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Evaluation of Tube Compensation in the Bennett 840 Ventilator—A New Ventilatory Mode to Support Spontaneous Breathing

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Hoshi, K., Ejima, Y., Hasegawa, R., Sasaki, C., Saitoh, K. and Matsukawa, S. *Evaluation of Tube Compensation in the Bennett 840 Ventilator—A New Ventilatory Mode to Support Spontaneously Breathing.* Tohoku J. Exp. Med., 2001, **195** (2), 65-72 —— Respiratory care patients frequently require intubation with an endotracheal tube (ETT). Unfortunately, the ETT introduces a pressure drop (⊿PETT) that depends on the respiratory flow rate, thus increasing the work of breathing (WOB). Pressure support ventilation (PSV) cannot adequately compensate for this added WOB, because the degree of inspiratory assistance by PSV is fixed. Therefore, a technique called tube compensation (TC) has been developed to address ⊿PETT. We examined the performance of TC and compared it with PSV of 5 cmH_2O. The experimental system was constructed from a simulator, a test-lung, flow sensors, and a Bennett 840, and the respiratory parameters were studied. ETTs with IDs 6.5 and 8.0 mm were used.

The quadratic approximation obtained for ⊿PETT in the 6.5-mm ETT was $2.316 \times \text{flow} + 7.910 \times \text{flow}^2$, while that for the 8.0-mm ETT was $1.881 \times \text{flow} + 3.353 \times \text{flow}^2$. The maximum inspiratory flow (MIF) increased significantly with increasing TC, but tidal volume and inspiratory time did not show marked changes. The MIF for TC of 100% was larger than that for PSV of 5 cmH_2O, when the 6.5-mm ID was used, but there was no significant difference between these modes when an ID of 8.0 mm was used. For both the 6.5 and 8.0-mm IDs, the PV loop corresponding to 100% TC was larger than that for PSV of 5 cmH_2O. TC only compensated for the WOB caused by the ETT, whereas PSV compensated for the WOB caused by the ETT and the demand valve system. In clinical use, the differences between TC and PSV will demand attention. ———— tube compensation; work of breathing; flow-dependent pressure drop

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In the intensive care unit (ICU), many patients need to be intubated with an endotracheal tube (ETT) for respiratory care. However, the ETT introduces a pressure drop (⊿PETT) that is dependent on the inspiratory flow rate, and increases the work of breathing (WOB) during spontaneous respiration (Shapiro et al. 1986; Bersten et al. 1989). It is widely accepted that the additional WOB caused by endotracheal tubes and the non-ideal demand-flow characteristics of the ventilator should be compensated for by using an appropriate level of inspiratory pressure assistance (Fiastro et al. 1988), and pressure support ventilation (PSV) is widely used as a standard ventilatory mode during the weaning and extubation process. It has been found that the pressure drop across the endotracheal tube has a nonlinear dependence on the flow generated by the ventilator (Guttmann et al. 1993). Since airflow varies from breath to breath and within each single breath, a constant level of pressure support is unable to compensate for the time varying ⊿PETT (Guttmann et al. 1997).

Tube compensation (TC) was developed as a new ventilatory mode that is able to compensate for ⊿PETT. This technique is based on continuous calculation of the flow-dependent ⊿PETT (Fabry et al. 1997). Haberthür et al. (1999) reported that TC was a suitable mode for compensating for added WOB at any level of ventilatory effort in tracheostomized patients. TC is considered to be a useful mode in respiratory care for maintaining the normal effort of spontaneous breathing (Mols et al. 2000). The first ventilator to employ TC—the Puritan Bennett 840 ventilator (Mallinckrodt Inc., St. Louis, MO, USA)—has been commercially available worldwide.

In this study, which preceeded clinical application of the Bennett 840 ventilator, a test-lung was used to examine the performance of TC as compared with a PSV of 5 cmH_2O, which is a commonly used ventilation mode before extubation in our ICU.

METHODS

The experimental setup is shown in Fig. 1. We serially connected a Puritan Bennett 7200ae ventilator (Mallinckrodt Inc.) to simulate spontaneous breathing, a DUAL ADULT TTL MODEL 1600 test-lung (Michigan Instruments, Inc., Grand Rapids, MI, USA), a flow sensor (OMR8101, Nihon Kohden Co., Tokyo), an endotracheal tube (Mallinckrodt Inc.), another OMR flow sensor, and a Puritan Bennett 840

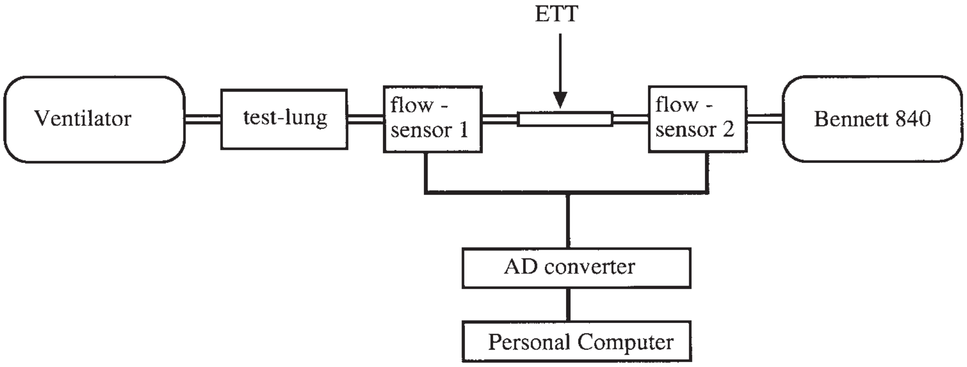

Fig. 1. Schematic diagram of the experiment.
The ventilator was used as a spontaneous breathing simulator with a deceleration wave (Tidal Volume = 600 ml, Respiratory Rate = 12 breaths/minute, Peak Inspiratory Flow = 60 or 100 liter/minute). The test-lung was a DUAL ADULT TTL MODEL 1600 (Resistance = 20 cmH_2O/liter/secend, Compliance = 50 cmH_2O/liter). The flow sensors were made by Nihon Kohden Co. The AD converter was a MacLab 8-ch. unit. ETT indicated the endotracheal tube (ID 6.0 or 8.0 mm).

ventilator.

The PB 7200ae ventilator settings were: a tidal volume (VT) of 600 ml, a respiratory frequency (f) of 12 breath/minute, peak inspiratory flow rates (PIF) of 60 and 100 liter/minute (a deceleration wave), FiO_2 of 0.21, inspiratory pause of 0%, and positive end-expiratory pressure (PEEP) of 0 cmH_2O. Because the waveform associated with spontaneous breathing is a deceleration wave, this type of waveform was used to stimulate spontaneous breathing in this study. The PIF of 60 liter/minute was used to create the typical maximum inspiration flow rate of an adult during normal spontaneous breathing, whereas the PIF of 100 liter/minutes was used to create the early, shallow, spontaneous breathing. The PB 7200ae was connected to one side of the test-lung so as to simulate spontaneous breathing on the other side. The resistance of the test-lung was set at 20 cmH_2O/liter/second, and the compliance at 50 cmH_2O/liter.

Airway pressure and flow were simultaneously measured at proximal and distal sites in the ETT. The flow, VT, and pressure signals were recorded on a PowerBook G3 (Apple Computer, Inc., Cupertino, CA, USA) via a MacLab AD converter (ADInstruments Pty Ltd., Sydney, Australia) at a sample rate of 100 Hz. We calculated and analyzed the maximum inspiratory flow rate (MIF), the area of the presure-volume loop (PV loop) at the distal site of the ETT, and the relationship between ⊿PETT and the inspiratory flow rates using Chart v3.6.4B5/s (ADInstruments Pty Ltd.), Microsoft Excel (Microsoft Corporation, Redmond, WC, USA), and StatView-J5.0 (SAS Institute Inc., Cary, NC, USA). The MIF and PV loops were calculated on a breath-by-breath basis. Data from six consecutive breaths were analyzed and averaged.

The actual WOB of a patient should be calculated from the intrapleural pressure, as in Fig. 2. Although this study was conducted in a test-lung, the PV loop yielded at the distal site of the ETT was used to analyze the WOB. The WOB represented by dots in Fig. 2A is equivalent to the amount of biological work due

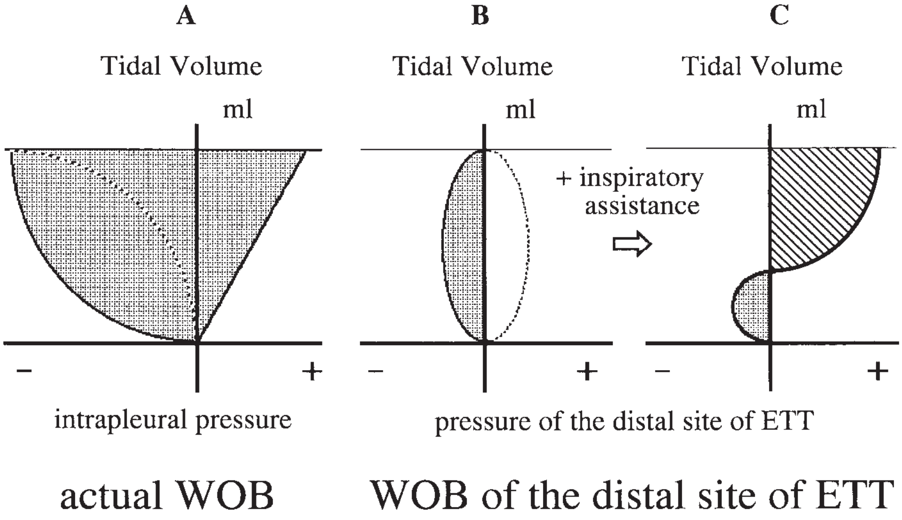

Fig. 2. Diffusion of WOB during inspiration (Fabry et al. 1997). The actual WOB of a patient should be calculated using the intrapleural pressure. The area represented by dots in Fig. 2A is equivalent to the biological work due to respiratory muscle effort alone, whereas the shaded area in Fig. 2C is mechanical work delivered by the ventilator and has no biological equivalent. The area represented by dots in Fig. 2C could be considered to express the WOB at the distal site to the ETT. The area represented by dots at the distal site to the ETT under the inspireatory assistance would become smaller, so that the inspiratory pressure assistance would become larger.

to respiratory muscle effort, whereas the shaded area in Fig. 2C represents the mechanical work delivered by a ventilator and has no biological equivalent (Fabry et al. 1997). The area represented by dots at the distal site of the ETT would become smaller while the inspiratory pressure assistance would become larger. The ability of a ventilatory mode to minimize WOB was, therefore, deemed to be of greatest importance.

The data from the distal site of the ETT were analogous to those arising from the trachea in the clinical situation, where the "trach" was postfixed. The data from the proximal site of the ETT were treated as arising from the Y-connector part, where the "air" was postfixed. An ETT with an inner diameter (ID) of 8.0 mm was used because it is a popular adult size, while the 6.5-mm version is a narrow adult size that possesses a larger resistance. Each ETT was wed by its original length, and connected to the original circuit, and the cuff was inflated by air without leakage.

The settings applied to the Bennett 840 assumed a body weight of 60 kg, PEEP of 0 cmH_2O, FiO_2 of 0.21, and a pressure triggering sensitivity of 1 cmH_2O. Before the application of the Bennett 840, the measurement was initially performed with the proximal end of the ETT opened to the atmosphere (open); the relationship between $\varDelta$PETT (where $\varDelta PETT = P_{air} - P_{trach}$) and the inspiratory flow rate was analyzed. Secondly, measurements of breathing pattern were performed during several modes of the Bennett 840 (TC=10, 50, and 100%, and PSV=5 cmH_2O).

The approximating formulas for the relationship between $\varDelta$PETT and the air flow rate was calculated by polynomial regression through zero using Stat View-J5.0. Statistical analyses was performed using repeated measures ANOVA within same conditions and using the unpaired *t*-test between different conditions. A *p*-value less than 0.05 was considered statistically significant. The numerical values are given as mean±S.D.

RESULTS

The quadratic approximating formulas obtained from the measured points during inspiration were: $\varDelta P_{ETT} = 2.316 \times flow + 7.910 \times flow^2$ ($r^2 = 0.99$, $p < 0.0001$) for the 6.5 mm ID, and $1.881 \times flow + 3.353 \times flow^2$ ($r^2 = 0.98$, $p < 0.0001$) for the 8.0 mm ID ($\varDelta$P: cmH_2O, flow: liter/second).

Fig. 3 shows the PV loop at the distal site during the inspiratory phase, for both IDs, with a PIF of 100 liter/minute. The area presented by dots expresses WOB in the distal site of the endotracheal tube. The size of this area decreased with increasing TC, while the area corresponding to the 5 cmH_2O PSV mode was smallest.

Table 1 shows the values of maximum inspiratory flow (MIF), VT, inspiratory mean flow (MF), inspiratory time (TI), and PV loop, for all test conditions. MIF increased significantly with increasing TC, under all conditions. The MIF of the 6.5 mm ID with 100% TC was larger than that for PSV of 5 cmH_2O, but, for the 8.0 mm ID there was no significant difference between two modes. VT in the 6.5 mm ID increased slightly from TC 10%, but VT in the 8.0 mm ID showed no such change. In both ETT sizes, the VT at 100% TC was smaller than that at 5 cmH_2O PSV. For both 6.5 and 8.0 mm IDs, TI showed no change with increasing TC, and TI for 100% TC was slightly shorter than that for 5 cmH_2O PSV. MF for both IDs increased slightly with increasing TC, but was smaller for 100% TC than for 5 cmH_2O PSV. The size of the PV loop during inspiration at the distal site of the ETT decreased significantly with increasing TC. The area of PV loop at any size of tube and settings was smallest with PSV 5 cmH_2O, except 6.5 mm tube at 60 liter/minute with TC 100%.

DISCUSSION

The presence of an ETT for the respiratory

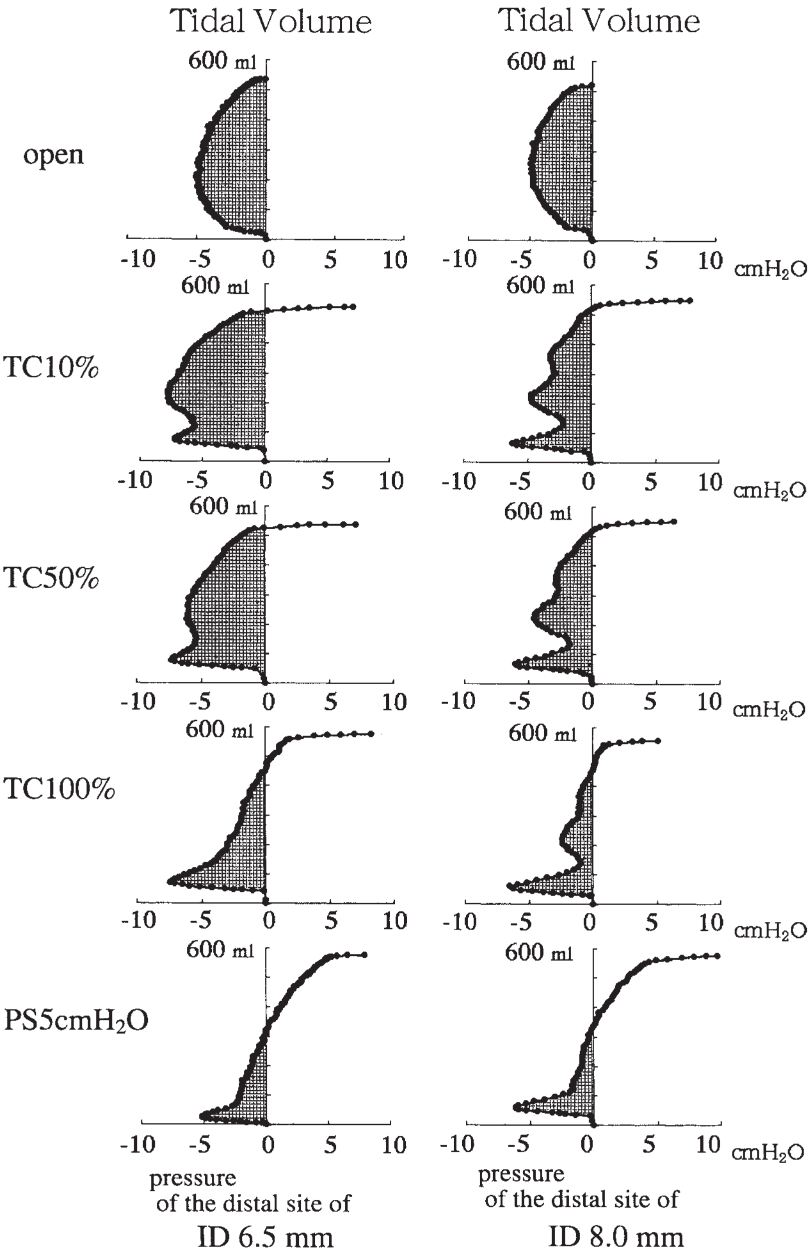

Fig. 3. PV loops during inspiration at the distal site to the ETT, at 100 liter/minute. The area represented by dots may express the WOB at the distal site to the ETT.

care increases the WOB during spontaneous breathing. It has been suggested that an appropriate level of inspiratory pressure assistance could theoretically compensate for this additional WOB. The flow, timing, and volume pattern of PSV were designed to adjust to patient demand with every breath. Both the work of breathing and the oxygen cost of breathing were diminished when PSV was applied (MacIntyre 1986). PSV has been widely used as the mode for respiratory care. However, the flow rate of the PSV mode causes the flow-dependent liter/minute pressure drop across the ETT and increases WOB.

There is the Hagen-Poiseuille formula for the relationship between the pressure and the flow. This formula is difficult to apply clinically, because this is the formula for the laminar

TABLE 1. *Values of MIF, TV, TI, MF, and PV loop*

		6.5 mm-100 L/min.	6.5 mm-60 L/min.	8.0 mm-100 L/min.	8.0 mm-60 L/min.
MIF L/sec.	Open	51.90 ± 1.08	40.10 ± 1.20	57.60 ± 0.90	42.60 ± 0.84
	TC 10%	58.86 ± 1.20^{a}	44.76 ± 0.60^{a}	63.12 ± 0.36^{a}	46.86 ± 0.48^{a}
	TC 50%	$61.62 \pm 0.36^{a,b}$	45.48 ± 0.30^{a}	63.48 ± 0.42^{a}	$47.76 \pm 0.78^{a,b}$
	TC 100%	$69.06 \pm 0.54^{a,b,c}$	$51.66 \pm 0.30^{a,b}$	$66.42 \pm 1.02^{a,b,c}$	$49.74 \pm 0.66^{a,b,c}$
	PSV 5 cmH_2O	$59.64 \pm 0.60^{a,c,d}$	$46.50 \pm 0.66^{a,b,d}$	$65.82 \pm 0.54^{a,b,c}$	$49.32 \pm 0.48^{a,b,c}$
VT ml	Open	598.8 ± 17.1	529.0 ± 3.8	456.7 ± 23.3	476.3 ± 15.1
	TC 10%	479.3 ± 6.5^{a}	503.4 ± 4.2^{a}	528.3 ± 3.1^{a}	547.6 ± 3.9^{a}
	TC 50%	480.1 ± 3.6^{a}	507.3 ± 5.6^{a}	527.7 ± 4.3^{a}	543.1 ± 4.5^{a}
	TC 100%	$507.9 \pm 6.3^{a,b,c}$	512.0 ± 4.8^{a}	535.8 ± 3.0^{a}	545.9 ± 2.6^{a}
	PSV 5 cmH_2O	$578.1 \pm 12.1^{a,b,c,d}$	$577.3 \pm 1.9^{a,b,c,d}$	$570.8 \pm 5.7^{a,b,c,d}$	$609.4 \pm 4.7^{a,b,c,d}$
TI sec.	Open	0.790 ± 0.006	1.168 ± 0.004	0.770 ± 0.000	1.160 ± 0.006
	TC 10%	0.808 ± 0.004^{a}	1.170 ± 0.006	0.795 ± 0.005^{a}	1.162 ± 0.004
	TC 50%	$0.800 \pm 0.006^{a,b}$	1.165 ± 0.005	0.790 ± 0.000	1.165 ± 0.005
	TC 100%	0.795 ± 0.005^{b}	1.165 ± 0.005	$0.785 \pm 0.005^{a,b}$	1.162 ± 0.004
	PSV 5 cmH_2O	$0.822 \pm 0.004^{a,b,c,d}$	$1.177 \pm 0.005^{a,c,d}$	$0.797 \pm 0.005^{a,c,d}$	1.163 ± 0.005
MF ml/sec.	Open	758 ± 18	453 ± 4	593 ± 31	411 ± 13
	TC 10%	593 ± 9^{a}	430 ± 5^{a}	665 ± 6^{a}	471 ± 5^{a}
	TC 50%	600 ± 6^{a}	435 ± 4^{a}	668 ± 5^{a}	466 ± 5^{a}
	TC 100%	$639 \pm 10^{a,b,c}$	$440 \pm 3^{a,b}$	$683 \pm 6^{a,b,c}$	470 ± 3^{a}
	PSV 5 cmH_2O	$704 \pm 16^{a,b,c,d}$	$491 \pm {}^{a,b,c,d}$	$716 \pm 5^{a,b,c,d}$	$524 \pm 4^{a,b,c,d}$
PV loop J/L	Open	0.198 ± 0.011	0.186 ± 0.001	0.168 ± 0.009	0.104 ± 0.002
	TC 10%	0.246 ± 0.009^{a}	0.124 ± 0.002^{a}	0.136 ± 0.001^{a}	0.048 ± 0.001^{a}
	TC 50%	0.212 ± 0.008^{b}	$0.119 \pm 0.001^{a,b}$	$0.131 \pm 0.001^{a,b}$	0.041 ± 0.001^{a}
	TC 100%	$0.111 \pm 0.001^{a,b,c}$	$0.047 \pm 0.001^{a,b,c}$	$0.074 \pm 0.002^{a,b,c}$	$0.029 \pm 0.002^{a,b,c}$
	PSV 5 cmH_2O	$0.051 \pm 0.002^{a,b,c,d}$	$0.052 \pm 0.000^{a,b,c,d}$	$0.056 \pm 0.003^{a,b,c,d}$	$0.015 \pm 0.001^{a,b,c,d}$

Mean±S.D.
$^{a}p < 0.05$ vs. Open.
$^{b}p < 0.05$ vs. TC 10%.
$^{c}p < 0.05$ vs. TC 50%.
$^{d}p < 0.05$ vs. TC 100%.
MIF, maximum inspiratory flow; VT, tidal volume; TI, inspiratory time; MF, mean flow (=VT/TI). L/min., liter/minute; sec., seconds, ml/sec., ml/seconds, J/L, Joule/liter.
The values are the means of the average values of six consecutive breaths.
A p-values are for comparisons in the same condition.

flow and the flow in the ETT is the turbulent flow. Guttmann et al. (1993) simultaneously measured the pressure in the airway and in the tip of an ETT in order to calculate the $\varDelta$PETT, and found that the relationship between $\varDelta$PETT and flow rate was quadratic: $\varDelta PETT = K1 \times flow + K2 \times flow^2$ (K1, K2 = coefficients). They pointed out that-in intubated and spontaneously breathing patients-the flow-dependent resistance of the endotracheal tube could increase the resistance of the respiratory system several fold, especially at high rates of gas flow. Tube compensation was developed as a new ventilatory mode that could compensate adequately for the endotracheal tube resistance during spontaneous breathing (Fabry et al. 1997; Haberthür et al. 1999). TC allows the patient's inherent breathing pattern, albeit with the inclusion of PEEP, and allows physicians to predict the patient's breathing pattern after extubation. Thus, TC was expected to become a useful mode for respiratory care. In this study, we examined the performance of TC using a test-lung, prior to clinical application. Furthermore, we compared TC with PSV of 5 cmH_2O, this being a commonly used mode before extubation in our ICU. The level of PSV required to compensate for the WOB was increased by the presence of an ETT, with or without a demand valve, from 3.4 to 14.4 cmH_2O (Brochard et al. 1991). We have often used PSV of 5 cmH_2O during the weaning and extubation process, referring to other clinical bedside parameters (such as VT, RR, and blood gas analysis data).

In the first results of this study, under the condition of 60 liter/minutes in the 8.0-mm ETT, we were able to simulate adult spontaneous breathing (VT of 476 ml, TI of 1.16 seconds, MIF of 42.6 liter/minute, and MF of 411 ml/second); under the condition of 100 liter/minute in the 8.0-mm ETT, we were able to simulate rapid, shallow breathing (VT of 457 ml, TI of 0.77 seconds, MIF of 57.6 liter/minute, and MF of 593 ml/second). The pressure drop across the ETT for both IDs can be expressed by a quadratic function of flow: $\Delta P = K1 \times flow + K2 \times flow^2$; $K1 = 2.316$, $K2 = 7.910$ at 6.5 mm ID, and $K1 = 1.881$, $K2 = 3.353$ at 8.0 mm ID.

The Bennett 840 ventilator measures flow and airway pressure inside the machine, and continuously calculates the pressure drop across the ETT using the original coefficients. The ventilator raises the airway pressure during inspiration depending on the inspiratory flow rate and controls the tracheal pressure at the PEEP level. In this study, when the TC was increased from 10 to 50 and 100%, the airway pressure increased accordingly. As the Bennett 840 ventilator increased the inspiratory flow in order to compensate for the $\Delta PETT$, MIF increased depending on the increase in TC. Changes in VT, TI, and MF following the increase in MIF varied significantly according to the size of the tube and PIF.

Comparing 100% TC with 5 cmH_2O PSV, MIF was larger for the 6.5 mm ID, but there was no significant difference for the 8.0 mm ID, while, for both modes, MIF for the 6.5 mm ID was larger than for the 8.0 mm ID. This demonstrated the larger resistance of the narrow ETT. VT for 100% TC was smaller than that for 5 cmH_2O PSV, while TI was slightly shorter and MF was also smaller. The PV loops for 100% TC in the 6.5 mm ID at 100 liter/minute and in the 8.0 mm ID at both flows were larger than those for 5 cmH_2O PSV, but the PV loop for 100% TC in the 6.5 mm ID at 60 liter/minute was smaller than that for the 5 cmH_2O PSV. This suggests that the inspiratory assistance provided by 5 cmH_2O PSV was greater than that provided by 100% TC, except under the 60 liter/minute condition in the ETT of 6.5 mm ID. This result differs slightly from another report (Haberthür et al. 2000), although that study was performed using patients while we used a test-lung without patient-ventilator interaction. Direct application of the data from the present study to the clinical setting is limited by the fact that this was a model lung. Furthermore, TC only compensated for the additional WOB caused by the pressure drop along the endotracheal tube, while PSV compensated for all additional WOB caused by the ETT and the ventilator demand valve system, including the ventilator circuit. Therefore, it is possible that the extra WOB caused by the resistance of the ventilator system resulted these findings.

CONCLUSIONS

TC is a new ventilator mode designed to compensate for the flow-dependent tube resistance during inspiration by decreasing the WOB. Furthermore, consideration must be paid to the differences between TC and PSV with regard to reduction of WOB.

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