A SINGLE COMPUTER-CONTROLLED MECHANICAL INSUFFLATION ALLOWS DETERMINATION OF THE PRESSURE-VOLUME RELATIONSHIP OF THE RESPIRATORY SYSTEM

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Svantesson C, Drefeldt B, Sigurdsson S, Larsson A, Brochard L, Jonson B. A single computer-controlled mechanical insufflation allows determination of the pressure-volume relationship of the respiratory system.

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ABSTRACT. Objective. To evaluate and further develop a method for determination and mathematical characterisation of the elastic pressure-volume (P_{el}-V) relationship in mechanically ventilated human subjects during one single modified insufflation with simultaneous determination of resistance of the respiratory system. **Subjects.** Eight adult non-smoking human subjects without heart, lung, or thoracic cage disease scheduled for non-thoracic surgery. The study was performed in anaesthetised and muscle-relaxed subjects. **Measurements and Main Results.** The P_{el}-V curve was determined with a computer-controlled Servo Ventilator 900C during a modified insufflation with either constant or sinusoidally varying flow. Pressure and flow were measured with the built-in sensors of the ventilator. Tracheal pressure (P_{tr}) was calculated by subtracting the pressure drop over the tracheal tube. The elastic recoil pressure in the peripheral lung, P_{el}, was obtained from the calculated P_{tr} by subtracting the pressure drop over the airways. P_{tr} was also directly measured through a catheter. The calculated P_{tr} gave similar results as the directly measured P_{tr}, thus indicating the reliability of the signal originating from the ventilator sensor for computation of downstream pressures. The inflection points of the sigmoidal P_{el}-V curve and the compliance of the linear segment were determined with high reproducibility. **Conclusions.** Using one single modified insufflation allows a fast and accurate determination of respiratory mechanics. The P_{el}-V curves were determined with high reproducibility and were adequately described by a three-segment model of the curve incorporating a linear segment between two asymmetrical non-linear segments.

KEY WORDS. Compliance, human, lung, mechanical ventilation, mechanics, resistance.

INTRODUCTION

The elastic pressure-volume (P_{el}-V) curve of the respiratory system is useful in ventilator setting so as to limit lung damage related to mechanical ventilation [1]. Traditional techniques for determination of the static P_{el}-V relationship, such as the super-syringe technique [2] and the flow interruption technique [3], are time-consuming and cumbersome. The dynamic methods based on the constant flow principle reflect both resistive and elastic properties unless flow rate is very low [4]. Most other methods based on this principle are limited to volumes within the tidal volume range [5, 6].

Recently, a new technique for fast and convenient determination of the P_{el}-V curve of the respiratory system, to be used in intensive care routine, was developed within the present research group [7]. Recording

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time was minimised by using observations during a single insufflation. The insufflation covered a volume range exceeding the tidal volume - smaller volumes were reached by modifying the preceding expiration - larger volumes by delivering an increased volume. Events before and during the insufflation were computer-controlled and standardised. To be suitable for routine use at bedside the measurement was automated. Extra equipment was minimised by using the built-in pressure and flow transducers of the ventilator. The elastic recoil pressure, P_{el}, was obtained by subtracting the resistive pressure drop in the tubes and airways from the pressure measured in the ventilator. In the referred study it was shown, in critically ill patients, that P_{el}-V curves recorded in accordance with the present single insufflation principle were similar to static P_{el}-V curves recorded with the flow interruption technique. In that study the resistance of the patient's respiratory system (R_{RS}), needed to calculate P_{el}, was estimated as the "effective resistance" during a complete ordinary breath [8].

In the present study we describe and evaluate a further developed method based on the single insufflation technique. Hence, the complete sigmoidal P_{el}-V curve was objectively characterised by applying a three-segment model of the curve. The reproducibility of the method was studied. Furthermore the validity of using the built-in pressure transducer of the ventilator was examined by comparing pressure-volume curves determined from directly measured tracheal pressure (P_{tr}) and those calculated from the pressure measured in the ventilator. A further objective was to develop the single insufflation technique to allow estimation of R_{RS} by the inspiratory resistance during the insufflation itself. To determine the inspiratory resistance during the single insufflation the flow rate during the insufflation was computer-controlled to be sinusoidally modulated at a low frequency (1 Hz). The study was performed in healthy human subjects.

MATERIALS AND METHODS

Materials

Eight adult human subjects (M = 3, F = 5; age: 33–60 yr.) were studied in the supine position before the start of non-thoracic surgery. The subjects were non-smokers, had normal body constitution, and were healthy with respect to heart, lungs and thoracic cage. They had a normal spirometry. Anaesthesia was induced intravenously with fentanyl and propofol, and muscle-relaxation for tracheal intubation was achieved with suxamethonium chloride. The trachea was intubated with a tube (PORTEX Ltd., Hythe, Kent, England), 7.5–8.0 mm I.D. After intubation pancuronium bromide was delivered for muscle-relaxation and anaesthesia was maintained with propofol. The subjects were connected to a Servo Ventilator 900C (Siemens-Elema AB, Solna, Sweden) and were ventilated in the volume-controlled mode. Minute volume (MV) was 100 ml/kg, breathing frequency 20 breaths/min, inspiratory and post-inspiratory pause time 33% and 5% of the respiratory cycle, respectively, the inspired fraction of oxygen (FIO_2) was 0.35, and the positive end-expiratory pressure (PEEP) was 5 cm H_2O. For standardisation and elimination of possible atelectasis after induction of anaesthesia the lungs were recruited according to previously described principles [9]; the lungs were inflated twice to an airway pressure of 40 cm H_2O maintained for 15 s. Between these two large insufflations six pressure-controlled breaths (6 breaths/min) were delivered with an airway pressure of 30 cm H_2O.

The study was approved by the Ethics Committee of the University Hospital of Lund, and informed consent was obtained from each subject.

Equipment and recording of signals

The measurement system consisted of the ventilator, a custom-built ventilator-computer interface (VCI) and a personal computer (PC) (Figure 1) [7, 10]. Data acquisition and analysis were performed with the PC that was equipped with a multi-function card comprising A/D converters, D/A converters, and a digital interface (PC-30, Boston Technology). Pressure in trachea distal to the tube was measured with a SenSym SX01DN pressure transducer ($\pm$ 70 cm H_2O, Sensortec GmbH, Germany) connected to a side-hole polythene catheter (XRO feeding tube, I.D. 0.8 mm, O.D. 1.5 mm) via a polythene pressure tubing (180 cm, I.D. 1.6 mm, O.D. 3.1 mm). The side-hole catheter was passed through the tracheal tube and ended 3 cm beyond its tip. Flow rate ($\dot{V}$) and pressure in the expiratory line (P_E) were measured with the flow and pressure transducers of the ventilator. All flow and pressure transducers were calibrated with the calibration system of the ventilator and checked against a super-syringe (1 liter) and a water manometer. Values of flow and volume are given at BTPS. To eliminate minor deviations in inspiratory and expiratory data we used a breath recorded at steady state to normalise the expiratory flow data so that the measured expired volume equalled the inspired volume. The signals representing P_{tr}, $\dot{V}$, P_E and a signal defining inspiration and expiration were fed to the VCI,

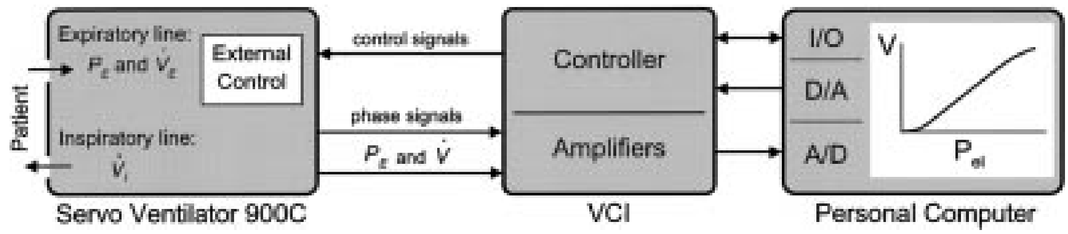

Fig. 1. A ventilator was computer-controlled via a custom-built ventilator-computer interface (VCI) to yield appropriate breathing patterns for automated determination of the pressure-volume relationship and the resistance of the respiratory system. The Servo Ventilator 900C is equipped with a socket allowing external control of ventilator function. The personal computer was provided with analogue-to-digital (A/D) and digital-to-analogue (D/A) converters and a digital interface (I/O) to enable control of the ventilator and recording of pressure in the expiratory line (P_E) and flow rate ($\dot{V}$).

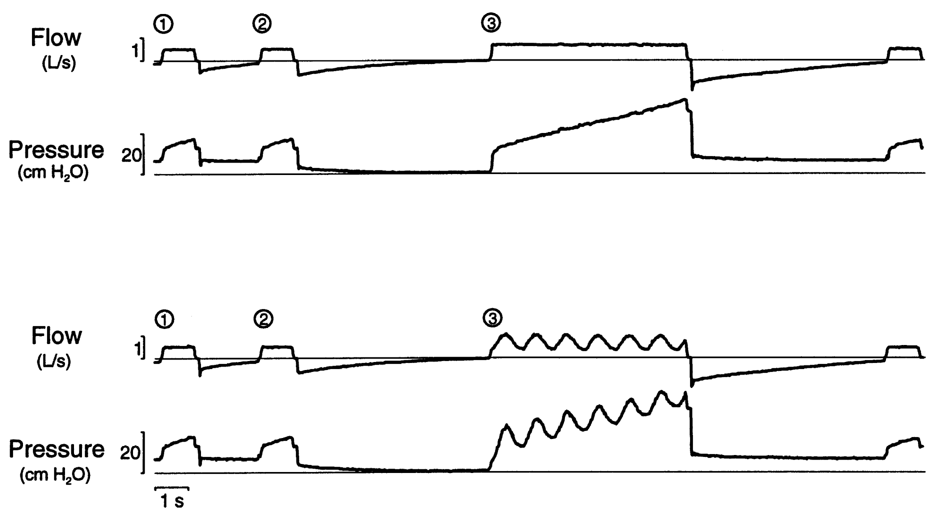

Fig. 2. Flow and pressure signals recorded during a measurement sequence containing the normal breath ①, the breath with a prolonged expiration and PEEP lowered to zero ②, and the breath with the insufflation ③. Top: The constant flow method. Bottom: The modulated flow method.

in which they were amplified before being fed to the A/D-converter of the PC. The signals were sampled at a frequency of 50 Hz.

The single insufflation technique

The settings of MV, breathing frequency and PEEP on the front panel of the ventilator can be overridden by applying analogue signals to the lines for external control of the ventilator which previously has been described in detail [7, 10]. The effects are immediate. By externally controlling for example MV, within ±100% of set value, the inspiratory flow can be altered for example to a sinusoidal form.

A recording comprised a sequence of computer-controlled breaths illustrated in Figure 2. The 1st breath is an ordinary breath as defined by the settings on the front panel of the ventilator. To obtain the desired starting point of the P_{el}-V curve, the expiration of the 2nd breath is adjusted by choosing the duration of expiration and the PEEP level. In this study a prolonged expiration of 6 s with PEEP lowered to zero was used to approach the volume at which the elastic forces within the respiratory system are equilibrated. The insufflation of the 3rd breath provides the basis for determination of the P_{el}-V curve over an extended volume range. As shown in Figure 2, the flow during the insufflation can be either constant or sinusoidally modulated. The expiration following the insufflation is prolonged to allow emptying of the large insufflated volume.

The complete computer-controlled sequence is defined in advance in a computer-operator dialogue. The volume ($V_{insuffl}$) and duration of the large insufflation,

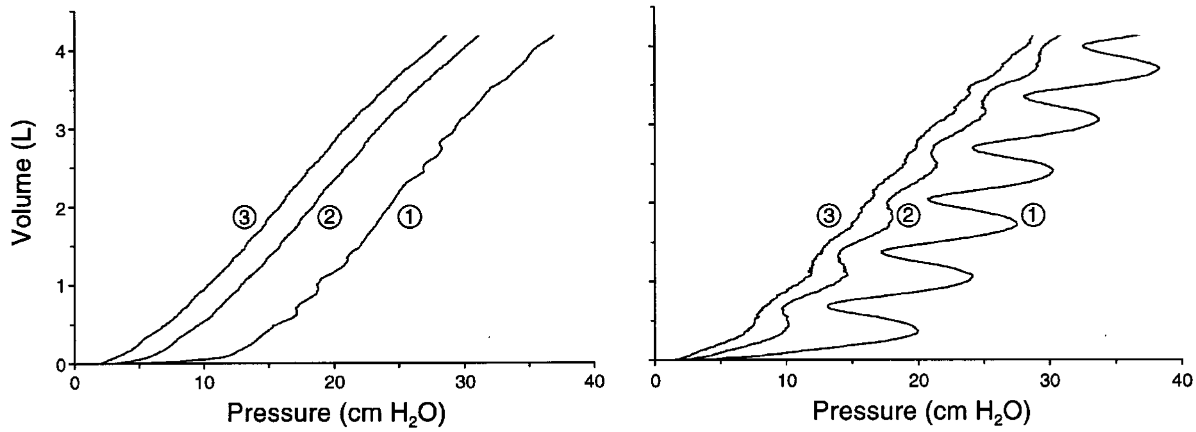

Fig. 3. The measured pressure ①, the calculated tracheal pressure ②, and the elastic recoil pressure ③, versus volume recorded in a representative patient. Left: The constant flow method. Right: The modulated flow method. In the latter the large sinusoidal pressure fluctuations are mainly due to the dominant resistance in the tracheal tube. The smaller fluctuations in tracheal pressure represent resistance below trachea. The elastic recoil pressure differs more from tracheal pressure in the left panel because resistance for the respiratory cycle determined with the constant flow method was higher than inspiratory resistance determined with the modulated flow method (right panel).

the desired duration and PEEP level of the preceding expiration, and the duration of the following expiration are entered as well as the value of the maximum pressure allowed during the test. Further, the settings of the ventilator are entered. These data are used for realisation of the test and are documented in the result files together with patient data.

Calibrated pressure and flow signals are continuously displayed on the computer screen. During the measurement sequence the computer program recognises each transition in ventilator phase and controls during each phase the needed parameters to obtain the desired ventilation pattern as described above. The computer terminates the large insufflation if and when $V_{insuffl}$ or the maximum pressure is attained before the predefined insufflation time has expired.

Data analysis

To correct the flow signal measured in the ventilator for changes of gas volume in the ventilator tubing, the product between the rate of pressure change and the tube compliance was subtracted from the measured flow signal. The volume was obtained by integration of the corrected flow signal. The tube resistance (R_{tube}) including the y-piece, a flexible connector and the tracheal tube, was defined as: $R_{tube} = K_1 + K_2 \cdot \dot{V}$. The coefficients, K_1 and K_2, were determined *in vitro*. The tracheal pressure, P_{tr}, was obtained by subtracting the resistive pressure drop over the connecting tubes from the measured pressure, P_E, according to Equation (1).

$$P_{tr} = P_E - \dot{V} \cdot R_{tube} \qquad (1)$$

In the muscle-relaxed subject, the elastic recoil pressure, P_{el}, equals the alveolar pressure. The tracheal pressure is in the muscle-relaxed subject the sum of P_{el} and the resistive pressure, which drives the pressure through the resistance of the respiratory system, R_{RS}. P_{el} is calculated by subtracting the resistive pressure from P_{tr}, according to Equation (2).

$$P_{el} = P_{tr} - \dot{V} \cdot R_{RS} \qquad (2)$$

The calculation of P_{el} can be followed in Figure 3.

For the constant flow method, R_{RS} was determined from the normal breath (No. 1 in Figure 2) as the quotient between the area of the P_{tr}-V loop and the area of the $\dot{V}$-V loop [8]. For the modulated flow method the estimation of R_{RS} depends on analysis of pressure changes synchronous to the modulations of flow (Figures 2 and 3) as further described below.

Characterisation of the pressure-volume curve using a three-segment model

The characterisation of the P_{el}-V curve was made using a three-segment model of the P_{el}-V curve (Figure 4) [11]. The model is based on the assumption that, at low volumes compliance is low and increases linearly with volume, it then assumes a constant high value and at larger volumes it falls linearly with volume. This corresponds to a P_{el}-V curve comprising a strictly linear

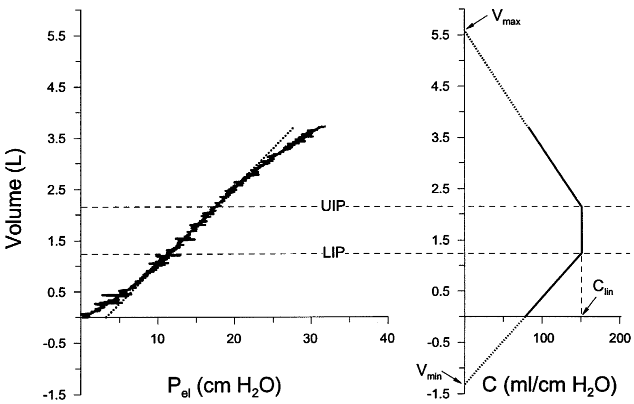

Fig. 4. The sigmoidal P_{el}-V curve was characterised using a three-segment model of the curve comprising a linear segment between two non-linear asymmetrical segments. The model was based on the assumption that compliance (C) is low and linearly increasing at low volumes, it then assumes a constant high value and at larger volumes it falls linearly with volume. The P_{el}-V curve asymptotically reaches V_{min} and V_{max} at its lower and upper ends, respectively.

segment with high compliance (C_{lin}) between two non-linear asymmetrical segments. The P_{el}-V curve asymptotically approaches V_{min} and V_{max} at its lower and upper ends, respectively. The linear segment is delineated by the lower and upper inflection points (LIP and UIP, respectively).

P_{el} as a function of volume was described as:

$$
\begin{aligned}
&\text{For } V_{min} < V \leq V_{LIP}: \\
&\quad P_{el} = P_{LIP} - \frac{(V_{LIP} - V_{min})}{C_{lin}} \cdot \ln\left(\frac{V_{min} - V_{LIP}}{V_{min} - V}\right) \\
&\text{For } V_{LIP} \leq V \leq V_{UIP}: \\
&\quad P_{el} = P_{LIP} + \frac{(V - V_{LIP})}{C_{lin}} \\
&\text{For } V_{UIP} \leq V < V_{max}: \\
&\quad P_{el} = P_{UIP} + \frac{(V_{max} - V_{UIP})}{C_{lin}} \cdot \ln\left(\frac{V_{max} - V_{UIP}}{V_{max} - V}\right)
\end{aligned}
\tag{3}
$$

V_{LIP} and P_{LIP} represent the values of volume and pressure at the lower inflection point and V_{UIP} and P_{UIP} represent the values of volume and pressure at the upper inflection point. The parameters, V_{min}, V_{LIP}, P_{LIP}, C_{lin}, V_{UIP} and V_{max}, were estimated with numerical technique. For the constant flow method, parameter estimation was achieved by minimising the sum of squared differences between P_{el} from measured data (Equation (2)) and P_{el} calculated using the model (Equation (3)). For the sinusoidally modulated flow method, the before mentioned parameters in addition to R_{RS} were determined by comparing P_{tr} from measured data (Equation (1)) with P_{tr} calculated using the model complemented by a resistive element (Equation (4)).

$$P_{tr} = P_{el} + \dot{V} \cdot R_{RS} \tag{4}$$

In this equation P_{el} is described by Equation (3).

Measurements

In each subject three recordings were performed. In order to evaluate the reproducibility of the method regarding determination of C_{lin}, P_{LIP} and P_{UIP} of the P_{el}-V curve two curves were recorded with the constant flow method with a time interval of 1–14 min. For validation of P_{tr} calculated from P_E according to Equation (1), P_{el}-V curves obtained from the calculated P_{tr} were compared to those obtained from the directly measured P_{tr}. A third recording was performed using the modulated flow method. Comparison of the first curves determined with the constant flow method and the curves determined with the sinusoidally modulated flow method were made to evaluate the method for determination of the respiratory system resistance, R_{RS}, during the large insufflation itself. For reference, the R_{RS} was also determined from the first breath (Figure 2) with the flow interruption technique [12].

Statistical analysis

The linear segment of the pressure-volume curve was described by the slope, i.e., compliance, of the linear segment, C_{lin}, and the position along the pressure axis of this segment determined as the pressure at the lower and upper inflection points of the curve, P_{LIP} and P_{UIP}, respectively. P_{el}-V curves were compared with respect to the values of C_{lin}, P_{LIP} and P_{UIP} using the Wilcoxon signed rank test. Differences were considered significant if $p < 0.05$. The data was expressed as mean ± SD.

RESULTS

The numerical estimation of model parameters always resulted in an adequate description of the entire P_{el}-V curve (Figure 5). The residual deviations between observations and the solution according to the three-segment model appeared to represent high frequency noise. As all features of the curves judged relevant were adequately described by the model, further analysis of P_{el}-V curves was based on the mathematical description.

The validity of using the built-in pressure transducer

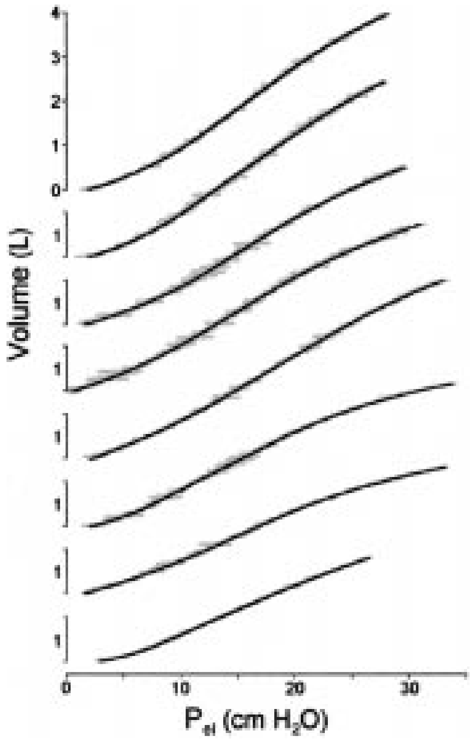

Fig. 5. The elastic recoil pressure, P_{el}, versus volume determined with the constant flow method in all patients. The smooth line represents the mathematical description of the curve, based on a three-segment model.

of the ventilator was examined by comparing P_{el}-V curves determined from directly measured P_{tr} and those calculated from the pressure measured in the ventilator. Compliance determined over the linear part of the P_{el}-V curve, C_{lin}, based on the directly measured P_{tr} was 4 ml/cm H_2O lower than C_{lin} of the P_{el}-V curves based on calculated P_{tr} ($p < 0.05$) (Table 1). The P_{LIP} of the P_{el}-V curves did not differ. The P_{UIP} of the P_{el}-V curves based on directly measured P_{tr} was 1.1 cm H_2O higher than that of the P_{el}-V curves based on calculated P_{tr} ($p < 0.05$).

The reproducibility in determination of P_{el}-V curves was examined by comparing C_{lin}, P_{LIP} and P_{UIP} estimated from two P_{el}-V curves recorded with the constant flow method. There were no significant differences between the two curves. Individual differences were small (Table 2).

The values of C_{lin}, P_{LIP} and P_{UIP} determined from the P_{el}-V curves recorded with modulated flow during the large insufflation did not differ from the values of the P_{el}-V curves recorded with constant flow (Table 1).

For the constant flow method, the respiratory system resistance, R_{RS}, determined from a complete ordinary breath was 4.1 ± 0.5 cm H_2O/(l/s). R_{RS} was 1.3 ± 1.8 cm H_2O/(l/s) as determined from the modulated flow insufflation and 2.0 ± 0.8 cm H_2O/(l/s) as determined with the flow interruption technique. The difference between the two latter values was not significant. The heart frequency was on average 74 min^{-1} (range: 65–81).

Table 1. Comparison of P_{el}-V curves

	Constant flow		Modulated flow
	Calculated P_{tr}	Measured P_{tr}	
C_{lin} (ml/cm H_2O)	146 ± 26	142 ± 25 *	143 ± 25 NS
P_{LIP} (cm H_2O)	11.3 ± 2.4	10.8 ± 2.9 NS	10.0 ± 2.2 NS
P_{UIP} (cm H_2O)	18.9 ± 2.4	20.0 ± 2.3 *	21.5 ± 3.9 NS

P_{el} – the elastic recoil pressure in the peripheral lung compartments; P_{tr} – tracheal pressure; C_{lin} – the compliance of the linear segment of the P_{el}-V curve; P_{LIP} – pressure at the lower inflection point; P_{UIP} – pressure at the upper inflection point.
The symbol * indicates significance ($p < 0.05$) compared with "Constant flow, Calculated P_{tr}"; NS – non-significant.

Table 2. Comparison of two P_{el}-V curves determined with the constant flow method

	C_{lin} (ml/cm H_2O)		P_{LIP} (cm H_2O)		P_{UIP} (cm H_2O)	
Subj.	1st curve	2nd curve	1st curve	2nd curve	1st curve	2nd curve
1	176	186	10.3	10.6	20.5	21.6
2	151	157	11.4	16.1	17.4	18.8
3	116	116	13.3	13.8	17.5	17.4
4	182	179	11.7	13.3	19.2	21.0
5	156	163	13.0	14.2	18.1	17.4
6	128	127	7.9	9.3	18.7	19.2
7	147	157	14.5	13.8	23.6	23.2
8	110	118	8.1	8.5	15.8	15.1
mean	146	150	11.3	12.5	18.9	19.2
SD	26	27	2.4	2.6	2.4	2.6

C_{lin} – the compliance of the linear segment of the P_{el}-V curve; P_{el} – the elastic recoil pressure in the peripheral lung compartments; P_{LIP} – pressure at the lower inflection point; P_{UIP} – pressure at the upper inflection point.

DISCUSSION

In this paper we describe a method for recording of the elastic pressure-volume relationship of the respiratory system. We have sought to combine the best properties of previously described principles for recording of static and dynamic P_{el}-V curves. The described method for determination of respiratory mechanics is first and foremost designed for studies of mechanically ventilated patients with critical lung disease. In intensive care a method needs to be fast. The recording (Figure 2) takes about 30 s and is fully automated. The following analysis is automated and takes no more than 2 min.

Measurement of the pressure in the expiratory circuit of the ventilator gave, after subtraction of tube resistive pressure, similar results as those based upon direct measurement of tracheal pressure. This reflects that during inspiration the expiratory ventilator tube serves as a catheter between the y-piece and the transducer. Further, the pressure and flow transducers in the Servo Ventilator have excellent frequency characteristics [12]. The resistance of the connectors and the tracheal tube was measured *in vitro* but under inspiratory sinusoidal flow similar to that applied during sinusoidal insufflations. In clinical practice tube resistance may increase because of accumulation of debris and phlegm. It is recommended that the tube is cleaned before measurements. Any water in the expiratory line must be removed. The tube may change its configuration *in situ* [13]. Using the value of tube resistance determined *in vitro* implies that an increase in resistance *in vivo* will be translated into the calculated value of resistance of the respiratory system. The results show that our method to subtract flow resistive pressures to calculate P_{tr} gives adequate estimates of C_{lin}, P_{LIP} and P_{UIP}. Accordingly, theoretical limitations of the method inferred by flow dependent resistance of the tracheal tube cause non-significant errors. The use of the ventilator transducers limits the need for extra equipment. The use of a single set of transducers implies that diagnostic data are directly applicable to treatment as calibration differences are eliminated.

The three-segment model of the pressure-volume curve with a linear segment between two asymmetrical segments, accurately described the findings in healthy humans. Ongoing studies show correspondingly good agreement between the model and the recorded data in various groups of patients. Particularly in patients with acute lung injury we have frequently observed P_{el}-V curves which are obviously non-symmetrical (unpublished). In such cases a lung model that allows non-symmetry gives a better fit than a model recently suggested by Venegas et al. [14]. Caution is advised in ascribing a particular physiological significance to the extent of the linear segment. Recent theoretical studies show that the position of a LIP and an UIP as well as compliance of the linear segment may be much dependent upon the opening forces of collapsed lung units [15, 16]. Still, principles according to which particularly the LIP may be used in setting the PEEP [2, 17] may be clinically useful according to a recent study of Amato et al. [1].

The mathematical description of the P_{el}-V curve allowed an unbiased comparison between duplicate determinations. The modest errors of the method indicate a high validity of obtained data.

The value of respiratory system resistance, R_{RS}, used in the constant flow method, represents "effective" resistance over the full respiratory cycle [8]. The use of sinusoidally modulated flow allows measurement of R_{RS} during the insufflation itself. The latter was lower than the values of R_{RS} determined over the complete respiratory cycle. Such a difference can be expected as inspiratory resistance is lower than expiratory particularly over the low volume segment. The flow interruption method showed that inspiratory resistance was indeed as low as the modulated flow method indicated. The low value of inspiratory resistance agrees with previous observations [12].

The use of the R_{RS} according to Varène and Jacquemin [8] which is larger than the inspiratory R_{RS} will lead to a displacement of the P_{el}-V curve which is the product of flow rate (around 0.5 l/s) and the overestimation of resistance (2–3 cm H_2O/(l/s)). Such an error may be acceptable. Indeed, the study of Servillo et al. shows adequate similarities between static P_{el}-V curves and P_{el}-V curves studied under constant flow [7]. In non-obstructive subjects the resistive pressures are small. Cardiac oscillations in airway pressure may then interfere with the measurement of resistance as has been discussed with reference to the flow interruption method [12]. In obstructive lung disease, measuring inspiratory resistance with the modulated flow method may be important. If heart beat oscillations in the measured pressure are synchronous with the flow oscillations at 1 Hz this will lead to errors in R_{RS} and a displacement of the P_{el}-V curve along the pressure axis. In the worst case, represented by sinusoidal heart beat oscillations at 1 Hz, the displacement would be about as large as the heart beat oscillations or about $\pm$ 0.5 cm H_2O.

The frequency of the flow modulations we used for determination of R_{RS} was 1 Hz compared to frequencies of usually 2–10 Hz or higher used for the forced oscillation method [18]. At our low frequency we consider that the obtained value of R_{RS} reflects the real part of impedance rather than its imaginary part, as reactance then plays little role. Another factor that relates to frequency of flow modulations is viscoelasticity. Considering that the time constant for viscoelastic recoil in healthy humans is about 0.8 s [12] viscoelastic resistance is in part incorporated in the resistance determined at 1 Hz.

The static P_{el}-V curve is often regarded as a golden standard. However, static situations never appear apart from during studies of static P_{el}-V curves. The static elastic pressure is from this aspect an artefact and may, accordingly, not be fully representative for the pressure that inflicts damage. Therefore, P_{el} determined under dynamic conditions is, at least in principle, more clinically relevant [15].

CONCLUSIONS

A method for fast and convenient bedside determination of respiratory mechanics with focus on the P_{el}-V curve is presented. Automated data acquisition ensures full reproducibility of a defined procedure. Parameters are calculated objectively. As a result the reproducibility was found to be good. In patients with obstructive lung disease the modulated flow method may be preferred as it gives a value of inspiratory resistance measured simultaneously with the P_{el}-V curve.

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