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A Ventilator Generating a Positive or Negative Internal Compliance

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ABSTRACT

This paper describes technical details of a ventilator for altering the resistive and elastic load placed on respiratory muscles during spontaneous breathing in intubated infants. Positive or negative values for ventilator resistance and/or ventilator compliance can be chosen by superimposing the weighted sum of the flow and/or the volume signal over the input to a pressure controller within the pressure feedback control system of the ventilator.

The aim of the study was to compare values of the ventilator's compliance (C_v), as measured with a ventilation mechanics calculator, with those C_v values set by the ventilator's C_v control knob on the front panel. Another aim was to compare measured values of total compliance of a combined ventilator-lung model system (C_t) with the values expected according to theory where $1/C_t = 1/C_v + 1/C_{lm}$ (Eq. α; C_{lm} is the lung model's compliance).

The C_v values set on the front panel were nearly identical to those measured ($C_{vm} = 0.97 * C_{vs} + 0.54$) over the whole tested range from -20 to +20 ml/kPa. Similarly, the measured C_t values were almost equal to those expected according to Eq. α; the standard deviation of the relative residuals was 2.7% for elastic loading and 12.4% for elastic unloading.

We conclude that the ventilator described in this study can effectively provide both elastic loading and elastic unloading of spontaneous breathing, as expected according to theory.

INTRODUCTION

Spontaneous breathing in preterm infants is often hampered by impaired lung mechanics, i. e. increased airway resistance and decreased alveolar compliance. This situation often results in respiratory muscle fatigue in preterm infants. Mechanical ventilation is then required to keep blood gases within the normal range.

Some undesired side effects of mechanical ventilation might be partially avoided by improving the lung mechanics of the infant so that respiratory muscle forces can cope with the total work load. There are limited facilities to improve the mechanics of airways and alveoli themselves and, therefore, we propose to influence the mechanics of the combined ventilator-lung system by resistive and elastic unloading. The total resistance of this combined biomechanical system is thereby reduced and the compliance elevated. In this way the work load of spontaneous breathing and the muscular effort can be reduced without ventilating the infant mechanically in the conventional sense. It also may be interesting for those involved in physiological studies of breathing mechanics, nerve activity etc. to use this method to increase the total resistance to breathing and/or to decrease the elasticity of the combined ventilator-lung system.

This study introduces a method for resistive and elastic loading and unloading of spontaneous breathing using a ventilator that offers an adjustable ventilator resistance and/or an adjustable ventilator compliance, i.e., a ventilator impedance. Resistive unloading with this method has been described elsewhere (1). The aim of this paper is to demonstrate the effects of elastic loading and unloading in a physical lung model.

Theoretical basics

During spontaneous breathing via a ventilator the pressure generated by the respiratory muscles (Prm) equals the sum of the pressure drop across the resistances of the airways and the endotracheal tube (Pr), the pressure across the lung and the thorax (Pc), and the pressure provided by the ventilator (Pv).

$$Prm = Pr + Pc + Pv \tag{1}$$

This leads to the known relationship

$$Prm = R*V' + V/C + Pv \tag{2}$$

where R is the resistance, V' the flow, C the compliance and V the volume.

In continous positive airway pressure (CPAP) mode, the ventilator pressure Pv remains constant. If the pressure of the ventilator Pv at the airway opening is changed proportionally to the inhaled/exhaled volume and flow (with the proportionality factors K1 and K2, respectively) we get

$$Pv = CPAP + K1*V' + K2*V \tag{3}$$

and from equation (2)

$$Prm = (R+K1)*V' + (1/C + K2)*V + CPAP \tag{4}$$

K1 has the dimension of a resistance and K2 the dimension of an inverse compliance. This suggests that the ventilator offers an additional resistance Rv and compliance Cv, i.e., a ventilator impedance Zv, imposed on the patient's breathing activity, with

$$Rv = K1 \quad \text{and} \quad Cv = 1/K2 \tag{5}$$

It then follows from equation (4) that

$$Prm = (R+Rv)*V' + (1/C + 1/Cv)*V + CPAP \tag{6}$$

$$Prm = Rtot*V' + V/Ctot + CPAP \tag{7}$$

with the total resistance Rtot

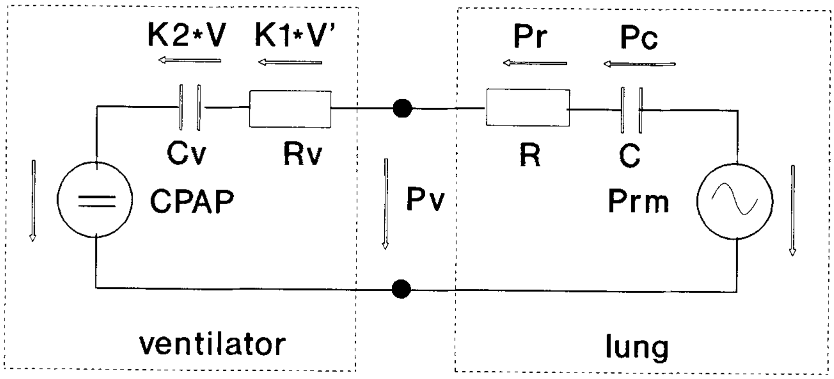
$$Rtot = R + Rv \tag{8}$$

and the total compliance Ctot

$$1/Ctot = 1/C + 1/Cv$$

or

$$Ctot=C*1/(1+ C/Cv). \tag{9}$$



<u>Figure 1:</u> Simplified schema of the combined ventilator-lung system. R, C, resistance and compliance of the patient; Rv, Cv, resistance and compliance of the ventilator; Prm, pressure generated by the respiratory muscles; Pv, ventilator pressure; K1, K2, preset values of the control knobs for ventilator resistance and ventilator compliance leading to additional ventilator pressures proportional to flow and volume.

Fig. 1 shows a simplified schema of the combined ventilator-patient system. Depending on the value and the sign of the factor K1, the magnitude of the ventilator's resistance Rv, can be small or high, and also positive or negative. That means that the total resistance Rtot can be increased or decreased in relation to the combined resistances of the airways and endotracheal tube.

In analogy, compliance is changed when varying the factor K2. With a positive ventilator compliance Cv the total compliance is decreased and with a negative Cv the total compliance is increased.

The calculation of the values of Rv and Cv is given in detail in the appendix.

Technical basics

1. The infant ventilator

The studies using ventilator impedance were performed using a ventilator that was developed at the Dresden Medical Academy and the Dresden Technical University (2-5). This ventilator was designed for respiratory therapy of newborn and preterm infants. It offers a great variety of modes, such as controlled mechanical ventilation (CMV), intermittent mandatory ventilation (IMV), high frequency oscillatory ventilation (HFOV) and continuous positive airway pressure (CPAP). CMV can be used in either controlled or assisted mode with flow controlled or pressure controlled inspiration and different inspiratory patterns for both flow and pressure.

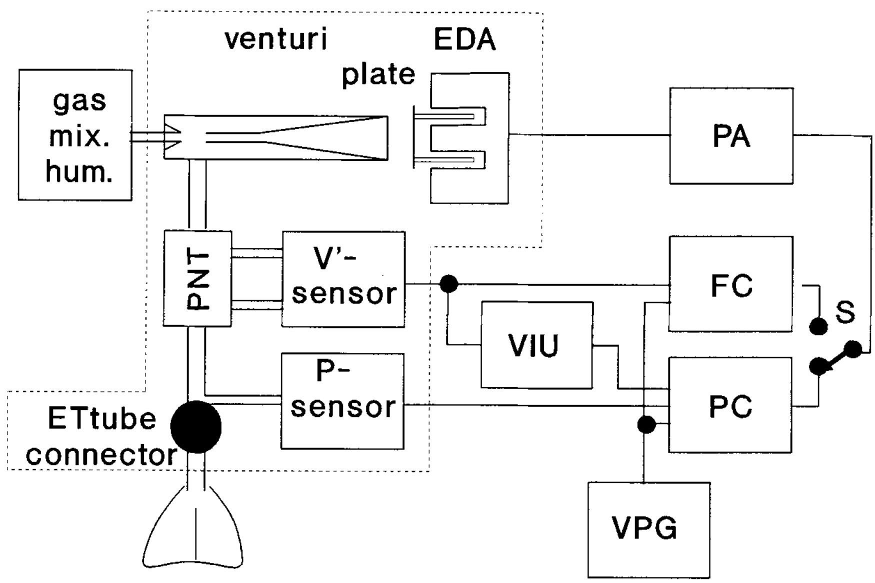


Figure 2: Operating principle of the ventilator.
PA, power amplifier; FC, flow controller; PC, pressure controller; VIU, ventilator impedance unit; VPG, ventilatory pattern generator; PNT, pneumotach; EDA, electrodynamic actuator; S, switch.

Fig. 2 shows the principle of one version of the ventilator especially used for animal experiments. The mixed and humidified gas enters a Venturi system. Its outlet is more or less obstructed by a plate which is controlled by an electrodynamic actuator (EDA). The distance of the plate from the main outlet of the Venturi determines the magnitude and direction of the flow through the lateral outlet of the Venturi. In other words, the jet flow within this system enables the unit to provide positive and negative pressures at the lateral outlet. This outlet is connected to the endotracheal tube via a miniaturized flow transducer (5) and short tubing. The total dead space of the patient circuit amounts to 1.5 ml. The pressure is measured with a pressure sensor next to the airway opening (endotracheal tube connector).

The ventilator uses two different kinds of feedback control loops, a flow-feedback-control loop and a pressure-feedback-control loop, which are used for generating different ventilatory patterns.

In flow-control mode the flow controller (FC) amplifies the difference between the actual flow value and the desired flow value provided by the ventilatory pattern generator (VPG). The FC controls the outlet valve so that the actual flow equals the desired flow value.

In pressure control mode, the difference between the actual pressure and the desired pressure is amplified by the pressure controller (PC) and the outlet valve is controlled as described for the flow-controlled mode.

The change from pressure control to flow-control is made using the switch S controlling the power amplifier. The ventilator impedance is generated in the pressure control mode. A ventilator impedance unit (VIU) is positioned between the output of the flow transducer and one input of the pressure controller.

2. The Ventilator Impedance Unit

According to equation (3) the ventilator pressure Pv is changed according to the volume and/or flow of spontaneous breathing. Thus the ventilator should be used in the pressure control mode with the control signal being the weighted sum of the flow and the volume signals. Weighting factors are K1 and K2 from equation (3). Fig. 3 shows the principle of the unit.

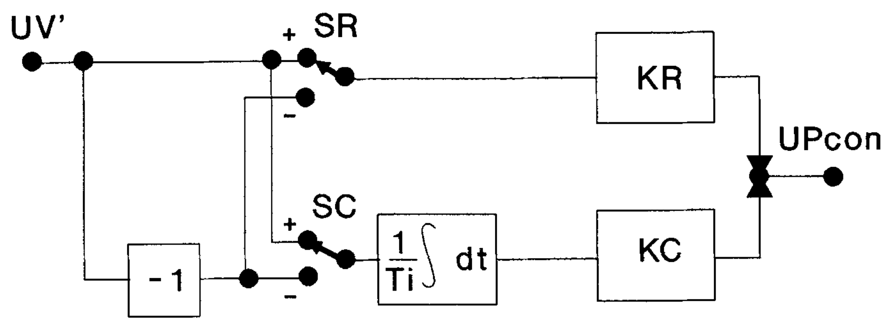


Figure 3: Principle of the Ventilator Impedance Unit (VIU). UV', output signal of the flow sensor; SR, switch for the sign of ventilator resistance; SC, switch for the sign of ventilator compliance; Ti, time constant of the integrator; UPcon, output signal of the VIU.

According to equations (3) to (5), the output signal UV' of the flow transducer should be integrated to obtain the volume signal and both signals should be amplified with the amplification factors KR and KC, respectively. A positive or negative ventilator resistance and/or ventilator compliance can be chosen using the switches SR and SC. The desired values for the ventilator resistance and ventilator compliance can be set using the control knobs for KR and KC. The output signal UPcon of the ventilator impedance unit is fed to one input of the pressure controller PC (c.f. fig.2).

We only studied the influence of a positive or negative ventilator compliance in our experiments, so KR was set at zero.

METHODS

The studies were performed using a physical lung model consisting of a "bag in bottle" system (fig. 4). A "Dräger test lung" was used as a bag. It was connected to the ventilator by a 2.5 mm endotracheal tube. The flow to and from the lung model was measured using a pneumotach head (Fleisch NEO 00). Another inlet of the bottle was connected via an injector to a rotameter. Different values of subambient pressure within the bottle could be generated by varying the rotameter flow. Spontaneous breathing was simulated by changing the flow of the rotameter periodically. Lung mechanical data were measured and computed using a Pulmonary Evaluation and Diagnostic System (PeDS unit; Medical Associated Services Inc., Hatfield, PA, USA). The pressure within the bottle was interpreted as the "oesophageal pressure" (Pe). The ventilator pressure (Pv) was measured at a T-piece between the outlet of the ventilator and the pneumotach head.

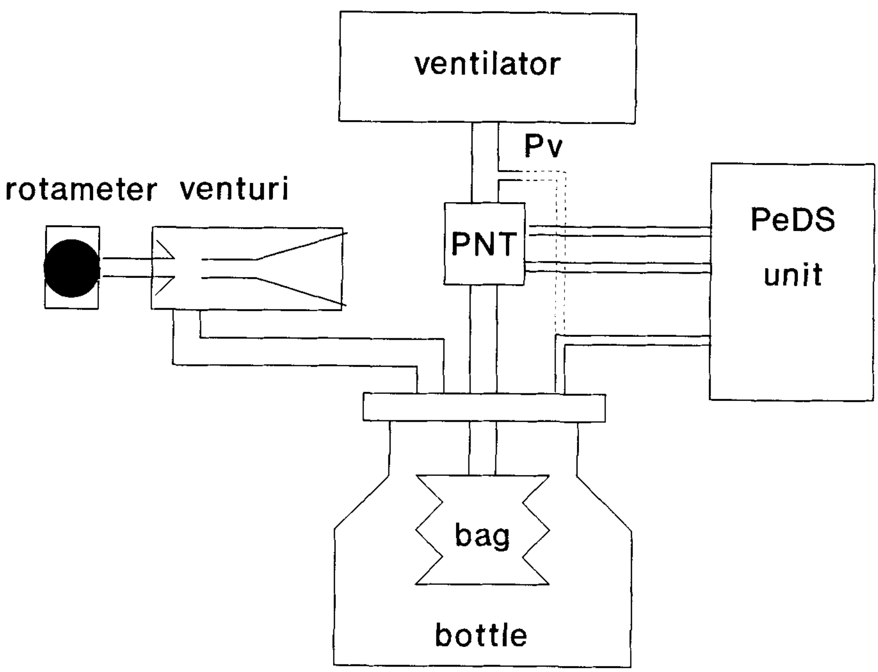


Figure 4: Physical lung model.
Measurement of the "oesophageal pressure" Pe within the bottle. For measurement of the ventilator pressure Pv the pressure transducer of the PeDS unit is connected with the pressure outlet of Pv (dotted lines) and the outlet of the bottle is clamped.

In the first set of experiments the ventilator compliance Cv was estimated by measuring the ventilator pressure Pv and the volume V. This study was performed using different settings of the control knob to adjust the preset ventilator compliance Cvs.

In a second set of experiments, the total compliance Ct of the combined lung model-ventilator system was estimated from the volume and the "oesophageal pressure" Pe for different settings of the ventilator compliance Cvs between 4.3 mL/kPa and -6.5 mL/kPa. Ct for ventilator compliance equal to infinity (KC=0) was regarded as baseline compliance of the lung model Clm. Each measurement was repeated five times to obtain representative results. The found results were compared to the theoretical compliance Cth computed according to equation 9. The standard deviation of the relative residuals was calculated according to the equation

$$s^2 = \frac{1}{n-1} * \frac{Ct(i)-Cth(i)}{Cth(i)}$$

Calculation and graphical representation of the data were performed by the program Matlab (The MathWorks Inc., USA).

RESULTS

The Pv-V loops of the ventilator itself indicate that it has the properties of compliance. Positive settings of the ventilator compliance (Cvs) lead to a pressure drop proportional to the inspired volume and a rise in pressure during expiration relative to the CPAP level. The lower the preset value of Cvs the smaller the change in volume for a given change in pressure Pv. With Cv equal to infinity there is no pressure change depending on the volume. With negative Cv an inspiratory effort induces a rise in pressure Pv and a pressure drop during expiration (fig. 5a).

The mean ventilator compliance Cvm can be assessed by calculating the slope of the Pv-V curve. There is a strong correlation between Cvm and the preset ventilator compliance Cvs, with Cvm = 0.97*Cvs + 0.54 (fig 5b).

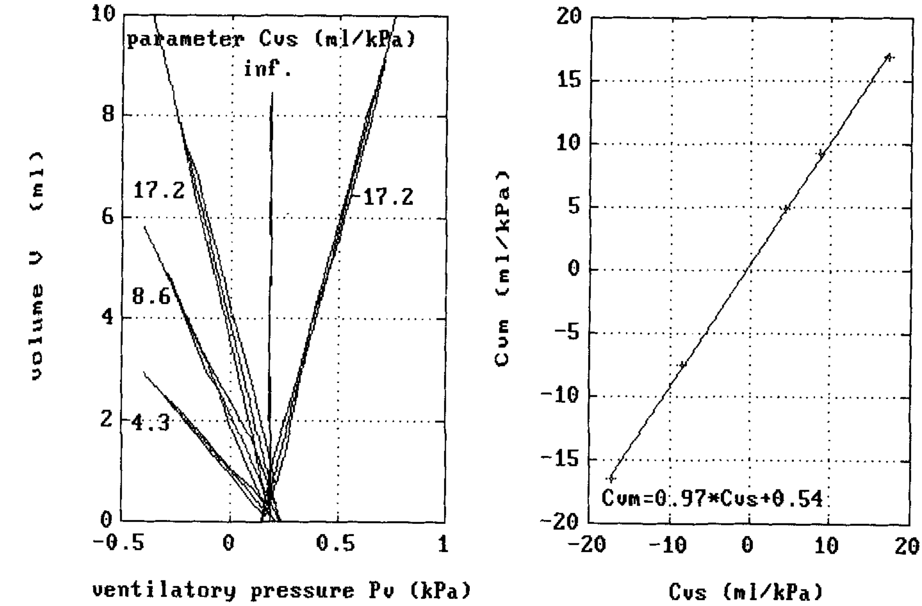


Figure 5: Volume V vs. Ventilator pressure Pv for different settings of the control knob for the ventilator compliance Cvs (fig. 5a). Measurements were done with CPAP (0.2 kPa). The slope of each PV-loop shows the measured ventilator compliance Cvm. Cvm is plotted against the ventilator's compliance Cvs (fig. 5b). The equation shows the best fit between the measured and the preset ventilator compliance.

Changing the ventilator's compliance produces changes in the total compliance of the combined lung model-ventilator system. The slope of the total compliance decreases, with Cvs = 4.3 ml/kPa (the total compliance is smaller than the lung model compliance). With Cvs equal to infinity the slope of the total compliance equals the slope of the lung model's compliance. When Cvs is equal to -8.6 ml/kPa the slope increases, which means that the total compliance is elevated (fig 6).

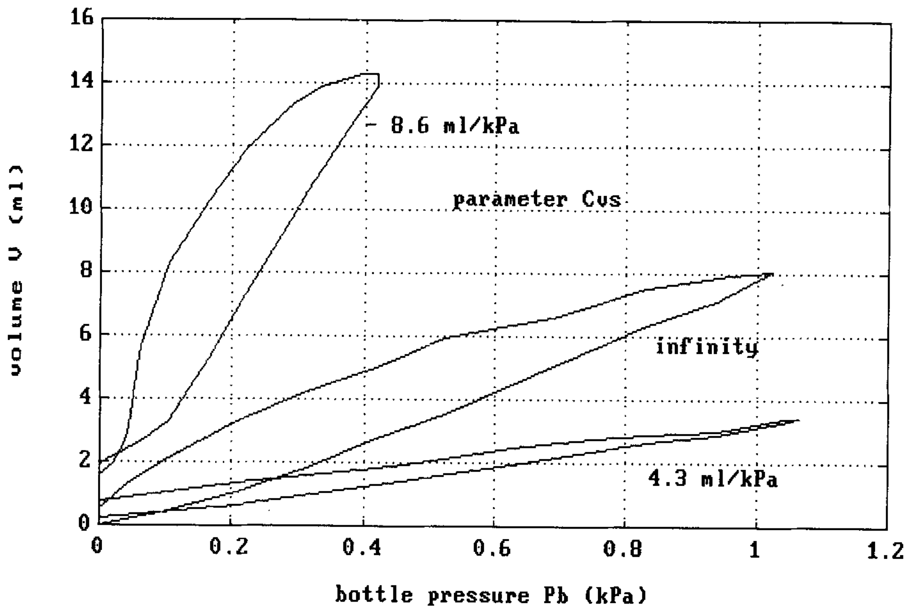


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Figure 6: PV-loops of the combined lung model-ventilator system as a whole for different settings of Cvs (-8.6 ml/kPa, infinity, 4.3 ml/kPa). The pressure Pb was measured within the bottle represents the "oesophageal pressure".

The total compliance of the combined lung model-ventilator system Ctm, measured with the PeDS unit, corresponds very well with the total compliance calculated from the baseline compliance of the lung model Clm and the preset ventilator compliance Cvs, according to equation 9, for both positive and negative values of Cvs (fig.7). The standard deviation of the relative residuals for elastic unloading is 12.4%, while that for elastic loading is 2.7%. The straight lines in figures 7c and 7d show the +/- 2 SD limits

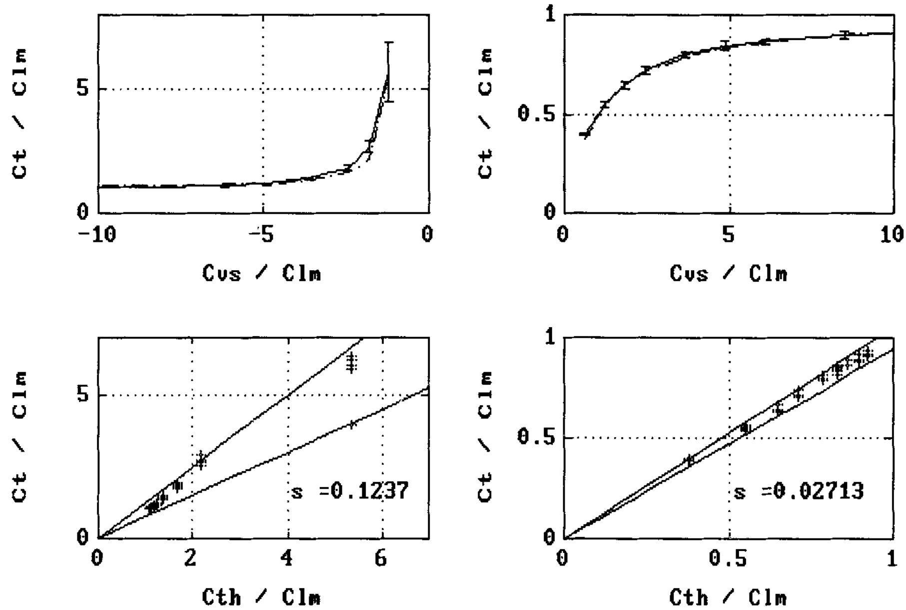


Figure 7: Upper figures: Total compliance Ct (solid line) and theoretical compliance Cth according to equation 9 (grid line) normalized for the baseline compliance of the lung model Clm against the normalized preset ventilator compliance Cvs/Clm. Negative values of Cvs (fig. 7a, left), positive values of Cvs (fig. 7b, right). Error bars show the highest and the lowest measured value for each setting of Cvs.

Lower figures: Normalized total compliance Ct plotted against the normalized theoretical compliance Cth for each experiment. Negative values of ventilator compliance (fig. 7c, left) and positive ventilator compliance (fig. 7d, right) are presented. Straight lines indicate the region of +/- 2 SD, where SD is the standard deviation of the relative residuals.

DISCUSSION

This study shows that it is possible to construct a ventilator offering either positive or negative compliance (c.f. fig 5a). The positive slope with negative Cv supports inspiration, because the ventilator's pressure Pv rises with increased inspiratory volume. During expiration the pressure Pv decreases again. With Cv equal to infinity there is no pressure change at all depending on the volume. With positive Cv, spontaneous breathing is hampered due to a fall in pressure Pv during inspiration and a rise during expiration. Thus both elastic unloading and elastic loading can be performed with this method.

There is a strong correlation between the mean ventilator compliance Cvm, as indicated by the regression line of the Pv-V loop, and the preset ventilator compliance Cvs, given by the settings of the control knob for the ventilator compliance (c.f. fig 5b).

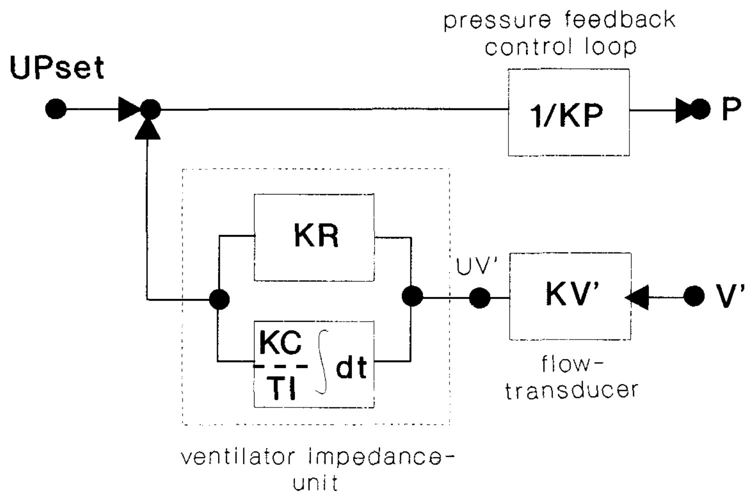
The total compliance Ct of the lung model-ventilator system can be changed by the ventilator compliance Cv. A positive Cv results in a decreased Ct and a negative Cv results in an increased Ct compared to the baseline compliance of the lung model Clm. With known model compliance and a preset ventilator compliance the total compliance can be calculated with a high degree of accuracy (c.f. fig. 7), especially with elastic loading.

It should be mentioned that stability problems arise if unsuitable values for Cv are chosen with respect to the compliance of the lung model. Small changes in Cv may induce large changes in Ct in the range where the negative ventilator compliance is approximately equal to the lung model's compliance Clm. The region where $0 < -Cvs/Clm < 1$ must be avoided because total compliance Ct attains an unstable negative value in that range. This can be dangerous because of the possible occurrence of overinflation even with small inspiratory efforts. It is therefore imperative to integrate a safety system for upper limits of tidal volume and ventilator pressure.

A similar but not identical respirator was described by Poon et al. (6, 7), Younes et al.(8, 12, 14, 15), and others from the latter group (9-11, 13). These authors used loading and unloading (both resistive and elastic) in healthy adult humans at rest and during exercise to study the effects on breathing pattern and regulation of breathing. Their system is a closed one using CO_2 absorbers within the breathing circuit, while our method is based on a strongly "open" system. Thus their respirator is not an allpurpose ventilator in the common sense, while ours offers all the possible modes of mechanical ventilation as well.

APPENDIX

A simplified block diagram is given in fig. 8 for the calculation of the values of Rv and Cv. The pressure feedback control loop (PFCL) can be characterized by a block with the transfer factor 1/KP, where KP is the transfer factor of the pressure transducer. The input of the PFCL is the voltage of the desired value and the output is the pressure. The flow is measured by a flow transducer with the transfer factor KV. Its output UV' is fed via the VIU to the input of the PFCL. The VIU contains two transfer blocks, one of which has the transfer function KR, while the other is an integrator that integrates the flow to a volume signal that has an amplification factor KC/TI (TI is the time constant of the integrator).



<u>Figure 8:</u> Block diagram for quantitative calculation of ventilation resistance (Rv) and ventilator compliance (Cv). (UPset is the control voltage for the present pressure).

If we now calculate the pressure as a function of flow we get

$$Pv = (KR*KV'/KP) * V' + KC/TI*KV'/KP * V \qquad (10)$$

Comparing this to equations 3 to 5 we can substitute for the ventilator resistance Rv

$$Rv = K1 = KR * KV'/KP \qquad (11)$$

and for the ventilator compliance

$$Cv = 1/K2 = TI/KC * KP/KV' \qquad (12)$$

As long as the transfer factors of the flow and pressure transducers KV' and KP are given, the resistance Rv can be influenced by KR and the compliance Cv by KC as well as TI.

As an example, the boundary values of Rv and Cv have been calculated for given transfer factors KP and KV' of the pressure and flow transducers and time constant TI of the integrator, with KR and KC changing between 0 and 1.

KP = 0.4 V/kPa	KR= 0...1	Rv= 0...30 kPa/l/s
KV'= 0.2 V/l/min		
TI = 0.15 s	KC= 1...0	Cv= 5... ml/kPa

Depending on the position of the switches SR and SC of fig.3 the ventilator resistance can be changed between 0 and +/-30 kPs/s/l and the compliance between +/- 5 ml/kPa and infinity.

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