Eur Respir J 2000; 16: 704–709
Printed in UK – all rights reserved

European Respiratory Journal
ISSN 0903-1936

Evaluation of a method for assessing respiratory mechanics during noninvasive ventilation

D. Navajas, J. Alcaraz, R. Peslin, J. Roca, R. Farré

Evaluation of a method for assessing respiratory mechanics during noninvasive ventilation. D. Navajas, J. Alcaraz, R. Peslin, J. Roca, R. Farré. ©ERS Journals Ltd 2000.
ABSTRACT: Noninvasive assessment of respiratory resistance (*R*rs) and elastance (*E*rs), which is not easy with conventional methods, could be useful in the optimization of pressure support ventilation. The aim of this study was to evaluate a simple noninvasive method (Delta-inst) of measuring *R*rs during nasal pressure support ventilation.

***R*rs and *E*rs (Delta-inst) were computed from inspiratory mask pressure, flow and volume recorded during pressure support ventilation. The Delta-inst method was compared with the forced oscillation technique (FOT) in seven patients with chronic obstructive pulmonary disease (COPD) and in eight healthy subjects without and with added resistance (3.1 $cmH_2O \cdot s \cdot L^{-1}$).**

***R*rs measured by Delta-inst (5.2±1.7, 7.2±0.5 and 6.9±1.2 $cmH_2O \cdot s \cdot L^{-1}$) and by FOT (5.0±0.7, 7.6±0.9 and 8.1±2.7 $cmH_2O \cdot s \cdot L^{-1}$) in healthy subjects without and with added resistance and COPD, respectively, were not significantly different ($p>0.05$). *R*rs measured by both techniques showed a significant coefficient of linear correlation ($r=0.70$ s) ($p<0.01$). In the COPD patients, the variability of Delta-inst *R*rs (30%) was greater than that of FOT *R*rs (21%). The agreement between *E*rs obtained by Delta-inst and by FOT was less than that found for *R*rs.**

Delta-inst is a noninvasive and simple method for reliably assessing resistance. Therefore, it is useful for monitoring airway obstruction and is potentially helpful in adapting the settings for pressure support ventilation in accordance with patient mechanics.
Eur Respir J 2000; 16: 704–709.

Unitat de Biofísica i Bioenginyeria, Facultat de Medicina, Universitat de Barcelona, and Servei de Pneumologia i Al-lèrgia Respiratòria, Hospital Clinic Provincial, Institut d'Investigacions Biomèdiques August Pi Sunyer, Barcelona, Spain.

Correspondence: D. Navajas, Laboratori Biofísica i Bioenginyeria, Facultat de Medicina, Casanova 143, E-08036 Barcelona, Spain. Fax: 34 934024516.

Keywords: Forced oscillation
monitoring
noninvasive ventilation
pressure support
resistance and elastance
respiratory impedance

Received: December 9 1999
Accepted after revision May 17 2000

This work was supported, in part, by Comision Interministerial Ciencia y Technologie (SAF99-0001), Direccion General Enseñanza Superior e Investigacion Cientifica (PM98-0027) and Fondo Investigacioues Sanitarias (FIS-98174).

One of the targets in optimizing ventilatory assistance is to provide a level of support suitable for maintaining sufficient inspiratory activity while avoiding muscle fatigue [1]. Such optimization would be of particular interest in patients undergoing changes in their respiratory system mechanics resulting in impairment of the patient/ventilator interaction. In these cases, adaptation of the magnitude of pressure support and the level of end-expiratory pressure to the respiratory system resistance (*R*rs) and elastance (*E*rs) at different lung volumes may be helpful in tailoring assisted ventilation. Nevertheless, assessment of *R*rs and *E*rs is particularly difficult during assisted ventilation given that the patient is not passive. As there is no way of measuring the contribution of patient muscles to the total driving pressure, methods based on analysis of the airway opening pressure (*P*ao), such as the classical interruption method or the analysis by linear regression of the relationship between *P*ao, volume (V) and flow V' [2], cannot be used. Alternatively, although lung mechanics can be studied by the oesophageal balloon technique, the method is generally regarded as too invasive for routine applications, particularly in noninvasive ventilation. Moreover, this method would not provide the mechanical properties of the overall system which are of interest for setting the ventilator.

One potentially interesting approach is the "Delta-inst" method recently described by CORTIS *et al.* [3]. This consists in increasing or decreasing the inspiratory pressure support for a single respiratory cycle. The assumption is that the time course of respiratory muscle activity during this inspiration is similar to that of the previous cycle. Thus, the change in the time course of *P*ao ($\Delta P_{ao}(t)$) between the two cycles would account for the change in the total pressure applied to the system, and *E*rs and *R*rs could be computed using the usual model from the relationship between $\Delta P_{ao}(t)$ and the resulting changes in the time courses of the flow ($\Delta V'(t)$) and V ($\Delta V(t)$) signals. In patients with acute respiratory failure, the method was found to provide, on average, *R*rs and *E*rs values similar to those of lung resistance and elastance obtained with the oesophageal balloon technique [3]. To the authors' knowledge, however, the performance of the method has not yet been compared with other independent means of estimation of *R*rs and *E*rs. Therefore, the aim of this investigation was to evaluate the Delta-inst method (increase in inspiratory pressure) by comparison with the forced oscillation technique (FOT) [4] in healthy subjects and in patients with severe chronic obstructive pulmonary disease (COPD).

Methods

The study was carried out in eight healthy volunteers (two female, 35±14 yrs, 173±5 cm, 68±10 kg) with spirometric values within the normal range and seven male patients with stable severe COPD (table 1). The study was approved by the Ethics Committee of the Hospital and informed consent was obtained from the patients.

The patient was connected to a bilevel positive airway pressure (BiPAP) support device (BiPAP; Respironics, Murrysville, PA, USA) by means of conventional tubing, whisper swivel and nasal mask. The BiPAP device was modified by including an electric switch to drive its valve coil either normally by the device internal signal (Delta-inst measurements) or by an external signal generated by a computer equipped with an analog/digital converter (AD/DA) system (FOT measurements). P_{ao} and V' were recorded [5], low-pass filtered at 16 Hz and sampled at 160 Hz.

During the Delta-inst method, the subject was assisted with bilevel pressure: 7 cmH_2O during inspiration and 3 cmH_2O during expiration. The Delta-inst manoeuvre consisted of manually changing the inspiratory pressure from 7 to 11 cmH_2O for one breathing cycle (fig. 1). The patient was not aware when this manoeuvre was performed. After the patient's adaptation to the assisted ventilation, eight manoeuvres were carried out, at intervals of ~30 s, over a measurement period lasting 4 min.

Assuming linearity, the P_{ao}, V' and V, signals corresponding to the inspiration prior to the Delta-inst manoeuvre ($P_{ao1}(t)$, $V'1(t)$ and $V1(t)$ respectively) and to the manoeuvre ($P_{ao2}(t)$, $V'2(t)$ and $V2(t)$ respectively) are related with R_{rs} and E_{rs} according to the following equations:

$$P_{ao1} + P_{mus1}(t) = R_{rs}\ V'1(t) + E_{rs}\ V1(t) + P_0 \quad (1)$$

$$P_{ao1} + P_{mus2}(t) = R_{rs}\ V1(t) + E_{rs}\ V2(t) + P_0 \quad (2)$$

where $P_{mus1}(t)$ and $P_{mus2}(t)$ are the pressures generated by the patient's muscles in the two consecutive inspirations, and P_0 is the static recoil pressure at end-expiration, *i.e.* total intrapulmonary positive end-expiratory pressure. On the assumption that this muscular action is unchanged

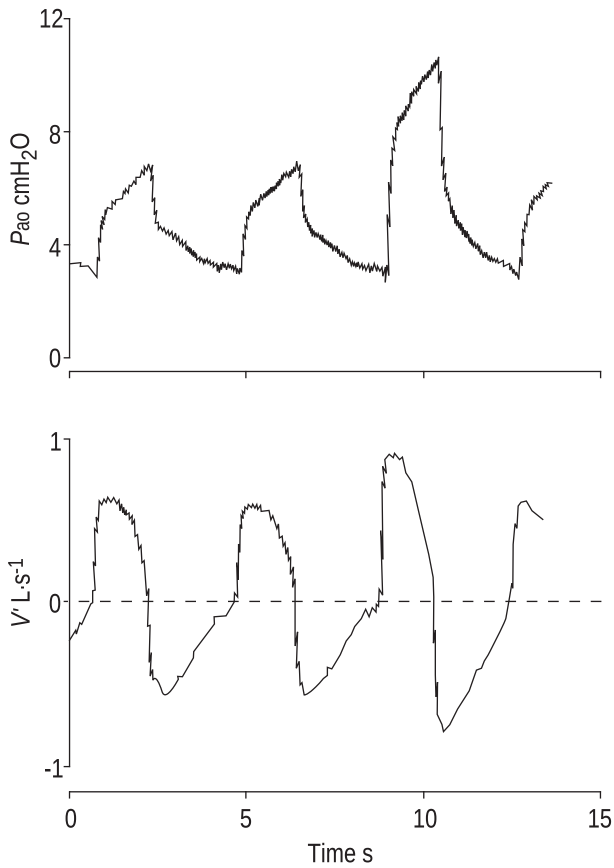

Fig. 1. – Example of nasal pressure (P_{ao}) and flow (V') recorded during a "Delta-inst" measurement. During stable bilevel pressure support (first two cycles in the figure), the patient was subjected to a breathing cycle with increased inspiratory pressure. The patient was not aware when this manoeuvre was performed.

($P_{mus1}(t)$=$P_{mus2}(t)$), as well as P_0, the differences (Δ) $\Delta P_{ao}(t)$=P_{ao2}-P_{ao1}, $\Delta V'(t)$=$V'2(t)$-$V'1(t)$ and $V\Delta(t)$=$V2(t)$-$V1(t)$ are related according to the following equation.

$$\Delta P_{ao}(t) = R_{rs}\ \Delta V'(t) + E_{rs}\ \Delta V(t) \quad (3)$$

This equation was used to compute R_{rs} and E_{rs} by least-square multiple linear regression of the measured data [5]. Fitting of Equation 3 was carried out for a period of time starting at the beginning of the inspiration (Δt), defined as the time when V' crossed the zero axis. V was computed by numerical integration of the recorded V' signal after

Table 1. – Anthropometric and lung function data of chronic obstructive pulmonary disease patients

Patient No.	Age yrs	Weight kg	Height cm	FEV_1 L	FEV_1 % pred	FEV_1/FVC %
1	71	73	170	1.54	47	46
2	72	65	174	1.13	33	35
3	70	83	172	0.64	19	34
4	65	73	170	1.80	53	63
5	66	52	159	1.22	44	54
6	72	84	168	1.74	56	62
7	71	89	158	0.78	30	42
Mean	70	74	167	1.26	40	48
SD	3	13	6	0.45	13	12

FEV_1: forced expiratory volume in one second; FVC: forced vital capacity; % pred: percentage of the predicted value.

correcting for the air leaks. This correction was based on the assumption that functional residual capacity was constant after several complete breathing cycles; leak V' was computed from the mean V' recorded over the entire cycles in each 30-s recording. To test the influence of the length of the fitting time, the analysis was carried out for different values of Δt: 0.25, 0.5 and 0.75 s. Delta-inst computations resulting in negative values of *R*rs or *E*rs or presenting a fitting error (normalized distance between model and data) of >15% were rejected as artefactual (8 and 14% rejected manoeuvres in healthy subjects and COPD patients, respectively). The mean, SD and coefficient of variation of the accepted *R*rs and *E*rs in each measurement were computed.

FOT measurements were carried out by externally driving the coil valve of the BiPAP device to generate a nasal pressure (±1 cmH_2O) consisting of three sinusoids at 2, 3 and 5 Hz superimposed to a constant value of 3 cmH_2O. Eight consecutive recordings, lasting 30 s each, were sampled in each FOT measurement. *R*rs and *E*rs were computed according to a common procedure [6]. Pressure and V' recordings were digitally high-pass filtered (8-pole Butterworth, 1 Hz) to isolate the FOT components. After rejecting the first and last second of data, each 30-s record was divided into six blocks of 8 s each (50% overlapping). Each block was multiplied by a Hanning window, their Fourier coefficients were computed and coherence (γ^2) was estimated by the cross-spectra method [6]. When γ^2 was <0.9, it was determined whether a specific data block was responsible for the low γ^2 and, if this was the case, the block was rejected. For each of the eight consecutive 30-s records, respiratory impedance (*Z*rs) at 2, 3 and 5 Hz was computed from the accepted blocks [6]. *R*rs and *E*rs were determined by fitting a simple resistance/inertance/elastance model to each *Z*rs. Finally, the corresponding mean and SD values of *R*rs and *E*rs were computed.

To test the adequacy of the Delta-inst method for detecting an increase in *R*rs, all of the measurements made in the healthy subjects were also performed with a mesh-wire resistance (3.1 $cmH_2O \cdot s \cdot L^{-1}$) connected between the pneumotachograph and the nasal mask. The order of the measurements in each subject was randomly determined.

Comparison of the *R*rs and *E*rs obtained by Delta-inst and by FOT was carried out using paired t-tests. Statistical significance was assumed at p=0.05.

Results

*R*rs and *E*rs measured by the FOT in the COPD patients (8.1±2.7 $cmH_2O \cdot s \cdot L^{-1}$ and 41.5±15.9 $cmH_2O \cdot L^{-1}$, respectively) were significantly greater than in the healthy subjects (5.0±0.7 $cmH_2O \cdot s \cdot L^{-1}$ and 22.8±5.2 $cmH_2O \cdot s \cdot L^{-1}$, respectively) (fig. 2). Loading the healthy subjects with an external resistance resulted in an increase in *R*rs measured by FOT. However, this increase (2.6±0.7 $cmH_2O \cdot s \cdot L^{-1}$) was lower than the added resistance (3.1 $cmH_2O \cdot s \cdot L^{-1}$) (p=0.07). As expected, *E*rs measured by the FOT in the healthy subjects did not significantly change when adding an external resistance.

Application of the Delta-inst method (Δt=0.25 s) both in the healthy subjects and in the patients resulted in *R*rs which were not significantly different from those taken as reference (FOT) (fig. 2). Moreover, analysis of the relationship between *R*rs obtained by both techniques in basal measurements in healthy subjects and in patients (fig. 3) showed a significant coefficient of linear correlation: *R*rs (Delta-inst)=2.9 $cmH_2O \cdot s \cdot L^{-1}$+0.48. *R*rs(FOT) (r=0.70, p<0.01). Nevertheless, as in the case of the FOT, the increase in *R*rs detected by Delta-inst (2.0±1.6 $cmH_2O \cdot s \cdot L^{-1}$) when the healthy patients were loaded with an external resistance was lower than expected (3.1 $cmH_2O \cdot L^{-1}$) (p=0.09). Figure 2 also shows that the *E*rs obtained using the Delta-inst method in the healthy subjects (37.0±9.9 $cmH_2O \cdot L^{-1}$) was significantly greater than that obtained by the FOT. *E*rs measured by the Delta-inst method was consistently greater in the COPD patients (49.6±28.0 $cmH_2O \cdot L^{-1}$) than in the healthy subjects (p<0.05). As expected, the addition of an external resistance in the healthy subjects did not result in a significant change in *E*rs measured by the Delta-inst method (fig. 2). Nevertheless, the *E*rs measured by the Delta-inst and by the FOT in basal measurements in healthy subjects and in patients did not show significant correlation.

The dependence of *R*rs and *E*rs estimates on Δt was similar in the measurements in healthy subjects and in COPD

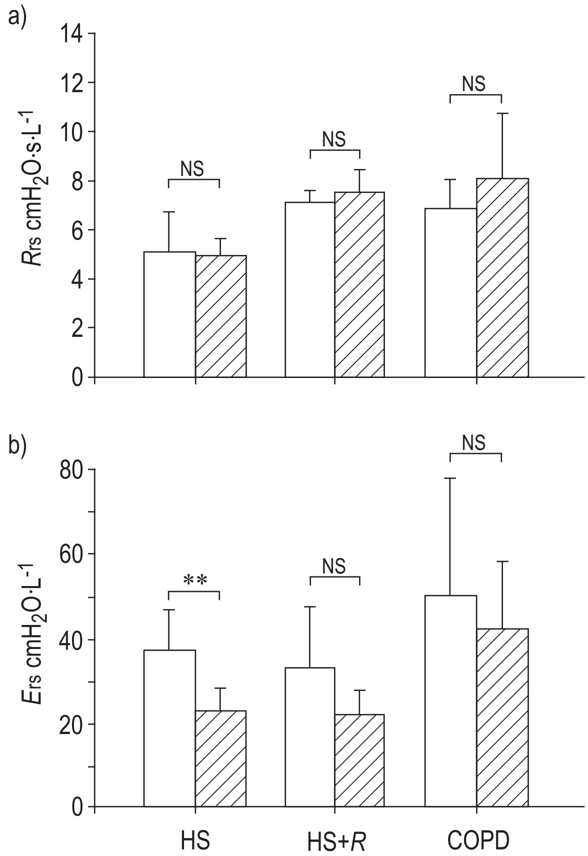

Fig. 2. – a) Respiratory resistance (*R*rs); and b) elastance (*E*rs) measured by the Delta-inst method (□; equation fitting time, starting at the beginning of inspiration=0.25s) and by the forced oscillation technique (FOT; ▨) in the healthy subjects without (HS) and with a load resistance (HS+R) and in the chronic obstructive pulmonary disease (COPD) patients. Data presented as mean±SD. **: p<0.01.

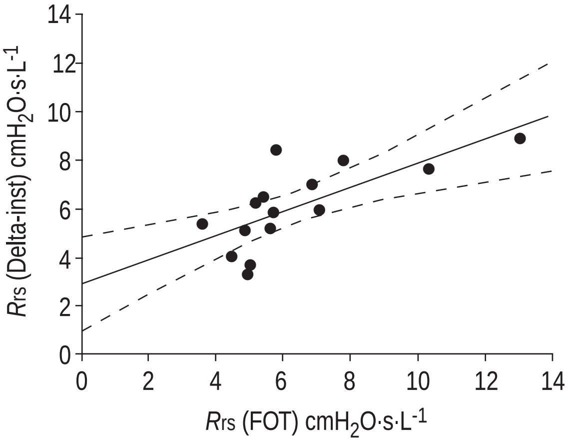

Fig. 3. – Relationship between the respiratory resistance obtained by the Delta-inst method (*R*rs (Delta-inst)) and by the forced oscillation technique (*R*rs (FOT)) in basal measurements. ——: regression line; ----: 95% confidence intervals.

patients. In the COPD patients, *R*rs obtained with Δt= 0.5 and 0.75 s were not significantly different (12 and -11% respectively) from those computed with Δt=0.25 s. However, in contrast to the results found with Δt=0.25 s, *R*rs measured by the Delta-inst method with Δt=0.50 and 0.75 s were not significantly correlated with the *R*rs obtained by the FOT. *E*rs estimates were considerably affected by Δt: *E*rs significantly decreased by 27 and 50% as Δt increased from 0.25 to 0.50 and 0.75 s, respectively. The coefficient of variation in the estimation of both *R*rs and *E*rs which were 30 and 53%, respectively, for Δt=0.25 s, rose sharply as Δt increased to 0.50 and 0.75 s. On average, in the COPD patients, the coefficient of variation in the assessment of *R*rs with the Delta-inst method (30%) was higher than that found when assessing *R*rs by the FOT (21%). By contrast, both methods showed a similarly high coefficient of variation in the measurement of *E*rs (53 and 54%, respectively).

Discussion

The Delta-inst method has recently been proposed as an easy procedure for the noninvasively estimation of *R*rs and *E*rs during assisted ventilation [3]. However, these authors did not include a detailed analysis of the performance of the method when compared with an independent measure of *R*rs and *E*rs. In the present work, the practical application of the method in healthy subjects and in patients with severe COPD was analysed and it was found that the Delta-inst method (Δt=0.25 s) was more suitable for measuring *R*rs than *E*rs. This method permitted the simple and reliable assessment of the degree of airway obstruction during assisted ventilation with pressure support through a nasal mask. Owing to the measurement variability, the computation of *R*rs and *E*rs from one single Delta-inst manoeuvre is not reliable, and, consequently, monitoring of patient′s mechanics requires determination of the moving average of several consecutive measurements. Hence, this noninvasive method can track an abrupt change in respiratory system mechanics only at low speed.

Evaluation of the Delta-inst method was carried out by comparison with FOT, which is the other noninvasive procedure applicable for assessing *R*rs than *E*rs during assisted ventilation. FOT data were interpreted in terms of a resistance/inertance/elastance model in the healthy subjects as well as in the COPD patients. Given the narrow frequency band explored (2–5 Hz), the real part of *Z*rs (*R*rs) was almost constant: in the COPD patients, who could show a greater frequency dependence, no significant difference in resistance was found from 2 (8.3±3.2 $cmH_2O \cdot s \cdot L^{-1}$) to 5 Hz (7.7±2.3 $cmH_2O \cdot s \cdot L^{-1}$). The usefulness of the FOT at a frequency of 5 Hz for assessing *R*rs in patients subjected to noninvasive nasal pressure support has been recently demonstrated [7]. Nevertheless, the suitability of this technique for determining *E*rs in nonapnoeic subjects has so far not been substantiated and remains debatable. To this end, the main difficulty stems from the fact that reliable estimation of *E*rs requires application of the FOT at frequencies as low as possible (<5 Hz). However, in this frequency band, the muscular breathing activity is not compatible with the basic FOT hypothesis that the patient is passive at the forced oscillation frequency. Another difficulty in estimating *E*rs by the FOT stems from the fact that, during assisted ventilation, the oscillations tend to trigger the ventilator and increase the respiratory frequency. This drawback would be avoided by using a ventilator generating bilevel pressure and applying the FOT simultaneously, as has been proposed in the case of applying continuous positive airway pressure and the FOT [8]. In the absence of such a device, the FOT measurements in the present study were performed during spontaneous breathing at the same end-expiratory pressure, and then, presumably, at approximately the same lung volume, as in the Delta-inst measurements. The reliability of the low-frequency (2–5 Hz) FOT procedure was indirectly demonstrated by the consistency of the *R*rs and *E*rs data obtained before and after loading healthy subjects with an external resistance: on average, the increase in *R*rs represented 84% of the added resistance and mean *E*rs was virtually unchanged (fig. 2). However, as regards the comparison between *R*rs and *E*rs measured by the Delta-inst method and by the FOT, it should be pointed out that the results obtained by both techniques were not expected to be the same. Indeed, although *R*rs is mainly determined by the airways and *E*rs by the tissues, both parameters are affected by other respiratory system phenomena such as tissue viscoelasticity, lung inhomogeneity and nonlinearities. As the contribution of these phenomena depends on frequency, airflow and lung *V*, the *R*rs and *E*rs obtained during early inspiration with the Delta-inst method and during the whole breathing cycle with the FOT could vary, particularly in patients [9].

According to the present results, the Delta-inst method was more suitable for estimating *R*rs than *E*rs. First, when comparing the values measured by this method and by the FOT, a significant correlation was found for *R*rs but not for *E*rs. Secondly, for Δt=0.25 s, estimation of *E*rs showed a coefficient of variation that was greater than that of *R*rs. As illustrated in figure 4, these results could be interpreted in terms of the sensitivity of parameter estimation. During the time period Δt=0.25 s, $\Delta P_{ao}(t)$ (Equation 3) is determined more by *R*rs than by *E*rs and, consequently, estimation of *R*rs is more robust than that of *E*rs. Improving estimation of *E*rs would require a prolongation of the time

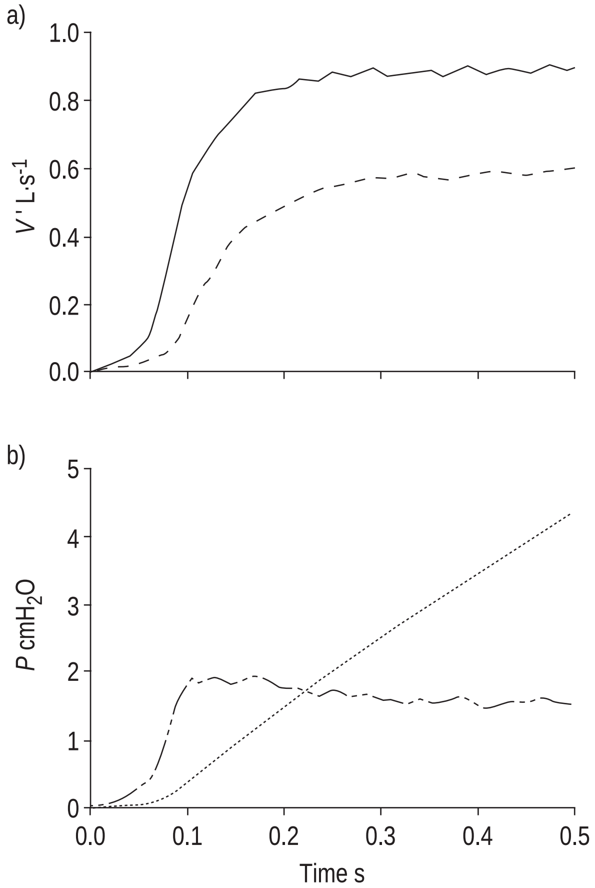

Fig. 4. – a) Detail of the inspiratory flows (V') measured before (----) and after (·····) the Delta-inst manoeuvre in the example in figure 1. b) Resistive (----; $\Delta P_{res}(t)=R_{rs}\,\Delta V'(t)$) and elastic (–··–; $\Delta P_{el}(t)=E_{rs}\,\Delta V(t)$) components of $\Delta P_{ao}(t)$ that would correspond to typical values of respiratory resistance (R_{rs}; 5 $cmH_2O \cdot s \cdot L^{-1}$) and elastance ($E_{rs}$;30 $cmH_2O \cdot L^{-1}$), where $\Delta P_{res}(t)$, $\Delta P_{el}(t)$, $\Delta V'$t, $\Delta V(t)$ and $\Delta P_{ao}(t)$ are the changes in the time courses of resistive and elastic pressure, V', volume and airway opening pressure, respectively. P: pressure.

period Δt given that, for Δt >0.25 s, the elastic term is markedly greater than the resistive term (fig. 4). Nevertheless, prolonging Δt would probably compromise the rationale of the Delta-inst method. Indeed, the main hypothesis of the Delta-inst method is that the patient does not modify their muscular inspiratory effort during the manoeuvre with increased inspiratory pressure. In accordance with published data [10], the time period Δt=0.25 s was considered for data fitting. However, to better characterize the Delta-inst method, the data were also analysed with Δt=0.5 and 0.75 s. As expected, prolonging Δt resulted in a loss of correlation between R_{rs} measured by the Delta-inst method and by the FOT, a considerable decrease in the estimated E_{rs} and a marked increase in the coefficient of variation of both R_{rs} and E_{rs}. These results could be attributed to a change in muscular pressure during the Delta-inst manoeuvre (fig. 1). Indeed, the response to increased airway pressure would be reduction in the muscular pressure during inspiration so as to keep tidal V' and V as constant as possible and, hence, Equation 3 would no longer be valid. In this case, apparent patient impedance (R_{rs} and E_{rs}) would be lower as the muscle pressure in response to the Delta-inst manoeuvre is reduced. The fact that muscle response affected E_{rs} more than R_{rs} could be ascribed to the greater sensitivity of E_{rs} to the last part of the Δt (fig. 4), where potential muscle response is increased. Moreover, the marked increase found in the coefficient of variation of R_{rs} and E_{rs} could be the result of the variability in the induced muscular response after the stimulus of the Delta-inst manoeuvre. In this regard, it should be mentioned that performing Delta-inst measurements with the two possible variants based on increasing or decreasing inspiratory pressure [3] could be useful for the indirect evaluation of the role of possible changes in muscular pressure.

In conclusion, the Delta-inst method may be of practical interest for the routine monitoring of the degree and evolution of airway obstruction in patients subjected to noninvasive pressure support ventilation. The method is noninvasive and the simple modification required in the ventilator pattern (slight increase in inspiratory pressure in one cycle) is not uncomfortable for the patient. When compared with the other methodological alternatives for monitoring assisted ventilation [11, 12], the Delta-inst method has the advantage that it may be easily implemented on any microprocessor controlled bilevel pressure device that is currently available since no hardware change is required. Indeed, only slight modifications in the software would allow the inclusion of a randomly distributed cycle with increased inspiratory pressure and the linear regression analysis of pressure and flow. Such a modified ventilator would allow routine automatic tracking of patient mechanics and potential adaptation of the ventilator settings during noninvasive pressure support. The Delta-inst method could also be of particular interest for setting proportional assist ventilation since this ventilatory mode requires the assessment of respiratory resistance and elastance [13]. However, the clinical usefulness of the Delta-inst method in helping to optimize pressure support ventilation in acute and chronic patients should be investigated in future studies.

Acknowledgements. The authors wish to thank M.A. Rodríguez for technical assistance.

References

1. Rossi A, Appendini L. Wasted efforts and dyssynchrony: is the patient-ventilator battle back? *Intensive Care Med* 1995; 21: 867–870.
2. Tobin MJ, Van de Graaf WB. Monitoring of lung mechanics and work of breathing. *In*: Tobin MJ, ed. Principles and Practice of Mechanical Ventilation. New York, NY, McGraw-Hill Inc., 1994; pp. 967–1004.
3. Cortis G, Olivei M, Hannemann U, *et al*. New method "Delta-inst" for noninvasive continuous assessment of the instantaneous mechanical activity of the respiratory muscles. *Am J Respir Crit Care Med* 1998; 157: A684.
4. Pride NB. Forced oscillation techniques for measuring mechanical properties of the respiratory system. *Thorax* 1992; 47: 317–320.

5. Peslin R, Felicio da Silva J, Chabot F, Duvivier C. Respiratory rnechanics studied by multiple linear regression in unsedated ventilated patients. *Eur Respir J* 1992; 5: 871–878.
6. Michaelson ED, Grassman ED, Peters WR. Pulmonary mechanics by spectral analysis of forced random noise. *J Clin Invest* 1975; 56: 1210–1230.
7. Farré R, Gavela E, Rotger M, Ferrer M, Roca J, Navajas D. Noninvasive assessment of respiratory resistance in severe chronic respiratory patients with nasal CPAP. *Eur Respir J* 2000; 15: 314–319.
8. Farré R, Rotger M, Montserrat JM, Navajas D. A system to generate simultaneous forced oscillation and continuous positive airway pressure. *Eur Respir J* 1997; 10: 1349–1353.
9. Farré R, Peslin C, Rotger M, Barberá JA, Navajas D. Forced oscillation total respiratory resistance and spontaneous breathing resistance in COPD patients. *Eur Respir J* 1999; 14: 172–178.
10. Horner RL, Innes JA, Murphy K, Guz A. Evidence for reflex upper airway dilator muscle activation by sudden negative airway pressure in man. *J Physiol (Lond)* 1991; 436: 15–29.
11. Alberti A, Gallo F, Fongaro A, Valenti S, Rossi A. $P_{0.1}$ is a useful parameter in setting the level of pressure support ventilation. *Intensive Care Med* 1995; 21: 547–553.
12. Flick GR, Bellami PE, Simmonds DH. Diaphragmatic contraction during assisted mechanical ventilation. *Chest* 1989; 96: 130–135.
13. Younes M. Proportional assist ventilation: a new approach to ventilatory support. *Am Rev Respir Dis* 1992; 145: 114–120.