Intensive Care Med (1990) 16:189–195

Intensive Care Medicine

Measurement of effective elastance of the total respiratory system in ventilated patients by a computed method

Comparison with the static method

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Received: 21 July 1988; accepted: 19 June 1989

Abstract. We have studied 28 patients mechanically ventilated for acute respiratory failure at different levels of externally applied positive end-expiratory pressure ($PEEP_e$). We describe and compare a computed method of measuring "effective" elastance of the total respiratory system ($E_{rs,eff}$) with the static values of elastance of the total respiratory system ($E_{rs,st}$), obtained with the end-inflation occlusion technique. $E_{rs,eff}$ was computed by an original device (Heres, R.P.A., Belgium), also the effective resistance of the total respiratory system was calculated. At zero end-expiratory pressure set by the ventilator (ZEEP). $E_{rs,eff}$ averaged 29.5 ± 13.5 cm $H_2O \cdot L^{-1}$ while $E_{rs,st}$ non-corrected for intrinsic PEEP ($PEEP_i$) averaged 36.4 ± 15.1 cm $H_2O \cdot L^{-1}$ and $E_{rs,st}$ corrected for $PEEP_i$ averaged 28.2 ± 13.4 cm $H_2O \cdot L^{-1}$. The small difference between $_{Ers,eff}$ and $_{Ers,st}$ corrected for $PEEP_i$ was statistically significant and these two values were highly correlated ($r = 0.98$). This significant difference disappeared rapidly with $PEEP_e$ and probably reflects a frequency-dependance due to pendelluft. We also observed that $PEEP_i$ was present in 21 of 27 patients at ZEEP. Our results also indicate that low levels of PEEP may improve E_{rs} in hyperinflated COPD patients, without inducing further hyperinflation. In conclusion, values of $E_{rs,eff}$ are very similar to static values corrected for $PEEP_i$ and permit an accurate and rapid approach to the management of ventilated patients.

Key words: Total respiratory system mechanics – Intrinsic PEEP – Mechanical ventilation

Despite the great number of patients requiring prolonged mechanical ventilation for acute respiratory failure, measurement of respiratory mechanics is not common practice in the intensive care unit. A reason for this lack of data is the absence of a simple, non-invasive and precise technique for the rapid quantitative determination of respiratory system mechanics in ventilated patients. Several different approaches have been proposed [1–7].

We have studied a computed method of measuring the "effective" elastance of the total respiratory system ($E_{rs,eff}$), which is the reciprocal of the "effective" compliance of the total respiratory system ($C_{rs,eff}$) in mechanically ventilated patients with acute respiratory failure. This computed method has been compared to the conventional static values obtained with the end-inflation occlusion technique [10]. This last method is the most commonly used in the intensive care unit and it provides the overall static elastance of the respiratory system [7]. When using the end-inflation occlusion method, allowance must be made for intrinsic positive end-expiratory pressure ($PEEP_i$) [10]. Jonson et al. [8] together with Pepe and Marini [9] have demonstrated the presence of $PEEP_i$ (or auto-PEEP) in some mechanically ventilated patients. Recently, Rossi et al. [10] have also demonstrated the importance and frequency of $PEEP_i$ in ventilated patients, which might be responsible for an important underestimation of $C_{rs,st}$ when $PEEP_i$ is not taken into account.

The aim of this study was to compare this computed method with the static method, corrected for $PEEP_i$ which offers a valid estimation of E_{rs} and is relatively easy to obtain when a modern ventilator, equipped with automatic end-inspiratory and end-expiratory facilities, is used. We have also studied the values obtained by these two methods when different levels of PEEP were applied by the ventilator ($PEEP_e$), to compare their relative evolution and to assess the mechanical effects of $PEEP_e$ in patients who are frequently characterized by dynamic hyperinflation.

Methods

Twenty-eight consecutive patients with acute respiratory failure of differing etiologies (Table 1) requiring mechanical ventilation were studied in the intensive care units of 2 hospitals (Hôpital Civil de Jumet and Clinique Louis Caty, Baudour, Belgium).

The patients were intubated transorally or transnasally with cuffed endotracheal tubes ranging from 7.5 to 9.5 mm internal diameter, except for one patient (patient 22) who had a cuffed tracheostomy tube. Vol-

Table 1. Patient's characteristics

Patient	Sex	Age	Baseline F_iO_2	VT (L)	$R_{rs,eff}$ $cmH_2O/L/s$	Diagnosis
1	F	75	40	0.48	14	Laparotomy, aspiration
2	F	75	36	0.52	12	Haemodynamic pulmonary oedema
3	F	40	40	0.63	46	Status asthmaticus
4	M	79	30	0.40	27	Silicosis, COPD
5	M	76	30	0.81	21	Silicosis, COPD
6	F	46	40	0.51	12	Septicaemia, ARDS
7	M	61	30	0.98	14	COPD
8	M	75	30	0.45	22	Abdominal surgery, COPD
9	M	58	40	0.58	25	COPD
10	F	47	50	0.84	8	ARDS
11	M	65	30	0.55	19	COPD
12	M	79	37	0.56	19	COPD, pulmonary embolism
13	M	65	30	0.62	25	COPD
14	F	72	40	0.31	16	Laparotomy, pneumonia
15	M	59	40	0.48	17	Silicosis, COPD, pneumonia
16	F	40	50	0.40	12	ARDS, pancreatitis
17	M	71	35	0.64	24	COPD
18	M	71	30	0.69	12	Left ventricular failure
19	M	66	30	0.57	6	COPD, left ventricular failure, pneumothorax
20	M	55	40	0.45	16	Silicosis, COPD, pneumonia
21	M	58	40	0.49	33	Silicosis, COPD
22	M	68	45	0.62	16	Bullectomy, COPD
23	M	83	40	0.48	48	Kyphoscoliosis, obstructed endotracheal tube
24	M	66	40	0.53	20	COPD, left ventricular failure
25	M	78	35	0.64	12	Silicosis, COPD, thoracic trauma
26	M	77	30	0.55	13	Laparotomy, diverticulitis, septic shock
27	M	80	30	0.78	17	Silicosis, COPD, left ventricular failure
28	M	67	30	0.50	35	Silicosis, COPD

Definitions of abbreviations. F_iO_2 = fractional inspired oxygen concentration; COPD = chronic obstructive pulmonary disease; ARDS = adult respiratory distress syndrome; VT = tidal volume; $R_{rs,eff}$ = effective resistance of the total respiratory system

ume-controlled ventilators (Siemens 900 C) were used for the study. The patients were clinically stable during the hours preceding the study. Tidal volume was set at a value prescribed by the primary physician. Respiratory rate ranged between 14 and 17 min-1. Inspiratory time and automatic end-inspiratory pause were set at 25% and 10% of the respiratory cycle respectively and these ventilatory parameters were kept constant, except during the occlusion manoeuvres (see below).

Extrinsic PEEP ($PEEP_e$) was set using the ventilator and used in twenty patients. The patients were studied in a supine or semi-recumbent position. Flow ($\dot{V}$) was measured with a heated Lilly pneumotachograph, connected to the endotracheal tube and to a differential pressure transducer (Validyne M.P. 45). Volume (V) was obtained by numerical integration of the flow signal (Heres, R.P.A., Belgium). Pressure at the airway opening (P_{ao}) was measured proximal to the pneumotachograph using a pressure transducer (Bentley, Trantec M 800). The occlusions were realised distally to the airway opening with the end-expiratory and end-inspiratory occlusion facilities of the ventilator. Computation of the effective respiratory parameters was made by a calculator (Heres) and based on an identification method. If the simplified form of the equation of respiratory motion is:

$$P(t) = R(t)\ \dot{V}(t) + E(t)\ V(t)$$

where P is the driving pressure which produces flow ($\dot{V}$) and volume displacement (V) at any instant t, the calculator computes, by analogy with an electrical system, an effective resistance (R_{eff}) and elastance (E_{eff}) based on an entire respiratory cycle given by the following formula:

$$R_{eff} = \frac{\int_{t_0}^{t_0+T} P(t)\ \dot{V}(t)\ dt}{\int_{t_0}^{t_0+T} \dot{V}^2(t)\ dt}$$

and

$$E_{eff} = \frac{\int_{t_0}^{t_0+T} P(t)\ V(t)\ dt}{\int_{t_0}^{t_0+T} V^2(t)\ dt}$$

where T is the respiratory period (see appendix).

Under controlled ventilation, the pressure signals are characterized by a sudden drop, during expiration, to atmospheric level (or to the selected level of PEEP), impeding precise computation of mechanical parameters during the expiratory period. Therefore, the computations are essentially made during the inspiratory period.

All signals were visualized on the screen of the calculator and recorded either with an eight-channel pen recorder (Gould, Brush) at a speed of 25 or 50 mm S^{-1}, or with a video-printer (Seikosha) at a speed of 50 mm S^{-1}.

Procedure and data analysis

Calibration of the pneumotachograph was carried out with the gas used for the ventilation of the patient.

After inserting the pneumotachograph into the circuit, mechanical ventilation was resumed on control mode with the initial ventilator settings. The relaxation of the patients was detected by the on-line inspection of $\dot{V}$, V and P_{ao} records [5, 7]. At this point, the different ventilatory variables and the effective mechanics parameters were recorded. These different values were averaged on three cycles and recordings were made only after satisfactory reproducibility was observed (less than 5% of breath-by-breath variability), reflecting perfect relaxation of the patient. An end-expiratory occlusion was performed by means of the end-expiratory hold button and permitted visualisation of an eventual end-expiratory "plateau" pressure, corresponding to auto-PEEP or intrinsic PEEP ($PEEP_i$), as previously described [9, 10] (Fig. 1). $PEEP_i$ represents the elastic recoil pressure of the total respiratory system ($P_{el,rs}$) at the end of the deflation and is due to the presence of expiratory flow limitation and failure to reach the elastic equilibrium point of the respiratory system during the preceding expiration. When $PEEP_e$ was applied, $PEEP_i$ was computed by subtracting $PEEP_e$ from the end-expiratory plateau pressure (Fig. 1). The $PEEP_i$ is thus defined as the difference between end-expiratory $P_{el,rs}$ and $PEEP_e$. These measurements were repeated at least three times in all patients and then averaged. Variations were sometimes transiently observed and probably reflected the effects of muscular participation. Following this, the airway opening was occluded for approximately two seconds, at the end of a mechanical inflation, using the end-inspiratory hold button of the ventilator, to measure the end-inspiratory plateau pressure. This end-inspiratory plateau pressure reflects the elastic recoil pressure of the total respiratory system at the end of the mechanical inflation. A true plateau was consistently demonstrated during relaxed expiration but variations, due to imperfect muscle relaxation, were sometimes observed. Hence, this end-inspiratory occlusion was maintained until a true plateau was observed in order to eliminate the components due to the stress relaxation properties of the total respiratory system and to confirm the perfect relaxation of the subject.

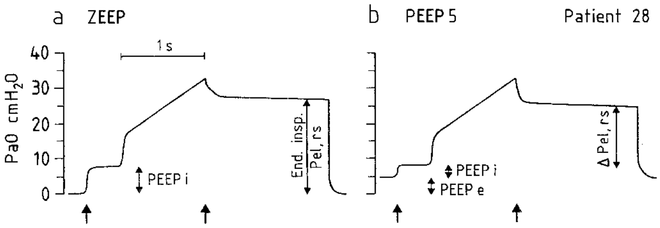

Fig. 1. **a** Record of pressure at airway opening (P_{ao}) in a mechanically ventilated COPD patient (N° 28). Airway occlusion at end-expiration was performed at the point indicated by the first *vertical arrow*. Following this occlusion, there is an end-expiratory "plateau" pressure reflecting $PEEP_i$. In this patient, $PEEP_i$ amounted to 8 cm H_2O. This first occlusion is released and at the end of the inflation, a second airway occlusion is carried out at the point indicated by the second *vertical arrow* and an end-inspiratory plateau pressure is obtained, reflecting the elastic recoil pressure of the total respiratory system ($P_{el,rs}$). **b** A record of the same patient with externally applied PEEP ($PEEP_e$) of 5 cm H_2O; the ventilator settings being unchanged. The end-expiratory plateau pressure shows a $PEEP_i$ of 3 cm H_2O and there is a slight decrease of $P_{el,rs}$ indicating an improvement of the elastance of the total respiratory system. Corrected static respiratory elastance is obtained by dividing $P_{el,rs}$ by the inspiratory tidal volume. For further explanation, see text

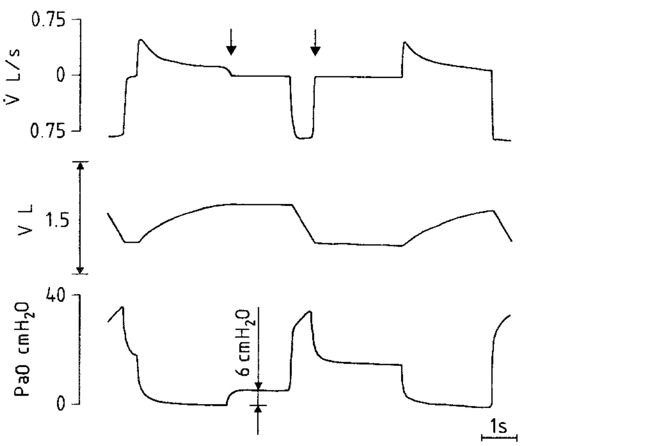

Fig. 2. Tracing of flow ($\dot{V}$), volume (V) and pressure at airway opening (P_{ao}) in a mechanically ventilated patient with COPD (patient 13) at ZEEP. Expiratory flow is present throughout the expiratory cycle. An end-expiratory occlusion is carried out and an end-expiratory plateau pressure of 6 cm H_2O appears, corresponding to the intrinsic PEEP (*first arrow*). Immediately thereafter, an end-inspiratory occlusion is performed and an end-inspiratory plateau pressure is obtained (*second arrow*). For further explanations, see text

Static respiratory elastance ($E_{rs,st}$) was obtained by dividing the difference between end-inspiratory $P_{el,rs}$ and the eventual $PEEP_e$ (Fig. 1) by the tidal volume (V_T), as conventionally described [1, 10]. In this connection, it should be noted that under ZEEP (Zero end-expiratory pressure set on the ventilator) the end-expiratory pressure did not always return to zero and often ranged at around 1 cm H_2O [10]. These elastance data are labelled non-corrected $E_{rs,st}$. In order to obtain a valid measurement of $E_{rs,st}$, $PEEP_i$ must also be subtracted from the end-inspiratory $P_{el,rs}$ and in this instance, $E_{rs,st}$ is measured by the difference between end-inspiratory $P_{el,rs}$ minus end-expiratory $P_{el,rs}$; this data is labelled corrected $E_{rs,st}$.

Three perfectly relaxed cycles were analysed and averaged out (Fig. 2). All these measurements were made initially at ZEEP (except in patient 18 who was studied initially at 5 cm H_2O $PEEP_e$). Thereafter, measurements were also made when $PEEP_e$ was applied at 5 cm H_2O, in twenty patients, 10 cm H_2O in twenty-three patients and 15 cm H_2O in ten patients. Other ventilatory parameters were kept constant during these measurements. Again, occlusions at end-expiration and at end-inspiration were carried out at each level of $PEEP_e$ and three cycles were analysed for the calculation of $PEEP_i$ and $P_{el,rs}$. Effective elastance was continuously computed and averaged over three respiratory cycles. Statistical analysis was performed using the *t*-test for paired data and regression analysis was carried out with the "least squares" method.

Results

Individual results of $R_{rs,eff}$ calculated at ZEEP are listed in Table 1. These values are the average of three cycles. The resistance of the endotracheal tube plus equipment was not substracted. Individual averaged values of corrected and uncorrected $E_{rs,st}$ and of $E_{rs,eff}$ at $PEEP_e$ values of 0, 5, 10 and 15 cm H_2O are listed in Table 2 and summarized in Fig. 3 at ZEEP and 10 cm H_2O PEEP. Patient 18 was studied only at a $PEEP_e$ of 5 and 10 cm H_2O. Also listed in Table 3 are the values of $PEEP_i$ (end-expiratory occlusion method) which were

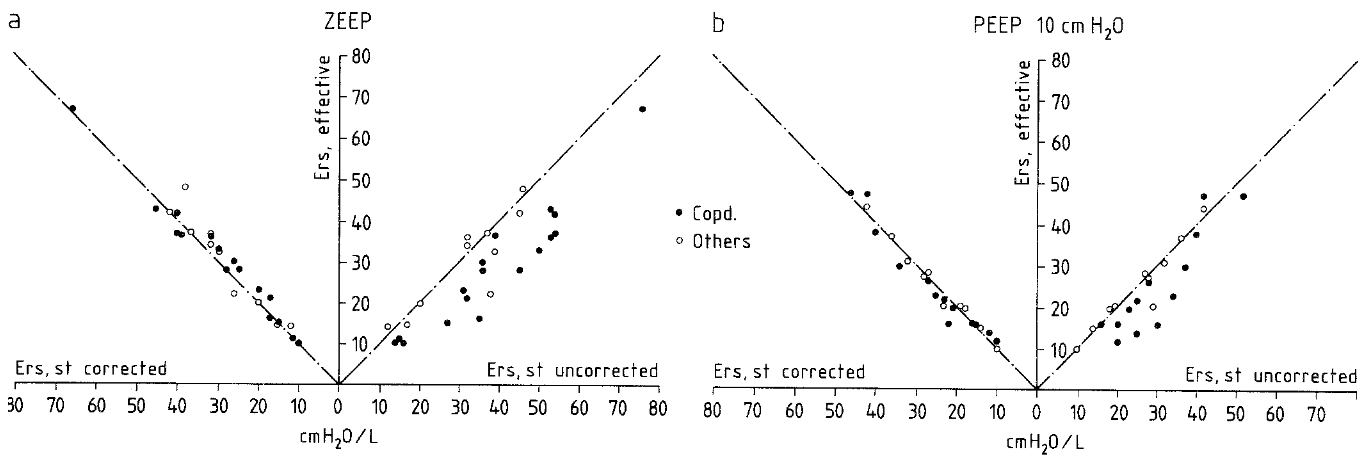

Fig. 3. Comparison between effective elastance of the total respiratory system ($E_{rs,eff}$) and static elastance of the total respiratory system ($E_{rs,st}$), corrected and uncorrected for $PEEP_i$. *Filled circles* are mean values of COPD patients. *Empty circles* are mean values for patients ventilated for ARDS or other aetiologies. *Interrupted lines* are lines of identity. *Left panel*: measurements with a zero end-expiratory pressure (ZEEP) applied by the ventilator. *Right panel*: measurements with a positive end-expiratory pressure of approximately 10 cm H_2O (PEEP = 10) applied externally

Table 2. Values of effective E_{rs} and static E_{rs} at ZEEP, $PEEP_e$5, $PEEP_e$ 10 and $PEEP_e$15 uncorrected and corrected for $PEEP_i$

	ZEEP			$PEEP_e$5 cmH_2O			$PEEP_e$10 cmH_2O			$PEEP_e$15 cmH_2O		
	$E_{rs,eff}$ cmH_2O/L	$E_{rs,st}$ uncorr. cmH_2O/L	$E_{rs,st}$ corr. cmH_2O/L	$E_{rs,eff}$ cmH_2O/L	$E_{rs,st}$ uncorr. cmH_2O/L	$E_{rs,st}$ corr. cmH_2O/L	$E_{rs,eff}$ cmH_2O/L	$E_{rs,st}$ uncorr. cmH_2O/L	$E_{rs,st}$ corr. cmH_2O/L	$E_{rs,eff}$ cmH_2O/L	$E_{rs,st}$ uncorr. cmH_2O/L	$E_{rs,st}$ corr. cmH_2O/L
1	32.5	39	30	30	34	29	28.4	27	27	31	30	30
2	37	37	37	33	32	32	31	32	32	–	–	–
3	36	53	32	–	–	–	–	–	–	–	–	–
4	28	45	25	20	37	25	14	25	12	16	19	15
5	15	27	15	15	24	15	12	20	10	11.5	19	9
6	36	32	32	–	–	–	27.4	28	28	31	30	30
7	10.2	16	10	–	–	–	–	–	–	–	–	–
8	16	35	17	–	–	–	15.6	30	22	–	–	–
9	23	31	20	21	25	21	22	25	23	25	26	26
10	20	20	20	19	19	19	20.3	19	19	25	25	25
11	33	50	30	25	35	26	23	34	25	22	28	21
12	36.5	39	39	–	–	–	38	40	40	–	–	–
13	28	36	28	26	33	26.5	26.3	28	27	–	–	–
14	48	46	46	–	–	–	37	36	36	42	42	42
15	43	53	45	40	44	40	–	–	–	–	–	–
16	42	45	42	42	45	45	44	42	42	52	51	51
17	21	32	17	18	29	21	16	20	15	20	23	20
18	–	–	–	10.4	10	10	10	10	10	–	–	–
19	14	12	12	–	–	–	15	14	14	–	–	–
20	67	70	70	43	44	44	47	42	42	–	–	–
21	42	54	40	49	56	45	47	52	46	–	–	–
22	11	15	11	–	–	–	–	–	–	–	–	–
23	22	38	26	18	24	19	20.5	29	23	–	–	–
24	30	36	26	23	31	26	20	23	21	–	–	–
25	10.3	14	10	11.4	14	12	16	16	16	–	–	–
26	14.4	17	15	18	15	15	20	18	18	–	–	–
27	34	32	32	38	38	38	–	–	–	–	–	–
28	37	54	40	32	40	31	30	37	34	–	–	–
Means	29.2	36.4	28.2	26.7	31.4	26.9	25.5	28.1	25.3	27.45	28.8	26.4
+ SD	13.5	15.1	13.4	11.0	11.7	10.9	11.0	10.2	10.7	12.1	10.3	12.4

Definitions of abbreviations. E_{rs} = elastance of the total respiratory system; ZEEP = zero end-expiratory pressure applied by the ventilator; $PEEP_e$ = externally applied positive end-expiratory pressure; $E_{rs,eff}$ = effective E_{rs}; $E_{rs,st,uncorr}$ = uncorrected for $PEEP_i$ static E_{rs}; $E_{rs,st,corr}$ = corrected for $PEEP_i$ static E_{rs}; Values are means of three measurements. – = not measured

Table 3. Individual values of intrinsic PEEP obtained with the end-expiratory occlusion method at ZEEP and at 5, 10 and 15 cmH_2O levels of externally applied PEEP

Patient	$PEEP_i$			
	ZEEP	$PEEP_e$		
	cmH_2O	5 cmH_2O	10 cmH_2O	15 cmH_2O
1	7.5	3	0	–
2	9	–	–	–
3	17	–	–	–
4	8	5	3	1.5
5	13	–	–	–
6	0	–	–	–
7	10	–	–	–
8	9	–	4	–
9	9	2.5	1	–
10	0	–	–	–
11	9	4	4	2.5
12	0	–	–	–
13	5	4	1	0
14	6.5	0	0	–
15	4	2	–	–
16	1	0	0	0
17	11	4	2.5	2.5
18	–	0	0	–
19	0	0	0	–
20	5	0	0	–
21	8	6.5	2.5	–
22	1	–	–	–
23	5	2.5	2.5	–
24	6.5	2.5	1	–
25	2.5	1	0	–
26	1.5	0	0	–
27	0	0	–	–
28	8	4	1	–

Definitions of abbreviations. ZEEP = zero end-expiratory pressure; $PEEP_e$ = externally applied positive end-expiratory pressure; $PEEP_i$ = intrinsic positive end-expiratory pressure

present in 21 of 27 patients at ZEEP. At ZEEP, the uncorrected values of $E_{rs,st}$ exceeded the corrected values in 21 subjects who demonstrated $PEEP_i$ and this difference was statistically significant ($p<0.001$). The relationship between uncorrected values of $E_{rs,st}$ and $E_{rs,eff}$ at ZEEP is summarized in Fig. 3. It can be seen that the uncorrected values of $E_{rs,st}$ often exceeded the values of $E_{rs,eff}$, particularly with the COPD patients. This difference was also statistically significant ($p<0.001$). The relationship between the corrected values of $E_{rs,st}$ and the values of $E_{rs,eff}$ at ZEEP are summarized in Fig. 3 and show a good correlation ($r=0.98$; $p<0.001$). The differences were relatively small but the effective values were often slightly superior to the corrected values of $E_{rs,st}$ and statistical analysis showed a small but significant difference ($p=0.043$; paired t-test). At levels of 5, 10, and 15 cm H_2O $PEEP_e$, the discrepancies between uncorrected $E_{st,rs}$ and corrected $E_{st,rs}$ values tended to diminish progressively. This was accompanied by a progressive decrease of $PEEP_i$ under the influence of $PEEP_e$ in all studied patients and $PEEP_i$ was completely substituted by $PEEP_e$ in six patients (see Table 3). As a result, the differences between uncorrected and corrected $E_{st,rs}$ showed a progressive reduction and became non-significant at $PEEP_e = 15$ cm H_2O ($p=0.064$; paired t-test).

The relationships between uncorrected $E_{rs,st}$ and $E_{rs,eff}$ showed the same evolution and are summarized in Fig. 3 at PEEP = 10 cm H_2O, where uncorrected $E_{rs,st}$ still exceeded $E_{rs,eff}$ but the difference was smaller. At $PEEP_e = 15$ cm H_2O, the statistical difference disappeared ($p=0.102$).

The analysis of $E_{rs,eff}$ and corrected $E_{rs,st}$ showed a strong similarity at PEEP 10 (Table 2) and this is summarized in Fig. 3. The significant statistical difference disappeared for each level of $PEEP_e$. Some $PEEP_i$ was present in 21 of 27 patients at ZEEP and most of these patients (except numbers 1, 14, 16, 23 and 26) presented with severe COPD. Of these five patients without COPD, patients 16 and 26 had only a very small $PEEP_i$ (1 cm H_2O) and patient 23 had an obstructed endotracheal tube. In this last patient $PEEP_i$ disappeared when the endotracheal tube was changed.

The influence of $PEEP_e$ on E_{rs} is shown in Table 2. Individual values of $E_{rs,eff}$ and corrected $E_{rs,st}$ showed a slight improvement with the lower levels of $PEEP_e$ even in patients with $PEEP_i$. Analysis of the subgroup of patients having $PEEP_i$ shows that eleven of these eighteen patients improved initially under the influence of $PEEP_e$. The influence of $PEEP_e$ on uncorrected $E_{rs,st}$ seems to be a result of the overestimation of these values due to $PEEP_i$.

Discussion

The purpose of the present study was to demonstrate the validity of a computed method for the determination of the elastic properties of the total respiratory system in ventilated patients. We have compared this computerized method to the static method which provides the overall static elastance of the respiratory system pertaining to the entire tidal volume. In addition this study confirms the presence of $PEEP_i$ in a large proportion of patients in the intensive care units. We have observed it in 21 of 27 patients when studied at ZEEP. The consequence of dynamic hyperinflation is $PEEP_i$ [14] and it is most frequently observed in COPD patients, where expiration is limited by dynamic airways compression.

The end-expiratory occlusion technique can be used to determine $PEEP_i$ when the ventilator has automatic end-expiratory occlusion. The occlusion must of course be performed at the same moment as the routine inflation. One must also be sure to avoid or to detect muscular participation during the occlusion and the presence of an end-expiratory plateau generally signifies relaxation of the patient if this repeated manoeuvre gives reproducible values for $PEEP_i$.

Our analysis of the measurement of the effective elastance of the total respiratory system shows that this method was strongly correlated with the static method, when a correction for $PEEP_i$ was made. In several cases, $E_{rs,eff}$ was virtually identical to the corrected $E_{rs,st}$ but several values of $E_{rs,eff}$ slightly exceeded the corrected static values and this difference was statistically significant. This discrepancy may reflect a frequency-dependence due to either non-uniformity of regional time constants and abnormalities in the distribution of ventilation [14], or due to lung visco-elasticity [15]. The disappearance of a significant difference with $PEEP_e$ and the decrease of E_{rs} with low levels of $PEEP_e$ is consistent with recruitment of new lung units and suggests that the small difference between $E_{rs,eff}$ and corrected $E_{rs,st}$ is due to pendelluft. These effective values are virtually the same as the static values and are susceptible to the potential errors related to $PEEP_i$. The $E_{rs,eff}$ and corrected $E_{rs,st}$ values of our patients are generally elevated (29.2 cm and 28.2 cm H_2O/l respectively) and similar to data from previous studies [1, 4, 7].

Regardless of the aetiology, the E_{rs} values of our patients tended to decrease under the influence of low values of $PEEP_e$ (5 and 10 cm H_2O). Even the patients demonstrating $PEEP_i$ in most cases did not show any alteration of respiratory mechanics under the influence of $PEEP_e$. Eleven of the eighteen patients with $PEEP_i$ tested under $PEEP_e$ showed an improvement of E_{rs}. This means that some of these patients with expiratory flow limitation and dynamic hyperinflation may be improved by $PEEP_e$. We interpret this slight decrease of E_{rs} as beneficial because it probably represents recruitment of collapsed alveoli and prevention of small airways closure suggesting improved oxygenation. It is also likely that $PEEP_e$ is most effective without adverse effects on the cardiac output if it is used within the best compliance range [1]. Furthermore, this effect of PEEP suggests a similar effect for CPAP in the same group of patients during weaning, by improving oxygenation and reducing the respiratory work during this process.

A potentially deleterious effect of PEEP on the end-expiratory lung volume must be considered, but we believe that this mechanism must be rather limited. Indeed, lung volumes are not likely to be increased to an important degree with $PEEP_e$ as shown by the progressive substitution of $PEEP_i$ by $PEEP_e$. Furthermore, E_{rs} decreased with $PEEP_e$, a finding unlikely to occur in the setting of exacerbated hyperinflation. This last point has been suggested by Simkovitz et al. [13] who studied the effect of PEEP in a group of ventilated COPD patients and did not observe an increase of end-expiratory lung volume with low levels of PEEP.

Our findings of the beneficial influence of PEEP on respiratory mechanics are consistent with those recently reported by Smith and Marini who describe very modest changes in end-inspiratory plateau pressures after application of PEEP in ventilated COPD patients. They estimated that the majority of the increase in lung volume occurred after the applied PEEP exceeded the initial $PEEP_i$ level [16].

With respect to the resistances of the respiratory system, a slight but significant difference was observed between the values obtained by the static and computed methods (unpublished results). The values of the R_{rs} obtained with the computed method were slightly higher but the precise cause of this difference remains unclear.

However, we estimate that this effective resistance offers a reliable quantification of the respiratory resistance although this does not provide strictly similar information when compared with the static method.

In conclusion, the present study indicates that the effective elastance of the total respiratory system is very similar to the corrected static values in ventilated patients. The slightly superior values of effective elastance are probably due to frequency-dependence related to pendelluft. It should be stressed, that with this computerized method allowance must not be made for $PEEP_i$, in contrast to the end-inflation occlusion method. The computerized method does not require either the measurement of end-inspiratory and end-expiratory occlusion or a constant flow-inflation for computation of the resistive properties.

Some of the inferior values of effective elastance compared to the corrected static values are probably due to errors of estimation of $PEEP_i$. This computerized method is precise, simple and non-invasive, it permits measurements of both elastic and resistive properties of the respiratory system in ventilated patients and it could also be employed for the study of lung and chest-wall mechanics if the transpulmonary pressure is used. An improvement of mechanical parameters is frequently observed induced by $PEEP_e$, even in hyperinflated patients. This observation, together with the decrease of $PEEP_i$ with $PEEP_e$, brings confirmation to the putative positive effects of continuous positive airway pressure (CPAP) in some patients with COPD. Indeed, $PEEP_i$ has to be considered in spontaneously breathing patients as an inspiratory threshold load, but it could be reduced during weaning by the application of CPAP. This effect, together with the probable improvement of pulmonary mechanics under CPAP in some COPD patients, could well be useful during the weaning process or as an alternative to mechanical ventilation.

Appendix

Modelling the mechanical behaviour of the lung system

Air flow in endotracheal tubes and also in the bronchial tree is not linearly related to the corresponding pressure drop and an important contribution of this pressure drop is in the form of $\dot{V}2$. Elastic recoil pressure drop is also a non-linear phenomenon (curvilinear law and hysteresis) that cannot currently be completely characterized. The problem to be solved can be viewed as a problem of identification which can be expressed in the following terms: given a mathematical model including some parameters, find the optimal values of these parameters that minimize the "distance" between the model and the real system. The choice of the mathematical model is essential, and it is generally governed by physical laws. The number of parameters has to be large enough to make the model valuable but must not be too large because inaccuracies due to measurement errors would then result.

One way of solving this difficult "identification" problem is to define experimental situations in which particular assumptions can be made in order to reduce the complexity of the observed phenomena; for example, experimentation at constant flow or in apnoea considerably simplifies the models involved and leads to more fundamental insight. This point of view is generally adopted in physiological studies and it generally takes much time for the experimentation.

Another way of solving the problem is to use the identification methods developed for the estimation of parameters for models which are linear in their parameters. One of these techniques is the so-called (recursive) least squares estimation method that can be introduced as follows:

Let us assume that the dynamical behaviour of the human ventilatory system is governed by an equation of the following type:

$$P(t) = R[\dot{V}(t),\ V_g(t)]\cdot\dot{V}(t) + E[V(t),\ V_g(t)]\cdot V(t) + (t) \quad (1)$$

with:

$P(t)$: driving pressure.

$R[\dot{V}(t),\ V_g(t)]\cdot V(t)$: resistive pressure drop including a non-linear resistance R that is a function of the air flow $\dot{V}(t)$ and of the gas volume $V_g(t)$.

$E[V(t),\ V_g(t)]\cdot V(t)$: elastic pressure drop including a non-linear elastance E that is a function of the volume variation $V(t)$ and of $V_g(t)$.

(t): all other terms necessary to make relation (1) true at all time t (cardiac pressure contribution, hysteresis phenomena, ...).

Then, the *measured* driving pressure P_m (t) can also be written as the following sum:

$$Pm(t) = R[\dot{V}m(t),\ V_g(t)]\cdot\dot{V}m(t) + E[Vm(t),\ V_g(t)]\cdot V_m(t) + P^o + m(t) \quad (2)$$

where m designates "measured" values, P^o represents the mean airway pressure level including the hydrostatic pressure and the pressure transducer offset and m(t) contains (t) and measurement errors for P_m, $\dot{V}m$ and Vm.

The Heres computer determines the optimal constant values R*, E* and P_o* that can be put in the following linear approximation of the calculated pressure drop $P_c(t)$:

$$P_c(t) = R^*\ \dot{V}m(t) + E^*\ Vm(t) + P_o^* \quad (3)$$

in order to minimize the following cost function:

$$J = \frac{1}{T}\int_0^{Tresp} [P_m(t) - P_c(t)]^2\, dt$$

The estimation time has been chosen as the respiratory period (T).

These optimal values R* and E* are also called effective resistance and elastance. They also can be associated with the classical r. m. s. (root mean squared) values used by electrical engineers. It is easily seen that the magnitude of J is related to the more or less non linear character of the dynamical behaviour. It is also interesting to note that the computation time is negligible so that measurements can be considered as "on line" ones.

When P_0 is negligible, the simplified solution minimizing J is

$$R_{eff} = \frac{\int_{t_0}^{t_0+T} P_m(t)\ \dot{V}(t)\ dt}{\int_{t_0}^{t_0+T} \dot{V}^2(t)\ dt}$$

and

$$E_{eff} = \frac{\int_{t_0}^{t_0+T} P_m(t)\ V(t)\ dt}{\int_{t_0}^{t_0+T} V^2(t)\ dt}$$

where T is the respiratory period.

The solution may be also expressed on his digitized form, as processed by the calculator:

$$R^{eff} = \frac{\sum_{i=1}^{N} P_m(i)\ \dot{V}(i)}{\sum_{i=1}^{N} \dot{V}^2(i)}$$

and

$$E_{eff} = \frac{\sum_{i=1}^{N} P_m(i)\ V(i)}{\sum_{i=1}^{N} V^2(i)}$$

where N is the number of samples during T.

Other non-linear models could be considered but the solving of these would be more difficult and would necessitate more sophisticated techniques (P. Young – Recursive estimation and time series analyses. – Springer Verlag, 1984).

Acknowledgements. The writers thank Pr. Jean-Claude Yernault for helpful criticism, Annette Fally for typing the manuscript and Jean-Pierre Peeters for the illustrations.

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