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Test of 20 similar intensive care ventilators in daily use conditions – evaluation of accuracy and performances

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Abstract. Infrequent control, aging of components, may compromise the accuracy of ICU ventilators. In order to assess the reliability of ventilators during their clinical use, we bench tested a group of 20 CPU_1 ventilators (Ohmeda) sampled at random in several ICU units. We found major leaks in 5 ventilators, attributable to the disposable tubings used in these systems. Mean error in expired tidal volume and corresponding standard deviation (precision) were greater than 100 ml in two. Positive end expiratory pressure measurement comprised a mean error higher than 2 cm H_2O in 40% of the ventilators tested. The valve opening pressure threshold was correlated to the inspiratory flow ($r = 0.81$) contrary to the valve opening delay (average 138 ± 40 ms). These two parameters did not correlate with the age of the ventilator. Our study addresses the need for periodic control of ventilator performance in order to minimize the risks of errors and malfunctions.

Key words: Artificial ventilation equipment – Critical care

The ventilators used in ICU are supposed to be accurate and reliable. Several studies in the literature report results of bench tests for different types and generations of ventilators [1–3] or some of their components [4, 5]. Nevertheless, all these studies considered new systems, properly calibrated and checked. To our knowledge, the stability of this accuracy has not been addressed during clinical use. Moreover, calibrations are sometimes infrequently performed and loosely controlled, leading to possible systematic spirometric errors or dysfunctions. To assess the reliability of ventilator performances in daily use conditions, we tested one type of ICU ventilator available in our institution, the Ohmeda CPU_1. This model of ventilator was chosen because its performances were previously published [1, 6] and also because it represents more than 40% of the ventilators used in our institution for ICU care. It belongs to the "old" generation of ventilators and some of the machines we tested were purchased 10 years ago. Therefore they many be more prone to age-related malfunctions especially concerning the demand valve performances.

Material and methods

Ventilators

Twenty ventilators (CPU_1, Ohmeda) were studied while they were actively used or just after withdrawal from patient support. They were sampled from 3 different ICU sites in our institution. This type of ventilator is a volume cycled generator using a pneumatic demand valve [1]. The inspiratory flow profile is square and tidal volume (V_T), respiratory rate (RR), and T_I/T_{TOT} can be set at will by modifying inspiratory flow ($\dot{V}$), inspiratory time (T_I), post-inspiratory pause (T_{PAUSE}), and expiratory time (T_E). Airway pressure (P_{AW}) is read on a manometer located on the front panel. Positive endexpiratory pressure (PEEP) is adjusted, according to the pressure manometer readings.

The front panel of the ventilator displays two sets of spirometric data. One includes expiratory data measured by a hot wire spirometric cell located downstream to the expiratory valve. It comprises minute ventilation ($\dot{V}_E$), expired V_T ($V_{T(e)}$), and patient respiratory rate (RR). Also available is the inspiratory spirometry which includes the same parameters as expiratory spirometry (inspiratory V_T is referred to as $V_{T(i)}$) plus the computed I/E ratio for the mechanical breaths. Noteworthy, these inspiratory spirometric values are not measured but calculated by the system from the ventilatory settings, according to an inner database resident in the system. It assumes that the pneumatic regulation of the inspiratory flow does not change with time. For technical reasons, a constant flow of 2 l/min permanently washes the circuitry to keep the demand valve in a semi-open state and reduce its opening delay and resistance. This flow is electronically subtracted from the raw spirometric expiratory values and is therefore unapparent to user. Only five of the ventilators tested included inspiratory pressure support.

All the ventilators in the study had disposable corrugated plastic tubes (22 mm ID) for both inspiratory and expiratory limbs. The in-

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spiratory branch of the circuit was equipped in 9 with a heated cascade humidifier (Fisher-Paykell, MR 428, New Zealand). It comprised a 30 cm tubing from the ventilator to the water cascade humidifier (if present) and another 70 cm tubing, including a water trap, from the humidifier to the Y piece. If no himidifier was present i.e. when the ventilator was used with a hygroscopic exchanger, the inspiratory line consisted of a single 1 meter tubing. The expiratory line was made of a 100 cm tubing joining the Y piece to the expiratory valve of the ventilator which included a water trap.

All the ventilators were routinely calibrated at various frequencies. Indeed, 55% of them came from clinical departments where controls were performed every 3 months, at the central engineering department of our institution. No other control occurred between, whatever the number of hours the ventilator was in function. In this group, the delay elapsed between the last control and the date of the test was in average 39 ± 23 days. The remaining ones (45%) came from an ICU where technicians performed maintenance and calibration after each cleaning and sterilization operation and before it was allocated to a new patient. Since the last calibration, the average duration of ventilation was 17 ± 5 days in this group. In all cases, the ventilators were intentionally not recalibrated before the test. This was aimed to reflect the actual spirometric accuracy and performances during or following a period of continuous utilization in one or several ICU patients.

Testing bench

The bench used for the test included a passive test lung (TTL, Medishield) with a compliance of 100 ml H_2O and a resistance of 1.5 cm H_2O/l/s. The test lung was attached to the Y piece of the circuit via a connecting piece which included a Fleish #2 pneumotachograph (Fleish, Lausanne, Switzerland) and a lateral port for the measurement of P_{AW} (Fig. 1). Pressure and flow were sensed by two Validyne DP 45 differential transducers (Validyne Corp. Nothridge CA, range ± 2.5 and ± 70 cm H_2O respectively). The pressure and flow signals were converted into numeric files by an A/D converter (Dash 16, Metrabyte) at a frequency of 200 Hz and stored in a computer (PC compatible), for off-line processing. Volume was obtained by numeric integration of flow. Calibrations of pressure and flow transducers were repeated 4 times before each test, using a water column and a 1 l air-filled syringe. The average value for the 4 calibrations of either P_{AW} or flow was retained as the respective calibration factor. A specially designed computer program was built to calculate for each sequence of the test; the ventilator spirometry i.e. tidal volume, RR, I/E ratio, peak P_{AW} and PEEP from the flow and P_{AW} signals.

Check of spirometry and leaks assessment

Each ventilator tested was connected to the calibrated bench. Total leaks were evaluated first, as the drop in P_{AW} during a 1 min occlusion of the expiratory port after the test lung had been inflated at 30 cm H_2O.

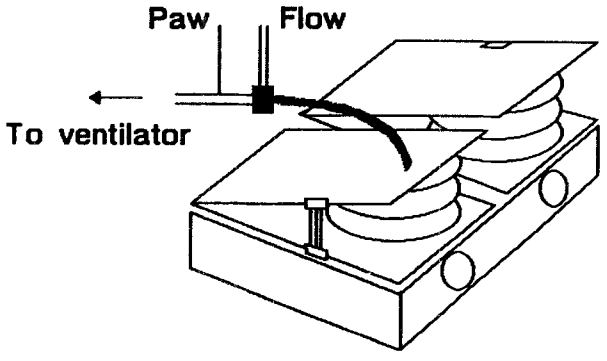

Fig. 1. The bench used for the test of leaks and spirometry, including a lung model, a pneumotachograph (Fleish #2) and two pressure transducers for flow and airway pressure measurements

When major leaks were recognized (drop greater than 30 cm H_2O in 1 min), the disposable tubings were replaced with leakproof silicone reusable 22 mm ID ones and the leak test was performed again. In this case, the value of leak reported corresponded to that measured with silicone leakproof tubings. Therefore, the residual leak reflected a loss of gas attributable to any part of the system except the tubings. In case of such a leak with silicone tubings, we removed all disposable furniture (humidifier, water traps, hygroscopic humidifier) and connected the ventilator straight to the test lung with leakproof tubings in order to check the ventilator itself and separate for internal and external leaks. Special attention was given to avoid any leak of the bench components. This point was checked before each test.

Five recordings were performed in controlled mode ventilation using five combinations of $\dot{V}$ (range $20-120\,l\,min^{-1}$), T_I (range 0.5–1.0 s), T_E (range 1.0–6.0 s), and a fixed 0.2 s T_{PAUSE}. This led to the following panel of ventilatory settings: 230–1300 ml for V_T, 0.17–0.60 for T_I/T_{TOT} and, $7.4-32.0\,min^{-1}$ for RR. The level of PEEP was incremented from 0–20 cm H_2O by 5 cm H_2O steps. The F_iO_2 was kept at 0.21. During each of the five recordings, 30 s of controlled ventilation on the test lung were sampled. The computer then calculated for 2 tidal breaths the average V_T, T_I, T_E, RR, Peak inspiratory pressure (P_{MAX}), and PEEP level actually measured by the bench (reference). The corresponding values of V_T, T_I, T_E, RR, Peak inspiratory pressure (P_{MAX}), and PEEP measured and displayed by the ventilator were noted.

Then, for each variable considered, we calculated the bias as the mean difference between the spirometric readings of the ventilator and that measured by the bench equipment. Hence the bias, i.e. the systematic error over the range tested, became positive when the ventilator overestimated the measurement [7]. The precision corresponded to the standard deviation of the bias values. It reflected the variability of the error in spirometry we observed over the range considered for each variable tested.

Demand valve testing

The ventilators were also tested in spontaneous ventilation mode with different levels of inspiratory pressure support (range 0–15 cm H_2O) when available. For this test, the ventilator was connected via a 3-way large bore swivel to a high impedance pump yielding an adjustable constant inspiratory flow, ($\dot{V}_I$) (0.250, 0.500, 0.750, 1.000, 1.500 l/s). The swivel which initially connected the pump to the atmosphere was quickly switched to the ventilator to trigger the demand valve with the preset constant inspiratory flow. During ths maneuver we recorded $\dot{V}$ and P_{AW} at 200 Hz. From this stepwise inspiratory demand, the valve opening time (V_{OP}) was assessed as the delay needed to reach a pressure plateau, starting from the onset of the simulated inspiration (Fig. 2). The

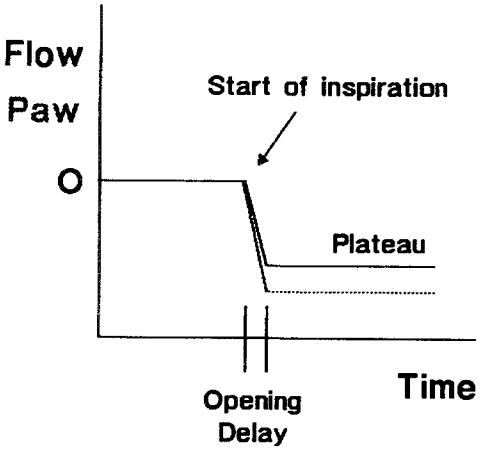

Fig. 2. Simulated recording of P_{AW} (*dotted line*) and inspiratory flow (*solid line*) versus time during the test of the valve. *Vertical lines* represent the beginning and the end of the valve opening process. The distance between lines corresponds to the valve opening delay (V_{OP}) The pressure plateau equals P_{NEG}

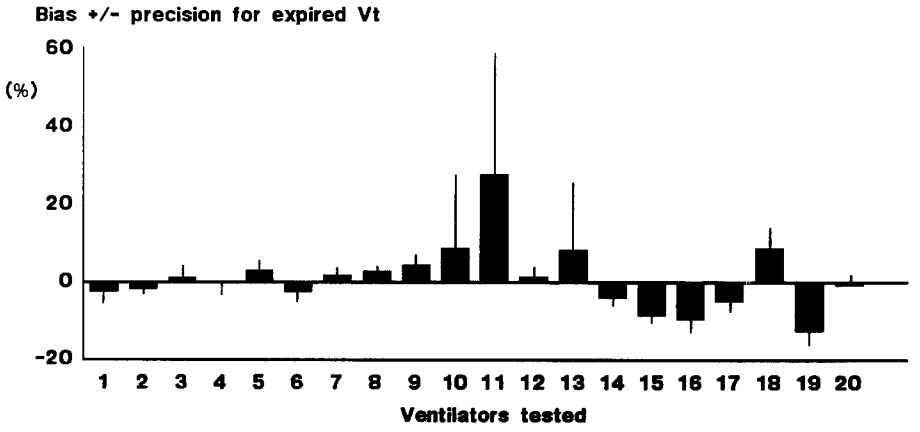

Fig. 3. Plot of the values of bias for $V_{T(e)}$ (in %) and precision (standard deviation of bias, *vertical bars*)

peak negative pressure (P_{NEG}) corresponded to the lowest level of P_{AW} achieved during the valve opening process. The work (W) required to open the valve and reach a stable level of P_{AW} was calculated as the area of a P_{NEG}/V_T loop [8] delimited by the starting point of inspiration and the first point of the P_{AW} plateau following valve opening. Additionally, the $P_{NEG}/\dot{V}_I$ curve was assessed for all ventilators. The slope of this linear pressure/flow relationship (Fig. 3) corresponded to the inspiratory impedance of the demand valve. Finally, the number of hours the ventilators had been used since they were manufactured was noted from the hour-counter of the systems, for correlations with experimental data.

The results are given as mean ± SD. For each variable the relative bias and precision between the measured and observed values are given [8]. The slopes of the $P_{NEG}/\dot{V}_I$ relationship and the correlation between working hours of each device and performances were calculated by means of least square linear regressions [9].

Results

The "age" of the ventilators at the time of the study ranged between 231 and 44,451 h (Table 1).

The compliance of the tubings including cascade humidifier was 1.5 ml/cm H_2O. Major leaks i. e. full depressurization from 30 cm H_2O over one min, were observed in 5 ventilators and were mainly related to the use of disposable tubings. When these tubings were replaced with thick leakproof reusable silicone tubes, the leaks returned to minimal values, with a pressure drop from the initial level of 30 cm H_2O never greater than 9.7 cm H_2O/min (corresponding in our experimental conditions to a volume escape of 14.5 ml/min, Table 1). These leaks were attributable to the equipments fitted to the tubings (humidifier, water traps...) because when all these external components were removed and the test for leaks resumed, we found no pressure drop and therefore no internal leak inside the ventilator. Of note, in one of the ventilators which was used in pressure support mode before the test, we found a 1 cm long tear in one of the disposable tubings. Because the mode used performs a dynamic pressure control to reach a definite inspiratory pressure plateau, the ventilator compensated for the leak and this failure passed unnoticed during clinical use. This leak source became obvious during the study and led as usual to the replacement of the defective tube before we resumed the tests.

The average bias for $V_{T(e)}$ in the 20 ventilators was low because negative values cancelled positive ones (Table 2, Fig. 4) in response to some non-linearities in the flow sensing by the ventilator. Therefore the mean value for precision which reflects non linearities, was not negligible and amounted to 56 ml. Moreover, two ventilators showed great errors in both bias and precision (no. 11 and 13) with major non-linearities expressed by the slope of the regression line between observed and measured $V_{T(e)}$ (1.60 and 2.27 respectively). By contrast, a positive systematic error was consistently observed for $V_{T(i)}$ with a dispersion of values (precision) similar to what was observed for $V_{T(e)}$. Accordingly, the average slope of the regression lines were almost identical for $V_{T(i)}$ and $V_{T(e)}$. Nevertheless, in considering $V_{T(i)}$, two ventilators showed especially large bias (no. 10 and 12).

Table 1. Number of hours the ventilators have been used since they were purchased, values of leaks. Bias (systematic error) for RR and I/E

No.	Hours of use (h)	Leak (cmH_2O/min)	RR (bias) (%)	I/E (bias) (%)
1	8291	3.0[a]	−3	−1
2	231	2.5	−2	−4
3	21110	6.4	−3	2
4	11964	7.8	−2	3
5	10669	9.7	−2	−6
6	1768	2.5	−1	−3
7	10309	3.6[a]	−2	−4
8	3554	3.9[a]	−3	−4
9	11906	4.5	−2	−2
10	4445	0.5	−3	7
11	15724	6.3	−1	−5
12	9346	1.8	−2	9
13	17409	1.9[a]	−1	3
14	26326	0.2[a]	−1	7
15	14873	3.2	−2	−9
16	25882	4.1	−2	8
17	32802	2.5	−2	5
18	44451	3.0	−3	4
19	9652	0.8	−2	1
20	9397	3.7	−1	1
Mean	14358	3.5	−2	1
SD	10748	2.5	1	5

[a] Existence of a major leak (>30 cmH_2O/min) leading to the change of tubings before the test. Values in the table are those obtained after the change of defective tubings

Table 2. Absolute bias, precision, and slope of the regression line between displayed and measured values for expired ($V_{T(e)}$) and inspired ($V_{T(i)}$) tidal volume

No.	$V_{T(e)}$			$V_{T(i)}$		
	Bias (ml)	Precision (ml)	Slope	Bias (ml)	Precision (ml)	Slope
1	−19	26	0.95	76	42	1.12
2	−12	8	0.99	82	74	1.16
3	8	23	1.01	90	60	1.13
4	0	27	0.97	80	93	1.07
5	23	30	1.00	99	44	1.09
6	−12	16	1.02	78	66	1.16
7	16	18	1.02	38	67	1.09
8	21	15	1.00	59	72	1.05
9	46	38	1.12	54	58	1.18
10	27	65	0.86	111	39	0.94
11	278	397	2.27	49	82	1.10
12	4	20	0.96	132	67	1.14
13	121	239	1.60	45	68	1.09
14	−33	18	0.96	59	67	1.17
15	−70	29	0.92	27	45	0.95
16	−74	10	0.98	65	27	1.05
17	−39	24	0.95	66	68	1.21
18	73	43	1.10	77	53	1.16
19	−99	53	0.83	20	47	1.04
20	−6	13	0.99	82	73	1.08
Mean	13	56	1.08	70	60	1.10
SD	78	92	0.31	27	15	0.07

Bias, represents the mean error observed between measured and displayed values; *precision*, represents the standard deviation of the bias, i.e. the dispersion of the values around the mean; *slope*, is that of the regression line between pairs of measured and observed values

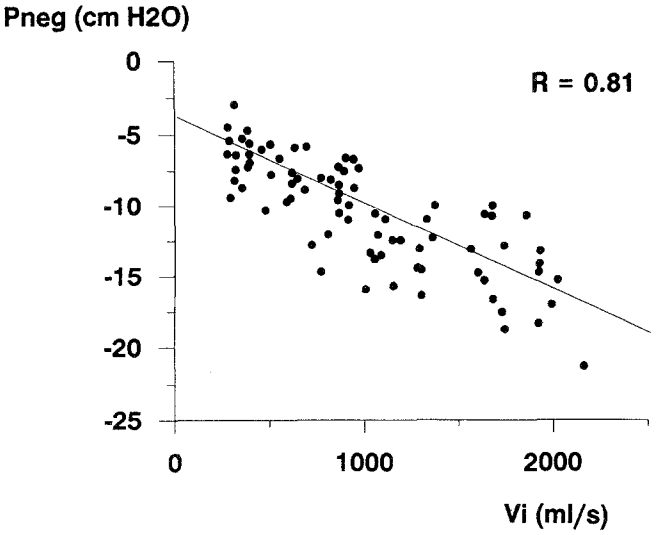

Fig. 4. Plot of the peak negative pressure reached during the valve opening process and the corresponding inspiratory flow during the test. The equation of the regression line is $P_{NEG} = -0.006\ \dot{V}_I - 4.2$, with P_{NEG} in cm H_2O and $\dot{V}_I$ in ml/s, respectively

Similar to volume assessment, some inaccuracies were also observed in the pressure display. The bias for PEEP was negative in all but one ventilator (Table 3) and reached values higher than 2 cm H_2O in 40% of the machines. Precision was good in all but two. P_{MAX} was less accurately measured than PEEP because it comprised larger and more scattered individual bias and precision values.

Finally, RR and I/E measurements were correctly supplied by the ventilator with a bias always in the range of ±9%.

The demand valve opening delay was constant ±10% for each ventilator with the 5 inspiratory flows used, with or without pressure support. It averaged 138±40 ms for the group. P_{NEG} which was the maximal depression recorded during the test of the demand valve, was highly correlated to the inspiratory flow (Fig. 3). The slope of the $P_{neg}/\dot{V}_I$ relationship, i.e. the impedance of the inspiratory limb, including the demand valve and the tubings was in average 6 cm H_2O/l/s. The work (W) needed to open the valve also correlated to inspiratory flow ($r = 0.75$). Interestingly, the work needed to open the valve at an inspiratory flow chosen at 1 l/s (33.6±12.4 mJ) and the valve opening delay did not correlate with the number of hours the ventilators had been used since its purchase ($r = 0.009$ and $r = 0.23$ respectively).

Considering the provenance of the ventilators, we did not find significant difference in accuracy for all variables studied between the ventilators which were checked between patients and those which were only controlled every 3 months.

Table 3. Absolute bias and precision, slope of the regression line between displayed and measured values for PEEP and P_{MAX}

No.	PEEP			P_{MAX}		
	Bias (cmH_2O)	Precision (cmH_2O)	Slope	Bias (cmH_2O)	Precision (cmH_2O)	Slope
1	0.70	0.32	0.95	0.98	1.60	1.03
2	−1.84	1.30	0.83	−0.89	3.00	1.00
3	−4.47	1.88	0.78	−3.19	1.30	1.04
4	−2.70	1.35	0.84	−3.60	0.94	0.97
5	−0.97	1.02	0.86	0.72	1.25	1.05
6	−4.50	1.40	0.83	−5.28	1.96	0.97
7	−1.16	0.74	0.89	1.49	2.52	1.06
8	−2.06	1.00	0.86	−5.21	2.64	1.01
9	−2.45	1.02	0.87	−2.11	1.68	0.97
10	−1.24	2.78	0.69	−1.54	3.04	0.81
11	−3.59	0.95	0.89	−1.15	3.56	0.95
12	−1.39	0.83	0.89	1.98	2.50	1.04
13	−0.36	0.57	0.93	3.47	3.56	1.10
14	−0.65	0.64	0.91	1.19	2.52	1.10
15	−1.44	0.80	0.90	2.58	3.80	1.13
16	−1.28	0.58	0.95	0.59	1.76	1.05
17	−4.77	0.84	0.94	−2.45	5.18	1.34
18	−1.91	0.31	1.04	3.60	3.79	1.20
10	−1.97	0.82	0.89	−0.64	0.85	1.05
20	−4.14	2.24	0.73	−3.45	5.31	1.03
Mean	−2.11	1.07	0.87	−0.65	2.64	1.05
SD	1.47	0.61	0.08	2.63	1.25	0.10

Bias, represents the mean error observed between measured and displayed values; *precision*, represents the standard deviation of the bias, i.e. the dispersion of the values around the mean; *slope*, is that of the regression line between pairs of measured and observed values

Discussion

This study was designed to test the accuracy of a group of comparable ICU ventilators in daily use conditions. We intentionally studied this type of respirator despite the availability of newer systems at our institution. Indeed, this group offered the opportunity to test use-related malfunctions. Moreover, we assessed the size of the spirometric errors observed during the clinical use of such ventilators.

The more important inaccuracies concerned V_T, P_{MAX}, and PEEP. In contrast, the time-related parameters (I/E and RR) were always accurate. Leaks turned out to be a striking problem in a large proportion of the ventilators tested. This failure was related to the use of disposable tubes which loosely fit the ventilators and were probably inadequately controlled before and during ventilation.

The performances of this model of ventilator have been previously published by Nunn and Lyle [1]. They were judged good in term of V_T measurement, despite a slope of the V_T-read/V_T-measured relationship higher than one. This was the case in 40% of the devices we tested but large individual variations were seen. The crossing point with the identity line was observed by Nunn, around 500 ml, and corresponds to a value of V_T close to tidal volumes used in clinical practice. We found the same phenomenon in our study. Therefore, the consequences of such a non-linearity of the flow transducer may be moderate and have sizeable effects only when V_T reaches extreme values. The bias in spirometry we report are related basically to uncorrected calibrations or failure of the spirometric cells. Indeed; these volume inaccuracies could be easily limited to the range of $\pm 5\%$ after a standard recalibration had been performed, due in one case to the replacement of the ventilator flow measurement device. Hence, even if these inaccuracies concerning V_T, are insufficient to jeopardize the patient safety or induce false low ventilation alarms, our data stress the need for periodic control and recalibration of the spirometric cell. However, we did not find an increased accuracy in ventilators which underwent a more frequent recalibration (between patients) compared to those which were controlled every 3 months. This stresses the limits of accuracy of electronic systems which do not include auto-calibrating procedures at each setup and are therefore much more prone to drifts and spirometric errors. Interestingly, the inspiratory volumes delivered by the ventilator were more accurate than expired ones despite the fact that they were calculated and not measured. Hence the mechanical characteristics of the flow regulation seems to be stable with time and not to require specific adjustments.

We chose to test the ventilators with air, not oxygen for practical considerations. The hot wire technique for flow measurement, like many other types of sensors [10–12] are subject to systematic errors when the composition of gas is changed. In a recent study, Synnott and Wren [4] in testing a hot wire spirometer reported an overestimation in flow of 30% when 100% oxygen was used instead of air. Accordingly, one may believe that changes in F_IO_2 during ventilation may induce an additional error. We did not specifically test this type of error.

The problems of imprecision in P_{AW} readings herein documented may be related to three major facts; firstly, the CPU_1 ventilator includes a mechanical pressure transducer which is read by visual observation, sometimes leading to observer-related readings errors; secondly, excepted a zero adjustment, no mechanical calibration procedure is possible; thirdly, the pressure was measured by the bench at the Y piece whereas the ventilator meter was more proximal. Differences in P_{AW} of about 2–3 cm H_2O have been reported between the ventilator pressure manometer and the Y piece [13]. Gas compression and dynamic characteristics of the system account for this difference. Incidentally, if the validity of the pressure manometer as an indicator of the true pressure of the circuitry is questionable, a calibration procedure could probably reduce the underlying errors due to transducer aging or drifting. Achieving a good precision level in the monitoring of P_{AW} is critical in the quest for a reliable titration of PEEP.

The leaks were essentially due to the type of tubings routinely used in our institution. We found that reusable sets which are tight and leakproof seem more reliable than disposable ones. Nevertheless, we observed that the moderate leaks which persisted after the tubings were replaced, disappeared when the ventilator water collector was bypassed. Thus, this component may also be considered as a part especially prone to leaks. A simple check which consists of connecting the ventilator to a test balloon would easily identify leaks and lead to the replacement of the defective components. This should not only be more regularly performed before clinical use, but also during a period of ventilatory support.

In the test of the demand valve, we chose to trigger inspiration by an artificially constant inspiratory flow instead of by a volume withdrawal [3]. This appeared to us to better represent reality. Indeed, patients generate an inspiratory flow which will in turn depressurise the circuit until a pressure threshold is reached and the valve opens. Because the early inspiratory flow may change in patients from one breath to another, the negative pressure achieved at the instant of valve opening will also vary. Hence, the assessment of the pressure/flow relationship of the valve, is the only way to anticipate the system behavior and simulate a large range of inspiratory demands. With this technique the demand valves disclosed a relatively high impedance, which varied linearly with flow and did not depend on the age of the ventilator tested. At an inspiratory flow of 1 l/s, P_{AW} was in average -10.2 cm H_2O below the pre-inspiratory level. This value is comparable to those reported by Katz et al. [14] with this type of ventilator at a CPAP level of 10 cm H_2O. When opened, and with a steady inspiratory flow, the pressure at the Y piece remained constant (Fig. 2) in the absence of pressure support. When pressure support was added, P_{AW} reached a positive plateau peaking at the preset support level. Pressure support did not change the opening pressure/flow relationship of the valve. At the same time, we found that the opening time was in the range of those measured in other ICU ventilators [3, 14–16] and did not vary consistently with the inspiratory flow. Consequently, in comparing our data to those of

other studies, we found that the valve had a relatively high impedance and an intermediate valve opening delay. Interestingly, these parameters were insensitive to mechanical wear.

In conclusion, systematic testing of a homologous group of ventilators revealed that recalibrations should be checked more often that than they are usually done in our institution. Therefore, spirometry should be controlled regularly in the course of clinical ventilation, by shortly ventilating the patient manually, during the time needed to recalibrate the system. This procedure would result in a V_T accuracy of $\pm 5\%$. New ventilators include auto-zero procedures and auto-tests which run at every setup. These systems may possibly be more accurate than those we studied. Other studies are needed to confirm this hypothesis.

Leaks accounted for the main sources of malfunction and could be easily attenuated if they were systematically searched for, particularly during the course of ventilatory support. Accordindly, the use of disposable tubing is questionable, even if the leaks are compensated for in clinical practice by increasing the tidal volume. In general, the inspiratory time is too short to allow a large volume of gas to escape during inspiration when a moderate leak is present. This was confirmed by the fact that patients were apparently efficiently ventilated despite the leaks we reported during the tests. Nevertheless, when facing difficult conditions, as in an adult respiratory distress syndrome, such malfunctions could induce critical problems if not diagnosed and corrected.

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