Innovations in Mechanical Ventilation

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New features of mechanical ventilators are frequently introduced, including new modes, monitoring techniques, and triggering techniques. But new rarely translates into any measureable improvement in outcome. We describe 4 new techniques and attempt to define what is a new invention versus what is *innovative*—a technique that significantly improves a measurable variable. We describe and review the literature on automated weaning, automated measurement of functional residual capacity, neural triggering, and novel displays of respiratory mechanics. *Key words: mechanical ventilation, weaning, neurally adjusted ventilatory assistance, ventilator, functional residual capacity.* [Respir Care 2009;54(7):933–947. © 2009 Daedalus Enterprises]

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Introduction

Mechanical ventilators are among the most sophisticated and most expensive devices in the intensive care unit (ICU). In fact, modern day ICUs can be characterized as the common space where mechanical ventilation of critically ill patients is provided. Ventilators have evolved from small, pneumatically powered and pneumatically controlled devices to marvels of microprocessor technology capable of closed-loop control.

The competition in the mechanical-ventilator marketplace is fierce and "new generation" ventilators are introduced frequently or new innovations are added. These innovations include new techniques, new modes, new monitoring, new displays, and new trigger and cycle variables.[1-4] Whether these represent real advances based on science or simply new "bells and whistles" to attract attention is often lost in the lexicon of strategic marketing and device-specific education.

According to Wikipedia, innovation is:

> a new way of doing something. It may refer to incremental, radical, and revolutionary changes in thinking, products, processes, or organizations. A distinction is typically made between *invention*—an idea made manifest—and *innovation*—ideas applied successfully. In many fields, something new must be substantially different to be innovative, not an insignificant change (eg, in the arts, economics, business, and government policy). In economics the change must increase value, customer value, or producer value. *The goal of innovation is positive change, to make someone or something better.*[5] [italics mine]

It is an interesting insight, to include separate definitions for innovation and invention. Clearly, invention has been the hallmark of mechanical ventilation evolution to date. While we might all agree that changes in ventilators have made them safer, easier to use, and more accurate in monitoring the patient, there have been few real innovations.

This paper reviews several new commercially available features of mechanical ventilators. We will try to keep in mind the difference between invention and true innovation. Not every new feature can be covered. In an effort to be complete, we discuss several features that are quite different from one another, including closed-loop control of ventilator weaning, automated measurement of functional residual capacity (FRC), neural triggering/cycling, and information display.

Closed-Loop Control of Weaning

Weaning from mechanical ventilation has undergone wholesale changes over the last decade. Improved under-standing of sedation, delirium, and weaning predictors has led to evidence-based development of ventilator-discontinuation guidelines based on the weaning-readiness screening and daily spontaneous breathing trails.[6] Discontinuation of mechanical ventilation does not always involve "weaning," which is the slow withdrawal of support, but the term "weaning" seems to be embedded in the respiratory-care lexicon and we will use it here.

History

Automated weaning is not a microprocessor-based invention; it has roots in the introduction of mandatory minute volume (MMV), in 1977, by Hewlett.[7] With an Engstrom ventilator, Hewlett devised a system to guarantee a pre-set minute volume ($\dot{V}_E$), based on a mechanical system. Conceptually, MMV would allow the patient to take over ventilation as lung mechanics improved and the patient assumed the work of breathing.[8-11] In that initial version of MMV, the $\dot{V}_E$ was guaranteed by an automatic increase in the set respiratory rate.

Subsequent versions of MMV have used increases in pressure support and/or the number of mandatory breaths to guarantee a minimum $\dot{V}_E$. More recently, ventilators have been developed that allow an increase of the rate or the pressure-support level, based on the underlying support mode.

Though MMV was introduced over 30 years ago, we have no evidence of its efficacy. Claure et al studied MMV in infants and found that, compared to intermittent mandatory ventilation, there were fewer mandatory breaths, lower peak airway pressure, and smaller tidal volume (V_T).[10] The only paper published on the use of MMV during the present decade was by Guthrie et al, who randomly applied synchronized intermittent mandatory ventilation and MMV in sequence for 2-hour periods. MMV resulted in fewer mandatory breaths and more spontaneous breaths, and, consequently, a lower mean airway pressure.[11] However, they did not evaluate the time to discontinuation of mechanical ventilation.

Adaptive support ventilation (ASV) is also a technique capable of automated weaning.[12-15] ASV can also choose initial ventilator settings and escalate ventilatory support when ventilation targets are not met. ASV uses a $\dot{V}_E$ target based on predicted body weight and a clinician-set percentage of the predicted $\dot{V}_E$. As an example, in adults, predicted $\dot{V}_E$ is 0.1 L/kg/min (eg, a 70-kg patient would receive a $\dot{V}_E$ of 7.0 L at a percent setting of 100%). Changing the percent $\dot{V}_E$ setting to 150% would change the target to 10.5 L/min. V_T and respiratory rate are portioned based on the minimum-work-of-breathing algorithm described by Otis et al.[13] A complete description of the nuances of ASV can be found elsewhere.[16,17]

ASV has been studied as a method to speed ventilator-discontinuation after cardiac surgery.[18-22] Those studies found that ASV is safe and effective, but the time to extubation differed substantially. In patients who had "fast-track" surgery, ASV shortened the time to extubation.[18-21] A recent trial by Dongelmans and colleagues found no difference in weaning time between ASV and traditional weaning in non-fast-track cardiac-surgery patients.[22] These conflicting data may well represent the importance of choosing the appropriate $\dot{V}_E$ target. If the percent $\dot{V}_E$ is too high the patient may be prone to increased pressure support, periodic breathing, and slower weaning. A more aggressive approach that is tailored toward pushing the patient toward spontaneous breathing may be warranted when ASV is applied as a weaning tool.

SmartCare/PS

Dojat et al initially described a closed-loop system for mechanical ventilation known as NeoGanesh in 1992.[12] Ganesh is the Hindu elephant-deity often depicted with 4 arms and riding a mouse. Ganesh is known as the remover of obstacles, the patron of arts and sciences, and the diva of intellect and wisdom.[23] NeoGanesh is perhaps aptly named for a technique to automatically withdraw mechanical ventilation, although, as we will see, the commercial version only changes the pressure-support setting, negating the need for many arms. Introduced as SmartCare/PS (Dräger, Telford, Pennsylvania) in 2008, this system uses measurements of the respiratory rate, V_T, and partial pressure of end-tidal carbon dioxide (P_{ETCO_2}) to control the pressure-support ventilation (PSV). Unlike previously introduced adaptive-control modes, this system uses several inputs in an effort to maintain patient comfort in a defined respiratory-rate range. Initially, the system adjusts pressure support to maintain a respiratory rate of 12–28 breaths/min, a V_T above a minimum clinician-set threshold (250–300 mL), and P_{ETCO_2} below a clinician-set threshold (55 mm Hg for normals, 65 mm Hg for patients with chronic obstructive pulmonary disease). If ventilation remains within the prescribed range for a predetermined period, the system automatically reduces the pressure-support setting in an effort to facilitate weaning.

The commercial version of SmartCare/PS identifies the patient's breathing pattern as normal ventilation, insufficient ventilation, hypoventilation, hyperventilation, and tachypnea. Table 1 describes the breathing patterns, the parameters that define those patterns, and the response of the SmartCare/PS algorithm to patterns outside the normal ventilation parameters.

The first trial of SmartCare/PS was with 19 patients. The system effectively maintained respiratory rate in the prescribed range for up to 24 hours.[24] The patients were classified as "weanable" or "unweanable" based on results of weaning parameters. The weanable patients remained in the prescribed "comfort range" for 95% of the duration of ventilation, whereas the unweanable patients were in the prescribed "comfort range" for 72% of the duration of ventilation. These initial results were considered a success for the technique. Dojat et al studied the SmartCare/PS system's ability to predict weaning success, compared to a conventional weaning technique,[24] and found that the SmartCare/PS system had a positive predictive value of 89%, compared to 77% for conventional weaning.

Another study by Dojat et al studied 10 patients who required PSV following acute lung injury (ALI).[25] PSV was implemented with a modified Veolar ventilator (Hamilton, Reno, Nevada), which continuously monitored patient respiratory rate, V_T, and airway pressure. A stand-alone, mainstream CO_2 monitor measured P_{ETCO_2}. All variables were measured every 10 seconds and averaged over a 2-min period.

The principle of control was to maintain ventilation within an "acceptable" range by automatic adjustments of pressure support. "Acceptable" ventilation was defined as a respiratory rate between 12 and 28 breaths/min, a V_T > 250 mL (> 300 mL in patients > 50 kg), and P_{ETCO_2} < 55 mm Hg (< 65 mm Hg in patients with chronic obstructive pulmonary disease). If the respiratory rate was 28–35 breaths/min and V_T and P_{ETCO_2} were in the acceptable ranges, the breathing pattern was denoted as "intermediate respiratory rate." In that situation, the pressure support was increased by 2 cm H_2O. If the respiratory rate was > 35 breaths/min, the pattern was denoted as "high respiratory rate" and pressure support was increased by 4 cm H_2O. If respiratory rate was < 12 breaths/min, the pattern was denoted as "low respiratory rate" and the pressure support was reduced by 4 cm H_2O. If V_T was low (< 250 mL) or P_{ETCO_2} was high, pressure support was increased by 2 cm H_2O. Apnea caused the ventilator to revert to continuous mandatory ventilation.

Additionally, if the ventilation was acceptable for 30 minutes, pressure support was reduced by 2 cm H_2O if the pressure-support level was < 15 cm H_2O, or by 4 cm H_2O if the pressure-support level was > 15 cm H_2O and ventilation had been stable for 60 minutes. The system tolerated transient instabilities of 2–4 min, depending on the pressure-support level. Tachypnea or inadequate ventilation longer than 2 minutes caused the pressure-support level to be increased by 2 cm H_2O if the pressure-support level was < 15 cm H_2O, or by 4 cm H_2O if the pressure-support level was > 15 cm H_2O. If 3 consecutive observation periods failed to have "adequate" ventilation despite changes in the pressure-support level, the ventilator displayed a message. When the pressure-support level had been < 10 cm H_2O for 2 hours and ventilation had been stable, the ventilator displayed a message suggesting a spontaneous breathing trial and discontinuation of mechan-

Table 1. Ventilation Patterns, Parameters That Define Those Patterns, and Response of the SmartCare/PS Algorithm

Breathing Pattern	Variable	Range	Qualifier	Response of SmartCare/PS
Normal ventilation	f (breaths/min)	15–30	No neurologic disease	No change
		15–34	Neurologic disease	
	V_T (mL)	> 300	> 55 kg	Pressure support decreased by 2–4 cm H_2O at 15-min, 30-min, or 60-min intervals, based on current level of pressure support
		> 250	36–55 kg	
	P_{ETCO_2} (mm Hg)	< 55	No COPD	
		< 65	COPD	
Insufficient ventilation	f (breaths/min)	Acceptable	None	Pressure support increased by 2–4 cm H_2O, depending on current level of pressure support
	V_T (mL)	< 300	Weight	
	P_{ETCO_2} (mm Hg)	> 55	COPD	
Hypoventilation	f (breaths/min)	< 15	None	Pressure support increased immediately by 4 cm H_2O
	V_T (mL)	> 300	Weight	
	P_{ETCO_2} (mm Hg)	> 55	COPD	
Central hypoventilation	f (breaths/min)	< 15	None	No change. If detected at 3 consecutive evaluations, alarm requests that clinician evaluate patient.
	V_T (mL)	< 300	Weight	
	P_{ETCO_2} (mm Hg)	> 55	COPD	
Tachypnea	f (breaths/min)	30–36	Neurologic disease	Pressure support increased immediately, by 2–4 cm H_2O, depending on current level f pressure support. If detected at 3 consecutive evaluations, alarm requests that clinician evaluate patient.
	V_T (mL)	> 300	None	
	P_{ETCO_2} (mm Hg)	< 55	None	
Severe tachypnea	f (breaths/min)	> 36	Neurologic disease	Immediate increase in pressure support. by 4 cm H_2O.
	V_T (mL)	> 300	Weight	
	P_{ETCO_2} (mm Hg)	< 55	COPD	If detected at 3 consecutive evaluations, alarm requests that clinician evaluate patient.
Hyperventilation	f (breaths/min)	< 15	Weight	Pressure support decreased immediately, by 4 cm H_2O
	V_T (mL)	> 300	COPD	
	P_{ETCO_2} (mm Hg)	< 55		
Unexplained hyperventilation	f (breaths/min)	> 30	Neurologic disease	No change. If detected at 3 consecutive evaluations, alarm requests that clinician evaluate patient.
	V_T (mL)	> 300	Weight	
	P_{ETCO_2} (mm Hg)	< 45	COPD	

f = respiratory frequency per minute
V_T = tidal volume
P_{ETCO_2} = end-tidal CO_2 pressure
COPD = chronic obstructive pulmonary disease
(Adapted from Reference 26.)

ical ventilation. Dojat et al found that closed-loop PSV resulted in more time in "acceptable" ventilation and less time in critical ventilation.[25]

Lellouche et al recently completed a randomized controlled trial of SmartCare/PS versus traditional weaning in 144 patients, in 5 centers in Europe.[27] They found significant reductions in the time to extubation, duration of mechanical ventilation until extubation, time to successful extubation, total duration of mechanical ventilation, and ICU stay. Criticisms of that trial include the fact that not all of the 5 centers used a weaning protocol, and at least 2 centers appear not to have used daily spontaneous breathing trials. In this instance it may be that the local care pattern was not up to the standard of care, which would bias the results in favor of the computer-directed weaning. However, given the knowledge we have about the time that elapses between the publication of new evidence and the translation of that evidence into practice, this study may depict the real standard across the world. Despite best-practice evidence, protocols are not always implemented or followed. The introduction of SmartCare/PS eliminates practice variations and implements the weaning protocol.

Bouadma and colleagues evaluated SmartCare/PS in 33 patients. They compared the times to recognition of weaning readiness by the algorithm and by the physicians.[28] They found that the algorithm detected weaning readiness earlier than the physicians in 17 patients, the physicians detected weaning readiness earlier than the algorithm in 4 patients, and detection was simultaneous in 11 patients.

They concluded that SmartCare/PS was successful in managing PSV for up to one week and proposed weaning readiness earlier than physicians. Though this study was not designed to evaluate duration of mechanical ventilation, it suggests that the elimination of practice variation may be of benefit and could lead to earlier discontinuation of mechanical ventilation.

Very recently, Rose et al compared SmartCare/PS to nursing-directed weaning in an Australian intensive care unit.[29] Australia does not include respiratory therapists in their medical system, and nursing care in that study was at a 1:1 ratio with mechanically ventilated patients. The study found no differences in time to extubation between the groups, but noted a trend toward earlier detection of weaning readiness by the experienced nursing staff. Rose et al concluded that SmartCare/PS did not reduce the duration of weaning, in stark contrast to the finding by Lellouche and colleagues. Rose et al suggested that SmartCare/PS has no obvious advantage over existing weaning methods, which consist of frequent assessment of weaning readiness and titration of ventilatory support by qualified and experienced nurses in a closed ICU model. These results give credence to the criticisms leveled at the European trial, where the care model was different. However, a larger trial of "current practice" of a large variety of ICU models would be required to determine when SmartCare/PS is likely to be of benefit.

Invention or Innovation

Closed-loop ventilation holds the promise of reducing practice variation and responding to changes in patient condition with a speed and vigilance not usually available from the ICU staff. There is no doubt that SmartCare/PS is innovative if judged against our definition of "making a positive change." As with the introduction of all new systems, we often learn more about our current practice than we do about the new article under review. SmartCare/PS challenges earlier versions of closed-loop control by allowing several minutes or longer in between changes, as compared to the typical breath-to-breath changes. While this is still far more frequent than the clinician can provide, we believe this less aggressive approach may be a key to the success. Despite the long history of SmartCare/PS, the available literature remains quite limited. Further research should help determine the scenarios where computer control of weaning will be beneficial.

Automated Measurement of Functional Residual Capacity During Mechanical Ventilation

History

Monitoring the mechanically ventilated patient has routinely revolved around the dynamic changes in airway pressure, volume, and flow. Blood gases and noninvasive measurements of oxygenation (transcutaneous oxygen, oximetry) and ventilation (P_{ETCO_2}, transcutaneous carbon dioxide) are also commonly used. Lung volume measurements during mechanical ventilation are typically limited to continuous monitoring of V_T and intermittent evaluations of vital capacity. The latter measurement has been considered as a parameter to evaluate cough effectiveness prior to discontinuation of mechanical ventilation.

FRC measurement during mechanical ventilation has been made with helium dilution, nitrogen washout, and tracer gases (sulfur hexafluoride).[30-35] In all of those studies the systems were homemade amalgamations assembled by the research team to evaluate the effects of positive end-expiratory pressure (PEEP) on end-expiratory lung volume.

East and colleagues produced the majority of work in this arena, following a research plan with the possibility of producing a commercially available system.[31,36,37] In their series of investigations they found that sulfur hexafluoride FRC measurements were accurate and aided in closed-loop control of PEEP in traditional ventilation and independent lung ventilation. But sulfur hexafluoride does not have regulatory approval for human use, and this and other issues precluded further advancement of this technique.

Automated FRC Measurements Using Nitrogen Washout

Measurement of FRC in the pulmonary function laboratory can be accomplished via body plethysmography, helium dilution, and nitrogen washout. In each case, a cooperative, spontaneously breathing patient is the subject. Helium dilution can be used in mechanically ventilated patients, but requires the addition of the helium cylinder and a helium analyzer. Additionally, the effects of helium on ventilator performance are quite variable and can lead to ventilator malfunction.[38-40]

Nitrogen washout is the simplest FRC measurement method in the mechanically ventilated patient, as there is no need for additional equipment or gases. However, the nitrogen-washout method requires accurate measurement of inspired and expired oxygen and carbon dioxide—in essence a metabolic monitoring module must be integral to the ventilator. Under normal conditions FRC is the result of the opposing forces of lung and chest wall compliance. In the mechanically ventilated patient FRC is a function of those forces, air-flow obstruction, and the application of PEEP. It may be more appropriate to consider FRC in the ventilated patient as end-expiratory lung volume (EELV). Equally important is the fact that the use of PEEP in the ventilated patient does not attempt to normalize FRC to predicted values. Measurements in the ventilated patient more likely are able to evaluate changes in EELV created

by alterations in therapy. For example, following a successful recruitment maneuver the EELV measurement may increase, compared to pre-procedure values. Similarly, following delivery of a bronchodilator in a patient with intrinsic PEEP (auto-PEEP), EELV may decrease.

Principle of Measurement

The calculation of EELV is based on a step-change in the fraction of inspired oxygen (F_{IO_2}) and the assumption that in the ventilator system, N_2 is the balance gas in the system, where $N_2 = 1 - F_{IO_2}$. The following is a simplified explanation of the measurement of EELV. Inspired and expired concentrations of N_2 are not measured, but determined from the presence of O_2 and CO_2. Inspired N_2 (F_{IN_2}) and end-tidal N_2 (F_{ETN_2}) are calculated as:

$$F_{IN_2} = 1 - F_{IO_2}$$

and

$$F_{ETN_2} = 1 - F_{ETO_2} - F_{ETCO_2}$$

where F_{ETO_2} is the fraction of end-tidal oxygen. Inspired alveolar V_T ($V_T^{alv(I)}$) and expired alveolar V_T ($V_T^{alv(E)}$) are calculated with energy-expenditure measurements for oxygen consumption ($\dot{V}_{O_2}$) and carbon dioxide production ($\dot{V}_{CO_2}$):

$$\dot{V}_{O_2} = (\dot{V}_{CO_2} / RQ)$$

where RQ is respiratory quotient
and

$$V_T^{alv(E)} = \dot{V}_{CO_2}/(F_{ETCO_2} \times f)$$

and

$$V_T^{alv(I)} = V_T^{alv(E)} + (\dot{V}_{O_2} - \dot{V}_{CO_2}) / f$$

where f is respiratory frequency per minute.

Remember, that the respiratory exchange ratio results in expired volumes that are typically smaller than inspired volumes, because the volume of oxygen consumed exceeds the volume of carbon dioxide produced. The single-breath nitrogen volumes associated with expiration and inspiration are then calculated as:

$$V_E^{N_2} = F_{ETN_2} \times V_T^{alv(E)}$$

and

$$V_I^{N_2} = F_{IN_2} \times V_T^{alv(I)}$$

The resulting values are then used to calculate the change in nitrogen concentration during a single breath:

$$\Delta V^{N_2} = V_E^{N_2} - V_I^{N_2}$$

At this point in the measurement, a baseline determination is made, which includes the values of $\dot{V}_{O_2}$, $\dot{V}_{CO_2}$, and F_{ETN_2} baseline. During this time the device assumes that the $\dot{V}_{O_2}$ and $\dot{V}_{CO_2}$ remain constant throughout the measurement, but that assumption may not always hold true: changes in $\dot{V}_{O_2}$ and $\dot{V}_{CO_2}$ associated with activity, agitation, or changes in patient condition can cause errors. A step-change in F_{IO_2} is then accomplished and the EELV is calculated as:

$$EELV = \Delta V^{N_2}/\Delta F_{ETN_2}$$

where F_{ETN_2} is the last recorded value following the step-change in F_{IO_2}. The breath-to-breath changes are calculated over approximately 20 breaths.

$$EELV = \frac{\sum_{breaths} \Delta V_{N_2}}{(baseline\ F_{ETN_2} - last\ F_{ETN_2})}$$

FRC INview

The EELV measurements performed by the Engström Carestation (GE Healthcare, Waukesha, Wisconsin), known as "FRC INview" can be made on demand or pre-programmed to make a series of measurements. Each EELV determination is made with 2 measurements in a series of 20 breaths. The step-change in F_{IO_2} required to determine EELV is typically 10%. Accuracy is best at F_{IO_2} of 0.4–0.65. Prior to the step-change the patient should be stable and the F_{IO_2} should be constant for at least 5 minutes. Because patient stability is required for accuracy, and V_{O_2} and V_{CO_2} are assumed to stay constant, the device will terminate the measurement if there is any change in the ventilation mode, ventilator settings, or performance of a procedure (eg, airway suctioning). Following the EELV measurement, the F_{IO_2} returns to the original setting. EELV is displayed numerically and graphically over the 20-breath measurement period. Table 2 lists the conditions and/or alterations in patient status that can affect the accuracy of the EELV measurement.

Accuracy and Application

The current EELV measurement technique (F_{IO_2} step-change then nitrogen washout) was pioneered by Stenqvist

Table 2. Factors That Can Affect the Accuracy of the End-Expiratory-Lung-Volume Measurement

Factor	Cause
Respiratory rate > 35 breaths/min	Agitation, anxiety
Variable respiratory rate	Head injury, metabolic syndromes
Large variations in V_T	Head injury, disorders of ventilation control
Febrile	Sepsis
Change in ventilator mode or settings	Settings result in alterations of EELV during the measurement
Position change	Position alterations of EELV during the measurement
Procedure	Suctioning, nebulization. Changes in EELV and gas concentrations alter measurement.
Leak	Circuit leak, cuff leak, chest-tube leak alter inspiratory and expiratory volume measurements, and gases are lost to the measurement system.
Humidity	Excessive humidity can decrease the accuracy of the flow sensor. A heat-and-moisture exchanger is preferred during measurement.
High F_{IO_2}	High F_{IO_2} alters accuracy of $\dot{V}_{O_2}$, and $\dot{V}_{CO_2}$ measurements and makes the step change in F_{IO_2} more difficult.
Elevated bias flow	Decreases accuracy of volume measurements

V_T = tidal volume
EELV = end-expiratory lung volume
F_{IO_2} = fraction of inspired oxygen
$\dot{V}_{O_2}$ = oxygen consumption
$\dot{V}_{CO_2}$ = carbon dioxide production

and colleagues,[41,42] and several early studies with lung models, animals, and patients found the method accurate and reproducible.[42-44] However, the evidence regarding the value of EELV in critically ill patients remains sparse. Bikker et al recently compared EELV in mechanically ventilated patients to reference equations for FRC.[45] In 3 groups (one group with normal lungs, one group with primary lung disorders, and one group with secondary lung disorders) they used 3 PEEP settings while measuring EELV and pulmonary compliance. They found that in mechanically ventilated and sedated patients, EELV was markedly lower than the predicted sitting FRC values. Importantly, they identified the critical issue in the use of EELV measurements to guide PEEP: PEEP-induced changes in EELV may not only detect recruitment or derecruitment: the change can result from inflation or deflation of already ventilated lung units. More simply, an EELV increase may represent overdistention of lung units that were already "open," as compared to recruitment of previously "closed" lung units. The EELV changes can be similar, but the effect on the patient is markedly different. Bikker et al

further suggested that the use of EELV measurements in conjunction with lung-compliance measurements is more likely to determine which EELV changes represent recruitment and which represent overdistention. We would add that, since metabolic measurements are integral to the nitrogen washout system, the addition of volumetric capnography and dead-space calculations along with EELV and compliance may be the optimal system for adjusting PEEP. This concept is supported by preliminary work by Rylander et al and others.[46-48]

Invention or Innovation

While the technique of monitoring EELV using the nitrogen-washout method is clearly inventive, clinical data regarding the usefulness of the technique are lacking. Admittedly, this is predominantly a function of its very recent introduction to the market. We predict that, when coupled with compliance measurement and capnography, this technique will provide data not currently available and will obtain the definition of innovative.

Neurally Adjusted Ventilatory Assist

History and Principle of Operation

Neurally adjusted ventilatory assist (NAVA) was introduced by Sinderby in 1999, and introduced on the Servo-i ventilator (Maquet, Bridgewater, New Jersey) in 2007.[49] NAVA is a mode of partial ventilatory support that uses the electrical activity of the diaphragm to control patient-ventilator interaction. The electrical activity of the diaphragm represents the final neural output of the respiratory centers to the diaphragm and is therefore able to both trigger and cycle a breath. Breaths remain pressure-controlled. With NAVA the pressure delivered during inspiration is proportional to the electrical activity of the diaphragm. The pressure level can be adjusted based on patient effort and the proportionality setting. Depending on the patient's response to the delivered pressure, an increase in the NAVA level may increase the delivered pressure (if electrical activity of the diaphragm is unchanged), or it could suppress electrical activity of the diaphragm and instead deliver a constant level of assist, or some level in between.

As NAVA is relatively new, a comparison to PSV may be helpful to highlight the differences. PSV delivers a fixed pressure, independent of changes in respiratory drive. PSV is patient-triggered and uses either flow or pressure measurements. The absolute pressure is set by the clinician and is unaffected by alterations in patient effort. PSV is normally cycled when flow reaches a predetermined percentage of the initial peak flow. PSV can also be cycled by time (in the presence of leaks) or pressure (in the presence

of patient expiratory effort).[50] When the pressure-support level is set too high, a large V_T may be delivered, which can cause hyperinflation and missed triggers because of auto-PEEP.[51]

NAVA requires an esophageal catheter that measures the electrical signal to the diaphragm (E_{di}). The E_{di} catheter is similar to a standard nasogastric tube in diameter and length, but has a series of electrodes that measure the E_{di}. The electrodes appear as black circumferential stripes. There is a reference electrode (the most proximal electrode) and 9 measuring electrodes. NAVA is triggered by the E_{di} signal, and the sensitivity can be set, like pressure triggering or flow triggering. The minimum E_{di} signal is recorded and the sensitivity is based on an increase in the signal above that reference value. NAVA cycling occurs when E_{di} drops to approximately 70% of its peak value. During NAVA, pressure triggering and flow triggering remain as redundant systems that activate if the catheter is not placed properly or if airway triggering is sensed first. A proportionality factor (known as the NAVA level) determines the delivered pressure for a given electrical activity of the diaphragm amplitude (ie, cm H_2O per unit of electrical activity of the diaphragm).

Based on the principle of operation then, NAVA may avoid the overinflation associated with PSV. Since NAVA uses the electrical activity of the diaphragm to trigger and cycle, patient-ventilator interaction should be improved. NAVA should be unaffected by the most common factors that confound traditional triggering: auto-PEEP and leaks (in noninvasive ventilation or uncuffed endotracheal tubes).

Application

The E_{di} catheter is placed with guidance from the E_{di} catheter positioning screen, which displays a series of 4 electrocardiograph leads and the E_{di} signal. Proper positioning of the E_{di} catheter is assumed when the E_{di} signal and the corresponding second and third electrocardiograph signals display in blue. Blue signals from the first or fourth lead suggest the E_{di} catheter needs to be either pulled back or advanced forward. This system simplifies placement, but even with perfect placement the signal can be absent or very low in the presence of excessive sedation, muscle relaxants, hyperventilation, high PEEP, or a neural disorder.

If used simultaneously for E_{di} measurement and nasogastric drainage, the conflicting goals can make the catheter fail to optimally perform either one of those functions. In our experience the catheter is not easily placed, nor is the signal always found. We have also seen that, though initial placement can achieve excellent signal quality, patient movement and routine ICU care can result in signal loss. These are preliminary observations and demonstrate that, despite proper engineering design and controls, clinical factors can confound implementation.

Setting the NAVA level is accomplished by selecting the Neural Access and NAVA preview screen. This screen allows the clinician to set the PEEP, F_{IO_2}, and E_{di} trigger level (in microvolts). The NAVA pressure-support and backup ventilation settings are set by the clinician and ensure safe ventilation in the absence of patient effort or loss of communication with the E_{di} catheter. The NAVA level is set by observing the airway pressure, delivered V_T, and E_{di}. The initial NAVA level is generally that which reduces E_{di} and corresponds to patient comfort.

Current Literature

Since the initial paper in 1999,[49] a series of original investigations have been published, all of which were authored or co-authored by the NAVA inventors. Allo et al evaluated the effects of NAVA on rabbits with ALI. They found that ALI caused a vagally mediated atypical diaphragm-activation pattern in spontaneously breathing animals. The addition of PEEP restored phasic activity, and NAVA efficiently maintained respiratory-muscle unloading while delivering safe V_T.[52] Those experiments lasted 4 hours.

Beck and colleagues found that NAVA reduced the number of missed triggers, compared to PSV, and reduced the transdiaphragmatic pressure-time product and E_{di} in a rabbit model.[53] At an excessive pressure-support level, only 66% of efforts triggered the ventilator, compared to 100% with NAVA. The large V_T resulted in auto-PEEP, missed triggers, and failure to adequately unload the respiratory muscles. In volunteers, Sinderby et al found that NAVA unloaded the respiratory muscles and that E_{di} was detectable and triggered the ventilator even at low amplitudes.[54]

In another animal study, Beck and colleagues found that NAVA can be successfully applied with noninvasive ventilation and that synchrony was maintained in the presence of large leaks.[55] Moerer et al found that NAVA successfully triggered and cycled the ventilator during the use of a helmet (noninvasive ventilation interface) in normal subjects.[56]

More recently, studies with adult and neonatal patients have been published.[57,58] Brander and colleagues studied 15 patients with ALI and systematically increased the NAVA level over a 3-hour period, which reduced the respiratory drive, unloaded the respiratory muscles, and allowed the clinician to identify an assist level that resulted in sustained unloading, appropriate V_T (5.4–7.2 mL/kg of predicted body weight), and normal hemodynamics.[57] In low-birth-weight infants, Beck et al found that NAVA improved patient-ventilator synchrony, even in the presence of leaks around the endotracheal tube (ET).[58]
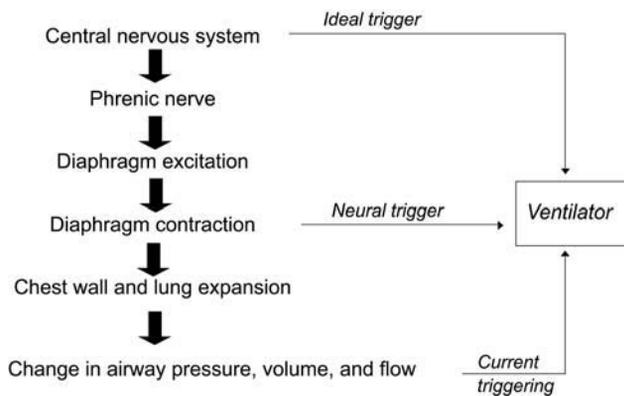
Fig. 1. Comparison of trigger signals in traditional mechanical ventilation and neurally adjusted ventilatory assist ventilation.

Invention or Innovation

NAVA is clearly an innovative design. Detecting respiratory-muscle effort earlier than changes occur at the airway is a positive change (Fig. 1). NAVA remains in a very early stage of development, and there have been no long-term studies to determine if NAVA has outcome benefits. However, we know that patients with greater patient-ventilator asynchrony have worse outcomes and longer duration of mechanical ventilation.[59] What we don't know is if that relationship is one of association or causality. If missed triggers and asynchrony lead to prolonged mechanical ventilation, then a system that reduces missed triggers might improve outcome. However, at present it is more likely that the severity of illness results in asynchrony, not the other way around.

Over the last 20 years traditional ventilator triggering has improved markedly, to the point where cardiac changes in intrathoracic pressure can cause triggering.[60] On several occasions cardiac activity has been responsible for triggering in patients with brain death and delayed organ donation.[61] This means that in many patients triggering is not the issue. Thille et al found that simply reducing the pressure-support level significantly reduced missed triggers.[51]

NAVA would seem to be ideal when factors confound traditional triggering, the most likely of these being in the presence of hyperinflation and/or leaks. This suggests that in a neonate with an uncuffed ET, particularly at a higher respiratory frequency, NAVA should be advantageous. Similarly, NAVA should have value in the patient with substantial auto-PEEP. Finally, in a patient with leaks and hyperinflation (eg, a patient with chronic obstructive pulmonary disease undergoing noninvasive ventilation), NAVA would seem ideal, but the NAVA requirement of a nasogastric tube may be counter to the noninvasive goal. Gaining the patient's trust and assuring patient comfort and cooperation are key to successful initiation of nonin-

vasive ventilation. How placement of a nasogastric tube would impact this is unknown. We await further research on NAVA and look forward to definitive answers about the proposed advantages.

Information Presentation

History and Principles

Every elementary school student has heard the adage that a picture is worth a thousand words, which is based on the Chinese proverb, "A picture's meaning can express ten thousand words."[62] This concept was embraced by the group led by Westenskow at the University of Utah[63] and commercially implemented as the "Ventilation Cockpit" on the G5 ventilator (Hamilton Medical, Reno, Nevada). The goal of information presentation is to build on the success of graphical displays of pressure, volume, and flow into pictorial representations of common clinical conditions.[64,65] Theoretically these displays might improve safety by allowing faster detection of untoward events. Given the current concern regarding errors in medicine, a system that alerts the clinician to changes in patient condition seems ideal.

Current Literature

Drews and Westenskow reviewed data regarding the use of graphical displays to depict changes in cardiovascular function, compared to numeric displays, and found that graphical displays can improve situational awareness, enhance clinician performance, and improve patient safety.[63] Wachter et al studied 19 clinicians' ability to manage 5 scenarios: obstructed ET; endobronchial intubation; auto-PEEP; hypoventilation; and a normal condition.[64] The graphical pulmonary display changed shape and color according to measures of pulmonary function. The display was generated from data from a respiratory monitor and a patient simulator. The monitor measured airway pressure, air flow, respiratory rate, P_{ETCO_2}, and V_T. Airway resistance and total lung compliance were calculated from measurements of airway pressure, volume, and flow. The participants were assigned alternately to one of 2 groups: intervention pulmonary display present, or control display. Both groups had access to standard displays of numerical data. In a simulated operating room, with a simulated patient, the participants assumed the role of an anesthesiologist called because of an unspecified problem midway through a surgery. At completion of each scenario, the volunteers responded to a National Aeronautics and Space Administration (NASA) Task Load Index questionnaire to assess perceived work load and performance.
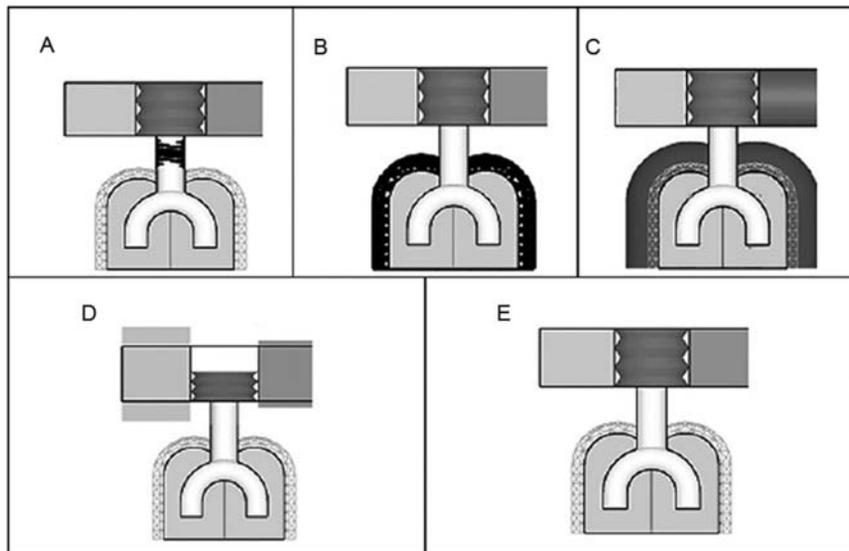
Fig. 2. Examples of the pulmonary display used by Wachter et al.[64] The display anatomically represents the bellows, airway, lungs, and inspired and expired gas. In each part of the figure, the left box represents the fraction of inspired oxygen (F_{IO_2}); the middle box is similar to the bellows of the ventilator and moves along the vertical axis, representing tidal volume; and the right box represents end-tidal carbon dioxide (P_{ETCO_2}). A: The obstructed endotracheal tube event; the upper airway has "black restrictive fingers." B: The endobronchial intubation event has a thickened compliance cage surrounding the lung icon. C: The intrinsic positive end-expiratory pressure event shows an overinflated lung icon, which extends past the normal boundary of the lung icon and the compliance cage. D: The hypoventilation event is shown with a short bellows icon that represents low tidal volume. E: The normal event: all parameters are within normal limits. (From Reference 63, with permission.)

Figure 2 shows the graphical displays used in the simulation. The graphical displays resulted in earlier detection of the airway-associated issues in 2 of the 4 abnormal conditions. During the normal scenario simulation, 3 clinicians using the graphical display, and 5 clinicians using the conventional display gave unnecessary treatments. Participants reported significantly lower subjective work load with the graphical display during the obstructed-ET and the auto-PEEP scenarios.

Implementation

The Hamilton G5 uses 2 main displays to reflect changes in pulmonary compliance and airway resistance. Figure 3 shows the normal display. This display is known as the "dynamic lung." In the normal display the lungs have smooth, rounded edges and the airways appear pink and of uniform diameter. As lung compliance falls, the lungs take on an angular appearance. Figure 4 demonstrates a fall in compliance to 50 mL/cm H_2O. The lungs have 6 sides and no longer fill the full volume. This display denotes the reduction in compliance and loss of lung volume. Figure 5 demonstrates a further reduction in compliance to 30 mL/cm H_2O and the lungs are reduced to 5 sides. At the lowest lung compliance ($\leq$ 20 mL/cm H_2O) the lungs have only 4 sides and appear the most angular (Fig. 6). These are in contrast to the display of hyperinflation or overdistention
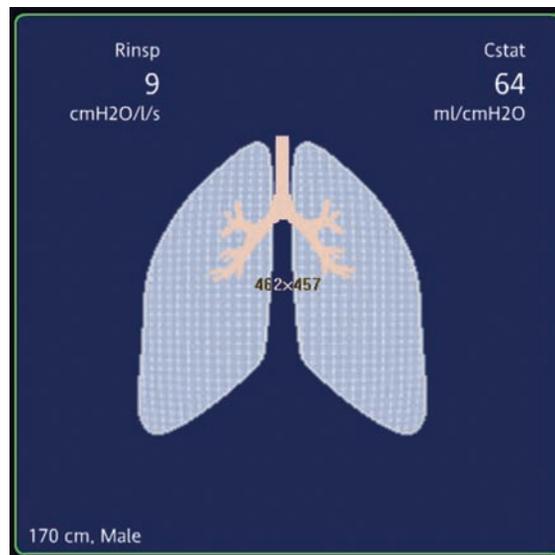

Fig. 3. Dynamic lung display on the Hamilton G5 ventilator, demonstrating normal compliance. The lungs are smooth, without any angles.

(Fig. 7). Figure 7 represents a pulmonary compliance of > 75 mL/cm H_2O. The lungs are overly rounded and exceed the normal boundaries of the lung display.

Figures 8–10 show changes in airway resistance. Normal airway resistance is displayed as pink airways with uniform walls (see Fig. 8). As resistance increases (see
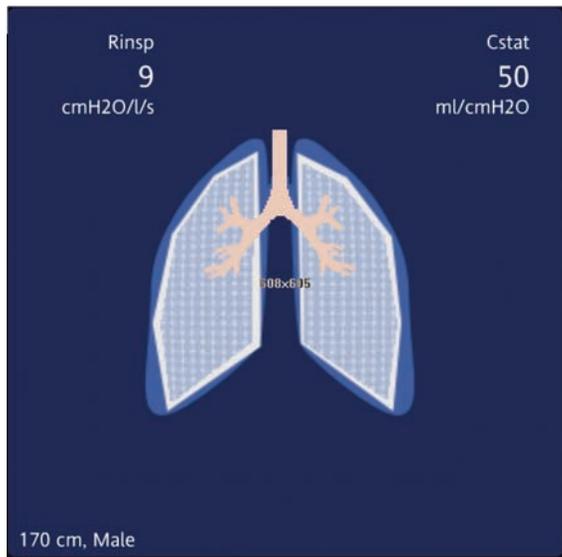
Fig. 4. Dynamic lung display on the Hamilton G5 ventilator, demonstrating a reduction in compliance to 50 mL/cm H$_2$O. Note the angular appearance and 6-sided lung.
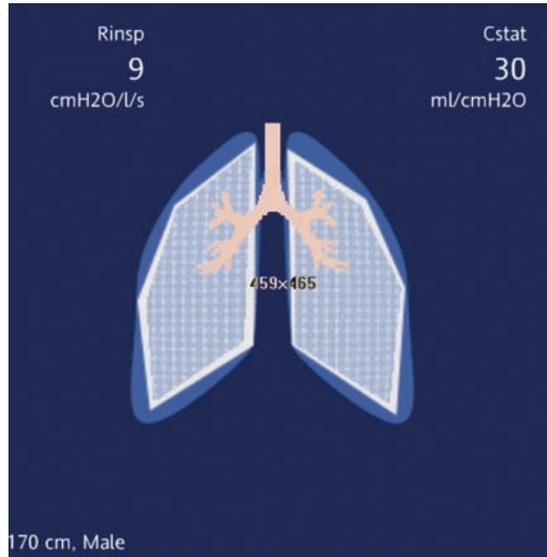

Fig. 5. Dynamic lung display on the Hamilton G5 ventilator, demonstrating a reduction in compliance to 30 mL/cm H$_2$O. Note the angular appearance and 5-sided lung.
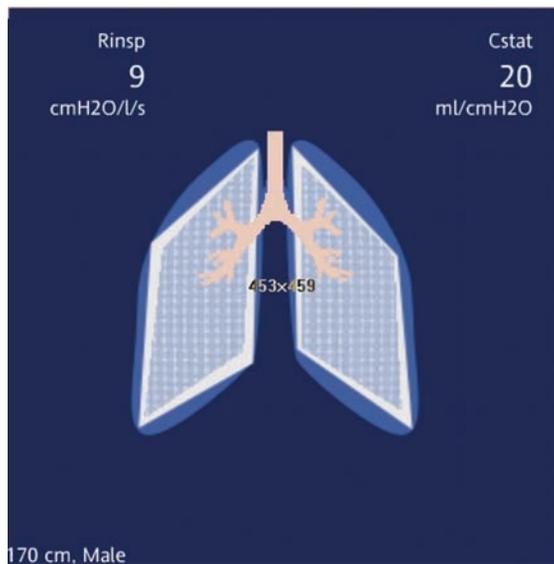

Fig. 6. Dynamic lung display on the Hamilton G5 ventilator, demonstrating a reduction in compliance to 20 mL/cm H$_2$O. Note the angular appearance and 4-sided lung.
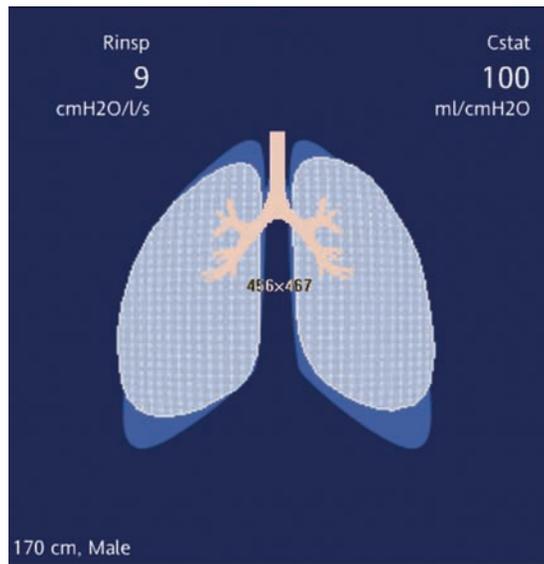

Fig. 7. Dynamic lung display on the Hamilton G5 ventilator, demonstrating an increase in compliance to 100 mL/cm H$_2$O. Note the ballooning appearance and rounded lungs stretching beyond the normal area.

Fig. 9), the color becomes deeper and the colored portion does not fill the gray space that outlines the airways. When airway resistance increases, the color becomes deeper and the gray space is noticeably larger (see Fig. 10). These changes alert the clinician of changes in pulmonary compliance and airway resistance without numerical values or the need for clinicians to remember the normal ranges for these values.

Figure 11 is a screen shot from the ventilator. Note the depiction of increased airway resistance. This is confirmed by the numerical display, which shows an airway resistance of 21 cm H$_2$O/L/s. Alerted to the change, the clinician found that the heat-and-moisture exchanger was occluded with secretions. Removal of the heat-and-moisture exchanger returned the display and numerical values to normal (Fig. 12)
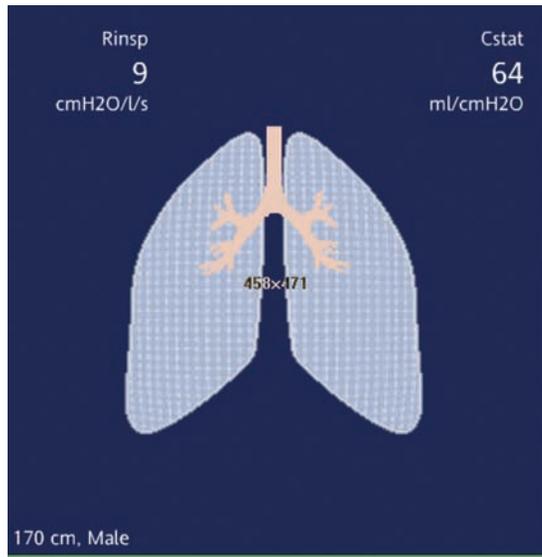
Fig. 8. Dynamic lung display on the Hamilton G5 ventilator, demonstrating normal airway resistance. Note the pink, uniform airways.
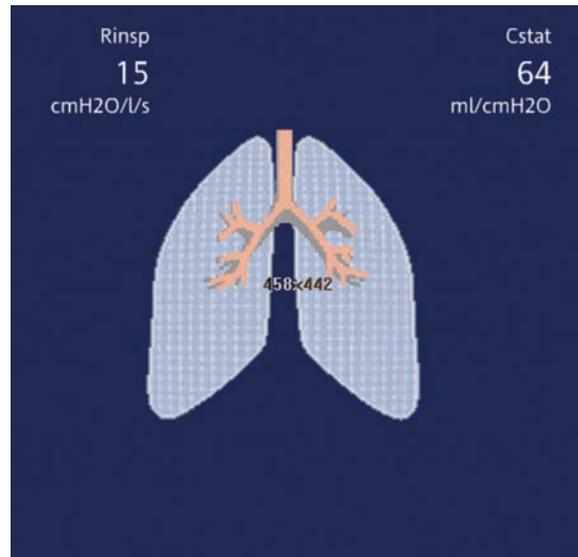

Fig. 9. Dynamic lung display on the Hamilton G5 ventilator, demonstrating an increase in airway resistance to 15 cm H_2O/L/s. Note the deepening color of the airway and the narrowing within the normal outlined gray space.
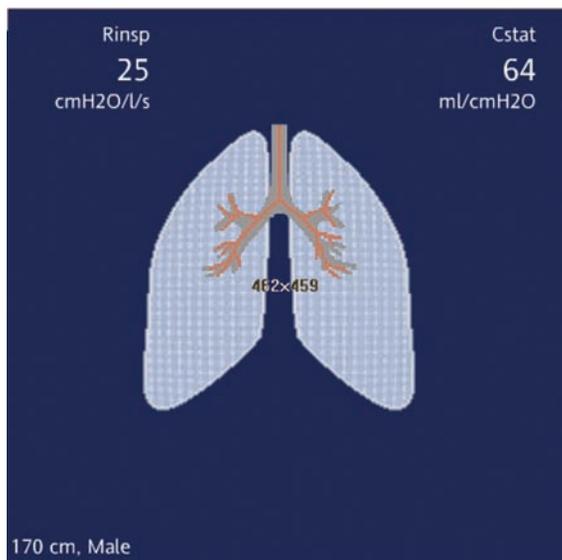

Fig. 10. Dynamic lung display on the Hamilton G5 ventilator, demonstrating an increase in airway resistance to 25 cm H_2O/L/s. Note the deepening color of the airway and the narrowing within the normal outlined gray space.

Clearly, increases in airway resistance and decreases in compliance are also associated with increases in peak inspiratory pressure and alterations in the shape of the pressure and flow waveforms. Importantly, a change in the graphical display that denotes reduced pulmonary compliance does not differentiate between pneumothorax and a distended abdomen. Both chest-wall and lung compliance can affect the display. Similarly, a change in airway resistance cannot distinguish between a kinked ET and bronchospasm. Despite these criticisms, given the overload of information and alarms in the ICU, simplified graphical displays may play a role in alerting clinicians of changes in patient condition earlier than traditional measures.

Invention or Innovation

Though a simple invention, compared to the other features we have discussed, we believe that these graphical displays are innovative. The presentation of information in a simple picture can enhance patient safety and reduce time to recognition of adverse events. One advantage is that such graphical display is applicable to every ventilated patient, whereas some other features are used only for weaning, certain disease states, or improving synchrony.

Summary

In our nearly 30 years of studying mechanical ventilation, many new techniques have been introduced. In most cases these were brought to market with little research or scientific background. It is refreshing to see the newest techniques being developed on stronger scientific footing. Each of the innovations we describe here is in fact early enough in development that the jury

Fig. 11. Screen shot demonstrating the plethora of information available on a typical ventilator. Note the increased airway resistance both graphically and numerically. (Courtesy of Paul Gabarini MSc RRT.)
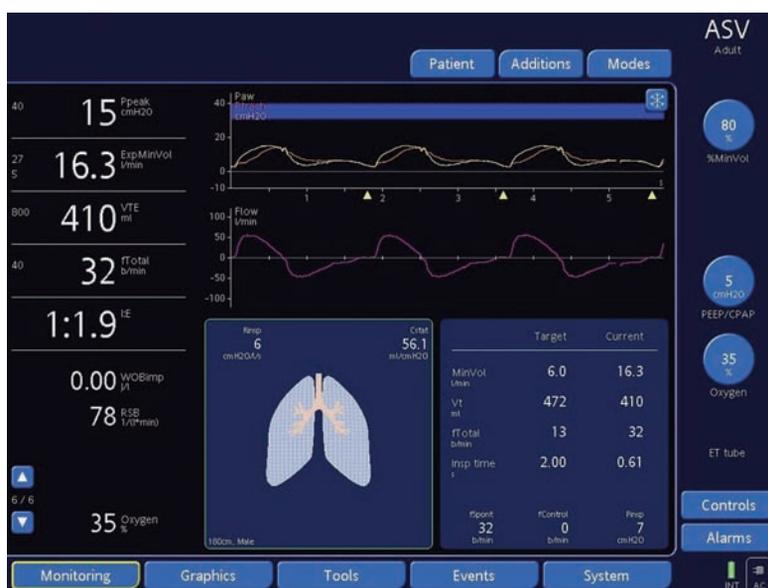

Fig. 12. Same simulated patient as in Figure 11, following removal of a plugged heat-and-moisture exchanger. (Courtesy of Paul Gabarini MSc RRT.)

is still out on the role and efficacy of each. Given our past skepticism about new ventilator techniques, this current group of offerings gives us hope that advantages will be proven with further study.[66]

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