ramps, and inspiratory or expiratory pressures, are modified.

Inspiratory-trigger related asynchronies: A highly sensitive inspiratory trigger is able to detect any change in pressure or flow in the system and interpret it as patient demand. The characteristic of the simulator that allows auto-cycling can only be observed if a ventilator is connected in support-pressure mode without PEEP as this is the only way in which the bag will be completely emptied, thus leading to the previously-mentioned change in the shape of the bag and generating a small inspiratory impulse that triggers a new cycle. This system provides an indefinite auto-cycling feature that can be tried in any desired situation. It is thus possible to study the effect of changing inspiratory thresholds or triggers. In terms of concept, the auto-triggering of the cycle would be what occurs in a patient under NIV when an over-sensitive inspiratory trigger is selected: the ventilator detects any minimal change in pressure or flow, even when it is not due to a real inspiratory effort and this produces an "auto-cycling" asynchrony. Auto-cycling or auto-triggering is a troublesome and highly common asynchrony that can be remedied simply by programming the inspiratory trigger for a lower level of sensitivity.[9,10] It may be observed how setting the inspiratory trigger to a lower level of sensitivity stops the ventilator from auto-cycling (Figure 3A). However, if the sensitivity of the inspiratory trigger is programmed at too low a setting, "ineffective effort" asynchrony may occur. This type of asynchrony requires considerable respiratory effort on the part of the patients and consists of the inability to activate the inspiratory cycle of the ventilator. It may lead to auto-PEEP, muscle exhaustion, excessively long inspiratory ramps, or incorrect programming of the inspiratory trigger.[11] Manual operation of the simulator syringe plunger demonstrates the pressure needed to activate the ventilator and shows that this pressure is higher in the event of auto-PEEP and that no response occurs until the positive pressure shown on the manometer (auto-PEEP) is canceled.[12] Another method of correcting auto-cycling involves adding a small external PEEP that prevents the bag from reaching rest point. This has the effect of cancelling the effect caused by the sudden change in the shape of the bag that gives rise to auto-cycling (Figure 3B).

Ramp-related asynchronies: The simulator also makes it possible to study the use of different ramps or pressurization rates. The simulator provides a graphic demonstration of the effect of filling the bag at different rates by selecting fast or slow ramps (Figure 3C). Observation of the calibrated syringe also shows that the inflation volume can be altered by varying the compliance of the system. If the elastic recoil on the plunger is increased—simulating a restrictive model—and short

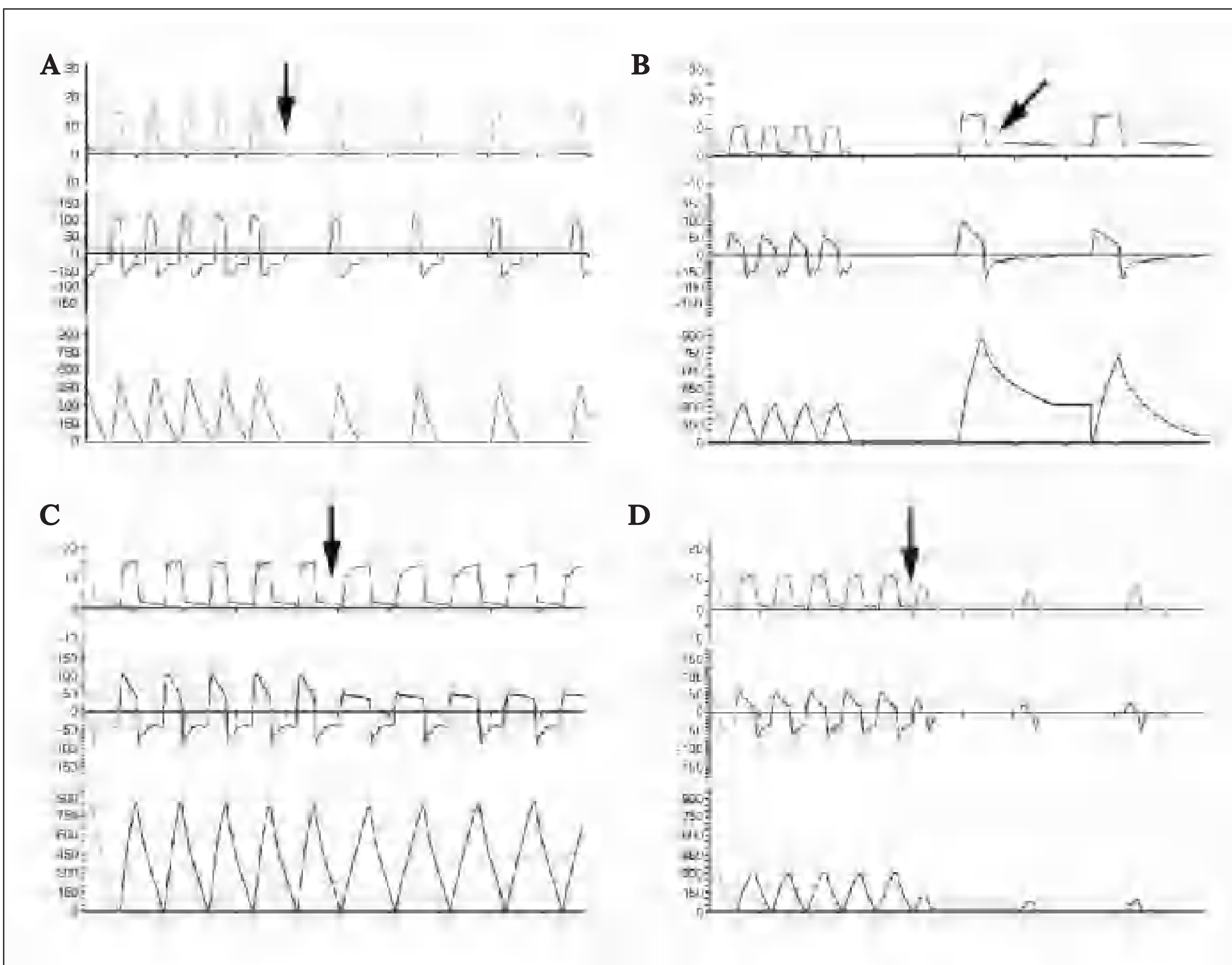

Figure 3. A: auto-trigger with correction using positive end-expiratory pressure (PEEP). Note that applying an extrinsic PEEP (arrow) cancels auto-cycling by preventing the sudden pressure change at the end of expiration that occurs in the model (see text). B: auto-trigger with correction using the trigger. note that applying a less sensitive inspiratory trigger (arrow) cancels auto-cycling. C: ramps. Note the high pressurization rate (ramp, 1/4) in the 5 initial cycles and the slow rate (ramp 4/4) in the 5 subsequent cycles (arrow indicates change of pressurization speed). D: note the presence of short cycles when the ramp is extended (arrow) in a restrictive system with low support pressure.

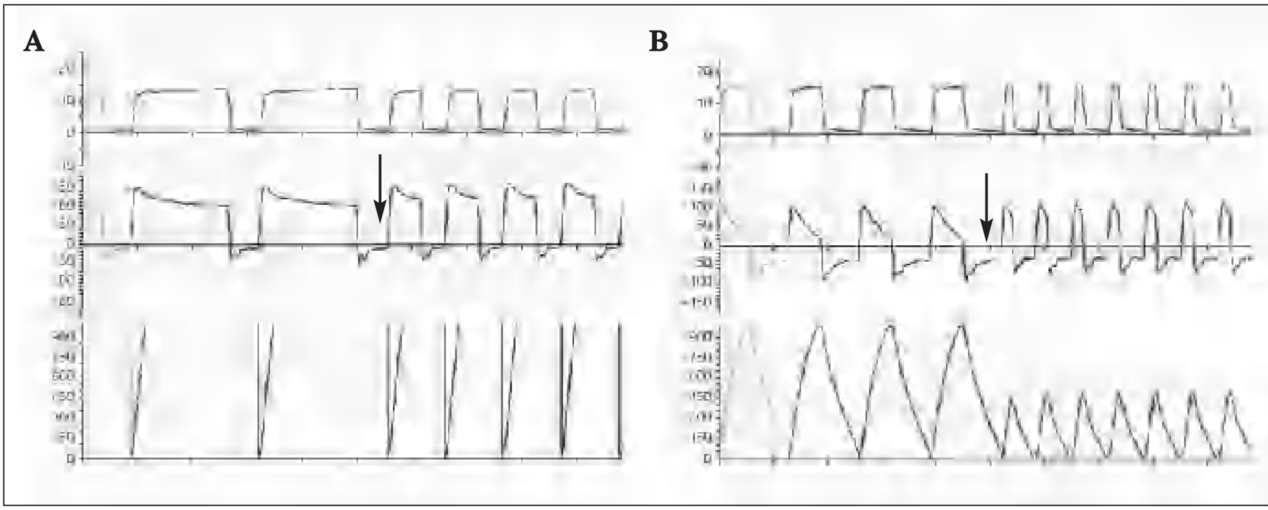

Figure 4. A: longer inspiration when a leak is opened in the tubes (inflation volume is very high and maximum inspiration time is set to 3 s). Asynchrony disappears when the maximum inspiration time is set to 1 s (arrow). B: expiratory trigger at 10%, causing the switch to the expiration cycle to take place when only 10% remains to complete the drop in inspiratory flow. The cycles are longer and respiration slower. However, when the expiratory trigger is set to 90% (arrow), the switch to the expiration cycle occurs earlier, thus increasing the breathing rate and reducing both the inspiratory time and the tidal volume.

ramps are used, the corresponding respiratory cycle will also be short (Figure 3D) in terms of inspiration time and the inspired volume will be small.[13] This asynchrony is called "short-cycle" asynchrony. By programming a longer ramp, the inspiration time and inspired volume will increase, thus reproducing what takes place in patients with a restrictive pattern who require ventilation using volumetric or barometric modes, with longer ramps. In these patients, however, if the ramp is too long, the short-cycle asynchrony may reappear.[14] Patients with acute obstructive patterns, on the other hand, usually require short ramps to improve adaptation to the ventilator.[15]

Leak-related asynchronies: Asynchronies secondary to the presence of leaks can also be easily simulated by opening to the outside a small orifice of variable size at the point where the ventilator tubes connect to the simulator. If a leak is caused during inspiration, the ventilator will not reach the preset inspiratory pressure and will continue to inflate indefinitely. This asynchrony is known as "prolonged inspiration"[16,17] and can be controlled by cancelling the leak, setting a maximum inspiration time, or increasing the sensitivity of the expiratory trigger (Figure 4A). This is a very common asynchrony that causes considerable anxiety in patients (as they cannot exhale), leads to hyperinflation, and generates auto-PEEP.[18] All of these effects can be shown in the simulator by gradually entrapping air in the bag and progressively increasing the pressure in the box at the end of expiration.

The expiratory trigger (Figure 4B) is the key element that makes it possible to switch from the inspiratory cycle to the expiratory cycle and is extremely important for synchronization.[19] When this trigger is incorrectly adjusted, it can lead to failure of NIV. If the sensitivity of the expiratory trigger is set too low (late switching to the expiratory cycle), the patient may not be able to expire fully and this can lead to hyperinflation. If, on the other hand, the sensitivity level is set too high (early switching to the expiratory cycle), the ventilator may switch to the expiratory cycle when the patient is still inspiring, thus increasing the breathing rate and leading to ineffective, superficial breathing.[20] A comparison of the inspiratory time (Figure 4A) and expiratory flow (Figure 4B) parameters shows that the model is capable of reproducing the behavior of patients who tend to present better conservation of expired tidal volume and curve shape when the respiratory cycle is controlled by the expiratory trigger rather than by the maximum inspiratory time. This data can be seen in the shape of the flow curves in Figure 4A and Figure 4B.

Limitations

The limitations of the simulator described here are due to the absence of an inspiratory impulse to activate the inspiratory cycle and examine the related problems for NIV. However, as described in the Procedures section, false impulses may occur due to the inertia of the bag and these can be used to study the sensitivity of the respirator inspiratory trigger (Figure 3A). As has been described, the simulator is also limited by the fact that the mechanical behavior cannot be recorded for subsequent analysis. Changes in volume can only be observed in the movement of the plunger and the pressure by means of the manometer. This model could be easily adapted to allow these data to be stored and subsequently analyzed, together with the signals from the ventilator, by fitting a pneumotachograph and a pressure transducer.

Conclusions

The data relating to the use of this simulator and the options it allows for selecting different variables show that ventilators used in situations of acute respiratory insufficiency should have a screen that allows the mechanical pressure, flow, and volume variables to be monitored so that all these asynchronies and the effect produced by each change of parameter can be seen and corrected.

In conclusion, we present a model for a lung simulator that is easy to assemble, readily available, and within the reach of any mechanical ventilation unit. It does not require any electronic devices, is easy to transport, and can be used with any single-tube or double-tube respirator. It is very easy to use and can help train physicians and nurses in respiratory mechanics in mechanical-ventilation situations, with special emphasis on patient-ventilator interaction.

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