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Estimation of respiratory parameters during mechanical ventilation: a simple method taking account of tubing compliance

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Abstract

A lumped parameter non linear model of a typical ventilator-patient circuit was used to simulate flow and pressure signals in a wide variety of respiratory disorders. Patient lungs were described by the series connection of a compliance (C_L = constant) and a resistance proportional to the airflow (K_L = proportionality constant). Aim of this work is to present an estimation algorithm which allows to reduce the error on the estimate of the patient respiratory parameters when using pressure and flow signals at the ventilator side instead of the patient side. The proposed algorithm is based on a simplified model of the ventilator-patient circuit, which incorporates the tubing compliance (C_{tube}) and neglects the pressure losses along the tubes during flow. C_{tube}, C_L and K_L are recursively estimated by minimizing a weighted pressure RMSE. The patient parameter estimates, thus obtained, are compared with the "true" values and those provided by the same algorithm when C_{tube} is omitted in the simplified model. Simulation results show that the respiratory circuit may significantly affect the estimate of C_L and K_L when using airflow and pressure at the ventilator side. The proposed estimation approach appears capable to greatly improve the estimate, in particular in the presence of an increased respiratory resistance.

1 Introduction

Knowledge of the respiratory viscoelastic parameters in mechanically ventilated patients is of great importance to assess status and progression of the disease and to optimize the therapy. Existing estimation methods [1] generally require the knowledge of pressure and flow values at the airway opening and therefore need the presence of measure devices near to the patient mouth. In Intensive Care Unit the introduction of further instrumentation at the patient bed can be an impediment to the clinical work and should be avoided. As a consequence often it should be preferred to collect pressure and flow signals directly at the ventilator. In this way, however, the signal distortion introduced by the respiratory circuit is neglected. The present simulation study shows that, sometimes, this choice may lead to a wrong estimate of the patient respiratory parameters. Aim of this work is to analyze when the estimation error is not negligible; moreover a simple algorithm is proposed in order to reduce this error in patients with high respiratory resistance.

2 Model description

A typical ventilator-patient circuit is simulated, consisting of three parts, namely (1) a mechanical ventilator, which supplies the respiratory flow and controls the inspiration and expiration valves, (2) the tubing system, consisting of the inspiration and the expiration leg, and (3) the patient.

The electrical analogue of the complete model for the ventilator-patient circuit is shown in figure 1.

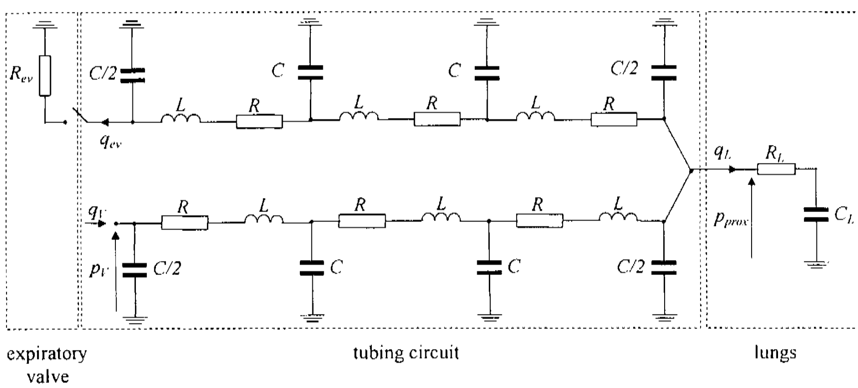

Figure 1: Model of the ventilator-patient circuit

2.1 Mechanical ventilator

The mechanical ventilator is modelled by an ideal flow generator. The transition from the inspiratory to the expiratory phase, and vice versa, is caused by the opening and the closing of the ventilator expiratory valve. In the present study an ideal occlusion device is considered, i.e. with instantaneous closing time and complete tightness in the closed state. Therefore the valve is described by a resistance (R_{ev}) connected with a switch. According to Valta et al. [2], the valve

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resistance includes a term proportional to the flow (K_{ev1} = proportionality constant) and a constant term (K_{ev2}), i.e.:

$$R_{ev} = K_{ev1}|q_{ev}| + K_{ev2} \tag{1}$$

where q_{ev} is the flow coming out of the expiratory valve. The signal (ev) controlling the switch is represented by a square wave: when the switch is on (ev=1) the expiration is allowed and the airflow proceeds from the respiratory circuit to the atmosphere through R_{ev}, when the switch is off (ev=0) the expiration is stopped and the airflow inflates the tubes and the patient lungs. The flow through the valve is:

$$q_{ev} = ev\frac{p_{ev}}{R_{ev}} \tag{2}$$

where p_{ev} is the pressure at the end of the expiration leg.

2.2 Tubing system

The airflow dynamics in the tubes is modelled using a lumped parameter description. Both the inspiratory and the expiratory leg are partitioned into three finite segments, each one described by a two-port network [3]. Using the π-configuration, shown in the central part of figure 1, the most significant characteristics of the breathing circuit are taken into account: the compliance of the tube, the resistance to the airflow and the inertia of the air. The model does not consider the air compression in the tubing circuit; in fact, pressure changes along the circuit remain low during routine mechanical ventilation.

The compliance (C) of each segment is held constant, according to [4], while the inertance is assumed inversely proportional to the tube internal section and directly proportional to the air density and to the segment length.

In accordance with previous experimental observations, the resistance of each segment is modelled by a non-linear "Rohrer resistor" [5] characterised by the following relation:

$$R_i = K_1|q_{rli}| + K_2 \tag{3}$$

where K_1 and K_2 are constant terms, while q_{rli} is the flow through R_i. The pressure drop over R_i is characterised by two components: a laminar flow term ($K_2\, q_{rli}$) depending on the viscosity of the gas and the geometry of the tube, and a resistance term ($K_1\, |q_{rli}|\, q_{rli}$) taking account of the airflow turbulence.

Each segment is thus described by two differential equations:

$$\frac{dp_i}{dt} = \frac{q_i - q_{rli}}{C/2} \tag{4}$$

$$\frac{dq_{rli}}{dt} = \frac{p_i - R_i q_{rli} - p_{i+1}}{L} \tag{5}$$

where p_i and q_i are the pressure and the flow at the i-th segment entrance, respectively.

2.3 Patient lungs

The computer simulation of lung mechanics is based on the description of patient lungs by a non-linear one-compartment model with lumped parameters. The model consists of the series connection of a resistance (R_L) and a compliance (C_L), where C_L is the total respiratory compliance and describes the elastic properties of lungs and chest wall, while R_L is the total respiratory system resistance and describes the flow resistance of both airway and pulmonary tissues. R_L was assumed proportional to the airflow (K_L = proportionality constant), i.e.:

$$R_L = K_L |q_L| \tag{6}$$

where q_L is the airflow entering the patient airway opening.

The lung model equations result:

$$q_L = \frac{p_{prox} - p_L}{R_L} \tag{7}$$

$$\frac{dp_L}{dt} = \frac{q_L}{C_L} \tag{8}$$

where p_{prox} and p_L are the pressure at the airway opening (proximal pressure) and the lung pressure, respectively.

3 Model validation

The mathematical model was implemented on a personal computer using the software packages MATLAB® (Version 5.3 release 11, MathWorks, Inc., USA) and Simulink® (Version 3, MathWorks, Inc., USA) for the integration of differential equations. The Runge-Kutta algorithm has been selected to numerically solve the model equations in the time domain.

To verify the model performance, a laboratory set-up was realised, consisting of a ventilator (Evita2dura, Dräger, Lübeck, Germany) connected via a conventional tubing system (DAR, Mallinckrodt, USA) to a training/test lung (TTL, METRON, Trondheim, Norway). The ventilator was set in the controlled ventilation mode with constant flow inflation. The ventilator settings were: tidal

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volume 0.5 l, inspiratory flow 0.5 l s^{-1}, I:E ratio 1:2 and respiratory rate 15 min^{-1}. Several clinical situations were reproduced by varying the tester parameters. The compliance was assumed equal to 0.03, 0.05, and 0.07 l cmH_2O^{-1}, while the TTL resistance was modified using different combinations of resistors (PneuFlo® Airway Resistors Rp5, Rp10, and Rp20, where Rp10 corresponds to two Rp5 resistors in series).

Estimation of model parameters was achieved by minimising a least-square criterion function of the differences between experimental data and model predictions. The criterion function was obtained as follows. In all the experimental tests, the Root Mean Square Error (*RMSE*) was calculated for both pressure and flow signals, measured at the ventilator side and at the airway opening. Each *RMSE* was then expressed in percent of a reference value ($\bar{f}$), i.e.:

$$RMSE_f\% = \frac{1}{\bar{f}}\sqrt{\frac{\sum_{k=1}^{N}\left[f_k - f_{s,k}\right]^2}{N}} \tag{9}$$

where N was the number of samples, and f and f_s were the experimental and the simulated signals respectively. For the pressure signals the reference value was chosen equal to the peak value, for the flow signals, instead, $\bar{f}$ was chosen equal to the mean inspiratory flow, in order to reduce the influence of possible measurement errors in the negative airflow peak. Finally, for each experimental test, we computed the mean value of the five $RMSE_f\%$. The value adopted for each model parameter was the mean of the values found in the experimental sessions. Model predictions of pressure and airflow, both at the patient and at the ventilator side, were in a good agreement with the corresponding measurements, being the mean *RMSE%* lower than 6% in all cases.

4 Simulation studies

The model was employed to reproduce a volume-controlled ventilation with constant flow inflation, passive expiration and post-inspiratory pause with airflow equal to zero. To evaluate the influence of the tubing system on the pressure and flow signals, each simulation was repeated for three different values of the tubing compliance (0.25, 0.5, and 1 ml cmH_2O^{-1} per meter length of tube), chosen according to the literature values [4]. A conventional 15 mm internal diameter tubing circuit was selected to represent commonly used non-disposable systems; the total length of tubes was held constant to 180 cm on both the inspiratory and the expiratory limbs. The values adopted for the model parameters concerning the ventilator and tubing system are shown in Table 1.

To reproduce a wide variety of respiratory disorders, K_L was changed between 8 and 90 $cmH_2O\ s^2\ l^{-2}$, while the respiratory compliance was varied from 0.02 to 0.09 l cmH_2O^{-1}.

Patient parameters (K_L and C_L) were estimated starting from the simulated pressure and airflow signals at the ventilator side instead of at the patient side. In order to limit the influence of the non linear effects of the tubing resistance and of the ventilator expiratory valve, all the estimations were performed using only the inspiratory data. The estimation procedure was repeated twice, first using the lung model (right side of figure 1), then using an algorithm based on a simplified model of the ventilator-patient circuit (see below).

Table 1: Parameter values for the ventilator and tubing system

Parameter		Value	Units
C		0.0006	l cmH_2O^{-1}
		0.0003	l cmH_2O^{-1}
		0.00015	l cmH_2O^{-1}
L		0.04	cmH_2O s^{-2} l^{-1}
R:	**K_1**	0.6	cmH_2O l^{-2} s^2
	K_2	0.0892	cmH_2O l^{-1} s
R_{ev}:	**K_{ev1}**	1.21	cmH_2O l^{-2} s^2
	K_{ev2}	1.24	cmH_2O l^{-1} s

4.1 Parameter estimation using the lung model

The simulator was first used to study the extent to which the patient parameter estimates are affected by the use of pressure and flow data at the ventilator. To this purpose, the patient parameters were first estimated by minimising the following criterion function [6]:

$$F(\theta) = \sum_{k=1}^{N} w(k)\left[p_{ev}(k) - p_m(k,\theta)\right]^2 \tag{10}$$

where $p_{ev}(k)$ is the simulated pressure coming out of the expiratory leg and $p_m(k,\theta)$ is the pressure provided by the lung model (right side of figure 1) with the parameter values, $\theta = [K_L, C_L]$, and the simulated airflow at the ventilator side (q_{LV}) as input. Therefore $p_m(k,\theta)$ is defined as follows:

$$p_m(k,\theta) = K_L\left|q_{LV}(k)\right|q_{LV}(k) + \frac{1}{C_L}\sum_{j=1}^{k} q_{LV}(j)T \tag{11}$$

where q_{LV} is given by the difference between the flow supplied by the ventilator (q_V) and the flow at the outlet of the expiratory limb (q_{ev}), and T is the sampling interval. $w(k)$ is a weighting factor chosen in order to better reproduce the proximal pressure plateau and, consequently, the lung pressure plateau too. In

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fact, a bad reproduction of p_L may lead to a wrong estimate of C_L (see eqn (8)) and, therefore, of R_L.

4.2 Parameter estimation using the proposed algorithm

In the second part of the work, the estimation model was derived as a simplified version of the ventilator-patient circuit shown in figure 1. The resulting model (figure 2) incorporates the compliance of the inspiratory and expiratory leg (C_{tube}), while neglects the pressure losses along the tubes during flow. A linear version of the same model (with constant R_L) was previously proposed to interpret data measured in ventilated patients [6]. Estimation of C_{tube}, K_L, and C_L was carried out by adopting a recursive procedure based on the weighted least-squares cost function shown in eqn (10), where $p_m(k,\theta)$ is the pressure provided by the proposed model (figure 2) with the parameter values, $\theta = [K_L, C_L, C_{tube}]$, and q_{LV} as input. Simulation studies have demonstrated that, in this case, a weighting factor different from one does not significantly modify the estimate, therefore $w(k)$ can be omitted.

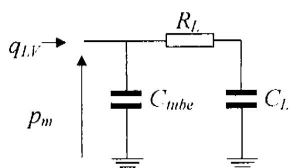

Figure 2: Three-element estimation model.

For both the estimation procedures, the accuracy of the estimates was assessed by computing the percentage coefficients of variation of K_L ($CV_{KL}[\%]$) and C_L ($CV_{CL}[\%]$) and the absolute value of the per cent error. Where the per cent errors for the i-th simulation are:

$$eK_{L,i}[\%] = 100\frac{\left|K_{L,i} - \hat{K}_{L,i}\right|}{K_{L,i}}, \quad eC_{L,i}[\%] = 100\frac{\left|C_{L,i} - \hat{C}_{L,i}\right|}{C_{L,i}} \tag{12}$$

where $K_{L,i}$ and $C_{L,i}$ are the patient resistance proportionality constant and the patient compliance, respectively, set in the numerical model during the i-th simulation ("true values"), while $\hat{K}_{L,i}$ and $\hat{C}_{L,i}$ are the same parameters estimated using the algorithm described above.

Furthermore the capacity to reproduce the real proximal pressure (p_{prox}) was evaluated by computing the root-mean-square error per cent ($RMSE\%$) defined as:

$$RMSE\% = \frac{100}{p_{proxpeak}} \sqrt{\frac{\sum_{k=1}^{N} \left[p_{prox}(k) - p_m(k,\theta)\right]^2}{N}} \tag{13}$$

where θ is the parameter vector, N is the number of simulation points, p_m is the pressure provided by the considered estimation model, and $p_{proxpeak}$ is the peak value of p_{prox}.

5 Results

Simply using the lung model with pressure and airflow at the ventilator side rather than the signals at the airway opening, the reproduction of the proximal pressure was fairly satisfactory, also assuming w=1. In fact the *RMSE%* was lower than 15% in all cases; in particular the error increased in the presence of a higher respiratory resistance. Also the coefficients of variation were always very good, being CV_{KL}[%] and CV_{CL}[%] lower than 2.9% and 4.1%, respectively. However, the parameter estimates were biased with respect to the "true" values, due to the signal distortion introduced by the patient-ventilator circuit. This effect on the lung parameter estimates was shown by the absolute value of the percent error in the estimate of K_L and C_L, i.e. eK_L[%] and eC_L[%], respectively.

An example of the results obtained considering a tubing compliance equal to 0.5 ml cmH_2O^{-1} per meter length of tube is shown in figure 3, where part (a) concerns the algorithm based on the lung model, while part (b) concerns the proposed algorithm. Each group of vertical bars corresponds to a different value of C_L ("true value"), while each bar corresponds to a different value of K_L increasing from left to right. As it is clear from the two graphs in figure 3 (a), the lung resistance is generally estimated worse than the lung compliance for low values of K_L, while the per cent errors are quite similar when considering high values of K_L. Moreover, results obtained with the other two values of tubing compliance show that the K_L and C_L estimate improves using a lower tubing compliance. The maximum value reached by eK_L[%] and eC_L[%] is, in fact, equal to 61.1% and 61.2%, respectively, for the highest value of tubing compliance, while it is equal to 22.7% and 37.8%, respectively, for the lowest value. Simulation studies show that assuming w=4 during the last part of the post-inspiratory pause, allows to partially reduce the error on the estimate of C_L, while it does not significantly improve the evaluation of R_L. In contrast, considering figure 3 (b) compared with figure 3 (a), it can be noticed that the proposed algorithm greatly improves the estimate of both K_L and C_L (see the different scales). In all cases examined, in fact, eK_L[%] and eC_L[%] are lower than 6.3% and 1.2%, respectively. Moreover the accuracy of the estimation remains good (CV_{KL}[%] < 1%, CV_{CL}[%] < 1.8%) and the proximal pressure is better reproduced (*RMSE%* < 0.5%).

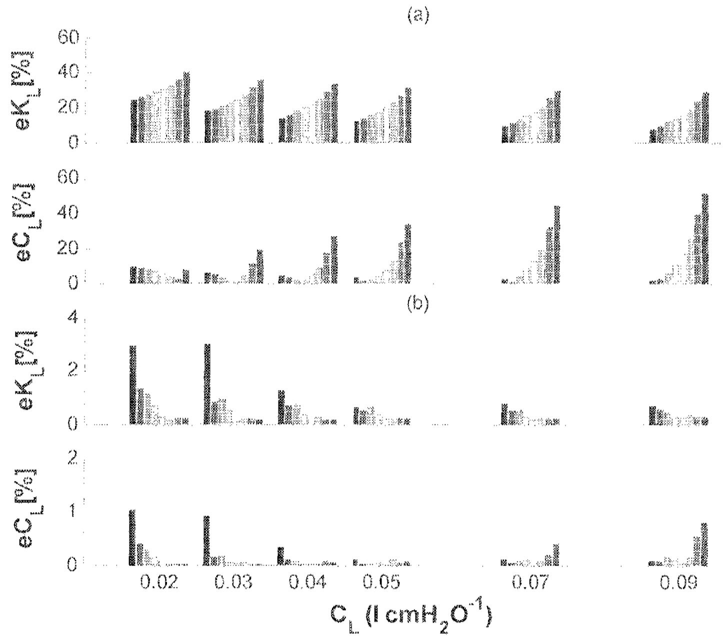

Figure 3: Absolute values of the percent error in the estimate of K_L (eK_L[%]) and C_L (eC_L[%]), obtained with the algorithm based on the simple two-elements model (a) and with the proposed algorithm which considers the compliance C_{tube} (b). The respiratory parameters were estimated starting from airflow and pressure at the ventilator, simulated with a tubing compliance equal to 0.5 ml cmH_2O^{-1} per meter length of tube. Each bar corresponds to a different value of K_L (8, 16, 24, 32, 40, 50, 70, 90 $cmH_2O\ l^{-2}\ s^2$) increasing from left to right.

6 Discussion and conclusion

This study explores by simulation the effects of the respiratory circuit on the pressure and flow signals. In particular, focus is on the error affecting the estimation of lung parameters when the tubing influence is neglected and the pressure and flow signals at the airway opening are assumed equal to the corresponding signals measured at the ventilator. During inspiration part of the volume of gas provided by the ventilator is used to distend the tubing system and therefore does not reach the patient [4]. The amount of gas lost into the tubes principally depends on the tubing compliance and on the inflation pressure. As it has been observed during simulation studies, when the tubing compliance increases, the flow at the airway opening raises more slowly and reaches a lower constant value with respect to the flow provided by the ventilator. Moreover, in the presence of a high K_L the pressure drop over the respiratory resistance increases and the difference between p_L and p_{prox} becomes greater; as a consequence a larger amount of air inflates the tubes. As it clearly appears from figure 3 (a), the sum of these effects may lead to non negligible errors in the estimation of lung parameters. The compliance (C_{tube}) included in the three-element model of figure 2 allows to take into account the respiratory circuit influence and the consequent airflow shape alteration. The proposed algorithm significantly improves the estimates (see figure 3 (b)), although it is successfully applicable only to patients with an increased airway resistance ($K_L > 8$ cmH_2O s^2 l^{-2}). In fact, in the presence of a low value of K_L, the two compliances are nearly in parallel and the simplified model becomes difficult to identify. In these cases the estimation results may remarkably depend on the starting value of parameters and the algorithm based on the simple two-element lung model may be preferred. Therefore, when signals at the airway opening are not available, the respiratory circuit components should be selected with great care in order to minimise the tubing influence.

In conclusion, the proposed algorithm could be a valid tool in Intensive Care Unit since it allows to estimate the respiratory parameters of patients with obstructive diseases without the need for special manoeuvres or the addition of measurement instruments at the patient bed.

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