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G. Bonmarchand
V. Chevron
C. Chopin
D. Jusserand
C. Girault
F. Moritz
J. Leroy
P. Pasquis

Increased initial flow rate reduces inspiratory work of breathing during pressure support ventilation in patients with exacerbation of chronic obstructive pulmonary disease

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G. Bonmarchand (✉) · V. Chevron
D. Jusserand · C. Girault
F. Moritz · J. Leroy
Intensive Care Unit, Hôpital Charles Nicolle, University of Rouen, 1, rue de Germont, F-76031 Rouen, France
FAX: + 33/3508 8128

P. Pasquis
Department of Physiology, Hôpital Charles Nicolle, University of Rouen, 1, rue de Germont, F-76031 Rouen, France

C. Chopin
Intensive Care Unit, University of Lille, F-59037 Lille, France

Abstract *Objective:* To investigate whether the level of initial flow rate alters the work of breathing in chronic obstructive pulmonary disease (COPD) patients ventilated in pressure support ventilation (PSV).
Design: Prospective study.
Settings: Medical ICU in University hospital.
Patients: Eleven intubated COPD patients.
Methods: We modulated the initial flow rate in order to achieve seven different sequences. In each sequence, the plateau pressure was reached within a predetermined time: 0.1, 0.25, 0.50, 0.75, 1, 1.25 or 1.50 s. The more rapidly the pressure plateau was achieved, the higher was the initial flow rate. In each patient, the pressure support level was an invariable parameter. The order of the seven sequences for each patient was determined randomly.
Measurements and results: Ten minutes after application of each initial flow rate, we measured the following parameters: inspiratory work of breathing, electromyogram (EMG) of the diaphragm (EMGdi), breathing pattern, and intrinsic positive end-expiratory pressure (PEEPi). Comparison between the means for each sequence and each variable measured was performed by two-way analysis of variance with internal comparisons between sequences by Duncan's test. The reduction of the initial flow rate induced a progressive increase in the values of the work of breathing, EMGdi, and mouth occlusion pressure (P 0.1). In contrast, the reduction of the initial flow rate did not induce any significant change in tidal volume, respiratory frequency or PEEPi.
Conclusion: As the objective of PSV is to reduce the work of breathing, it seems logical to use the highest initial flow rate to induce the lowest possible work of breathing in COPD ventilated patients.

Key words Pressure support ventilation · Initial flow rate · Inspiratory pressure rise time · Work of breathing · COPD

Introduction

Pressure support ventilation (PSV) is a mode of assisted ventilation used in the initial treatment of patients with acute respiratory failure and for weaning mechanically ventilated patients [1, 2]. PSV acts by increasing the efficacy of each spontaneous breath and by reducing inspiratory muscle activity [3]. The current settings for this mode of ventilation essentially concern the pressure support level, as the higher the support level, the lower the work of breathing performed by the patient [4, 5]. Moreover, the pressure support level is the only possible adjustment with the

majority of ventilators. The inspiratory rise time initially used to achieve the desired pressure support is usually preset by the ventilator design, corresponding to an initial flow rate ranging from 58 to 100 l/min [6].

This fixed setting is questionable, as two recent studies have demonstrated that variations in the initial flow rate could induce modifications in the breathing pattern [7, 8]. These studies were conducted on heterogeneous patient groups without measurement of the work of breathing. We conducted this study to assess the influence of initial flow rate variations in a homogeneous group of COPD patients and to measure the work of breathing by Campbell's diagram method.

Material and methods

Patients

We studied 11 COPD patients (nine men) who were admitted to our Intensive Care Unit because they required ventilatory support. All were intubated (Portex cuffed endotracheal tube, internal diameter 7.5 mm) and mechanically ventilated with volume-controlled mode (CMV) using constant inspiratory flow (Evita Dräger 1, Drägerwerk AG, Lübeck, Germany). At the time of this study, all patients were clinically stable and considered to be ready for weaning by the primary physicians. The diagnosis of COPD was established by clinical history, chest X-rays and physical examination combined, in seven patients, with previous pulmonary function tests. The precipitating cause of acute respiratory failure was acute pneumonia in five cases and was thought to be primarily related to respiratory tract infections in six cases. The patients were studied, at the bedside in the Intensive Care Unit, in a semi-recumbent position, the upper half of the bed tilted to 45°. A physician not involved in the procedure was always present to provide patient care.

The clinical and respiratory characteristics of the patients are listed in Table 1. The measurements of static respiratory system compliance ($C_{st,rs}$), inspiratory resistance (RI, aw), and intrinsic positive end-expiratory pressure (PEEPi,stat) were evaluated under static conditions just prior to the study. $C_{st,rs}$ was calculated as the ratio between tidal volume (V_T) and the difference between the end-inspiratory plateau pressure obtained after more than 2 s of airway occlusion and PEEPi,stat. RI,aw was obtained by dividing the difference between peak airway pressure and plateau pressure by the inspiratory flow provided by the ventilator. As the inspiratory flow was constant, it was calculated from volume and inspiratory time. PEEPi,stat was measured as the plateau level of airway pressure reached within 0.5–1.5 s of expiratory occlusion at the end of exhalation.

The protocol was approved by the local ethics committee and all patients gave their written informed consent.

Apparatus and calculations

All patients were put on the same type of commercially available ventilator: Evita Dräger 1, a microprocessor-controlled ventilator equipped with software no. 11 (Drägerwerk AG, Lübeck, Germany). In PSV, this ventilator has an adjustable gradient of pressure rise. The demand valve therefore modulates the initial flow rate according to the gradient chosen, the inspiratory resistance, and the patient's inspiratory activity. The pressure plateau is reached after a variable interval of time depending on the gradient selected and each patient's individual characteristics. In general when the slope of the gradient selected is the steepest, the plateau pressure is reached in approximately 0.1 s, and the initial flow rate is maximal. When the selected slope is the most gentle, the same plateau pressure is reached in 1.5 s and the initial flow rate is minimal. Inspiration is flow triggered by a flow sensor, which was always set at its lowest level 1 l/min. When the selected pressure level is reached, the airway pressure plateau is maintained by a feedback loop that continuously adjusts the flow rate. The inspiratory pressure ceases when the flow falls below 25% of the peak flow or when the patient begins active expiration. The fractional concentration of inspired oxygen was identical during both PSV and CMV. This concentration was never greater than 40%.

The protocol was conducted with the same 1 m long (internal diameter 22 mm), corrugated disposable plastic tubes for inspiratory and expiratory limbs. No water humidifier was used, in order to avoid additional circuit resistance. Measurements of airway pressure and flow were obtained from an experimental set-up inserted between the Y connection of the ventilator tubes and the endotracheal tube. This device was composed of an inspiratory pathway and an expiratory pathway separated by two one-way valves. This

Table 1 Clinical and respiratory characteristics of patients (PaO_2 arterial blood oxygen tension, FIO_2 fractional inspired concentration of oxygen, PS_{SET}, selected pressure level, $C_{st,rs}$ static respiratory system compliance, RI, aw, inspiratory airway resistance, $PEEP_{i,stat}$ static intrinsic positive end-expiratory pressure)

Patient No.	Sex/Age (year)	PaO_2/FIO_2 (mm Hg)	PSSET (cm H_2O)	$C_{st,rs}$ (ml cm H_2O^{-1})	RI, aw (cm $H_2O\,l^{-1}\,s$)	$PEEP_{i,stat}$ (cm H_2O)	Duration of ventilation (days)
1	M/65	270	20	59	15	19	4
2	M/70	161	15	74	18	13	18
3	F/69	363	20	47	35	12	14
4	M/53	183	20	67	18	11	16
5	M/59	180	12	109	24	11	8
6	F/64	319	17	56	30	8	9
7	M/63	270	15	83	24	14	6
8	M/62	306	10	66	19	13	7
9	M/71	224	20	44	52	15.5	6
10	M/70	336	25	83	51	10.5	13
11	M/69	319	15	/	30	/	3

separation avoided rebreathing and allowed measurement of mouth occlusion pressure (P 0.1) by a pneumatic occlusion system. The dead-space of the inspiratory component of this set-up was 105 ml, and the resistance was 3.5 cmH_2O/l per s at a flow of 1 l/s. When an intubation tube with an internal diameter of 7.5 mm was added to the exploration circuit, the resistance was 11 cmH_2O/l per s at a flow of 1 l/s.

Flow ($\dot{V}$) was measured with a pneumotachograph (Fleisch no. 1) and a differential pressure transducer (Validyne 45 MP; Validyne Corp., Northridge, California). The response of the pneumotachograph was linear in the full range of $\dot{V}$ rates observed in the study. Electrical integration of $\dot{V}$ allowed determination of V_T. The inspiratory (T_i) and expiratory time (T_e), and total breath cycle duration (T_{tot}), were also measured. Minute ventilation ($\dot{V}_E$) was defined as the product of V_T and respiratory frequency (f). Another similar differential pressure transducer was used to measure the pressure at the airway opening via a side port mounted close to the endotracheal tube. From airway pressure tracings, we estimated whether or not the patient had reached the selected pressure support level and maintained a plateau, and we measured P 0.1, an inspiratory neuron afferent activity index [9].

Esophageal pressure (Pes) was measured with a third differential pressure transducer. The esophageal balloon was placed at a distance of between 40 and 50 cm from the nares and was inflated with 0.5 ml of air [10]. The validity of Pes was assessed using the occlusion test [11]. We used Pes tracings to measure the dynamic intrinsic PEEP (PEEPi,dyn), which is the pressure difference between the terminal part of the end-expiratory Pes plateau and the initiation of the inspiratory flow, as well as to measure work of breathing.

The patient inspiratory mechanical work performed per breath on the respiratory system (W_I, joules) was calculated using Campbell's diagram method, by integration of the area subtended by the dynamic change in P_{es} and volume during the inspiratory phase and the relaxation pressure-volume curve of the chest wall (C_w) [12]. The beginning of the inspiratory phase was graphically determined as the point of sharp inspiratory inflection on P_{es} tracings, at the beginning of each cycle studied. As we did not measure C_w, we assumed that it was in the normal range for our COPD patients [13]. The slope of C_w was calculated as 4% of the theoretical vital capacity per cmH_2O [14]. The C_w line was placed upon the Campbell diagram at the beginning of the inspiratory phase as derived from the P_{es} tracing. Work performed per litre of ventilation (W_I/l, $J \cdot l^{-1}$) was obtained by dividing W_I by V_T. Work per minute (W_I/min, $J \cdot min^{-1}$) was obtained by multiplying W_I by the corresponding respiratory frequency. The V_T and f necessary to calculate W_I/l and W_I/min were measured on the same breaths used to compute W_I. The value of W_I was taken as the mean of the calculation on five loops, extracted at random from a record of 10 cycles performed at a paper speed of 50 mm/sec. Surface areas of the loops were measured using a computerized planimetry program.

Electromyogram of the diaphragm (EMGdi) was recorded using two surface electrodes placed in the seventh or eighth intercostal space, the first on the right midclavicular line, and the second 50 mm behind the first, close to the anterior axillary line. The last electrode, placed on the left shoulder, served as the earth. The signal was analyzed as follows: the EMG signal was filtered (20–300 Hz), amplified, rectified and integrated by a "leaky integrator" with a time constant of 100 ms to obtain the moving time average [15, 16]. The peak activity of the moving time average quantified the diaphragmatic muscle activity. The peak activity in the seven sequences was expressed as a percentage of the peak activity observed in the highest initial flow rate. The end-tidal carbon dioxide pressure ($PetCO_2$) was continuously monitored by a rapidly responding analyzer (LB2 medical gas analyzer, Sensor Medics Corp. Anaheim, California).

All signals were conditioned and displayed on a Gould model ES 2000 V12 recorder (Gould Inc., Cleveland, Ohio). Visual examination on a screen allowed selection of sequences of regular breaths acceptable for paper recording and analysis. V_T, T_i, T_e, T_{tot} and $PetCO_2$ were measured at a paper speed of 10 mm/s. EMGdi, W_I, and PEEPi were obtained from tracings performed at a paper speed of 50 mm/s, P 0.1 was the only parameter measured at a paper speed of 100 mm/s.

Protocol

Prior to the study, the patient was breathing spontaneously through the ventilator circuits. We introduced an esophageal balloon and placed surface electrodes on the skin to record EMGdi. We then selected the pressure support level to induce a V_T of 7–10 ml/kg with a regular respiratory rate of 12–25 breaths/min, without any clinical evidence of sternocleidomastoid muscle contraction or patient discomfort. During selection of the pressure support level, the sensitivity of the inspiratory trigger system was set to its minimal value (1 l/min) and the initial flow rate was set close to the maximal value to reach the preset pressure plateau in 0.1 s. Finally, the patient was randomly attributed to one of seven sequences in which only the initial flow rate was modified, while the pressure support level selected remained the same. We varied the initial flow rate in order to reach the plateau pressure in 0.1, 0.25, 0.50, 0.75, 1, 1.25 and 1.50 s. The pressure plateau was reached more rapidly the higher the initial flow rate. For convenience, these seven different ventilatory settings were termed T 0.1, T 0.25, T 0.50, T 0.75, T 1, T 1.25 and T 1.50. In each sequence, measurements were obtained after at least 10 min of observation when the patient was in a stable ventilatory condition.

The effects of the various initial flow rates were evaluated by measuring the breathing pattern, PEEPi,dyn, work of breathing, EMGdi, P 0.1, and $PetCO_2$. The study could have been interrupted either at the patient's request or in the presence of any signs of poor clinical tolerance (i.e., either f above 30 breaths/min, evidence of excessive activity of sternocleidomastoid muscle, or a 20 b.p.m. increase in heart rate).

Statistical analysis

The values are expressed as the mean ± standard deviation. Comparison between means for each sequence and each variable measured was performed by two-way analysis of variance (ANOVA) with internal comparisons between sequences by Duncan's test.

Results

The level of pressure support applied was 17 ± 4.3 cmH_2O (range: 12–25 cmH_2O). V_T, generated at the maximal initial flow rate, was 0.505 ± 0.130 l (range: 0.295–0.706 l). Over the experimental range of initial flow rate, $\dot{V}$ peak decreased from $0.98 \pm 0.27\ l \cdot s^{-1}$ to $0.60 \pm 0.10\ l \cdot s^{-1}$ with decreasing initial flow rate. Sequence T 1.5 was interrupted in three cases, at the patient's request in two cases and following the decision of the attending physician in one case. In these three cases, the level of pressure support selected was not achieved. The level of pressure support selected was achieved in all other cases. In only four out of eleven patients, was the pressure plateau maintained beyond sequence T 1. The influence of the various initial flow rates on ventilatory parameters and parameters

reflecting the activity of the inspiratory muscles are described in Tables 2 and 3, and in Figs. 1 and 2.

Between the first (T 0.1) and the last sequence (T 1.50), the values for $\dot{V}_E$ and V_T/T_i decreased progressively from 9.68 ± 1.89 to $8.50 \pm 1.81\,l \cdot min^{-1}$ ($p < 0.05$) and from 0.54 ± 0.11 to $0.36 \pm 0.06\,l \cdot s^{-1}$ ($p < 0.05$), respectively. As the values for V_T remained stable, the values for V_T/T_i decreased as a result of the increase in T_i values. The values for T_i increased progressively from 0.94 ± 0.18 s to 1.38 ± 0.17 s ($P < 0.05$), while T_e remained stable from 2.24 ± 0.66 s to 2.19 ± 0.79 s. This resulted in a progressive increase of T_i/T_{tot} values from 0.30 ± 0.06 to 0.40 ± 0.08 ($p < 0.05$). Between the first and the last sequence, the values for PEEPi remained stable. In contrast, the values for P 0.1 and $PetCO_2$ increased progressively. The values for P 0.1 increased to a proportionally greater extent (from 2.2 ± 1.4 to 3.6 ± 2.5 cmH_2O, $p < 0.05$) than the values for $PetCO_2$ (from 39 ± 6.9 to $40.6 + 7$ mmHg, $p < 0.05$).

In parallel with the reduction of initial flow rate, the values for work of breathing increased progressively, regardless of their mode of expression: W_I, W_I/l or W_I/min (Table 3). In parallel, the values for EMGdi also increased progressively from 100 to $162.4 \pm 38\%$ ($p < 0.05$).

Discussion

Our results show that work of breathing increases when the initial flow rate of PSV decreases. Whereas V_T, f and PEEPi are not affected by the time to reach the plateau pressure.

These results could be associated with methodological problems. Before developing the discussion,

Table 2 Changes in ventilatory parameters with decreasing levels of initial flow rate (V_T tidal volume, f respiratory frequency, $\dot{V}_E$ minute ventilation, T_i inspiratory time, T_e expiratory time, T_{tot} total breath duration , $P\ 0.1$ mouth occlusion pressure, $PEEP_i$ dynamic intrinsic positive end-expiratory pressure, Pet_{CO_2} end-tidal carbon dioxide pressure, $\dot{V}_{peak}$ peak flow). Data are means ± standard deviation in parentheses. Comparison between means for each sequence and measured variable was performed by a two-way analysis of variance (ANOVA), when a significant difference was found, comparison was completed by internal comparisons between sequences by Duncan's test

	Time at which the selected PS level was reached (s)							ANOVA	Duncan's test		
	T 0.1	T 0.25	T 0.5	T 0.75	T 1	T 1.25	T 1.5		T 0.1 versus T 0.75	T 0.75 versus T 1.50	T 1.50 versus T 0.1
No.	11	11	11	11	11	11	8				
V_T (l)	0.505 (0.130)	0.519 (0.103)	0.523 (0.092)	0.534 (0.115)	0.502 (0.113)	0 .460 (0.102)	0.492 (0.068)	NS	–	–	–
f (breaths · min^{-1})	19.7 (4.4)	19.1 (4.9)	1.8 (3.5)	18.1 (4.7)	18.4 (4.7)	19 (3.8)	17.6 (4.2)	NS	–	–	–
$\dot{V}_E$ ($l \cdot min^{-1}$)	9.68 (1.89)	9.61 (1.47)	9.29 (1.63)	9.33 (1.99)	8.89 (1.61)	8.58 (1.66)	8.50 (1.81)	< 0.01	NS	NS	0.05
T_i (s)	0.94 (0.18)	1.07 (0.20)	1.17 (0.22)	1.23 (0.25)	1.26 (0.35)	1.18 (0.21)	1.38 (0.17)	< 0.001	0.05	NS	0.05
T_e (s)	2.24 (0.66)	2.23 (0.64)	2.28 (0.63)	2.31 (0.88)	2.20 (0.75)	2.10 (0.74)	2.19 (0.79)	NS	–	–	–
T_{tot} (s)	3.19 (0.73)	3.31 (0.78)	3.45 (0.68)	3.54 (0.99)	3.46 (0.88)	3.27 (0.74)	3.57 (0.79)	NS	–	–	–
T_i/T_{tot} (s)	0.30 (0.06)	0.33 (0.05)	0.35 (0.07)	0.36 (0.08)	0.37 (0.08)	0.37 (0.09)	0.40 (0.08)	< 0.001	0.05	NS	0.05
V_T/T_i ($l \cdot s^{-1}$)	0.54 (0.11)	0.49 (0.06)	0.46 (0.09)	0.44 (0.09)	0.41 (0.10)	0.39 (0.08)	0.36 (0.06)	< 0.001	0.05	0.05	0.05
P 0.1 (cm H_2O)	2.2 (1.4)	2.2 (1.6)	2.4 (1.3)	3 (1.7)	3 (1.6)	3.6 (1.7)	3.6 (2.5)	< 0.001	0.05	NS	0.05
$PEEP_i$ (cm H_2O)	5.7 (3.1)	5.9 (2.6)	6 (2.5)	6.6 (2.5)	6.3 (2.8)	6.3 (2.3)	6.2 (2)	NS	–	–	–
Pet_{CO_2} (mm Hg)	39 (6.9)	39 (7)	39.5 (6.6)	39.7 (6.7)	40.9 (6.5)	41.4 (6.8)	40.6 (7)	< 0.01	NS	NS	0.05
$\dot{V}$ peak ($l \cdot s^{-1}$)	0.98 (0.27)	0.75 (0.10)	0.73 (0.11)	0.69 (0.11)	0.65 (0.12)	0.62 (0.12)	0.60 (0.10)	0.05	0.05	NS	0.05

Table 3 Changes in work of breathing, and parameters reflecting the activity of the inspiratory muscles with decreasing levels of initial flow rate (W_I inspiratory mechanical work performed per breath on the respiratory system, W_I/l work performed per liter of ventilation, W_I/min work per minute, P_{es} *plat*–P_{es} *peak* the pressure difference between the terminal par of the end-expiratory esophageal pressure (P_{es}) plateau and the peak inspiratory P_{es}, *EMG di* magnitude of the diaphragm EMG signal expressed as a percentage of the peak activity observed in T 0.1). Data are means ± standard deviation in parentheses. Comparison between means for each sequence and measured variable was performed by a two-way analysis of variance (ANOVA); when a significant difference was found, comparison was completed by internal comparisons between sequences by Duncan's test. The expression of comparisons was limited to three sequences for the purposes of simplicity. Analysis of all the results shows that, from one sequence to another, the values for all parameters gradually increased. This progressive increase meant that, in general, the results of one sequence were not significantly different from the two sequences on either side. In contrast, the differences were generally significant when two sequences were separated from each other by two sequences

	Time at which the selected PS level was reached (s)							ANOVA	Duncan's test		
	T 0.1	T 0.25	T 0.5	T 0.75	T 1	T 1.25	T 1.5		T 0.1 versus T 0.75	T 0.75 versus T 1.50	T 1.50 versus T 0.1
n	11	11	11	11	11	11	8				
W_I (J)	0.42 (0.17)	0.42 (0.16)	0.52 (0.20)	0.59 (0.22)	0.63 (0.30)	0.66 (0.21)	0.68 (0.27)	< 0.001	0.05	0.05	0.05
W_I/l ($J \cdot l^{-1}$)	0.87 (0.30)	0.82 (0.26)	0.99 (0.27)	1.16 (0.35)	1.18 (0.35)	1.35 (0.31)	1.24 (0.26)	< 0.001	0.05	0.05	0.05
W_I/min ($J \cdot min^{-1}$)	7.97 (2.98)	7.99 (4.39)	9.01 (3.25)	10.14 (3.49)	10.98 (4.14)	11.73 (3.28)	14.49 (5.32)	< 0.001	NS	0.05	0.05
P_{es} plat – P_{es} peak (cm H_2O)	8.8 (4.1)	9.1 (3.7)	11 (3.5)	14.1 (5.5)	14.1 (5.1)	15.4 (5.5)	14.7 (4.6)	< 0.001	0.05	0.05	0.05
EMG di (%)	100 (0)	104.5 (24.2)	110.4 (19.3)	132.6 (54)	152.8 (52.7)	143.6 (43.3)	162.4 (38)	< 0.001	0.05	0.05	0.05

the possible biases related to the use of Campbell's diagram must be examined. Errors can be related to the difficulty of precisely determining the start of the inspiratory phase on P_{es} tracings. This difficulty is essentially observed under conditions of low inspiratory activity. To minimize this possible error, we averaged the values of five randomized breathing cycles. An underestimation of the work performed by the inspiratory muscles can also be related to the calculation of C_w as, after calculating C_w, we assumed that the configuration of the respiratory system and the chest wall relaxation curve remained the same in all seven sequences. In fact, the configuration of the respiratory system has probably been modified, since the activity of the patients' respiratory muscles increased from the sequence T 0.1–T 1.5. The increased inspiratory muscle activity had induced a rib cage-abdominal distortion from its passive relaxed configuration [17, 18] and, consecutively, a decrease in the value of C_w and an increase of the work performed against the distortional forces. Nevertheless this effect could not be responsible for a 50% increase in W_I.

Moreover, we probably underestimated the work performed by the inspiratory muscles by measuring the volume changes at the airway opening with a pneumotachograph. We therefore did not take into account the trapped intrathoracic gas, which is not in direct communication with the airways and which may be appreciable in COPD patients. As the trapped gas does not participate in ventilation, the mechanical work performed in compressing and expanding these excluded territories is not included in the Campbell diagram [19]. Use of the Campbell diagram may consequently underestimate the true work and the real difference between each sequence.

The increased W_I resulting from the initial flow rate decrease could be related to both an increase in elastic and resistive components of W_I. In the spontaneously breathing patient, it is possible to partition W_I into its components [20]. This computation cannot be calculated during assisted ventilation, like PSV, since the patient is pulling at the same time the ventilator is pushing. The elastic component of W_I, which is commonly very high in COPD patients with acute respiratory failure and severe hyperinflation [21–23], depends mainly on the compliance of the respiratory system, end-expiratory volume and PEEPi [24]. As, in the present study, the values for PEEPi, V_T, f and T_e varied minimally from one sequence to another, we could postulate that the elastic component of W_I is similar over the different sequences. The observed increase in W_I is probably related to an increase in its resistive component and to pressure dissipation caused by time constant inequalities and the viscoelastic behaviour of the pulmonary tissue [13, 20, 23, 25].

On the basis of our data, it is impossible to distinguish and quantify the respective proportions of resistances due to the patient and to the ventilator in the

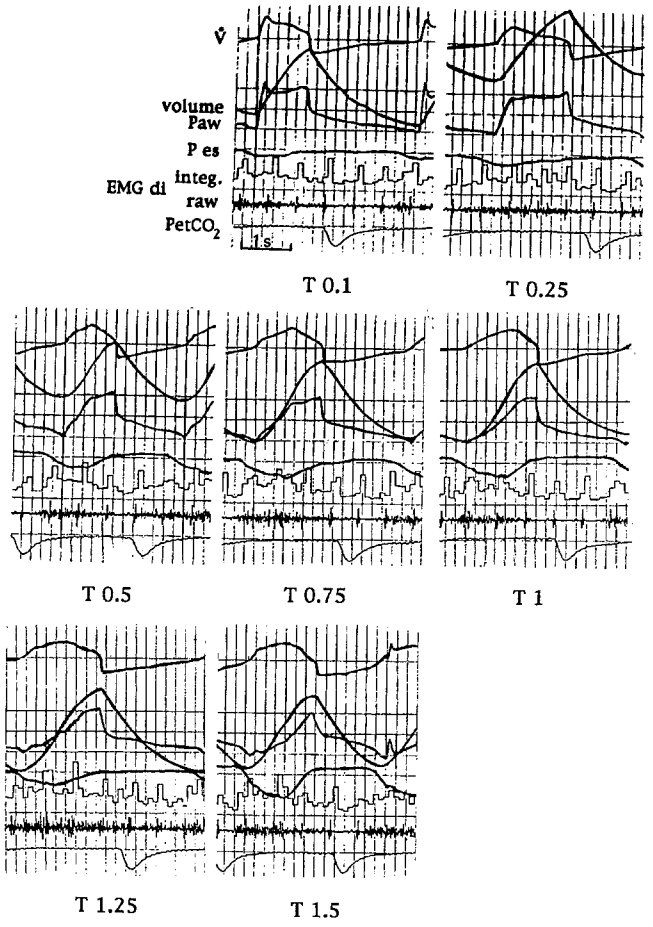

Fig. 1 Effects of decreasing initial flow rate during PSV on tracings of respiratory flow ($\dot{V}$), volume, airway pressure (P_{aw}), esophageal pressure (P_{es}), diaphragm EMG (EMGdi) signals (raw and integrated), end-tidal carbon dioxide pressure ($PetCO_2$) in a representative patient (case no. 4) after 10 min of treatment, with a constant pressure support level (20 cmH_2O). The initial flow was set so that the plateau pressure was reached after a time ranging from 0.1 s (T 0.1) to 1.5 s (T 1.5). The plateau pressure was reached more rapidly as the initial flow rate increased. As the initial flow rate decreased, inspiratory effort increased: as reflected on P_{es} tracings by increased depth of deflection below the baseline, while the V_T mobilized remained at similar values. The pressure support level selected was reached in all sequences, but the duration of the plateau pressure dec reased as the initial flow decreased

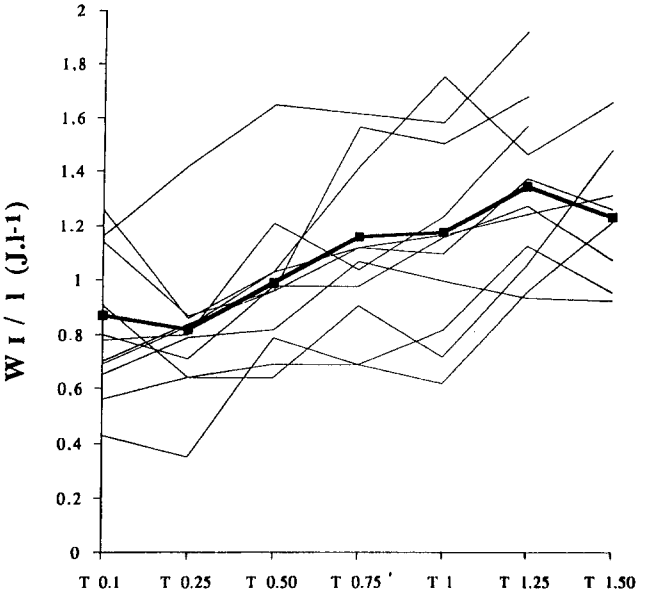

Fig. 2 The effect of initial flow rate decrease on the work performed per litre of ventilation (W_I/l). Data are shown for each of the eleven patients, the *bold line* and *squares* indicate variation of the means. The last sequence could not be performed in three patients

increase of W_I. The main determinant of the increase in resistances and W_I could be the ventilator, which delivers an insufficient flow, i.e., an inspiratory flow delivered at less than the peak value demanded by the patient. Our findings therefore confirm those previously reported in patients ventilated by assist-control ventilation. The work of breathing in assist-control ventilation varies inversely to peak flow when the gas delivery is slowed more than is customary for critically ill subjects [26, 27]. The inspiratory flow should generally be adjusted to exceed the peak value demanded by the patient; a true flow rate delivered in the range of 60–70 l/min is sufficient in most cases [28].

Whatever the origin of the increased activity of inspiratory muscles, it was related to an increased activity of inspiratory centres. This increase, investigated by measuring P 0.1, cannot be explained by the very moderate increase of carbon dioxide [9]. The increase of P 0.1 was due more to a mechanical origin, comparable to that observed in response to bronchospasm or the increased resistance of an inspiratory circuit [29]. Treatment of a patient using PSV with a ventilator delivering a low initial flow rate, therefore, appears to constitute similar conditions to those in which the patient breathes through a circuit of insufficient diameter. Whatever the chemical or mechanical origin of the increase in ventilatory drive, once initiated, it can only accentuate the inadequacy between the demand flow and the delivery flow.

Finally, we stress that our observations were made in COPD patients, using only one type of ventilator and carried out over relatively brief intervals. It may therefore be inappropriate, at this time, to extrapolate our findings directly to all types of ventilators and clinical settings. However, to induce the lowest work of breathing possible it seems logical to adjust the initial flow rate to exceed the peak value demanded by the patient. Whether the selected pressure level is reached in 0.1 or 0.25 s does not appear to be of any great importance, as all parameters studied in these two sequences had similar values. In particular, the values for W_I/l were close to those observed by Fleury et al. in

spontaneously breathing normal elderly subjects (0.9 ± 0.2 J/l) [20].

On the other hand, it is more difficult to define precisely the flow rate beyond which the patient is committed to a potentially fatiguing mode of ventilation. We could have achieved this by using parameters such as the tension time index of the diaphragm defined by Bellemare and Grassino [30], or we could have investigated a shift toward low-frequency components on the EMGdi signal [31]. However, these calculations, as well as the comparison of our data with those reported in the literature examining the relationship between work of breathing and weaning outcome are questionable [32–34]. As it can be assumed that if, in each sequence, a higher pressure support level than those applied was selected, the values for work measured or the fatigability indices calculated would have been lower than those actually measured. Conversely, if a lower level of pressure support had been applied, or if the diameter of intubation tubes had been smaller [35], these same values would probably have been increased, and the threshold beyond which the flow rate would have been potentially fatiguing would also have been increased.

In conclusion, we have demonstrated that in COPD patients requiring ventilatory support, in PSV the use of high initial flow rate decreases the work of breathing and P 0.1 without alteration of the spirometric data or PEEPi. As the objective of PSV is to take over, partially or completely, the patient's work of breathing, it seems logical to use the highest initial flow rate in COPD patients to induce the lowest possible work of breathing.

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