Correction of Error in Respiratory Resistance Measurements Made With the Flow-Interruption Technique During Mechanical Ventilation: Evaluation of the Puritan Bennett 7200 and 840 Ventilators

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BACKGROUND: Calculation of total inspiratory resistance (R_{tot}) for patients on ventilatory support is typically based on measurement of airflow velocity and airway opening pressure during end-inspiratory occlusion by the inspiratory valve in the ventilator. Systematic error is introduced into R_{tot} measurements because the inspiratory valve closes over a period of time (not instantaneously, so gas continues to flow into the circuit while the valve is shutting) and because the circuit tubing is a distensible compartment between the occluding valve and the respiratory system. The R_{tot}-measurement error can be minimized with a rapidly-shutting occlusion valve positioned at the airway opening, or, alternatively, by mathematical correction that accounts for the valve-closure period and circuit tubing characteristics. METHODS: In a bench study we measured R_{tot} with the Puritan Bennett 7200 and 840 ventilators (using the inspiratory valves that are built into those ventilators) and compared those measurements to measurements made with a rapidly-shutting valve at the airway opening. We deemed the rapid-occlusion-valve measurements the best available (benchmark) values. We also studied the closure characteristics of the ventilators' inspiratory occlusion valves and created equations for mathematical correction of R_{tot} values measured with those valves. RESULTS: Compared to the benchmark measurements, the measurements from the Puritan Bennett 7200 averaged 23.2% relative error and 2.6 cm H_2O/L/s absolute error. Measurements from the Puritan Bennett 840 averaged 7.3% relative error and 1.0 cm H_2O/L/s absolute error. Mathematical correction for the circuit tubing and valve-closure time reduced the average relative and absolute error to 3.0% and 0.4 cm H_2O/L/s, respectively, for the Puritan Bennett 7200, and to 4.5% and 0.3 cm H_2O/L/s, respectively, for the Puritan Bennett 840. CONCLUSIONS: The Puritan Bennett 840 measures R_{tot} more accurately than the Puritan Bennett 7200. Our equations to mathematically correct R_{tot} measurements made with the PB7200 and PB840 are useful in settings where very accurate R_{tot} measurements are necessary. *Key words: ventilator, mechanical ventilation, monitoring, airway resistance, measurement error, respiratory physiology*. [Respir Care 2004;49(9):1022–1028.

Introduction

Respiratory resistance in mechanically ventilated subjects can be calculated from measurements of airflow rate ($\dot{V}$) and airway opening pressure (P_{AO}) during an end-inspiratory occlusion maneuver.[1–3] Total inspiratory resistance (R_{tot}) is defined as:

$$R_{tot} = (P_{peak} - P_{plat}) \div \dot{V} \quad (1)$$

in which $\dot{V}$ is the flow rate immediately preceding flow-interruption, and P_{peak} and P_{plat} are the peak and plateau pressures of the P_{AO} waveform. In clinical practice, R_{tot} measurements are performed by interrupting inspiration

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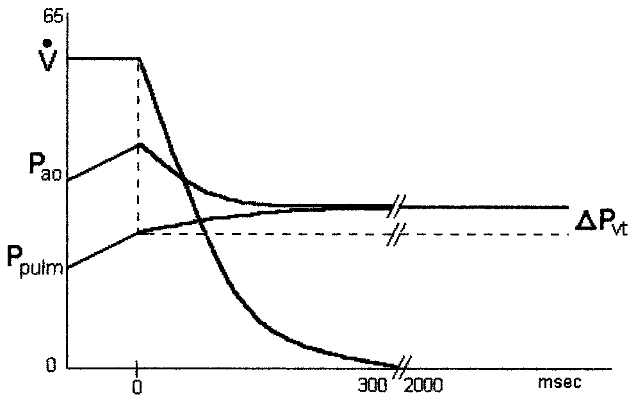

Fig. 1. Theoretical depiction of airflow rate ($\dot{V}$), airway occlusion pressure (P_{AO}), and intrapulmonary pressure (P_{pulm}) during the valve-closure period of end-inspiratory occlusion. Valve closure begins at time 0. Instantaneous closure of a valve at the airway opening would immediately decrease P_{AO} to P_{pulm}, and the plateau pressure (P_{plat}) would be represented by the dashed line. However, the occluding valve closes over a period of time and the occlusion valve is positioned within the ventilator (not at the airway opening), so flow continues while the valve is shutting (ΔV_{valve}) and the additional gas volume is injected into the respiratory system from emptying of the distended tubing circuit (ΔV_{tube}), so P_{pulm} continues to rise during valve closure and P_{plat} is overestimated by amount ΔP_{VT} (due to the combination of ΔV_{valve} and ΔV_{tube}).

under constant flow conditions, by occlusion of a valve positioned within the ventilator (not at the airway opening). In that arrangement R_{tot} is systematically underestimated because the inspiratory-valve closes over time (not instantaneously) and because the circuit tubing is a distensible compartment between the occluding valve and the respiratory system. Specifically, those 2 factors allow the injection of additional volume (ΔV_{valve} and ΔV_{tube}) into the respiratory system, which increases P_{plat} (by amount ΔP_{VT} [due to the combination of ΔV_{valve} and ΔV_{tube}]) and therefore reduces the value of the term $P_{peak} - P_{plat}$ in Equation 1 (Fig. 1).[4–8] Reports of the size of that error differ but range as high as 34%, depending on the valve-closure characteristics, the tubing compliance, the flow rate, and the static compliance and resistance of the respiratory system.[4–6,8]

In research settings where very accurate resistance measurements are necessary, the error can be minimized by using a rapidly-shutting valve placed at the airway opening (not the valve inside the ventilator).[4] Alternatively, a mathematical correction for ΔV_{valve} and ΔV_{tube} can correct for the error introduced by the valve-closure period and the circuit tubing. The offset of P_{plat} can be defined as:

$$\Delta P_{VT} = (\Delta V_{valve} + \Delta V_{tube}) \div C_{RS} \quad (2)$$

in which C_{RS} is static respiratory system compliance. Correction for ΔV_{tube} requires knowledge of the ventilator tubing compliance, which is easily measured.[7,9–11] Correcting for ΔV_{valve} requires mathematical characterization of the occluding valve and is therefore specific to each ventilator model. To date, the only conventional ventilator valve thus described in the literature is that in the Siemens Servo 900C.[5]

Since valve closure characteristics differ among ventilator models, the accuracy of R_{tot} measurements performed with the ventilator valve also differ by model. Clinicians and researchers should be aware of the size of error associated with specific ventilator models. We conducted a bench study in which we measured R_{tot} with 2 ventilators, a Puritan Bennett 7200 (PB7200, Puritan Bennett, Carlsbad, California) and a Puritan Bennett 840 (PB840), and compared those measurements to measurements made with a rapid-occlusion valve positioned at the airway opening. We also developed a method to measure and mathematically determine ΔV_{valve} for the PB7200 and PB840 and we used that method to obtain mathematically corrected R_{tot} values.

Methods

Resistance Measurements

The bench model for measuring R_{tot} consisted of the ventilator connected to a 2-chamber lung model (TTL, Michigan Instruments, Grand Rapids, Michigan) via a standard adult respiratory tubing circuit (MR850, Allegiance, McGraw Park, Illinois). We studied 4 different respiratory resistances. Resistance was created by placing constricted-orifice resistors (Pneuflo, Michigan Instruments, Grand Rapids, Michigan) into the circuit at the entrance to the lung model. Resistors were used in combination to approximate R_{tot} of 5, 10, 15, and 25 cm H_2O/L/s. Those values were chosen to represent resistances that occur in normal and diseased respiratory systems.[12–15] The lung model compliance was set constant at 0.023 L/cm H_2O. The compliance was determined by measuring static respiratory system pressure with a variable-reluctance pressure transducer (MP45 ± 50 cm H_2O transducer, Validyne Engineering Company, Northridge, California) after injecting 100-mL increments of air with a volumetric syringe. A relatively low compliance was used, because (by Equation 2) larger errors in measurement of resistance are expected with low-compliance systems. The compliance value was chosen to fall within the range encountered clinically, as with severe acute respiratory distress syndrome.[12]

With both the PB7200 and PB840 we performed 2-s end-inspiratory interruption maneuvers with each of the 4 tested resistances, while measuring P_{AO} and $\dot{V}$ at the airway opening. Flow was measured with a thermal mass flow meter (model 4000, TSI, St Paul, Minnesota) with a 4-ms response time, and P_{AO} was measured with the pres-

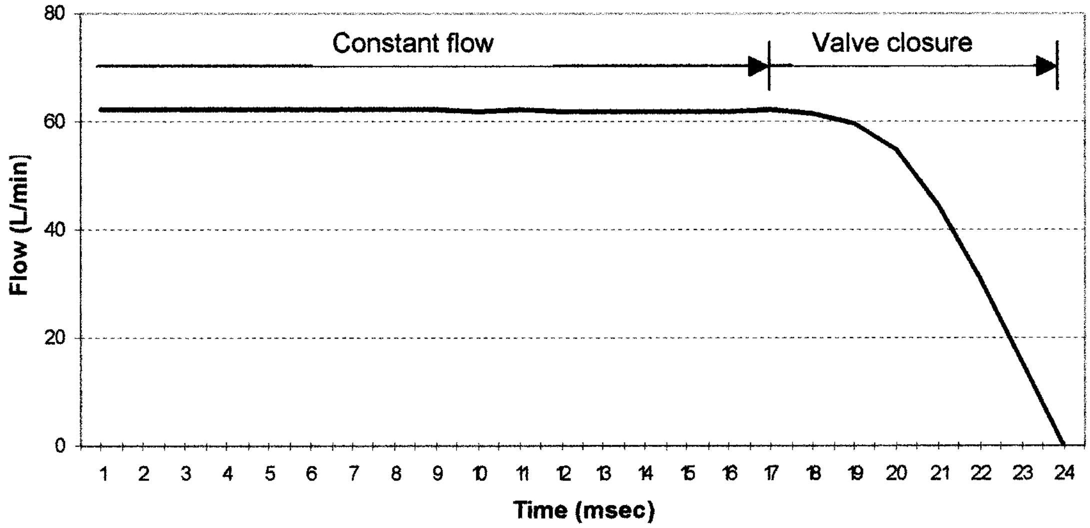

Fig. 2. Flow immediately prior to and during closure of the inspiratory valve during end-inspiratory occlusion, with a setup that included a rapidly closing valve. Data were acquired at 1,000 Hz. The valve-closure time was 7 ms.

sure transducer. Data were collected in a laptop computer (ThinkPad 380, IBM, White Plains, New York) equipped with a data acquisition card (DAQ Card 1200 and BNC208 board, National Instruments, Austin, Texas) and software (BioBench 1.0, National Instruments, Austin, Texas). Data were recorded at 200 Hz.

The pressure transducer was calibrated with a water manometer. We leak-tested the system by observing the P_{AO} waveform during a 7–8-s end-inspiratory pause maneuver at a plateau pressure of approximately 20 cm H_2O. Ventilator settings were maintained constant; we used continuous mandatory ventilation mode with the PB7200 and we used assist control mode (equivalent to continuous mandatory ventilation mode) with the PB840. Inspiratory flow was 60 L/min, with a square-wave setting, tidal volume was 0.5 L, fraction of inspired oxygen was 0.21, respiratory rate was 4 breaths/min, and positive end-expiratory pressure was zero.

To collect benchmark resistance measurements we performed end-inspiratory-occlusion tests with a rapidly shutting, helium-driven, pneumatic, sliding occlusion valve (series 4220, Hans Rudolph, Kansas City, Missouri) that had a closure time of 7 ms (Figure 2), placed at the airway opening, with both the PB840 and the PB7200. During those measurements we used the same ventilator settings, but flow was interrupted by the rapid-occlusion valve rather than by the inspiratory valve within the ventilator. To ensure that inflation time was precisely the same as with the ventilator's valve, the rapid-occlusion valve was controlled by a digital storage oscilloscope (Classic 6000, Gould Instruments, Valley View, Ohio) that tracked the time since onset of flow through the flow meter. Bench testing with the oscilloscope indicated that inflation times were within 15 ms of one another. We deemed the R_{tot} values measured with the rapid-occlusion valve the best available (benchmark) R_{tot} values, because the valve-closure time of 7 ms and positioning the valve at the airway opening minimizes the contributions of ΔV_{valve} and ΔV_{tube}. Differences from the benchmark R_{tot} values were considered to be error on the part of the ventilators' inspiratory valves.

Mathematical Characterization of Valves

To develop an equation to describe ΔV_{valve} as a function of $\dot{V}$, we measured flow during valve closure, with the flow meter placed at the ventilator inspiratory port. A square-wave flow pattern was used and $\dot{V}$ was varied between 40 and 100 L/min, in increments of 5 L/min. The $\dot{V}$ data were collected at 500 Hz, via the laptop computer's serial port and communication software (HyperTerminal 6.3, Hilgraeve, Monroe, Michigan) and then, using statistical software (SigmaPlot 5.0, SPSS, Chicago, Illinois), computationally integrated for the valve-closure period to quantify ΔV_{valve}. We also performed linear regression analysis with that software, to derive an equation relating $\dot{V}$ at the initiation of valve closure to ΔV_{valve}.

Mathematical Correction of R_{tot}

In addition to characterizing ΔV_{valve}, mathematically correcting R_{tot} (by Equation 2) also requires characterizing ΔV_{tube}, which is related to the tubing compliance (C_{tube}) and the pressure gradient driving redistribution:[7]

$$\Delta V_{tube} = C_{tube} \times (P_{peak} - P_{plat}) \qquad (3)$$

Table 1. Total Inspiratory Resistance

	Benchmark R_{tot} (cm H_2O/L/s)	Uncorrected R_{tot} (cm H_2O/L/s)	p for Benchmark vs Uncorrected	Corrected R_{tot} (cm H_2O/L/s)	p for Benchmark vs Corrected
Puritan Bennett 7200	4.1 ± 0.26	2.7 ± 0.11	0.019	4.2 ± 0.11	0.62
	9.5 ± 0.13	7.5 ± 0.21	0.00062	9.4 ± 0.22	0.26
	13.4 ± 0.22	10.5 ± 0.27	0.0096	12.7 ± 0.29	0.16
	24.6 ± 0.29	20.7 ± 0.17	0.0031	23.8 ± 0.19	0.068
Puritan Bennett 840	4.6 ± 0.05	4.3 ± 0.17	0.095	5.1 ± 0.18	0.042
	10.1 ± 0.20	9.4 ± 0.09	0.037	10.6 ± 0.092	0.083
	14.5 ± 0.13	13.4 ± 0.2	0.017	14.8 ± 0.22	0.12
	25.9 ± 0.083	23.8 ± 0.16	0.0017	26.0 ± 0.18	0.49

Values are mean ± SD
R_{tot} = total inspiratory resistance

To measure C_{tube} we occluded all the openings of the tubing and used a volumetric syringe equipped with a 1-way valve to inject 10-mL volumes of air into the tubing while measuring intratube pressure. We calculated ΔV_{valve} and ΔV_{tube} for both the PB7200 and the PB840 and determined the mathematically corrected R_{tot} values.

Statistical Analysis

Each R_{tot} measurement was performed 3 times. We used the 2-tailed Student's *t* test to compare the benchmark R_{tot} values to the corrected and uncorrected R_{tot} measurements, to assess for absolute error, relative error, and significant differences. Difference were considered statistically significant when $p < 0.05$.

Results

Resistance Measurements

Table 1 shows the benchmark, uncorrected, and corrected R_{tot} values, and the p values for the differences. The benchmark values ranged from 4.1 to 25.9 cm H_2O/L/s. The benchmark values were slightly different between the PB7200 and the PB840, because they were set up at different times. The largest difference was 1.3 cm H_2O/L/s.

Figures 3 and 4 show the relative and absolute errors for the R_{tot} values from Table 1. With the PB7200 the uncorrected relative R_{tot} error averaged 23.2% (range 15.9–34.1%) and the absolute error averaged 2.6 cm H_2O/L/s (range 1.4–3.9 cm H_2O/L/s). Absolute error increased with increasing resistance, whereas relative error was greatest with the lowest resistance. The differences between the benchmark values and uncorrected PB7200 values were statistically significant in all cases.

With the PB840 the relative R_{tot} error averaged 7.3% (range 6.5–8.1%) and the absolute error averaged 1.0 cm H_2O/L/s (range 0.3–2.1 cm H_2O/L/s). Absolute error also increased with increasing resistance (see Figs. 3 and 4). In contrast to the PB7200, with the PB840 the relative error was *not* greatest with the small resistances, but instead remained within a narrow range for all R_{tot} values. Differences between the benchmark and the uncorrected PB840 resistances were statistically significant except in the case of the lowest resistance ($p = 0.095$).

Mathematical Characterization of Valves

To determine ΔV_{valve} we made 26 $\dot{V}$ measurements at the inspiratory port of the PB7200 and PB840 during valve closure. The ventilators used were not the same units we used for the R_{tot} measurements. Over the $\dot{V}$ range used, ΔV_{valve} averaged 32.8 mL for the PB7200 and 12.4 mL for the PB840. In both cases the ΔV_{valve} values were linearly distributed, as a function of $\dot{V}$, as would be predicted for a solenoid-driven proportional valve. By linear regression:

$$\Delta V_{valve,7200} = 0.031s \times \dot{V} - 0.0025L \tag{4}$$

and

$$\Delta V_{valve,840} = 0.0102s \times \dot{V} - 0.00086L \tag{5}$$

Mathematical Correction of R_{tot}

After mathematical correction with equations 2 through 5 the average relative error of R_{tot} measurements from the PB7200 was reduced to 3.0% (range 1.1–5.2%) (see Fig. 3) and the absolute error was reduced to an average of 0.4 cm H_2O/L/s. In the model with the smallest R_{tot} the mathematically corrected value was higher than the benchmark value, by 0.1 cm H_2O/L/s, whereas the other corrected values were still lower than the benchmark. After mathe-

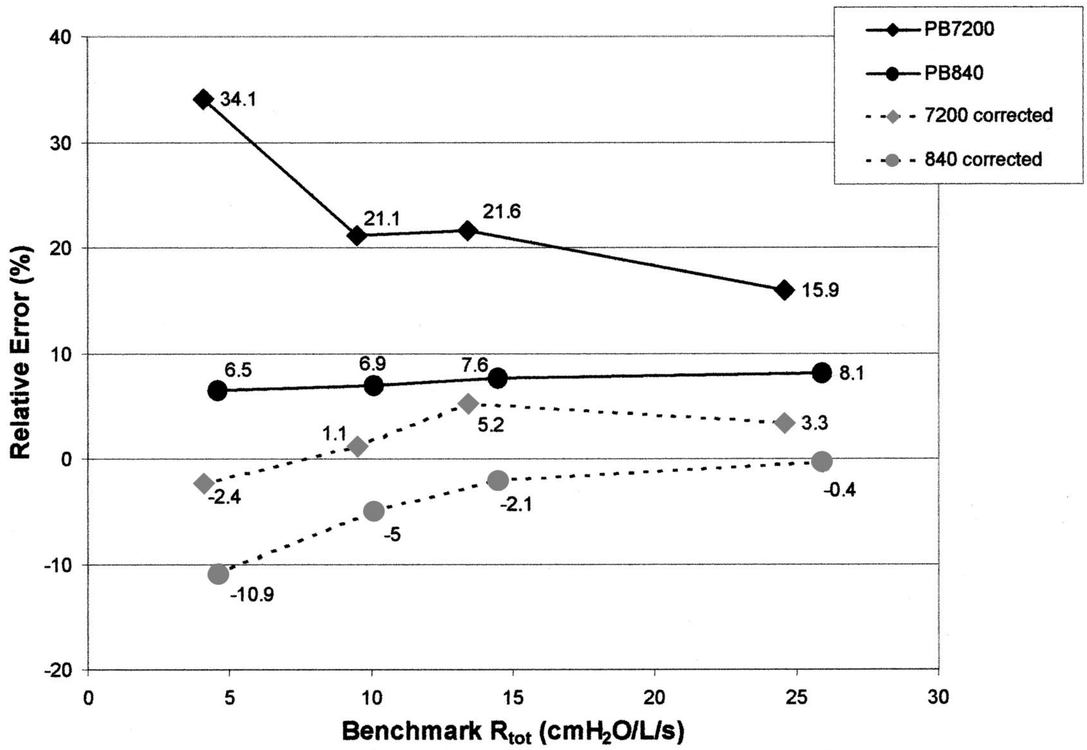

Fig. 3. Benchmark total inspiratory resistance (R_{tot}) values versus relative error in R_{tot} measurements (before and after mathematical correction) made with the Puritan Bennett 7200 ventilator (PB7200) and the Puritan Bennett 840 ventilator (PB840).

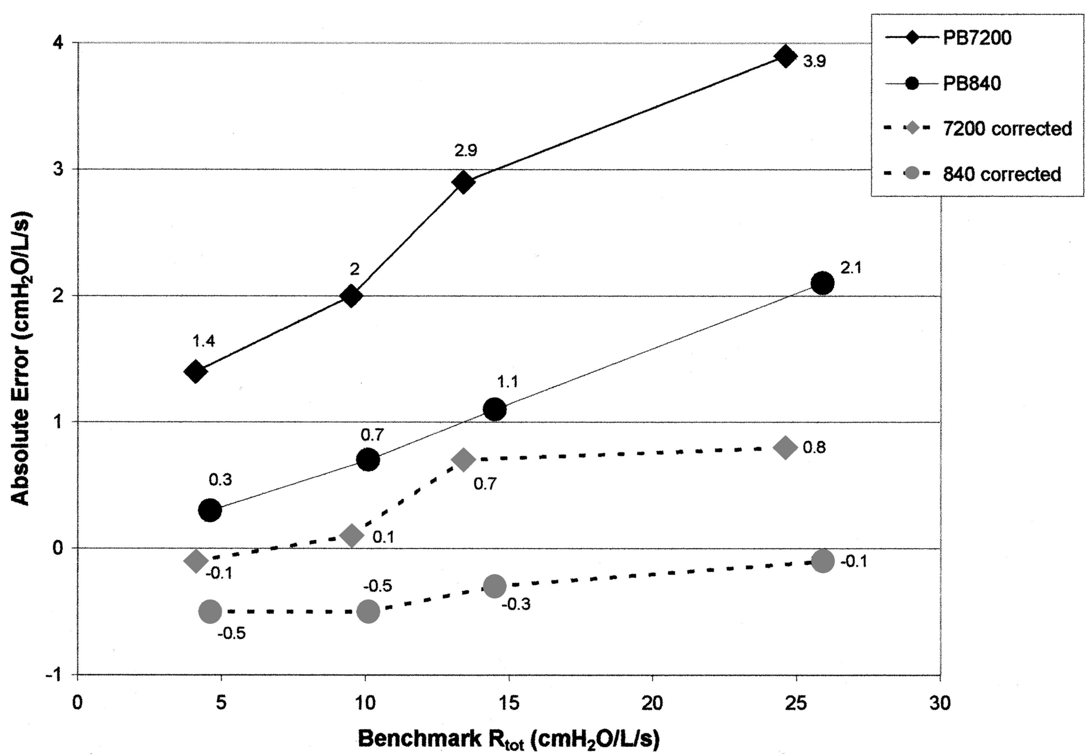

Fig. 4. Benchmark total inspiratory resistance (R_{tot}) values versus absolute error in R_{tot} measurements (before and after mathematical correction) made with the Puritan Bennett 7200 ventilator (PB7200) and the Puritan Bennett 840 ventilator (PB840).

matical correction the differences from benchmark were not statistically significant.

Mathematical correction also reduced the average R_{tot}-measurement error with the PB840 (see Figs. 3 and 4). The average relative error was 4.6% and the average absolute error was 0.3 cm $H_2O/L/s$. In all cases with the PB840 the corrected R_{tot} values were higher than the benchmark measurements. In the model with the lowest R_{tot} the mathematical correction actually *increased* the error, from 6.5% lower than benchmark to 10.9% higher than benchmark. However, the absolute error remained small; it was 0.3 cm $H_2O/L/s$ before correction and 0.5 cm $H_2O/L/s$ after correction. The difference from benchmark after correction was statistically significant ($p = 0.042$). The other differences were not significant.

Discussion

Our objectives were:

1. To determine and compare the respiratory resistance measurement errors of the PB7200 and PB840 ventilators
2. To derive mathematical characterizations of the PB840 and PB7200 ventilators' inspiratory valves to determine a mathematical correction for R_{tot} measurements
3. To use a bench model to determine the validity of the mathematical correction

With our model we found an average R_{tot}-measurement error of 23.2% with the PB7200. This aspect of the PB7200 had not been studied previously, but that magnitude of relative error is consistent with the report by Sly et al, who found an error range of 10.4–26.2% with a Siemens Servo 900C ventilator.[6] That study, however, was performed with a bench model of a pediatric respiratory system, with ventilation parameters substantially different than the adult ventilation parameters we used in the present study.

Our model found considerable relative errors, but the corresponding absolute values might not be clinically important. For instance, the 34.1% error measured with a resistance of 4.1 cm $H_2O/L/s$ represents an absolute error of only 1.4 cm $H_2O/L/s$. That magnitude of error may not be clinically important, but it could be problematic in research settings.

R_{tot} measurements from the PB840, which is a newer-generation ventilator, were considerably more accurate. In our low-compliance model, which represents a worst-case scenario, the relative error was never greater than 8.1% and the absolute error was $\leq$ 2.1 cm $H_2O/L/s$. Thus, ΔV_{valve} was less with the PB840 than with the PB7200, and R_{tot} measurements made with the PB840 are more accurate than those with the PB7200.

Bates et al showed with a computational model that measurements made with a valve that shuts in 12 ms may still give R_{tot} values that are as much as 7% lower than benchmark.[4] Benchmark measurements made with a rapid-occlusion valve may underestimate the measured error by a similar degree. However, Bates and Milic-Emili stated that a valve that closes in $\leq$ 10 ms is sufficient for accurate measurements.[16] Valves with that closure speed are the fastest available to physiologists and those valves are the benchmark for occlusion technique. Our intention was to compare R_{tot} measurements made with the ventilators' inspiratory valves and measurements that would be obtained in a standard physiology laboratory with the occlusion technique, and so the benchmark is appropriate for this study.

Others have described and used mathematical characterization of valve closure for correcting respiratory resistance measurements,[5,17–19] but those corrections have not been directly validated. A theoretical analysis predicted that mathematical correction would not be feasible, based on the large magnitude of relative error.[8] But our bench model findings support that mathematical correction of R_{tot} measurements can be useful. Our corrected R_{tot} values were within 1 cm $H_2O/L/s$ of the benchmark in all cases. All the uncorrected R_{tot} measurements were lower than the benchmark values. Some of the corrected values were higher than benchmark, but the degree of error was still reduced in all cases except one. We do not believe that the overestimation of the benchmark in some instances or the case where the mathematical correction increased the error indicates a fault with the method. The post-correction overestimation was small (always $\leq$ 0.5 cm $H_2O/L/s$), and this may reflect the small underestimation of the true R_{tot} that is inherent to the benchmark, as noted above. Similarly, in the case where the correction increased the error, the magnitude of error was small and the slight underestimation of the true R_{tot} by the benchmark may again obscure the result. By way of example, in this case the benchmark resistance measurement was 4.6 cm $H_2O/L/s$ and the error was 0.3 cm $H_2O/L/s$ (underestimated) before correction and 0.5 cm $H_2O/L/s$ (overestimation) after correction. If the benchmark underestimated the true resistance by 2% (4.7 cm $H_2O/L/s$ instead of 4.6 cm $H_2O/L/s$), which is possible, then the correction method would have decreased the error instead of increasing it.

The differences in mechanics between our bench model and animal or human subjects could affect the validity of a mathematical R_{tot} correction. R_{tot} is actually a combination of 2 resistances: R_{min} (the immediate decrease in P_{AO} [from P_{peak} to P_1] at the end of inspiration) and R_{dif} (the slower, small-amplitude drop from P_1 to P_{plat} that follows cessation of flow). R_{min} represents airway resistance, whereas R_{dif} is due to gas redistribution in the lung and viscoelastic properties of the respiratory system. R_{dif} is not often measured directly in clinical situations, because the measurement is technically difficult: a curvilinear backward extrapolation of the P_{AO} waveform to the time of valve closure must be performed.[3] More commonly, R_{tot} is measured and often used as an inference of airway resistance, for diagnosis or assessing treatment efficacy (eg, following suctioning, β agonists, or steroids). In certain disease states, such as obstructive lung disease or acute lung injury, R_{dif} may be a substantial component of R_{tot} and may be responsible for observed R_{tot} changes.[20–22] Our bench model is limited in that it does not include physiologic properties that contribute to R_{dif}. In principle, injecting ΔV_{valve} and ΔV_{tube} should offset P_1 and P_{plat} by the same amount, and thus the mathematical correction should accurately determine both R_{tot} and R_{min}. Based on that principle, mathematical corrections similar to ours have been applied to R_{min} and R_{tot} in human and animal models.[5,17–19] However, further investigation is needed to determine the applicability of measured and corrected R_{tot} measurements in animal and human subjects, particularly diseased subjects.

We chose to measure ΔV_{valve} in different individual ventilator units than those we used to measure R_{tot} in the bench model because of the reliable performance characteristics of the microprocessor-controlled servoid valves used in Puritan Bennett ventilators. In fact, ΔV_{valve} values from several different PB7200 units produced identical plots. We did not present those data here because they are unnecessary to the objectives of this report.

Conclusions

We found R_{tot}-measurement errors with the PB7200 and PB840. The error was less with the PB840. The absolute error with either ventilator seems unlikely to influence clinical decision making but may be important in research settings. The error due to valve-closure characteristics is predictable, and our mathematical descriptions of ΔV_{valve} allow correction of R_{tot} values. Our equations corrected R_{tot} measurements so that they were not significantly different than the benchmark values. In settings where very accurate R_{tot} measurements are necessary, using a rapid-occlusion valve at the airway opening is the benchmark method, but mathematical correction of measurements made with a ventilator is an alternative that requires less sophisticated equipment. These methods can easily be adapted to study other ventilator models.

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