The Effect of Inspiratory Effort on Circuit Compensation for Volume-Targeted Modes

Ping-Hui Liu and Robert L Chatburn

BACKGROUND: Critical-care ventilators provide patient circuit compensation (CC) to counteract the loss of volume due to patient circuit compliance. No studies show the effect of inspiratory efforts (indicating maximal value of the muscle pressure waveforms [P_{max}]) on CC function. The goal of this study was to determine how P_{max} affects volume delivery with or without CC for both volume control continuous mandatory ventilation with set-point targeting scheme (VC-CMVs) and pressure control continuous mandatory ventilation with adaptive targeting scheme (PC-CMVa) modes on the Servo-u ventilator. METHODS: A breathing simulator was programmed to represent an adult with moderate ARDS with different P_{max}. It was connected to a ventilator set to VC-CMVs or PC-CMVa. The change in tidal volume (ΔV_T) was defined as the difference between V_T with CC on versus off. V_T error was defined as the difference between the simulator displayed V_T and the set V_T with CC on versus off. RESULTS: For both VC-CMVs and PC-CMVa modes, ΔV_T decreased as P_{max} increased. The V_T error decreased as P_{max} increased for VC-CMVs. In contrast, V_T error increased on PC-CMVa mode as P_{max} increased and peaked 39.0% for P_{max} = 15 cm H_2O. For both modes, the difference in V_T errors for CC on versus CC off decreased as P_{max} increased. CONCLUSIONS: CC corrected the delivered V_T for volume lost due to compression in the patient circuit as expected. This compensation volume decreases as airway pressure drops due to patient P_{max}. *Key words: medical simulation; mechanical ventilation; patient circuit compensation; error; ARDS*. [Respir Care 2022;67(7):857–862. © 2022 Daedalus Enterprises]

Introduction

One of the key elements of lung-protective ventilation strategy for patients with ARDS is to control tidal volume (V_T) within 4–8 mL/kg of predicted body weight (PBW).[1,2] Previous studies showed that the compressible volume of the ventilator circuit can have important effects on the display of delivered V_T.[3,4] Therefore, the automatic compensation for compressible volume by the ventilator is a critical concern.

To date, many critical-care ventilators provide patient circuit compensation (CC) to counteract the loss of volume in ventilator circuits based on circuit compliance and airway pressure. In general, the volume required to compensate for circuit compliance is calculated by the ventilator as the change in airway pressure (during inspiration) times the circuit compliance (determined during pre-use check). But the specific algorithms used in such compensation are never described in ventilator manuals, making observed behavior difficult to interpret. Inspiratory efforts (indicating maximal value of the muscle pressure waveforms [P_{max}]) decrease inspiratory pressure for volume-targeted modes, that is, volume control continuous mandatory ventilation with set-point targeting scheme (VC-CMVs) and pressure control continuous mandatory ventilation with adaptive targeting scheme (PC-CMVa).[5] Therefore, P_{max} may affect the way CC works.

To the best of our knowledge, previous studies did not investigate the effects of P_{max} on the CC function. The goal

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of this study was to assess how P_{max} affect volume delivery with or without CC for both VC-CMVs and PC-CMVa modes on the Servo-u ventilator.

Methods

We evaluated the performance of CC on delivered V_T using different modes of ventilation using a critical-care ventilator and a breathing simulator.

Breathing Simulator

This study was conducted using the Active Servo Lung 5000 (ASL 5000) (sw3.6; IngMar Medical, Pittsburgh, Pennsylvania) programmed to simulate an adult patient of 64 kg of PBW with moderate ARDS and different P_{max}. The ASL 5000 simulator was connected to a computer to analyze and record respiratory parameters.

Simulations using the ASL 5000 have 2 components: (1) a lung model comprising resistance (R) and compliance (C) and (2) an effort model that is a representation of the muscle pressure (P_{mus}) as a function of time. The lung model parameters are fairly easy to obtain from studies that record respiratory system mechanics. The effort model is more problematic because very few studies have measured P_{mus}. Furthermore, P_{mus} parameters can be derived in a variety of ways, such as a digitized signal (eg, esophageal pressure or electrical activity of the diaphragm signal from a neurally-adjusted ventilatory assist catheter). Usually, the ASL 5000 is programmed to represent P_{mus} as simply a modified sinusoidal function with parameters of frequency, P_{max} (the maximal value of P_{mus} waveform), Increase percentage (time to reach P_{max}, expressed as a percent of the total cycle time), Hold percentage (period of no-flow during the inspiratory phase, expressed as a percent of the total cycle time), and Release percentage (time to reach $P_{mus} = 0$, expressed as a percent of total cycle time). A passive patient is modeled as $P_{max} = 0$ (making all other settings for P_{mus} inactive).

For this study, the lung model was programmed as a single constant airway resistance (including endotracheal tube and normal airway resistance) and single linear respiratory-system compliance using evidence-based values for an adult patient with moderate ARDS.[6] Kallet et al[7] suggested a model for P_{mus} that we believe is unrealistic because it included a Hold percentage > 0, which we believe is not present in normal breathing. Therefore, the effort models for our study were set in terms of the parameters of the simulated P_{mus} waveform based on data from a previous study.[8] Briefly, actual P_{mus} waveforms from that study were analyzed to determine representative values for Increase percentage and Release percentage (Fig. 1). A "ruler" was constructed from a group of lines in an Excel spreadsheet, and the timing of Increase percentage and Release percentage was measured by hand. P_{max} values were arbitrarily set based on our estimate about how much effort would be needed to generate a normal V_T given the lung model parameters. The resultant simulation parameters are shown in Table 1.

> **QUICK LOOK**
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> **Current knowledge**
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> The circuit compensation (CC) function in critical-care ventilators is designed to compensate for the loss of volume due to circuit compliance. There is no study investigating the effects of inspiratory effort on CC function.
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> **What this paper contributes to our knowledge**
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> Increased simulated patient effort resulted in a change in the calculated circuit compressible volume. During VC-CMV the difference in CC were small. At excessive effort (−15 cm H_2O) the error in delivered tidal volume during adaptive pressure modes peaked at 40%. Measurement of CC during patient effort alters accuracy if delivered V_T in some modes.

Ventilator

The Servo-u ventilator (Getinge, Gothenburg, Sweden) was used for all experiments. The ventilator precheck was performed with a standard-length heated wire circuit. The experiment was performed with a filled heated humidifier that was turned off (ie, the experiment was conducted at room temperature). In addition, because the ventilator corrects delivered volume for body temperature and pressure saturated (BTPS) with water conditions, the simulator was set to correct measured values to BTPS.

Two modes were used: volume control, classified as VC-CMVs; and pressure regulated volume control, classified as PC-CMVa.[5] Ventilator settings for each mode are shown in Table 2. For all experiments, F_{IO_2} was set at 0.21 and PEEP at 12 cm H_2O.

Outcome Variables

V_T measured by the ASL 5000 was designated as patient inspiratory V_T, corrected for BTPS. The effect of P_{max} on CC was defined in 2 ways. First, the change in V_T due to CC was calculated as the difference between V_T with CC on versus off (ΔV_T expressed as a percent) for different levels of P_{max} as shown in Equation 1:

$$\Delta V_T(\%) = (V_T \text{with CC on} - V_T \text{with CC off})/V_T \text{with CC off} \times 100\%$$

Second, we calculated the effect on V_T delivery error, where the target V_T was that set on the ventilator ($V_{T\ set}$)

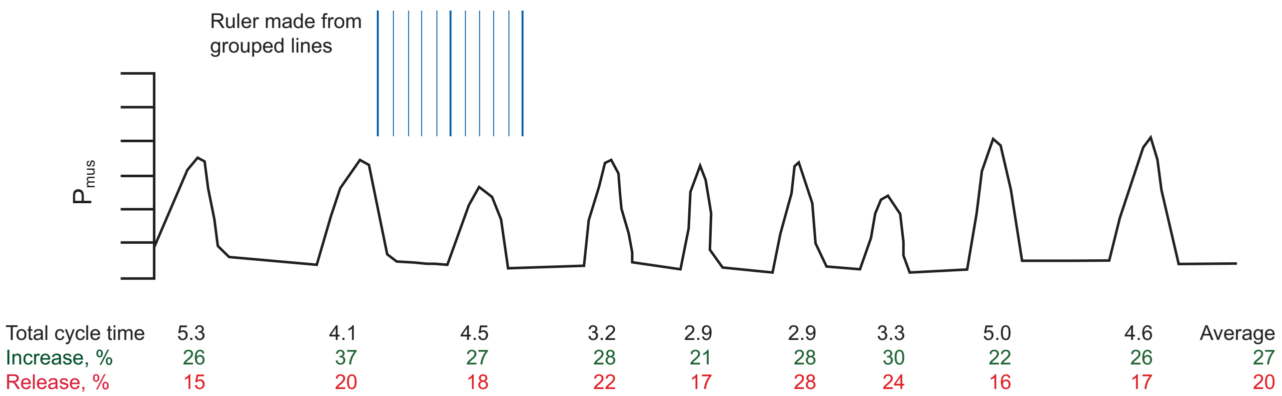

Fig. 1. Method to estimate effort model parameters for simulation with the ASL 5000 breathing simulator. Muscle pressure (P_{mus}) waveforms are modified from Kondili et al.[8]

Table 1. Simulation Parameters

Lung Model	
R_{insp} (cm H_2O/L/s)	10
R_{exp} (cm H_2O/L/s)	10
C (mL/cm H_2O)	40
Effort Model	
Frequency (breaths/min)	20
P_{max}	0, 5, 10, 15
Increase	27%
Hold	0%
Release	20%
Pause	0%

R_{insp} = inspiratory resistance
R_{exp} = expiratory resistance
C = compliance
P_{max} = maximum value of the muscle pressure waveform

Table 2. Ventilator Settings

	Mode Settings		
Mode Classification	V_T mL	f breaths/min	T_I s
VC-CMVs	420	20	0.9
PC-CMVa	420	20	0.9

V_T = tidal volume
f = breathing frequency
T_I = inspiratory time
VC-CMVs = volume control continuous mandatory ventilation with set-point targeting scheme
PC-CMVa = pressure control continuous mandatory ventilation with adaptive targeting scheme

that was compared to that actually delivered as measured by the simulator ($V_{T\ meas}$) expressed as a percentage as shown in Equation 2:

$$V_T\text{Error}(\%) = (V_{T\ meas} - V_{T\ set})/V_{T\ set} \times 100\%.$$

Defined this way, error represents the deviation from the desired V_T such that positive numbers for error indicate that the actual delivered volume is in excess of what is expected, a situation that could increase risk of V_T overdosage. V_T error was calculated for both CC on and CC off.

Procedure

The ASL 5000 simulator was calibrated according the manufacturer's instructions. The pre-use check including internal leakage test, patient circuit test with patient circuit compliance, and flow and pressure transducer calibration on the Servo-u ventilator was performed.

Each combination of mode and P_{max} value was considered an individual experiment (ie, 8 experiments).

Data Analysis

We recorded the mean value of patient inspired V_T as recorded by the ASL 5000 for 10 breaths after stabilization. All calculated values were rounded to the nearest 0.1%.

Results

The effects of simulated P_{max} on pressure, volume, and flow waveforms are shown in Figure 2. The first waveform (P_{max} = 0) shows the effects for passive inflation. As P_{max} increased, inspiratory pressure decreased as indicated by the distorted P_{vent} waveforms.

Figure 3 shows the effects of P_{max} on ΔV_T. During VC-CMVs, the ΔV_T was 8.0%, 7.0%, 4.0%, and 1.6% on P_{max} of 0, 5, 10, and 15 cm H_2O, respectively. During PC-CMVa, the ΔV_T was 6.8%, 4.3%, 2.0%, and 0% on P_{max} of 0, 5, 10, and 15 cm H_2O, respectively. When P_{max} increased, ΔV_T during both VC-CMVs and PC-CMVa decreased (Fig. 1).

Figure 4 shows the effects of P_{max} on V_T error. During VC-CMVs with CC off, the V_T error was −3.4%, −1.0%,

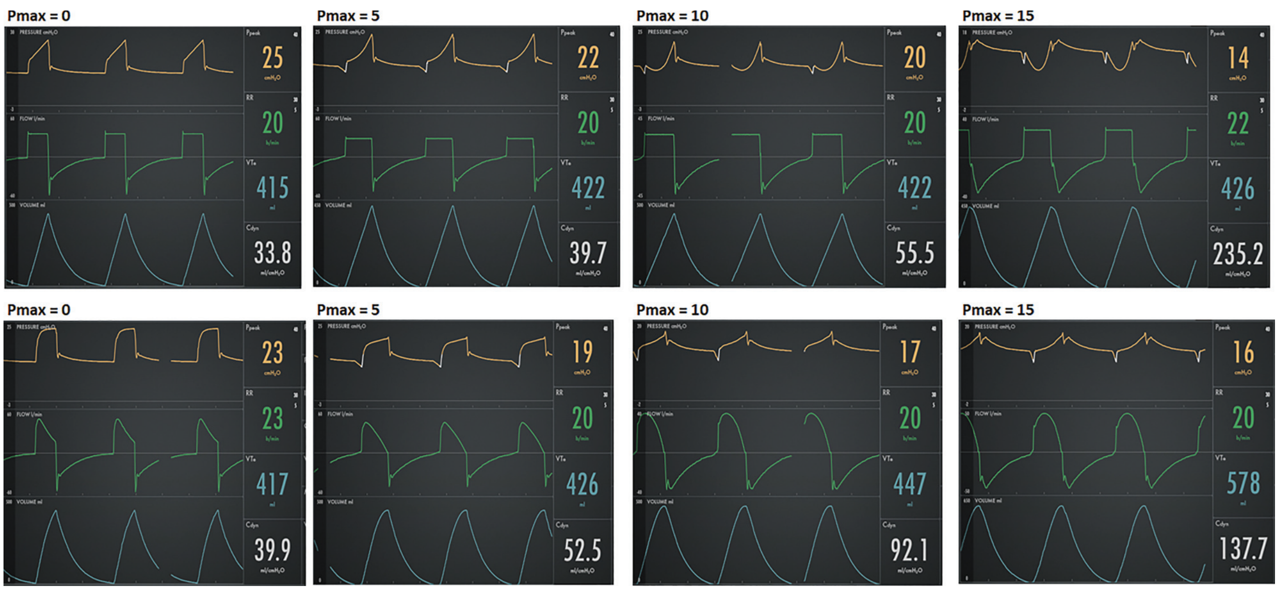

Fig. 2. Representative pressure, volume, and flow waveforms showing effects of different levels of inspiratory effort, simulated as maximal value of the muscle pressure (P_{max}) in cm H_2O. The top row is volume control continuous mandatory ventilation with set-point targeting scheme (VC-CMVs) and the bottom row is pressure control continuous mandatory ventilation with adaptive targeting scheme (PC-CMVa). See Table 2 for ventilator settings.

0.6%, and 2.4% for P_{max} of 0, 5, 10, and 15 cm H_2O, respectively. During VC-CMVs with CC on, the V_T error was 4.3%, 5.9%, 4.8%, and 4.0% for P_{max} of 0, 5, 10, and 15 cm H_2O, respectively. During PC-CMVa with CC off, the V_T error was −1.8%, 0.7%, 8.3%, and 39.0% for P_{max} of 0, 5, 10, and 15 cm H_2O, respectively. During PC-CMVa with CC on, the V_T error was 4.9%, 5.0%, 10.7%, and 39.0% for P_{max} of 0, 5, 10, and 15 cm H_2O, respectively.

The V_T error during VC-CMVs trended in different directions for CC off versus CC on, but the absolute value remained within 5%. In contrast, V_T error on PC-CMVa increased with increased P_{max} and peaked at 39.0% for P_{max} = 15 cm H_2O. For both modes, the difference in V_T error between CC on and CC off (ie, V_T error with CC on minus V_T error with CC off) decreased as P_{max} increased.

Discussion

Change in Tidal Volume (ΔV_T)

This is the first study to investigate the effect of CC on different modes with the presence of P_{max} in a simulator setting. Both VC-CMVs and PC-CMVa modes are volume targeted (ie, allow $V_{T\ set}$). The compliance of the breathing

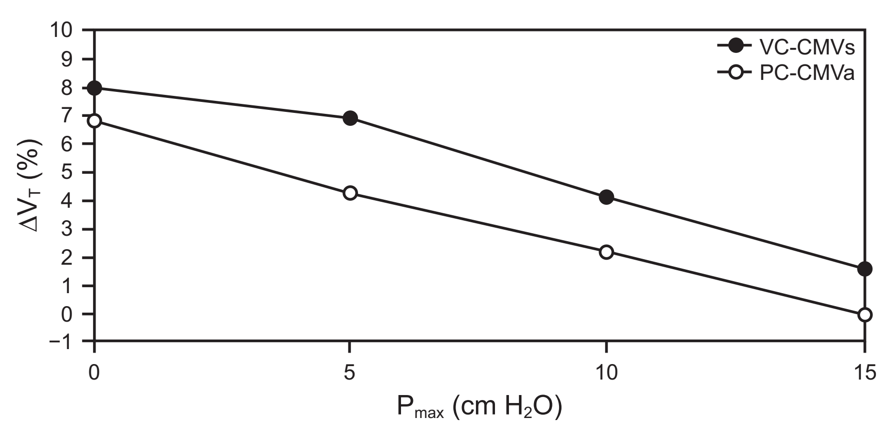

Fig. 3. The effect of muscle pressure on change in tidal volume. ΔV_T = change in tidal volume; VC-CMVs = volume control continuous mandatory ventilation with set-point targeting scheme; PC-CMVa = pressure control continuous mandatory ventilation with adaptive targeting scheme; P_{max} = maximal value of the muscle pressure.

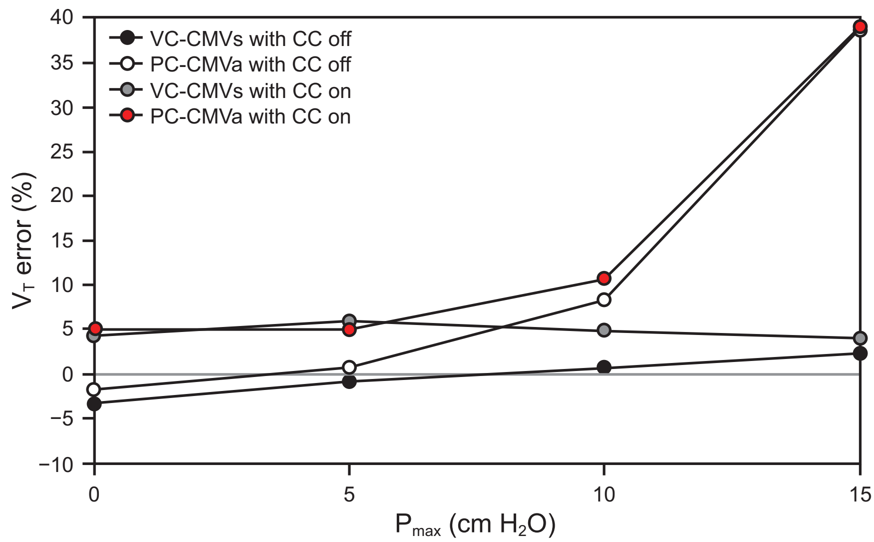

Fig. 4. The effect of maximal value for muscle pressure on tidal volume error. V_T = tidal volume; VC-CMVs = volume control continuous mandatory ventilation with set-point targeting scheme; CC = circuit compensation; PC-CMVa = pressure control continuous mandatory ventilation with adaptive targeting scheme; P_{max} = maximal value of the muscle pressure.

circuit ($C_{circuit}$) was defined by the change in volume (V) divided by the change in pressure (P), as shown in Equation 3:

$$C_{circuit} = \Delta V / \Delta P$$

Therefore, as airway pressure increases during inspiration the compressible volume increases and the greater the difference between target and delivered V_T. Rearranging Equation 3, we get the equation for the volume lost due to compression during inspiration, as shown in Equation 4:

$$\Delta V_{lost} = C_{circuit} \times \Delta P$$

The equation of motion for the respiratory system is shown in Equation 5, where P_{vent} is the pressure provided by the ventilator, P_{mus} is the pressure provided by the patient's respiratory muscle, C is the compliance of the respiratory system, V is the volume, R is the resistance of airway, and $\dot{V}$ is the flow over time.[9]

$$P_{vent} + P_{mus} = V/C + R \times \dot{V}$$

In our experiments, C, R, and V were held constant. P_{max} simulates the peak P_{mus}. Therefore, as P_{max} increases, P_{vent} decreases to maintain the right-hand side of Equation 5 constant. As a result of the decrease in pressure, the lost volume decreases according to Equation 4. This explains the pressure waveform deformations in Figure 2 and the decrease in ΔV_T shown in Figure 3. This is an example of work shifting as described elsewhere.[10,11]

V_T Error

The operator's manual for the Servo-u ventilator specifies inspiratory V_T error of ± (4 mL + 7% of actual volume) for V_T range of 100–4,000 mL.[12] The International Standards Organization has specified an error standard of ± (5 + 10% of the set volume).[13]

For VC-CMVs, V_T error was within this range with CC on or off for all simulated P_{max}. Hence, there are no clinical implications of our findings for this mode.

For PC-CMVa, V_T error was within this range with CC on or off only for low simulated P_{max} (ie, $P_{max} \leq 5$ cm H_2O). The adaptive targeting scheme of this mode attempts to automatically decrease P_{vent} as P_{max} increases to maintain the average V_T at the set target value.[14] However, the ventilator can decrease P_{vent} only to the PEEP level. Beyond that, increases in P_{max} result in increased V_T. The clinical implication is that patients with high P_{max} (eg, patients with COVID-19) V_T overdosage is possible and perhaps common.

The major limitation of this study was that it was based a simulation of a single kind of subject, one with moderate ARDS. The function we used to P_{max} was evidence based, but real P_{max} can take a wide variety of forms that varies randomly. Finally, we did not study situations of patient-ventilator asynchrony.

Conclusions

CC corrected the delivered V_T for volume lost due to compression in the patient circuit as expected. This compensation volume decreases as airway pressure drops due to patient P_{max}. The difference between set and delivered V_T was minimal for VC-CMVs but increased to excessive amounts during PC-CMVa in the presence of large P_{max}.

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