VENTILATOR SETTING DEPENDENCE OF PARAMETERS FOR DETECTING EXPIRATORY FLOW LIMITATION IN MECHANICAL VENTILATION: A SIMULATION STUDY

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Abstract: The present study examines, through simulation, four parameters proposed in literature to discriminate between normal and obstructive conditions, i.e. the three time constants calculated from flow-volume curves at the last 75%, 50%, and 25% of the expiratory tidal volume and the portion of tidal expiration over which there is no change in flow when a negative expiratory pressure is applied to the expiratory circuit outlet (flow-limited portion, FLP). Aim of the work is to evaluate the ability of these parameters to detect and quantify expiratory flow limitation (EFL) in mechanically ventilated patients, paying particular attention to how parameters are influenced by a change in the tidal volume or in the exhalation valve resistance. Respiratory mechanics of artificially ventilated patients has been reproduced using a non linear and non-homogeneous dynamic morphometric model of the tracheobronchial tree, incorporating the main mechanisms limiting expiratory flow. This study shows that the three time constants and FLP could be a valid tool to detect severe EFL, while only FLP seems suitable in presence of partial EFL. Moreover the considered parameters appear suitable to quantify the degree of EFL, even though they can be affected to some extent by changes in ventilator setting.

Introduction

The importance of detecting and quantifying an obstructive pathological condition in mechanically ventilated patients is well known, especially when the pathology leads to the expiratory flow limitation (EFL) phenomenon [1, 2]. Four parameters have been proposed in literature to discriminate between normal and pathological conditions: the time constants calculated from flow-volume curves when the 75%, 50%, and 25% of tidal volume remains to be exhaled (RCfv75, RCfv50, and RCfv25, respectively) [1] and the portion of tidal expiration over which there is no change in flow when a negative expiratory pressure (NEP) is applied to the outlet of the expiratory circuit (flow-limited portion FLP) [2]. However, these four parameters are generally computed in different experimental conditions, since the mechanical ventilator employed and/or the ventilator set-up may vary from patient to patient. Aim of this work is to examine, through simulation, the four above-mentioned parameters in normal and obstructive conditions, paying particular attention to how they are influenced by a change in the tidal volume or in the exhalation valve resistance. In order to reproduce the respiratory mechanics of artificially ventilated patients with both homogeneous and non-homogeneous obstructive pulmonary conditions, we used a modified version [3] of a non linear, dynamic, and homogeneous model recently proposed [4].

Methods

Simulation model: The employed mathematical model [3] is based on the Weibel's dichotomous representation of the tracheobronchial tree. In the model each main bronchus (generation 1) is represented separately and forks into two distinct regions of the same number of branches. In each region all branches of the same generation are assumed identical, allowing the description of each region to be simplified as done in the previous work [4]. Since the two main bronchi and the four regions can be differently characterized, this model allows mechanical non-homogeneities of the lungs to be reproduced. Moreover it is non linear, dynamic, and incorporates the main mechanisms limiting expiratory flow (wave speed limitation and viscous flow limitation).

The model, including a representation of the mechanical ventilator, has been extensively described in previous works [3, 4]; here we report the expression of the resistance of the expiratory circuitry:

$$R_{ec} = K_{1ec} + K_{2ec}|q| \tag{1}$$

where q is the tracheal flow and K_{1ec} and K_{2ec} are constant parameters taking account of resistance laminar and turbulent components, respectively. K_{1ec} and K_{2ec} have been modified in the simulations to reproduce different exhalation valves.

Patient simulation setting: In this study we simulated four respiratory conditions: normal (a), partially obstructed (b), and severely obstructed (c and d). Normal respiratory condition (a) was simulated using transmural pressure-diameter curves similar to the curves proposed by Lambert et al. [5] and assuming the

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resistance and the compliance of the respiratory zone of each region equal to 6 cmH_2O s/l and 0.02 l/cmH_2O, respectively. The chest wall compliance was set equal to 0.1 l/cmH_2O [3]. To simulate the obstructed conditions we doubled the resistance of the respiratory zone and decreased the airways section of generations from 8 to 16 (b and c) or from 5 to 16 (d), in all the four regions (c and d) or in two regions only (b). In particular we reduced the maximum diameter of generations from 8 to 16 of 50% of its normal value in cases b and c, while in case d we reduced the maximum diameter of generation from 5 to 11 and of generations from 12 to 16 of 25% and of 50% of its normal value, respectively. Moreover in all pathological cases we modified the transmural pressure-diameter curves of generations, from 8 to 16 (cases b and c) or from 4 to 16 (case d), in all the lungs to account for an easier collapsibility of these airways (see Figure 1).

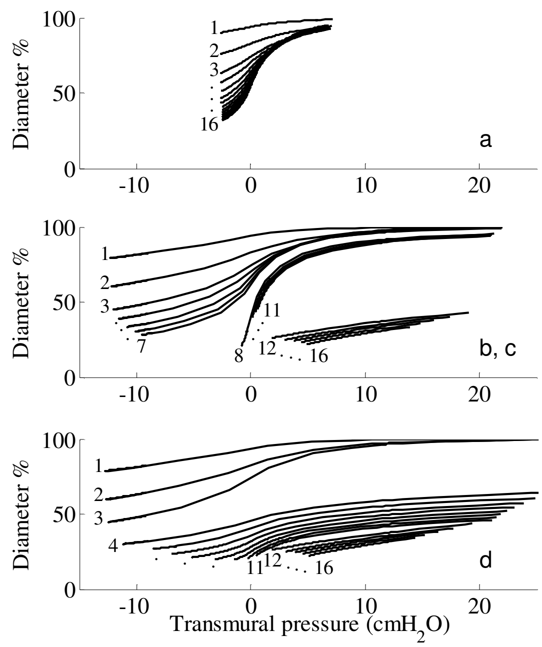

Figure 1: Branch diameter, as a percentage of its maximum value, plotted against transmural pressure for airways of the conductive zone of a generic region. Top: normal case (a); middle: partially obstructed (b) and severely obstructed (c) case; bottom: severely obstructed case (d).

Ventilator simulation setting: All cases were first simulated using the ventilator setting proposed in [3], assumed as basic setting, i.e. tidal volume (V_T) of about 570 ml, respiratory frequency of 15 breath/min, K_{1ec} and K_{2ec} equal to 0.5 cmH_2O s/l and 8.7 cmH_2O s^2/l^2, respectively. All cases were then simulated changing ventilator parameters, as follows: (1) decreasing (K_{1ec} and K_{2ec} equal to 0.36 cmH_2O s/l and 1.8 cmH_2O s^2/l^2, respectively [2]) or increasing (K_{1ec} and K_{2ec} equal to 1 cmH_2O s/l and 15 cmH_2O s^2/l^2, respectively, experimental result not shown) the exhalation valve resistance; (2) decreasing or increasing of 25% the tidal volume (minute volume and I:E ratio being equal to the basic setting). When considering the expiratory flow ranging between 0 and 1.2 l/s, the modification of parameters K_{1ec} and K_{2ec} changes the mean value of the exhalation valve resistance of about 75% of the mean value obtained using the basic ventilator setting.

Parameter calculation: In all the performed simulations we evaluated the time constants RCfv75, RCfv50, and RCfv25, using the following expressions:

$$\mathrm{RCfv75} = 0.75 * V_T / (q_{75} - q_{end}) \tag{2}$$

$$\mathrm{RCfv50} = 0.5 * V_T / (q_{50} - q_{end}) \tag{3}$$

$$\mathrm{RCfv25} = 0.25 * V_T / (q_{25} - q_{end}) \tag{4}$$

where q_{75}, q_{50}, and q_{25} are the tracheal flow at the last 75%, 50%, and 25% of exhaled volume, respectively, and q_{end} is the tracheal flow at end-expiration [1].

Moreover we simulated the application of a NEP of -5 cmH_2O to the outlet of the expiratory circuit during a test cycle and we computed the flow limited portion FLP, expressed as:

$$\mathrm{FLP} = 100 * V_{FLP} / V_T \tag{5}$$

where V_{FLP} is the portion of tidal expiration over which there is no change in tracheal flow with NEP [2]. At the same expiratory volume, the expiratory flow in the test cycle was assumed identical to the flow in the preceding control cycle, when the flow variation is lower than 3% of the peak expiratory flow of the control cycle.

Results

Figure 2 shows the expiratory flow-volume curves in the four cases under study when using the basic ventilator setting. Test curve (broken line), obtained using the NEP technique, is compared by superimposition with the control curve (continuous line) obtained during the preceding breathing cycle. Symbols represent the point corresponding to q_{75} (*) q_{50} (x), q_{25} (○) and to the beginning of the curves superimposition (●). In cases c and d the two curves are superimposed, except for the initial flow peak, thus indicating the presence of expiratory flow limitation in most of expiration. In case b, instead, superimposition interests only the final part of expiration, corresponding to middle-low volume. Varying the exhalation valve resistance, the flow-volume curves of cases c and d remain nearly unchanged, except for the peak expiratory flow, in accordance with Lourens et al. [6]. In cases a and b it can be noted a modification of the curve shape in the first part of expiration, corresponding to middle-

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high volumes (data not shown). The tidal volume modification, instead, produces a change in volume range and, in the severely obstructed cases, it modifies the volume at end-expiration. The changed expiratory flow-volume curves when decreasing or increasing tidal volume are shown in Figure 3.

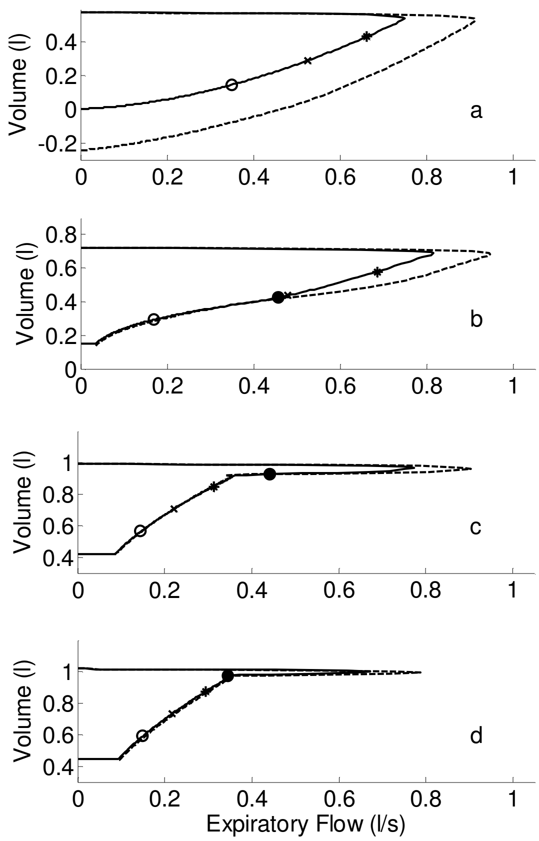

Figure 2: Expiratory flow-volume curve of test cycle (broken line) and preceding control cycle (continuous line) obtained by NEP technique using basic ventilator setting.

Figure 4 shows the three time constants for the different cases computed using the basic ventilator setting (symbol) and the interval determined by the maximum and the minimum parameter value obtained with the modified ventilator settings. In cases a and b, time constants are similarly influenced by the changes in tidal volume or in the exhalation valve resistance; the highest parameter values are reached using the highest resistance, except for case b where the highest RCfv25 value is reached using the highest tidal volume.

Finally FLP values are reported in Figure 5. They show a significant modification only in case b, where FLP increases when decreasing the exhalation valve resistance or, in a minor way, when decreasing tidal volume.

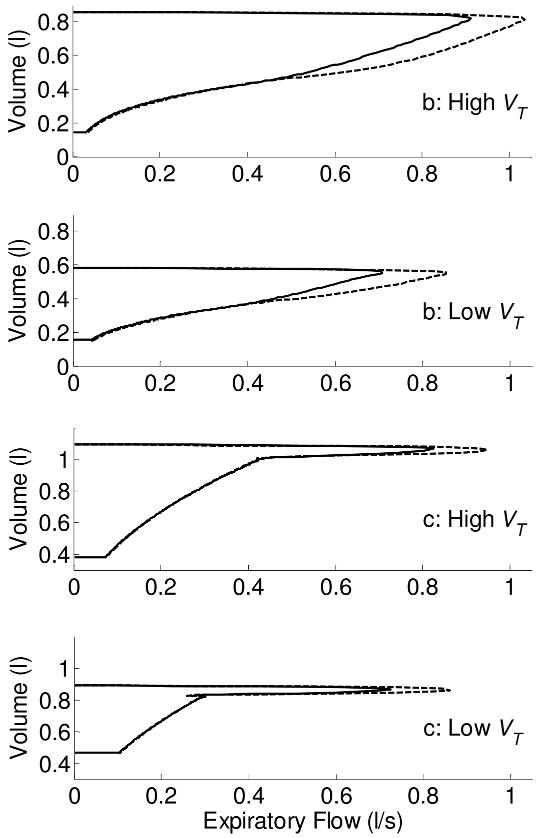

Figure 3: Expiratory flow-volume curve of test cycle (broken line) and preceding control cycle (continuous line) obtained by NEP technique in partially obstructed (b) and severely obstructed condition (c). High V_T: increased tidal volume; Low V_T: decreased tidal volume.

Discussion

The NEP technique shows the presence of EFL in cases b, c, and d (Figure 2), in accordance with patient simulation setting. Despite all the examined parameters appear able to discriminate between normal and severely obstructive conditions, only RCfv25 and FLP can distinguish between normal and partially obstructive condition (Figures 4 and 5, symbols). This may be explained by the fact that expiratory flow limitation is evidenced by the NEP manoeuvre only after the 50% of the volume has been exhaled (see Figure 2, case b).

A variation in ventilator setting changes the working point of the subject, but it is difficult to predict its effect on flow-volume curves, because of the high complexity of the respiratory system. For example, an increase in tidal volume with constant minute volume and I:E ratio implies no change in end-expiratory volume in cases a and b, while determines a lower end-expiratory volume in cases c and d (e.g. Figure 2 and 3, cases b and c).

The simulation results show that, despite a modification in the flow-volume curves, the flow limitation condition is not substantially affected by the ventilator setting, especially in the most severely

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obstructive cases. In fact the ability of the considered parameters to discriminate between normal and pathological cases seems independent of the ventilator setting (Figure 4 and 5). However, especially in cases a and b, parameters change with ventilator setting. In cases a and b the maximum variation from values obtained with basic setting is about 20% and 15%, respectively. In cases c and d, instead, parameter variations appear more moderate, being always lower than 5%. These changes of parameters with ventilator setting should not be neglected by the clinician if his purpose is not only to discriminate between normal and EFL condition, but also to quantify the degree of flow limitation.

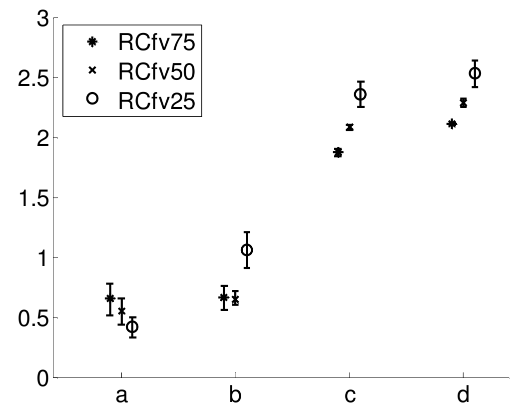

Figure 4: Time constants (in s) calculated from the flow-volume curves for the four examined cases. See text for abbreviation and definition.

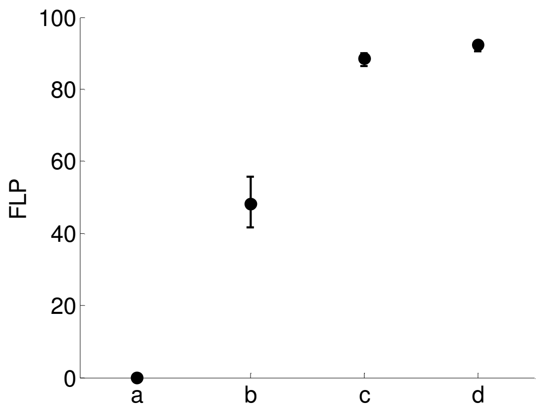

Figure 5: Flow limited portion (FLP), expressed as a percentage of tidal volume, for the four examined cases.

Conclusions

This study shows that FLP and the three RCfv may be a valid tool to detect the presence of severe EFL, while RCfv75 and RCfv50 seem insufficient when considering subjects with a partial EFL. FLP and the three time constants appear suitable also to quantify the degree of flow limitation, within a confidence interval depending on the ventilator setting. Of course, when considering experimental tests instead of simulation tests, further aspects have to be taken into account, such as noise and measurement errors. Since the flow values involved in RCfv25 computation are low, this time constant is probably affected significantly by both these complications. At the contrary, FLP calculation should be less influenced by noise and measurement errors, even though it requires an external manoeuvre and its value critically depends on when the flow is assumed equal in test and control curves.

The presented preliminary simulation results encourage to proceed with an appropriate clinical validation.

Acknowledgments

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