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Automatic compensation of endotracheal tube resistance in spontaneously breathing patients

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Abstract

The considerable additional ventilatory work needed to overcome the resistance of the endotracheal tube (ETT) is flow-dependent. In spontaneously breathing intubated patients this additional ventilatory work is therefore dependent on the flow pattern and cannot be adequately compensated for by support with a constant pressure. We propose a method to fully compensate for the ETT resistance during inspiration and expiration by regulating tracheal pressure (P_{trach}). P_{trach} is calculated at a rate of 500 Hz by measurement of flow and pressure at the outer end of the ETT and from coefficients describing the flow-dependent ETT resistance. The calculated tracheal pressure is fed into a modified demand-flow ventilator which can then control tracheal pressure to a target value ($P_{trach,targ}$). Tracheal pressure can either be kept constant (automatic tube compensation, ATC), or changed in any chosen fashion. We tested our system on a laboratory lung model simulating a spontaneously breathing patient. Even under the simulation of extreme conditions the maximum deviation of P_{trach} from $P_{trach,targ}$ was smaller than 2.5 mbar. We evaluated our system in 10 spontaneously breathing intubated patients breathing at ATC with or without volume proportional pressure support (VPPS) by measuring P_{trach}. The mean maximum deviation of P_{trach} from $P_{trach,targ}$ was 2.9 mbar. The rms-deviation was 1.1 mbar (inspiration and expiration considered) and 1.7 mbar (inspiration alone). The accuracy of the control of P_{trach} is thus comparable to the control of airway pressure afforded by the unmodified demand-flow ventilator.

Key words: Ventilator weaning; Respiration; Intubation

1. Introduction

During spontaneous breathing the respiratory muscles accomplish the work necessary for the distension of the lung and thorax. During controlled ventilation the respiratory muscles are entirely passive while the lung and thorax are inflated by the positive pressure applied to the patient's airways via the endotracheal tube (ETT). Between spontaneous breathing and controlled ventilation lie the various methods for assisted spontaneous breathing. The technique closest to spontaneous breathing is continuous positive air-

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way pressure (CPAP) where a continuous pressure is applied to the patient's airways (or, more accurately, to the outer end of the ETT) and is kept constant over inspiration and expiration. Also commonly used is patient triggered inspiratory pressure support (IPS), where the ventilator attempts to sense and subsequently support the patient's inspiration by a constant positive pressure being applied to the outer end of the ETT. The inspiratory pressure support aims to partly relieve the patient of his ventilatory work, arising predominately from viscous resistances and the elasticity of the respiratory system. The latter is only dependent on tissue properties, whereas the viscous resistances are caused not only by the patient's conductive airways but also by components of the ventilator (resistance of ventilator tubing, humidifier and expiratory valve) and most importantly by the ETT [4,12]. This problem has previously been recognized and various authors have claimed to be able to compensate for the additional ETT resistance by giving a suitable level of inspiratory pressure support [5,2]. But compensating the ETT resistance by a constant pressure support does not take into account that the pressure drop across the ETT is not constant. Specifically, the ETT resistance is flow-dependent, and the pressure drop across the ETT increases much more than proportionally with flow.

It has been suggested that in spontaneously breathing patients, the additional ventilatory work can be quantified by the difference between airway pressure (P_{aw}) measured at the outer end of the ETT and the positive end-expiratory pressure (PEEP) or the CPAP-value [7]. Fig. 1 shows the pressure/volume diagram of one breath of a spontaneously breathing patient intubated with an ETT of 7.5 mm ID and 24 cm long, with CPAP was set at 5 mbar. The difference between P_{aw} and the set CPAP-value was between -2.5 mbar (inspiration) and $+3.3$ mbar (expiration). However, a much larger difference between tracheal pressure (P_{trach}) and the set CPAP-value was found, ranging from -10.4 mbar (inspiration) to $+8.8$ mbar (expiration). Any difference between P_{aw} and P_{trach} during inspiration represents additional ventilatory work, which is caused only by the ETT. Assessing the patient's additional ventilatory work by using the airway pressure registration, does not take into account the ETT resistance and is therefore of only limited value [9,2].

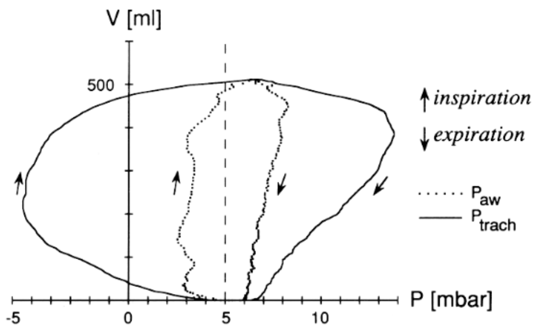

Fig. 1. Pressure/volume loops of a spontaneously breathing intubated patient with a CPAP of 5 mbar. Airway pressure (P_{aw}), measured at the outer end of the ETT, is shown by the dotted line. The deviation from the set CPAP-value (shown by the dashed line) lies between -2.5 mbar (inspiration) and $+3.3$ mbar (expiration). Tracheal pressure (P_{trach}) is shown by the continuous line. The deviation from the set CPAP-value lies between -10.4 mbar (inspiration) and $+8.8$ mbar (expiration) and is due to the ETT resistance.

Conventional ventilators measure P_{aw} at the outer end of the ETT and regulate P_{aw} at the target airway pressure ($P_{aw,targ}$) by controlling the gas flow (demand-flow ventilator). A major problem with assisted spontaneous breathing lies in the fact that the patient's effort during inspiration cannot be known in advance and therefore it is not possible to preset the timing and pattern of the inspiratory flow. During a forced inspiration, for example, the ventilator is ideally capable of increasing the necessary inspiratory flow with a minimal possible delay. If P_{aw} falls below the preset $P_{aw,targ}$ the ventilator will increase gas flow, if P_{aw} exceeds $P_{aw,targ}$, gas flow will be reduced. The ventilator's feedback control needs to minimize the deviation of P_{aw} from $P_{aw,targ}$ as quickly as possible, at the same time trying to avoid oscillations which can arise due to the ventilator's response delay and due to the compliance of the tubes, humidifier etc. Conventional demand-flow ventilators calculate the necessary gas flow (V') as a function of target deviation (P_{err}) and time (t):

$$V' = f(P_{err}, t) \tag{1}$$

where P_{err} is the deviation of P_{aw} from $P_{aw,targ}$.

Under CPAP the demand-flow ventilator controls P_{aw} to the desired CPAP-value so that the patient breathes at a constant P_{aw} which is greater than atmospheric pressure. The ventilator thus compensates for any resistances up to the point where the pressure measurements are taken, i.e. the outer end of the ETT. If the ventilator measured P_{trach} instead of P_{aw} then the ETT resistance would also be compensated for. However, direct measurement of P_{trach} via a catheter is problematical and cannot be performed continuously [6].

We have presented a method to continuously calculate tracheal pressure from measurements of P_{aw} and V' [6]. Briefly, we first determined the relation between V' and pressure drop across the ETT (ΔP_{ETT}) in the laboratory, using an artificial trachea, and approximated this relationship by

$$\Delta P_{ETT} = K_1 * V' + K_2 * V'^2 \tag{2}$$

where K_1 and K_2 are coefficients which are dependent on the inner diameter and length of the ETT and differ with inspiration and expiration. Continuous measurements of P_{aw} and V' thus allowed us to continuously calculate tracheal pressure ($P_{trach,calc}$) according to Eq. 3:

$$P_{trach,calc} = P_{aw} - \Delta P_{ETT} \tag{3}$$

As our results revealed an excellent correspondence between the tracheal pressure measured by catheter and the calculated tracheal pressure [6], we decided to use $P_{trach,calc}$ to control the ventilator. An electronically operated demand-flow ventilator was then modified so that P_{trach} could be controlled to any desired pattern.

The present study aims firstly to describe the technical realization of a demand-flow ventilator which controls tracheal pressure, secondly to describe its properties on a laboratory lung model simulating a spontaneously breathing intubated patient, and thirdly to show its practicability for controlling tracheal pressure in patients.

2. Method

We modified a commercially available ventilator (EVITA, Dräger, Lübeck/Germany) in such a way that the difference between the target tracheal pressure ($P_{trach,targ}$) and the calculated tracheal pressure ($P_{trach,calc}$) can be fed into the demand-flow control unit of the ventilator. The expiratory valve of the ventilator is controlled with a current that is proportional to a pressure which is the sum of the target tracheal pressure, the pressure drop across the ETT and the pressure drop across the expiratory branch of the ventilator. Fig. 2 shows the circuit diagram of the ventilator and our technical modifications.

2.1. Regulation system

Unmodified state: A differential pressure transducer measures the deviation of P_{aw} from the servo pressure of the expiratory valve (P1) which is regulated at $P_{aw,targ}$. This target deviation (P_{err}) is represented by the voltage U_{err} which is fed into the demand-flow controller of the ventilator ($V'C$) that controls the two high pressure servo valves (HPSV) for oxygen and compressed air. The resulting gas flow (V') is a function of P_{err} and time (Eq. 1). The demand-flow controller of the ventilator approximatively can be described as a "proportional integral controller" with a response delay of 20 ms. Fig. 3 shows the step response of V' to sudden changes in P_{err} of a duration of 0.5 s and of varying magnitudes.

The EVITA-ventilator has a pneumatic expiratory valve which opens when the pressure in the expiratory limb is greater than the servo pressure P1 applied to the valve membrane. P1 is produced by the PEEP-valve. Atmospheric pressure acts as the reference pressure P3 of the PEEP-valve. The current I_{exp} controls the PEEP-valve in such a way that the pressure difference P1-P3 is proportional to I_{exp}.

Modified state: We made use of the fact that the entire ventilator can be described as operating as a function of the two electrical parameters U_{err} and I_{exp}. In the modified state the external control unit generates U_{err} and I_{exp} and feeds them into the ventilator. The external unit uses V' and P_{aw}, both measured at the outer end of the ETT, as input parameters from which it determines $P_{trach,calc}$ and generates U_{err} and I_{exp} as output parameters in such a way that the ventilator produces a P_{trach} that is equal to $P_{trach,targ}$, in other words P_{aw} is automatically increased or decreased

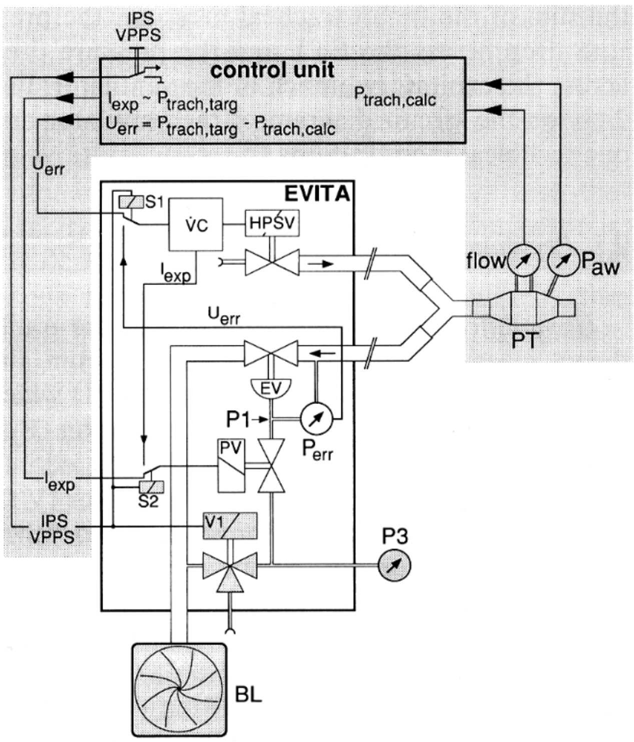

Fig. 2. Circuit diagram of the ventilator: Unmodified state: The PEEP-valve (PV) produces a servo pressure (P1) that regulates the expiratory valve (EV). P1 is regulated to the target airway pressure ($P_{aw,targ}$). The deviation of the airway pressure (P_{aw}) from $P_{aw,targ}$ can thus simply be measured by a differential pressure transducer. According to this deviation (P_{err}) the demand-flow controller of the ventilator (V′C) then controls the two high pressure servo valves (HPSV) for oxygen and compressed air (only one valve shown) for the inspiratory flow (V′). The blower (BL) is not in use, the expiratory limb is open to atmospheric pressure. The valve V1 connects the reference entry of the PEEP-valve with atmospheric pressure. Modified state (shaded grey): The gas flow produced by the HPSV is controlled by the voltage U_{err} that is fed into the demand-flow controller of the ventilator from an external control unit. The same unit produces a current I_{exp} that controls the servo pressure (P1) to the value of $P_{aw,targ}$. The blower (BL) produces a subatmospheric pressure of −14 mbar that is connected to the expiratory valve (EV), while also acting as reference pressure P3 for the PEEP-valve. V′ and P_{aw} are determined at the outer end of the ETT. The two relays S1 and S2 and the valve V1 are used to switch from the modified to the unmodified state whenever desired.

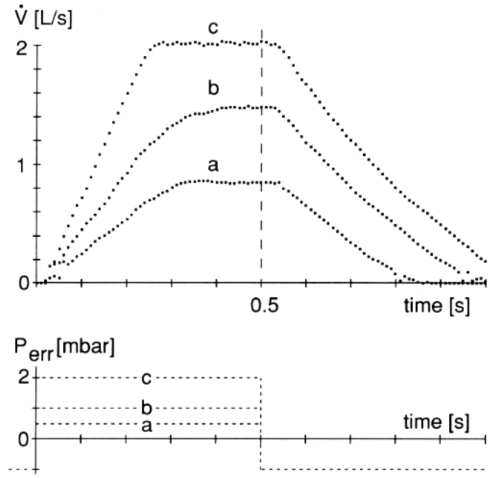

Fig. 3. Step response of the inspiratory flow to sudden changes in P_{err} of a duration of 0.5 s and of varying magnitudes (a) -1.0 mbar $\rightarrow +0.5$ mbar $\rightarrow -1.0$ mbar (b) -1.0 mbar $\rightarrow +1.0$ mbar $\rightarrow -1.0$ mbar c) -1.0 mbar $\rightarrow +2.0$ mbar $\rightarrow -1.0$ mbar.

by the pressure drop across the ETT. When only compensation for ETT resistance without any further pressure support is desired, called automatic tube compensation (ATC), then P_{trach} is regulated to a constant value over the entire breath:

$$P_{trach,targ} = PEEP. \tag{4a}$$

If it is desired that the patient is relieved not only of the ventilatory work necessary to overcome the ETT resistance but also of part of his elastic and/or resistive ventilatory work, then $P_{trach,targ}$ is calculated by the external control unit according to:

$$P_{trach,targ} = PEEP + A * V(t) + B * V'(t) \tag{4b}$$

Eq. 4b represents volume and/or flow-proportional pressure support of the tracheal pressure (ATC with VPPS and/or FPPS with the factors of proportionality A and B resp.). Eq. 4b is only effective during inspiration; during expiration it was considered better to keep $P_{trach,targ}$ constant at PEEP; i.e. to apply Eq. 4a.

Also when generating U_{err} and I_{exp} the external control unit uses different algorithms for inspiration and expiration. During inspiration U_{err} is kept proportional to the deviation of $P_{trach,calc}$ from $P_{trach,targ}$

$$U_{err} \sim P_{trach,calc} - P_{trach,targ}. \qquad (5)$$

I_{exp} is chosen so that P1 remains equal to the sum of $P_{trach,targ}$ and ΔP_{ETT}

$$I_{exp} \sim P_{trach,targ} + \Delta P_{ETT} \qquad (6)$$

During expiration the additional resistance of the expiratory branch (ΔP_{eb}, tubing and valve) has to be taken into account also [1]:

$$I_{exp} \sim P_{trach,targ} + \Delta P_{ETT} + \Delta P_{eb} \qquad (7)$$

ΔP_{ETT} and ΔP_{eb} are calculated according to Eq. 2, U_{err} is chosen so that the ventilator will not deliver gas flow.

With the unmodified ventilator the pressure at the expiratory valve can be lowered only to atmospheric pressure. With small PEEP-values and high gas flows, the resistance of the ETT can therefore not be fully compensated for during expiration. To overcome this restriction we connected a blower (MFO 1000, Meidinger, Allschwil/Switzerland) to the expiratory gas exit. This blower maintains a subatmospheric pressure of −14 mbar, independent of gas flow. The reference input of the PEEP-valve is connected to this negative pressure source. This reference pressure (P3) is measured in order to correct I_{exp} so that P1 is independent of changes of the reference pressure.

The modified ventilator can be used not only for automatic tube compensation with optional volume and/or flow proportional inspiratory pressure assist (modified mode), but also for all forms of conventional mechanical and pressure support ventilation (unmodified mode). The relays S1 and S2 and the valve V1 enables one mode to be switched to the other at any time.

2.2. Measuring technique

V′ and P_{aw} were measured at the outer end of the ETT. Gas flow was measured with a Fleisch No. 2 pneumotachograph (Metabo, Epalinges/Switzerland) connected to a differential pressure transducer CPS1 (Hoffrichter, Schwerin/Germany) via two silicone tubes (20 cm long, 4 mm ID). The measuring tubes between the pneumotachograph and the differential pressure transducer needed to be very short for the regulation of the tracheal pressure, because the dynamic error in the flow signal had to be kept to a minimum, even during rapid changes in airway pressure. Airway and tracheal pressure were measured with a differential pressure transducer 1210A (ICSensors, Milpitas/USA). The transducers were connected to the measuring sites by polyethylene tubes (180 cm long, 1.5 mm ID). In the artificial trachea, tracheal pressure was measured in a ring channel that was situated 6 cm below the tip of the ETT. When measuring the tracheal pressure of patients, a small catheter which is closed at the end and with two sideholes (K-31 Baxter, Trieste/Italy), was introduced through the ETT so that its tip was situated 3 cm below the tip of the ETT. Tracheal pressure was only measured to assess the accuracy of the regulating system, however in regulation itself we only used the calculated tracheal pressure. The analog flow and pressure signals were sampled at a rate of 500 Hz, digitized (12 bit) and stored at a rate of 100 Hz (for patient measurements) or at a rate of 250 Hz (for laboratory measurements). The external control unit calculated the tracheal pressure and the parameters U_{err} and I_{exp} at a rate of 500 Hz. The ventilator registered incoming data at a rate of 125 Hz. For patient measurements the pneumotachograph was heated, and gas flow was corrected for viscosity under the assumption that inspiratory and expiratory air was 100% H_2O-saturated.

3. Investigation

3.1. The laboratory lung model

The modified ventilator was tested with the humidifier and original tubing on an active mechanical lung model (LS4000, Dräger, Lübeck/Germany). The model can simulate a variable compliance and resistance for any chosen flow pattern (for details see [9]). The flow pattern was generated by a wave form generator. The lung model was connected to the ETT by an artificial trachea (21 mm ID) that contained multiple side-holes and a ring channel for measurement of tracheal pressure (as described above, for details see [6]).

The model trachea was "intubated" with an ETT of 8 mm ID and 32 cm length. The ETT was bent to anatomical shape and connected to the ventilator tubes by a swivel connector. For our investigations we simulated a resistance of 5 mbar * s/l and a compliance of 15 ml/mbar. The wave form generator produced a sinusoidal signal at a rate of 30/min the amplitude of which was chosen so as to result in a tidal volume of 750 ml.

3.2. Measurements in patients

We tested the modified ventilator on 10 spontaneously breathing intubated patients after open-heart surgery. The clinical situation of the patients is listed in Table 1.

The study was approved by the Hospital Ethics committee. After controlled mechanical ventilation of a variable duration (average 11 ± 6 hours [1]) all patients were ventilated with synchronized intermittent mandatory ventilation (SIMV). Once spontaneous breathing was clearly recognizable we switched to patient-triggered inspiratory pressure support (IPS) (20 ± 8 hours after intubation). Following observation of the patient's ventilatory pattern in the IPS-mode, we switched to the modified mode and applied a volume proportional pressure support of tracheal pressure of 10 mbar/l (VPPS with ATC). Within the first few breaths VPPS was adjusted so that the tidal volume returned to the value under ventilation with IPS in the unmodified mode. In patient No.3 VPPS needed to be increased to 15 mbar/l, in patients No. 2 and No. 10 it was reduced to 8 mbar/l, patient No. 9 did not need any pressure support at all and was able to breathe with ATC immediately. After recording flow and pressure continuously for 3–10 min VPPS was gradually reduced while observing respiratory frequency, tidal volume, T_I/T_{tot}-ratio and end-tidal CO_2. When a VPPS of zero was reached (1.0 ± 0.7 hours after begin of VPPS) continuous recordings of flow and pressure were again taken for 3–10 min. No patient needed VPPS again. After 1.1 ± 0.7 hours of spontaneous breathing with ATC without VPPS extubation was performed. The duration between switching to the modified mode (i.e. the commencement of ATC together with VPPS) and extubation took on average 2.2 ± 1.2 hours.

[1] Mean ± SD.

Table 1
Clinical situation of the patients

No.	Age (yr)	Sex	ETT [1] ID, length	Intubation duration [2] (hr)
1	53	f	7.5, 24	13
2	74	m	8.5, 26	33
3	66	m	8.5, 26	20
4	60	f	7.5, 24	14
5	66	f	7.5, 24	21
6	73	f	7.5, 24	33
7	69	m	8.5, 26	16
8	65	m	8.5, 26	14
9	65	m	8.5, 27	12
10	49	f	7.0, 22	22
mean				

[1] Mallinckrodt 107 intermediate tubes ID in mm, length in cm. [2] Total number of hours of intubation prior to commencement of investigation.

4. Results

Fig. 5 shows the pressure/volume loops of P_{aw} and P_{trach} obtained with the physical lung model which was connected to the model trachea and an ETT. P_{trach} was measured in the ring channel of the model trachea. The ventilator was operating under automatic tube compensation (ATC) mode without any additional pressure support. $P_{trach,targ}$ was set at 5 mbar over the entire breath. The rms-deviation of P_{trach} from $P_{trach,targ}$ was 1.3 mbar. The maximum deviation was 2.5 mbar.

Table 2 shows the results of 10 patients breathing spontaneously with ATC and no additional pressure support. Clinical conditions are listed in Table 1.

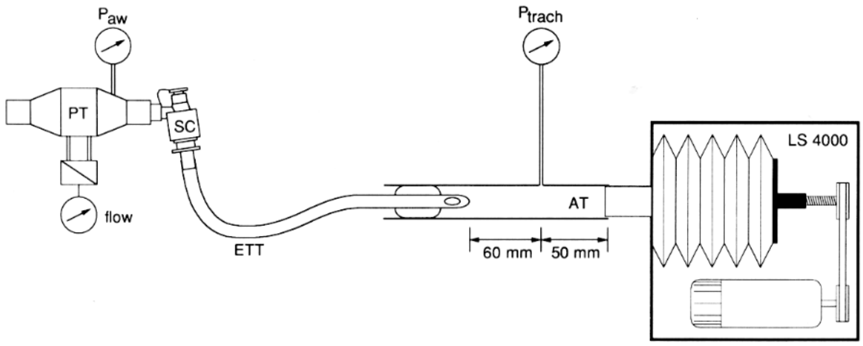

Fig. 4. Measuring configuration in the laboratory lung model: The lung model (LS4000) produces a sinusoidal flow at a resistance of 5 mbar*s/l and a compliance of 15 ml/mbar. Tracheal pressure can be measured in the model trachea (AT). Airway pressure is measured outside the ETT and the swivel connector (SC) in the pneumotachograph (PT).

Data are average values from consecutive data sequences of several minutes duration. Table 2 shows the rms-deviation of the measured P_{trach} from $P_{trach,targ}$.

Under ATC without any additional pressure support $P_{trach,targ}$ is constant and is equal to the chosen end-expiratory pressure (PEEP) during the entire breath. Due to a small, but unavoidable delay in the ventilator response (see Fig. 3), tracheal pressure falls below the target pressure in the early part of inspiration (rising flow), and exceeds the target pressure in the later part of inspiration (falling flow). In contrast, tracheal pressure exceeds target pressure in the early part of expiration (rising flow), and falls below the target pressure in the later part of expiration (falling flow). The negative target deviation arising in early inspiration exceeds that in late expiration, and the positive target deviation arising in early expiration exceeds that in late inspiration. The maximum negative target deviation arising in early inspiration was determined breath by breath, and an average of these deviations is shown in column "$\Delta P_{trach,min}$". The maximum positive deviation arising in early expiration was determined breath by breath, and an average of these deviations is shown in column "$\Delta P_{trach,max}$". Table 2 also shows the maximum deviations of P_{aw} from $P_{trach,targ}$ for inspiration and expiration ($\Delta P_{aw,min}$ and $\Delta P_{aw,max}$ resp.). The average maximum deviation of P_{aw} from $P_{trach,targ}$ of -5.5 mbar (expiration) and $+10.3$ mbar (inspiration) is equivalent to the maximum expiratory pressure relief and inspiratory pressure support respectively (i.e. the

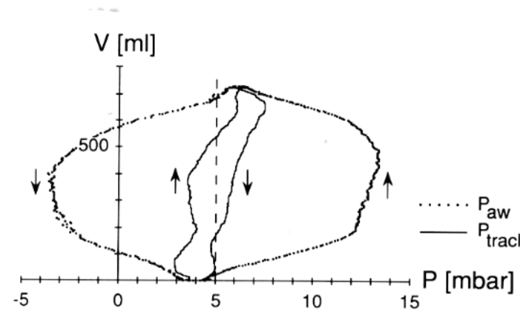

Fig. 5. Pressure/volume loops in the physical lung model connected to the model trachea and an ETT. The ventilator is operating under the mode ATC with a PEEP of 5 mbar (shown by the dashed line). Upward-pointing arrows signify inspiration, downward-pointing arrows signify expiration.

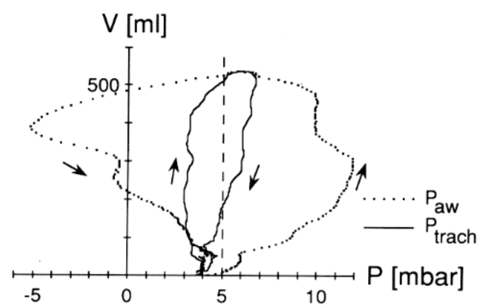

Fig. 6. Pressure/volume loops in patient No.1 breathing spontaneously under ATC with a PEEP of 5 mbar (shown by the dashed line). Under ATC without any pressure support the PEEP chosen corresponds to $P_{trach,targ}$. Upward-pointing arrows signify inspiration, downward-pointing arrows signify expiration.

Table 2
Results of 10 patients breathing spontaneously with ATC and no additional pressure support

No.	n [1] br	rms [2] [mbar]	ΔPtrach min [2,3] [mbar]	ΔPtrach max [2,4] [mbar]	ΔPaw min [2,4] [mbar]	ΔPaw max [2,3] [mbar]
1	72	1.1	−2.0	2.6	−8.1	6.8
2	232	0.9	−1.6	2.4	−3.3	5.3
3	58	0.9	−2.1	1.8	−2.6	7.6
4	122	1.0	−2.6	2.2	−4.9	10.2
5	46	1.4	−4.4	4.1	−7.7	22.4
6	71	1.2	−2.8	2.9	−5.9	8.2
7	35	0.8	−1.9	2.6	−5.1	5.5
8	64	0.9	−1.7	2.4	−2.9	3.1
9	43	1.2	−2.7	4.7	−7.3	13.0
10	66	1.6	−2.8	3.9	−7.4	20.6
mean		1.1	−2.5	2.9	−5.5	10.3

[1] Number of analyzed breaths. [2] Deviation from PEEP-level. [3] During inspiration. [4] During expiration.

maximum pressure applied for compensation of ETT resistance).

Fig. 6 shows the pressure/volume loops of an intubated patient breathing spontaneously under ATC without any additional pressure support. $P_{trach,targ}$ is equal to PEEP during the entire breath and was set at 5.0 mbar. P_{trach} was measured by a catheter introduced into the trachea through the ETT. The rms-deviation of P_{trach} from $P_{trach,targ}$ was 1.2 mbar. The maximum deviation was 2.0 mbar.

Table 3 summarizes the data of the 9 patients breathing spontaneously with automatic tube compensation (ATC) and an additional volume proportional pressure support of tracheal pressure (VPPS). When calculating the rms-deviation and the maximum deviations of P_{trach} from $P_{trach,targ}$ only the inspiratory samples were taken into account. This was done because $P_{trach,targ}$ rapidly drops at the end of inspiration from end-inspiratory pressure support to PEEP and naturally P_{trach} cannot follow this change equally rapidly. The rms-deviation of P_{trach} from $P_{trach,targ}$ was 1.7 mbar and is thus clearly larger than under ATC without VPPS due to quiet expiration not being taken into account. The maximum target deviations (between −2.6 mbar and +2.3 mbar in average) correspond to the ones under ATC without VPPS.

Fig. 7 shows the pressure/volume loops of P_{aw} and P_{trach} of an intubated patient breathing spon-

Table 3
Results of 9 patients breathing spontaneously with ATC and VPPS

Patient no.	VPPS mbar/l	n [1] br	rms [2] [mbar]	ΔPtrach min [2] [mbar]	ΔPtrach max [2] [mbar]	ΔPaw min [3] [mbar]	ΔPaw max [2] [mbar]
1	10	52	1.3	−2.1	1.6	−10.1	7.0
2	8	103	1.2	−1.5	2.7	−6.2	6.0
3	15	59	1.7	−2.5	2.9	−4.8	6.3
4	10	125	2.1	−3.5	1.5	−8.0	7.5
5	10	24	2.6	−4.1	4.5	−13.1	18.5
6	10	154	2.3	−3.5	1.6	−3.3	4.9
7	10	31	1.6	−2.3	0.9	−7.0	9.3
8	10	47	0.9	−1.4	1.2	−5.1	2.4
10	8	51	1.8	−2.4	3.8	−10.7	24.1
		mean	1.7	−2.6	2.3	−7.6	9.6

[1] Number of analyzed breaths. [2] Deviation from Ptrach,targ during inspiration. [3] Deviation from Ptrach,targ during expiration.

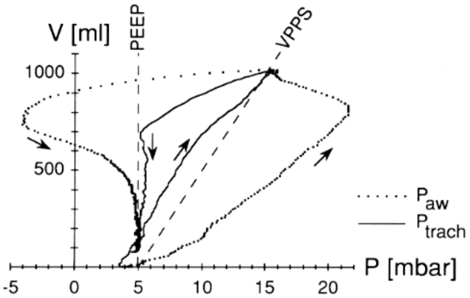

Fig. 7. Pressure/volume loops of P_{aw} and P_{trach} in patient No.7 breathing spontaneously under ATC with an inspiratory volume proportional pressure support (VPPS) of 10 mbar/l and a PEEP of 5 mbar. The dashed lines show $P_{trach,targ}$ during inspiration (labelled VPPS) and expiration (labelled PEEP). Upward-pointing arrows indicate inspiration, downward-pointing arrows indicate expiration.

taneously under ATC with a VPPS of 10 mbar/l and PEEP of 5 mbar. The oblique dashed line shows $P_{trach,targ}$ during inspiration. During expiration $P_{trach,targ}$ was equal to PEEP (vertical dashed line). The inspiratory rms-deviation of measured P_{trach} from $P_{trach,targ}$ was 1.4 mbar. The maximum target deviation was 2.1 mbar.

5. Discussion

Conventional demand-flow ventilators control the airway pressure (P_{aw}) to a desired target value independent of the patient's ventilatory efforts: In the CPAP-mode P_{aw} is kept constant; in the IPS-mode the ventilator produces a constant support of P_{aw} during inspiration, while during expiration P_{aw} is equal to the PEEP-value – various trigger criteria have been used to enable the machine to recognize the patient's inspiratory efforts and the end of inspiration and thus switch from inspiratory to expiratory P_{aw} and vice versa. The uniform and constant inspiratory pressure support and the inadequacies in the trigger mechanism disturb the ventilatory pattern of the spontaneously breathing patient and can lead to desynchronisation between the patient's ventilatory efforts and the pressure support produced by the ventilator and can lead to a sensation of dyspnea and respiratory discomfort [15]. The resistance of the ETT, expiratory tubing and expiratory valve heavily restrict expiratory air flow and can lead to dynamic hyperinflation. Various attempts have been made to develop forms of inspiratory pressure support that should automatically adapt to the patient's breathing effort. Younes et al. [14], Poon and Ward [8] and Schaller [10,11] proposed the continuous control of inspiratory P_{aw} in proportion to inspiratory flow (called "negative resistance", "negative impedance ventilation (NIV)", "flow proportional pressure support (FPPS)" or "resistive assist") and/or in proportion to inspired volume (called "proportional assist ventilation (PAV)", "volume proportional pressure support (VPPS)", "negative compliance" or "elastic assist") and developed various technical solutions for these techniques.

However, these techniques do not take into account the fact that the total resistance of the respiratory system (resistance of airways, ETT and ventilator tubing) is nonlinear and strongly flow-dependent in intubated patients. Furthermore, the resistive assist of Younes and Poon/Ward is only effective during inspiration [8,16]. Our proposed solution for a spontaneously breathing intubated patient is to compensate for the ETT resistance during inspiration as well as during expiration, i.e. have the patient breathe as if he were not intubated and to offer additional inspiratory pressure assistance when necessary. Consequently the ventilator should not control airway pressure, but tracheal pressure. In order to do this we modified a demand-flow ventilator (EVITA). This ventilator has the advantage of having not only very fast inspiratory and expiratory valves, but also that all regulating parameters within the ventilator are available as electrical signals. This enables us to easily feed modified parameters from an external control unit into the feedback system of the ventilator. The ventilator, when so modified, can be switched back to its original working mode at any time.

The simplest mode of controlling tracheal pressure is to control it to a constant value. We call this "automatic tube compensation" (ATC). If ATC is used with a target tracheal pressure of zero then the patient is "breathing fresh air", i.e. he is breathing at atmospheric pressure as though he were not intubated (under the assumption that the resistance of the vocal cords is negligible). An increase in inspiratory pressure support gives the patient relief from the ventilatory work that he spends not on the ETT but on his own respiratory system. Any pattern of pressure support ($P_{trach,targ}$) can be chosen in the external control unit in order to achieve this. In the present study we chose to provide pressure support only in proportion to inhaled volume (VPPS), but $P_{trach,targ}$ can be just as simply rendered flow proportional (FPPS) or a combination of the two, depending on the factors A and B in Eq. 4b.

One of the major difficulties in regulating P_{trach}, is in its determination. Our experience is that long-term measurement of P_{trach} via a catheter introduced through the ETT is complicated and prone to disturbances. We therefore developed a method to calculate P_{trach} continuously from measurements of P_{aw} and V' according to Eq. 3 under the assumption that the ETT is not obstructed (see [6] for details). This method for calculating P_{trach} has proved reliable and easy to perform even under "rough" clinical conditions.

We used a laboratory lung model to optimize our algorithms in the control of P_{trach}, so that deviations of P_{trach} from $P_{trach,\ targ}$ could be corrected promptly and without large oscillations. The physical lung model simulated active ventilation and allowed the setting of the most important passive mechanical properties of a ventilatory system (resistance, compliance and the ETT resistance). To evaluate the limits of our system and to be prepared for potential disturbances during patient measurements that could not be simulated by our model (e.g. flow oscillations due to secretions in the ETT) we simulated extreme situations such as very small compliance, a very high respiratory ratio and minute ventilation. At a compliance of 15 ml/mbar, sinusoidal flows of up to 1.2 l/s, a respiratory frequency of 30/min and a minute ventilation of 22.5 l/min the maximum deviation of measured P_{trach} from $P_{trach,targ}$ was smaller than 2.5 mbar.

Until present we have tested our modified ventilator on 10 patients ventilated post-operatively. The rms-deviation of P_{trach} from $P_{trach,targ}$ was 1.1 mbar under ATC (entire breath) and 1.7 mbar under ATC plus VPPS (only inspiration considered); therefore regulation of P_{trach} with our system is actually of the same accuracy as the regulation of P_{aw} with a traditional demand-flow ventilator in CPAP-mode. We found large deviations of P_{trach} from $P_{trach,targ}$ only during forced expiration or forced inspiration, when the airway pressure necessary for ETT-compensation exceeded the limits of -14 mbar or $+40$ mbar that our ventilator can provide or when the inspiratory flow demanded by the patient exceeded the maximum flow of 2 l/s.

For ATC without any additional pressure support (P_{trach} remains constant during the whole breath) our patients needed inspiratory airway pressures of between 3 and 24 mbar in excess of $P_{trach,targ}$ and expiratory airway pressure of between 3 and 13 mbar below $P_{trach,targ}$. The pressure support that we offered the patient to compensate for ETT resistance can therefore be easily larger than the inspiratory pressure support usually applied.

One advantage of ATC lies in the capability of being able to simulate the mechanical properties of the non-intubated ventilatory system before extubation is actually performed. This is particularly useful in patients who required long-term ventilation whereby the decision whether to extubate was thus facilitated and the various, not altogether reliable weaning-indices [13] could be complemented with this "electronic extubation".

To summarize, we have shown that it is possible to compensate for the ETT resistance by regulating tracheal pressure without direct measurement of tracheal pressure, using a modern electronic demand-flow ventilator and an external control unit.

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