

Non-invasive detection of respiratory muscles activity during assisted ventilation

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Abstract – The instantaneous pressure applied by the respiratory muscles [$P_{\text{mus}}(t)$] of a patient under ventilatory support may be continuously assessed with the help of a model of the passive respiratory system updated cycle by cycle. Inspiratory activity (IA) is considered present when P_{mus} goes below a given threshold. In six patients, we compared IA with (i) inspiratory activity (IA_{ref}) obtained from esophageal pressure and diaphragmatic EMG and (ii) that (IA_{vent}) detected by the ventilator. In any case, a ventilator support onset coincides with an IA onset but the opposite is not true. IA onset is always later than IA_{ref} beginning ($(0.21 \pm 0.10 \text{ s})$) and IA end always precedes IA_{ref} end ($(0.46 \pm 0.16 \text{ s})$). These results clearly deteriorate when the model is not updated. **To cite this article: L. Heyer et al., C. R. Biologies 325 (2002) 383–391.** © 2002 Académie des sciences / Éditions scientifiques et médicales Elsevier SAS

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Résumé – Détection non invasive de l'activité des muscles respiratoires en ventilation assistée. La pression exercée par les muscles respiratoires (P_{mus}) d'un patient sous assistance ventilatoire est estimée en continu à partir d'un modèle du système respiratoire passif réactualisé cycle à cycle. Une activité inspiratoire (AI) est détectée lorsque P_{mus} passe sous un certain seuil. Chez six patients, nous avons comparé AI avec (i) l'activité inspiratoire (AI_{ref}), obtenue à partir de la pression œsophagienne et de l'EMG diaphragmatique, et (ii) celle (AI_{vent}) détectée par le ventilateur. Dans tous les cas, un déclenchement du ventilateur coïncide avec le début d'une AI , mais certaines AI ne déclenchent pas le ventilateur. Le début de AI est toujours situé après le démarrage de AI_{ref} ($-0,14 \pm 0,09 \text{ s}$) et la fin de AI précède celle de AI_{ref} ($0,46 \pm 0,16 \text{ s}$). Ces résultats sont nettement dégradés avec un modèle non réactualisé. **Pour citer cet article : L. Heyer et al., C. R. Biologies 325 (2002) 383–391.** © 2002 Académie des sciences / Éditions scientifiques et médicales Elsevier SAS

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Version abrégée

Des études récentes sur l'interaction entre un patient et son ventilateur d'assistance montrent le besoin d'une détection fine de l'activité respiratoire du patient. Nous proposons une méthode alternative des méthodes invasives classiques (électromyographie diaphragmatique EMGdi, pression œsophagienne), permettant d'estimer en continu la pression exercée par les muscles respiratoires (P_{mus}) à partir des signaux débit et pression à l'entrée des voies aériennes. Cette méthode repose sur l'utilisation d'un modèle de la mécanique respiratoire du patient, réactualisé à chaque cycle ventilatoire.

$P_{\text{mus}}(t)$ est obtenue en soustrayant à la pression des voies aériennes mesurée une pression calculée à partir des variations de débit (F) et de volume (V), selon la formule :

$$P_{\text{RS}} = P_0 + E V + (\alpha |F| + R_0) F$$

où P_{RS} représente la pression nécessaire pour imposer au système respiratoire passif les mouvements observés, P_0 est une estimation de la pression régnant dans les poumons à la fin de l'expiration, E est l'élastance du système respiratoire et α et R_0 les coefficients de la relation linéaire entre résistance et débit du système respiratoire.

Les paramètres (P_0 , E , α et R_0) de ce modèle sont estimés cycle par cycle à partir des signaux de pression et de débit à l'entrée des voies aériennes par une méthode des moindres carrés appliquée sur une partie du cycle. Cette partie retenue correspond aux zones du cycle où la probabilité d'occurrence d'une activité inspiratoire est la plus faible et où le modèle peut être identifié. Sont donc exclus le tout début de l'insufflation, le passage de l'insufflation au dégonflement et la fin du cycle, lorsque celle-ci est caractérisée par une phase à débit nul. Le calcul ne nécessite aucune modification du régime ventilatoire du patient. On considère qu'il existe une activité inspiratoire du patient (AI) lorsque $P_{\text{mus}}(t)$ (négative en inspiration) passe en dessous d'un seuil, qui est fonction de la qualité d'ajustement du modèle sur la partie retenue.

Nous avons étudié six patients de réanimation sous assistance ventilatoire équipés de capteurs de pression œsophagienne et gastrique ainsi que d'un cathéter d'EMGdi. Ces derniers permettent d'établir l'activité inspiratoire de référence (AI_{ref}) basée sur la présence d'une activité électromyographique confirmée par l'évolution de la pression œsophagienne. Au cours d'une phase initiale de ventilation contrôlée, les paramètres de mécanique passive des patients ont été mesurés par la méthode de l'insufflation à débit constant, ainsi que la pression positive de fin d'expiration intrinsèque à partir de la pression œsophagienne.

Ensuite, dix enregistrements ont été effectués (entre un et trois par patient) dans les conditions de ventilation assistée en pression déterminées par le médecin en charge du patient. Entre sept et 40 cycles ont été obtenus par enregistrement, en fonction de la durée entre deux ajustements des paramètres de la ventilation assistée.

Nous avons alors comparé les résultats des mesures de AI_{ref} avec les activités détectées par notre méthode (AI) et avec la détection réalisée par le ventilateur, lorsqu'il est déclenché par le patient. Pour cela, nous avons mesuré le temps écoulé entre le début de AI et le déclenchement du ventilateur (Δ_{ven}), la différence de temps entre le début de AI et le début de AI_{ref} (Δ_{beg}) et la différence de temps entre la fin de AI et la fin de AI_{ref} (Δ_{end}). Par ailleurs, nous avons réalisé les mêmes calculs à partir d'une pression musculaire estimée par un modèle simple non réactualisé de la mécanique ventilatoire, utilisant les paramètres de mécanique passive mesurés en ventilation contrôlée.

Dans tous les cas (209 cycles au total), un déclenchement du ventilateur coïncide avec le début d'une activité détectée par notre méthode (AI) et le début de AI précède le déclenchement du ventilateur : $\Delta_{\text{ven}} = 0,15 \pm 0,10$ s (moyenne \pm écart type). Il arrive qu'une activité détectée par notre méthode ne parvienne pas à déclencher le ventilateur.

Le début de AI est toujours situé après le démarrage de AI_{ref} : $\Delta_{\text{beg}} = -0,14 \pm 0,09$ s. La fin de AI précède celle de AI_{ref} : $\Delta_{\text{end}} = 0,43 \pm 0,19$ s. La faible variabilité observée globalement se retrouve au niveau de chaque enregistrement et démontre une bonne reproductibilité de la méthode.

Les résultats obtenus avec une méthode basée sur un modèle simple non réactualisé, sont nettement moins bons : les différences alors obtenues, $\Delta_{\text{beg,o}}$ et $\Delta_{\text{end,o}}$ ont des valeurs aberrantes ($\Delta_{\text{beg,o}} > 0$ et $\Delta_{\text{end,o}} < 0$) et une variabilité importante ($\Delta_{\text{beg,o}} = 0,16 \pm 0,67$ s ; $\Delta_{\text{end,o}} = -0,14 \pm 0,49$ s).

La comparaison entre les valeurs des paramètres de mécanique passive obtenues au préalable et celles obtenues en cours de ventilation assistée en pression montre la nécessité de la réactualisation de ces paramètres : les valeurs obtenues au préalable diffèrent notablement de celles observées en cours de ventilation assistée ; ces dernières évoluent d'un enregistrement à l'autre chez un même patient.

L'utilisation d'un modèle de la mécanique ventilatoire permet donc d'obtenir une détection non invasive de l'activité inspiratoire d'un patient sous assistance ventilatoire. Ceci passe cependant par une réactualisation des paramètres de ce modèle, réactualisation que nous avons obtenue par une estimation par moindres carrés à chaque cycle, sur une partie seulement du

cycle. L'estimation indirecte de l'activité inspiratoire d'un patient par l'observation de grandeurs mécaniques reste soumise aux limitations de la transmission de la commande ventilatoire au système respiratoire passif.

En particulier, la perte de performance de notre méthode en fin d'inspiration peut s'expliquer par la baisse d'efficacité de l'action du diaphragme à haut volume pulmonaire.

1. Introduction

Management of acute respiratory failure implies usually the use of mechanical ventilation. Assisted modes of mechanical ventilation were developed to address the need for maintaining patient's intrinsic respiration, a necessary condition for eventual weaning from the ventilator. Assisted ventilation aims at reducing excessive respiratory efforts while improving gas exchange by applying positive pressure to the airway thereby unloading the respiratory muscles. Recent approaches even tend to let the patient totally control the ventilator either mechanically [1] or through the neural drive issued from the patient's respiratory centers [2], but these techniques are still under development. In assisted ventilation, although the patient's inspiratory effort triggers the mechanical breath, coordination between spontaneous breathing and mechanical assistance is not guaranteed, owing to poor interaction between the patient and the ventilator. A solution would be to monitor the patient's inspiratory activity over some breaths and to subsequently and interactively adjust the ventilator settings so as to improve this interaction. This points to the need for non invasively detecting respiratory muscle activity during ventilatory support. The present study aims to demonstrate the feasibility of a proposed non invasive method.

Usually, respiratory muscles activity is detected or quantified via esophageal and gastric pressure measurement or diaphragmatic EMG obtained from an esophageal probe. These techniques are invasive and moreover not reliable when applied during long periods in intensive care conditions [3]. Recently, a theoretical method has been proposed that can determine the pressure developed by the respiratory muscles (P_{mus}) in

partially supported ventilation [4]. $P_{\text{mus}}(t)$ is obtained as the difference between the observed airway pressure and a pressure calculated from observed flow and volumes variations via a model of the passive respiratory system.

We have adapted this method to the long-term continuous detection of inspiratory activity in pressure support ventilated patients. Two modifications of the method are needed and a detection rule has to be defined. First, the initial model of the passive respiratory system is adapted to current patients by taking into account a possible intrinsic P_{EEP} (P_{EEPi} , positive end expiratory pressure due to air trapping in the alveoli, the expiration duration being too short for the lung to empty) and a non linearity of the resistance parameter. The modified model includes then four parameters instead of two. Second, these four parameters are estimated cycle-by-cycle from airway pressure and flow signals with a selective least squares method [5]. The detection rule is as simple as possible: an inspiratory activity is 'detected' when P_{mus} goes below a given threshold. We present a comparison of our results with reference data obtained with invasive measurements as well as results obtained with the original method in six patients.

2. Materials and methods

2.1. Patients

After institutional approval and informed consent were obtained, six tracheally intubated patients were enrolled in the study (Table 1). All of them were admitted to the intensive care unit for acute respiratory failure of chronic obstructive pulmonary disease

Table 1. Demographic and passive mechanical data.

Patient	Age (yr)	Weight (kg)	Height (cm)	ETT (mm)	E_p (cm H ₂ O l ⁻¹)	R_p (cm H ₂ O l ⁻¹ s)	P_{EEPi} (cm H ₂ O)
C	71	53	155	8	36	28	5.3
D	63	45	175	9	41	26	8.5
F	72	65	170	9	36	18	6.5
G	60	79	170	9	22	30	6.1
P	77	88	160	8	21	27	5.5
Q	70	50	160	8	14	8	5.2

ETT: endo-tracheal tube caliber. E_p , R_p ; respectively, elastance and resistance of the respiratory system measured during a period of passive controlled ventilation. P_{EEPi} : intrinsic PEEP.

Table 2. Assisted ventilation data.

Recording	P_{EEP} (cm H ₂ O)	Pressure support (cm H ₂ O)	<i>n</i>
C1	2	28	12
D1	7	26	25
D2	12	26	36
D3	10	27	28
F1	6	15	6
G1	3	13	28
G2	3	13	19
P1	5	16	18
P2	5	15	17
Q1	3	15	20

P_{EEP} : pressure imposed by the ventilator during deflation; pressure support: inspiratory pressure given by the ventilator; *n*: number of recorded cycles.

(COPD). At the time of the study, all patients were receiving partial ventilatory support (PSV, Dräger Evita II) and were able to sustain spontaneous breathing for at least 5 min.

During PSV, each breath is assisted by an inspiratory pressure generated by the ventilator at a preset value. This is synchronized with the patient's effort to breathe. The assisted breath is initiated when the spontaneous inspiratory flow reaches a preset threshold value of 3 l min⁻¹ (flow triggering level). The insufflation is stopped when the instantaneous flow is lower than 25% of the maximal value of the inspiratory flow.

Throughout the study, the applied positive end-expiratory pressure and the inspiratory pressure were set by the attending physician and were not modified for the study (Table 2). All patients were in a semi-recumbent position during measurements. Some patients were recorded several times because of a change in the assisted ventilation parameters (Table 2).

2.2. Measurements

Esophageal (P_{es}) pressure was measured with a micro pressure transducer-tipped catheter (MTC P3FC 3F; Dräger ME, Best, The Netherlands). The airway pressure (P_{ao}) was recorded 1 cm from the oral end of the endotracheal tube by using another external transducer (Sims, Kirchseeon, Germany). The validity of P_{es} measurement was assessed by performing 'occlusion tests', as proposed by Baydur et al. [6]. The gas flow (*F*) was measured by using a Fleish (No. 2) pneumotachograph connected to a differential pressure transducer (Validyne MP45; ± 2 cm H₂O). The diaphragmatic electromyogram (EMGdi) was recorded by an esophageal probe positioned at the level of the gastro-esophageal junction (MCT-Cond.M 8F; Dräger ME). This probe consists of eight steel rings at a distance of

9 mm from each other, two adjacent electrodes forming a pair. With the esophageal probe in place, the optimal pair giving the best-quality signal was chosen. The raw EMG was amplified and the band pass-filtered between 20 and 500 Hz. All pressure, flow, and EMGdi signals were digitized by an analog-to-digital converter with a 16-bit resolution at a sampling frequency of 1 000 Hz (MP100 Biopac System Inc., Santa Barbara, CA). The removing of cardiac artifacts from the EMGdi tracing was done manually on the computer screen. Then, the "intEMGdi" signal was obtained by a moving average (30 ms) upon its absolute value.

Neural inspiratory activity (IA_{ref}) was obtained from intEMGdi and validated from P_{es} signal in the following way (Fig. 1): the onset of intEMGdi, which coincides with the point of rapid decline in P_{es} , was used to define the onset of IA_{ref} , and the end of IA_{ref} was defined as the onset of the rapid decline in intEMGdi, contemporary of the return of P_{es} to the base line [7].

All patients underwent a period of passive mechanically controlled ventilation. This was achieved by increasing the respiratory rate of the controlled ventilation. During this period, passive elastance E_p and resistance R_p of the respiratory system were measured according to Rossi and associates [8] during constant flow inflation. The intrinsic PEEP (P_{EEPi}) was measured as the difference in P_{es} between the beginning of the inspiratory effort and the start of inspiratory flow. These characteristics are gathered in Table 1. Measurements during PSV were performed after patients had a 2-min period of spontaneous breathing, and recordings started once the decrease in ventilatory drive associated with PSV onset was completed [9]. Recordings were

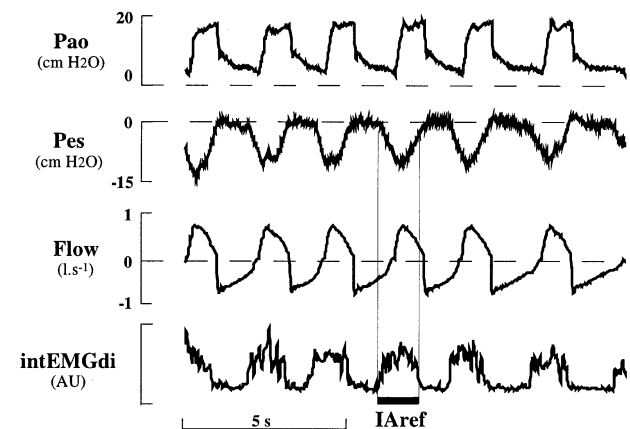


Fig. 1. Individual tracings of airway pressure (P_{ao}), esophageal pressure (P_{es}), flow (*Flow*) and integrated electromyodiaphragmatic activity (intEMGdi) in arbitrary units (AU) in a patient (patient P) during application of partial support ventilation. Neural inspiratory activity (IA_{ref}) is indicated by the large horizontal bar (see text for its determination).

continued until 30 cycles were acquired or any ventilator setting was changed.

2.3. Signal analysis and calculation

Digitized signals were transferred as text files and then processed using programs written in MATLAB™ (The MathWorks®) language. Flow signal was first corrected for a possible departure of zero flow from zero value. The digital integration of flow over time provided then the volume (V) data.

2.3.1. Model

The analysis is based on a mathematical model involving two main components in the pressure, P_{RS} , needed to communicate to the passive respiratory system the observed movement: (i) the elastic pressure ($P_{el} = P_0 + E V$) depending both on a constant elastance E and on a residual pressure P_0 at the end of expiration (depending on the imposed ventilation pattern), (ii) the resistive pressure ($P_{res} = (\alpha |F| + R_0) F$), which points out a flow-dependent resistance (R_0 and α are respectively the constant and slope parameters of the resistance–flow relationship). The motion equation of the passive system, is then:

$$P_{RS} = P_0 + E V + (\alpha |F| + R_0) F \quad (1)$$

In this equation, the residual pressure at the end of expiration is labeled P_0 , instead of P_{EEPi} , as it is a parameter of the model to be identified, while P_{EEPi} is measured. This mathematical model is likely to correspond to the system made up of one patient and his tracheal tube, because tubes exhibit such a flow-dependent resistance.

This system is submitted to two pressure sources, namely the respiratory muscles, supplying a pressure P_{mus} , and the ventilator, supplying a pressure P_{ao} . From the arrangement of these pressure sources, one can deduce that the relationship between P_{ao} , P_{mus} and P_{RS} is:

$$P_{ao} = P_{RS} + P_{mus} \quad (2)$$

P_{mus} is then obtained as:

$$P_{mus} = P_{ao} - P_0 - E V - (\alpha |F| + R_0) F \quad (3)$$

2.3.2. Selection

The four parameters (P_0 , E , α and R_0) are obtained for each cycle by a least squares algorithm that minimizes the mean squared difference between P_{ao} and P_{RS} signals on a part of the respiratory cycle (Fig. 2) most likely (i) to include little respiratory muscle

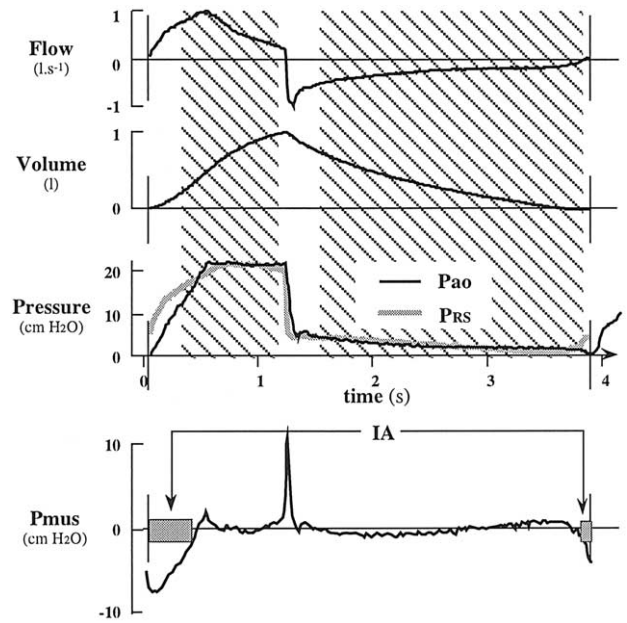


Fig. 2. Selective least squares and detection of inspiratory activity procedures illustrated on a respiratory cycle (patient C). Hatched areas correspond to the zone where parameters of the model are estimated by the least squares method. Vertical lines delimitate the cycle. P_{RS} obtained from parameter estimation is superimposed to P_{ao} . P_{mus} is the difference between P_{RS} and P_{ao} . Grey rectangles on the P_{mus} curve indicate periods where inspiratory activity (IA) has been detected ($P_{mus} < IA_{thr}$).

activity (this excludes the transition between expiration and inspiration) and (ii) to fit closely the proposed model (which excludes the transition between inspiration and expiration, where fast transients cannot be taken into account by a model without inertia coefficient).

The first selected zone starts 0.3 s after beginning of insufflation and finishes 0.1 s before its end; the second selected zone starts 0.3 s after beginning of deflation and finishes at the end of the cycle or as soon as the absolute value of flow goes below 0.1 l s^{-1} .

2.3.3. Automatic detection of activity

Inspiratory muscle activity (IA) is considered present when $P_{mus}(t)$ goes down below a predetermined threshold (IA_{thr}). This threshold has been chosen as a function of the standard deviation observed between P_{RS} and P_{ao} on the zone selected for parameter estimation. Namely $IA_{thr} = 1.5 \text{ SD}$. As is the case in Fig. 2, in most cycles IA is detected at the beginning and at the end of the cycle. Our automatic algorithm was designed to provide the end of the first detected IA and the beginning of the last detected IA of the cycle. Other detected IAs were not taken into account, as they cannot be faced to ventilator detection.

The same procedure has been carried out with the original estimation of P_{mus} based on the passive mechanical parameters obtained before PSV [4].

$$P_{\text{mus,o}} = P_{\text{ao}} - P_{\text{EPI}} - E_p V - R_p F \quad (4)$$

In this case, the threshold ($IA_{\text{thr,o}}$) has been set at 1.0 cm H₂O, which corresponds to the mean value of all IA_{thr} . The corresponding detected IA will be called IA_o .

2.4. Comparison criteria

The time differences between detected and reference estimate of IA onset ($\Delta_{\text{beg}} = IA_{\text{ref}}$ onset time – IA onset time) and end ($\Delta_{\text{end}} = IA_{\text{ref}}$ end time – IA end time) indicate the precision of our method and their standard deviations give an estimation of its reproducibility. The difference between detected IA onset and ventilator triggering ($\Delta_{\text{ven}} = \text{inflation start time} - IA$ onset time) evaluates the sensibility of our method compared to that of the ventilator trigger. All these differences are illustrated in Fig. 3, where IA (and the pressure signal from which it originates) and IA_{ref} (and the corresponding intEMGdi signal) are represented for the same neural inspiration in patient P.

Similar comparisons have been done with $P_{\text{mus,o}}$: for each detected IA , $\Delta_{\text{beg,o}} (= IA_{\text{ref}}$ onset time – IA_o onset time) and $\Delta_{\text{end,o}} (= IA_{\text{ref}}$ end time – IA_o end time) were obtained.

We calculated a mean value (and standard deviation) of each time difference for each recording (for graphi-

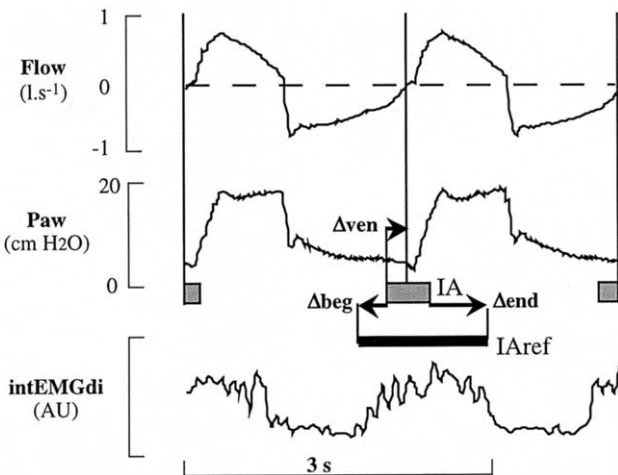


Fig. 3. Comparison between detected IA and IA_{ref} . In this enlarged part of Fig. 1 (P_{es} signal has been excluded for sake of simplicity), the corresponding IA is positioned relative to the IA_{ref} given in Fig. 1 (remaining IA of the two cycles are mentioned without corresponding IA_{ref}). Long vertical lines delimitate cycles from flow signal. IA representation: same as in Fig. 2.

cal presentation) and over all recordings. Statistical comparison has been carried out by paired or univariate t -test over all recordings.

3. Results

3.1. Sensitivity

In all cycles, ventilator triggering coincides with an IA detection (inversely, some detected IA do not trigger the ventilator, but we did not take them into account). Our method detects inspiratory activity in advance (Δ_{ven} significantly > 0 , t -test, $p < 0.01$) on the ventilator (mean difference \pm SD calculated on all recordings: $\Delta_{\text{ven}} = 0.15 \pm 0.10$ s).

3.2. Comparison with the reference method

IA onset is almost always (198/209) late with respect to IA_{ref} onset. When this is not the case, the difference is small (< 0.01 s). This evidences that we detect inspiratory activity **after** it appears on EMGdi, but the difference as well as the standard deviation are small: $\Delta_{\text{beg}} = -0.14 \pm 0.09$ s (mean \pm SD, calculated on all recordings). The end of IA is detected prematurely in almost all cases (205/209) by our method. This demonstrates that our method does not indicate existence of inspiratory activity **after** it disappears on EMGdi. The difference and the standard deviation are greater: $\Delta_{\text{end}} = 0.43 \pm 0.19$ s (mean \pm SD, calculated on all recordings).

Individual recording results are shown in Fig. 4. Differences vary from one recording to another and between patients, but they stay comparable for Δ_{beg} as well as for Δ_{end} .

3.3. Results obtained with the original P_{mus} estimation

In eight cycles, detection of IA with original P_{mus} fails to detect any IA . In the remaining cycles, the advance on the ventilator is erratic ($\Delta_{\text{ven,o}} = 0.51 \pm 0.78$ s).

The difference $\Delta_{\text{beg,o}}$ has an unrealistic statistically significant (t -test, $p < 0.01$) positive mean value ($\Delta_{\text{beg,o}} = 0.16 \pm 0.67$ s) corresponding to a detection of IA **before** the onset of IA_{ref} in seven out of ten recordings ($p < 0.01$). Similarly, the difference $\Delta_{\text{end,o}}$ has an unrealistic statistically significant (t -test, $p < 0.01$) negative mean value ($\Delta_{\text{end,o}} = -0.14 \pm 0.49$ s) corresponding to a detection of IA **after** the end of IA_{ref} in two recordings ($p < 0.05$). As illustrated in Fig. 5, which gathers the results of both methods for each recording, the detection of IA with P_{mus} calculated in

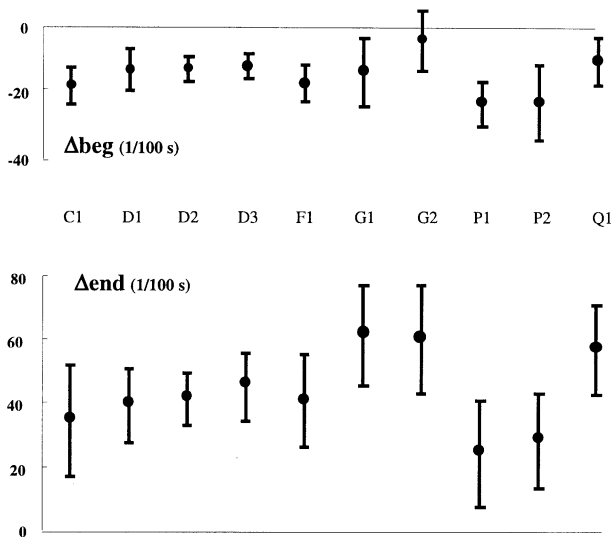


Fig. 4. Differences between detected IA and IA_{ref} for individual recordings. Δ_{beg} negative indicates a delay taken by our method to detect IA onset relative to the reference based on diaphragmatic EMG. Δ_{end} gives the same information for the end of IA , a positive value indicates an advance in detecting IA end. Vertical bars represent one SD.

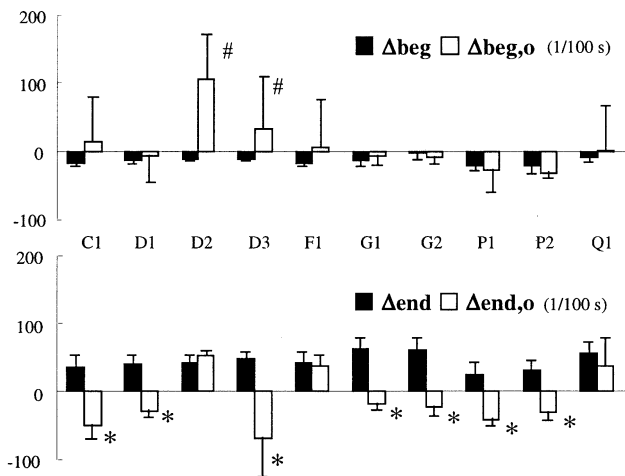


Fig. 5. Comparison of the results of our method (full rectangles) with that obtained with the original computation of P_{mus} (empty rectangles) for individual recordings. Vertical bars represent one SD. #: Significantly > 0 ($p < 0.05$); *: significantly < 0 ($p < 0.05$).

the original way gives unrealistic results (mean $\Delta_{\text{beg,o}} > 0$, or mean $\Delta_{\text{end,o}} < 0$) in numerous recordings and exhibits a high variability as compared to our method.

3.4. Comparison of the mechanical parameters

Our method gives a cycle-by-cycle estimation of respiratory mechanics parameters. In order to compare these values to the ones calculated during passive

controlled ventilation, we first calculated for each cycle a mean resistance value (R_m) and we calculated then mean values of elastance (E), resistance (R_m) and P_0 for each recording. Table 3 summarizes these data together with the passive mechanical data of Table 2. While P_0 is generally close to P_{EEPi} , other mechanical parameters exhibit great significant ($p < 0.05$) differences between passive controlled situation and PSV and even between recordings on the same patient (D for example).

4. Discussion

As an alternative to invasive methods, we used an estimation of the instantaneous pressure applied by the respiratory muscles [$P_{\text{mus}}(t)$] derived from [4] to detect inspiratory activity in pressure support ventilated patients. In [4], this estimation was used for the analysis of work of breathing under various levels of pressure support ventilation (PSV) and has been validated in patients who did not exhibit intrinsic PEEP. However, the work of breathing obtained by this method exhibited a poor correlation with the work of breathing obtained from P_{es} measurements. This original method has two main drawbacks: (i) the model of the respiratory mechanical system on which it relies is too simple and (ii) the parameters of this model, once determined in controlled ventilation mode, are never reevaluated.

Our COPD patients all exhibit intrinsic PEEP (Table 3) and this had to be included in the computation of P_{mus} . Moreover, to take into account the behaviour of tracheal tube, we introduced a non-linearity on flow resistance. The resulting model of passive respiratory mechanics has four parameters instead of two, which is not problematic, as the least squares technique allows to evaluate any multi-linear model [10]. Moreover, Peslin et al. [10] conclude that this 4-parameter model is satisfactory in all their patients; then, this model, even if not the only plausible one, remains the most likely to provide satisfactory results on many patients. Our patients are not numerous enough to allow us to assert that the proposed method will always work for any patient; however, their respiratory mechanical characteristics as well as the used PEEP and Pressure Support cover a wide range of values (Tables 1 and 2).

This least squares technique is well adapted to the second modification we needed to bring to the original P_{mus} estimation method: a continuous reevaluation of the model parameters. For each successive cycle, we apply this technique selectively to parts of the cycle where inspiratory activity is not likely to occur during PSV (our results do not contradict this hypothesis). Another selection of favorable zones for least squares

Table 3. Mechanical parameters obtained during passive controlled ventilation (E_p , R_p , P_{EEP_i}) and by least squares estimation (E , R_m , P_0) during PSV.

Patient	Recording	E_p	E	R_p	R_m	P_{EEP_i}	P_0
C	C1	36	15 (1)	28	18 (2)	5.3	5 (2)
D	D1	41	11 (2)	26	16 (2)	8.5	11 (1)
	D2		27 (4)		36 (2)		15 (1)
	D3		8 (2)		11 (1)		13 (3)
F	F1	36	24 (2)	18	9 (1)	6.5	7 (1)
G	G1	22	15 (7)	30	11 (4)	6.1	4 (2)
	G2		7 (2)		13 (2)		6 (1)
P	P1	21	20 (7)	27	10 (1)	5.5	7 (2)
	P2		20 (5)		9 (2)		6 (2)
Q	Q1	14	7 (1)	8	3 (1)	5.2	7 (1)

Elastances are expressed in $\text{cm H}_2\text{O l}^{-1}$, resistances are in $\text{cm H}_2\text{O l}^{-1} \text{ s}$, and pressures in $\text{cm H}_2\text{O}$. Estimated values are mean (SD).

fitting has already been used in other circumstances [11, 12]. Such a continuous method has at least two advantages: (i) it is totally non-invasive, as the estimation of respiratory mechanics parameters is obtained without the need for any maneuver from the ventilator, and (ii) it allows a follow-up of the state of the patient's passive respiratory system.

In order to validate our results, we had to call for a reference method. Recently, Parthasarathy et al. [13] evaluated the concordance of neural inspiratory time measurements based on flow, esophageal pressure and transdiaphragmatic pressure with a more direct measurement of neural activity. This implied the use of esophageal electrode recordings of diaphragmatic EMG that we also used as references. These authors conclude that indirect estimates of onset and duration of neural inspiratory time displayed poor agreement with diaphragmatic EMG measurements. In fact, their results exhibit a huge variability that we do not observe in our results. This comes probably from the fact that their indirect evaluation of neural inspiratory time relies on robust but too simple criteria.

The interpretation of P_{mus} variations into an inspiratory activity needs the use of a threshold. We decided to adapt this threshold that we use to the local quality of the model and of the pressure and of the flow signal that we get (standard deviation between measured and modeled pressures). This can be discussed on the basis that this threshold always stays around 1 $\text{cm H}_2\text{O}$ and is only significantly modified when the patient exerts an inspiratory effort not detected by the ventilator. In such a circumstance, the standard deviation increases and consequently increasing the threshold will not bring any robustness but will increase the risk of missing this effort already undetected by the ventilator. The 1.5 coefficient we used ($IA_{thr} = 1.5 \text{ SD}$) was chosen to give a compromise between sensitivity and specificity. The method is relatively robust as regards this coefficient:

results are almost identical (less than 10% variation in all estimates) when this coefficient takes values between 1.0 and 2.0.

On the whole, our method gives satisfactory results in the detection of inspiratory activity. The major defect resides in the detection of the end of this activity: our method does not 'see' this activity up to its end as defined by diaphragmatic EMG. This seems rather unavoidable, since presence of activity on diaphragmatic EMG does not imply that diaphragm contraction is efficient. Indeed, at high lung volume, which is the case in ventilated patients with high intrinsic PEEP at the end of insufflation, diaphragm contraction may have no efficiency at all. This may explain the discrepancy between our indirect method and the reference one based on diaphragmatic EMG. The fact that the lag of our method on EMGdi is stable on one patient nevertheless allows detecting a significant change in inspiratory effort duration.

The comparison of our results with those obtained with a simpler evaluation of P_{mus} underlines the need for a continuous reevaluation of respiratory mechanics; this is even more obvious when putting together parameter evaluations: Table 3 evidences the fact that for the same patient, respiratory mechanics evolve significantly from one situation to another. Incidentally, the fact that evaluation methods differ cannot afford the difference observed. Such observations should revive development of computer programs for automatic measurement of respiratory mechanics in ventilated patients [14].

5. Conclusion

We conclude that non-invasive detection of respiratory muscle activity is possible during pressure support ventilation. This implies the use of a model of passive respiratory mechanics; this model has to be reevaluated

continuously. Simple methods are available for this and have been satisfactorily applied on clinical data. A non-negligible spin-off of this study is the follow-up of respiratory mechanics in pressure support ventilation.

The shorter detected inspiratory activity as compared to EMG diaphragmatic activity may be predicted by the lower mechanical output of diaphragm at high lung volume.

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References

- [1] M. Younes, Proportional assist ventilation, a new approach to ventilatory support, *Am. Rev. Respir. Dis.* 145 (1992) 114–120.
- [2] C. Sinderby, P. Navalesi, J. Beck, Y. Skrobik, N. Comtois, S. Friberg, S.B. Gottfried, L. Linström, Neural control of mechanical ventilation in respiratory failure, *Nature Medicine* 5 (1999) 1433–1436.
- [3] M.J. Tobin, Respiratory monitoring in the intensive care unit, *Am. Rev. Respir. Dis.* 138 (1988) 1625–1642.
- [4] Y. Yamada, M. Shigeta, K. Suwa, K. Hanaoka, Respiratory muscle pressure analysis in pressure support ventilation, *J. Appl. Physiol.* 77 (1994) 2237–2243.
- [5] P.Y. Carry, D. Gallet, Y. François, J.P. Perdrix, A. Sayag, F. Gilly, A. Eberhard, V. Banssillon, P. Baconnier, Respiratory mechanics during laparoscopic cholecystectomy: the effects of the abdominal wall lift, *Anesth. Analg.* 87 (1998) 1393–1397.
- [6] A. Baydur, P.K. Behrakis, W.A. Zin, M. Jeager, J. Milic-Emili, A simple method for assessing the validity of the esophageal balloon technique, *Am. Rev. Respir. Dis.* 126 (1982) 788–791.
- [7] S. Field, S. Sanci, A. Grassino, Respiratory muscle oxygen consumption estimated by the diaphragm pressure-time index, *J. Appl. Physiol.: Respir. Environ. Exercise Physiol.* 57 (1984) 44–51.
- [8] A. Rossi, S.B. Gottfried, B.D. Higgs, L. Zocchi, A. Grassino, J. Milic-Emili, Respiratory mechanics in mechanically ventilated patients with respiratory failure, *J. Appl. Physiol.* 58 (1985) 1849–1858.
- [9] J.P. Viale, S. Duperret, P. Mahul, B. Delafosse, C. Delpuech, D. Weismann, G. Annat, Time-course evolution of ventilatory responses to inspiratory unloading in patients, *Am. J. Respir. Crit. Care. Med.* 157 (1998) 428–434.
- [10] R. Peslin, J. Felicio da Silva, F. Chabot, C. Duvivier, Respiratory mechanics studied by multiple linear regression in unsedated ventilated patients, *Eur. Respir. J.* 5 (1992) 871–878.
- [11] L. Eberhard, J. Guttman, G. Wolff, W. Bertschmann, A. Minzer, H.J. Kohl, J. Zeravik, M. Adolph, J. Eckart, Intrinsic PEEP monitored in the ventilated ARDS patient with a mathematical method, *J. Appl. Physiol.* 73 (1992) 479–485.
- [12] J. Guttman, L. Eberhard, G. Wolff, W. Bertschmann, J. Zeravik, M. Adolph, Maneuver-free determination of compliance and resistance in ventilated ARDS patients, *Chest* 102 (1992) 1235–1242.
- [13] S. Parthasarathy, A. Jubran, M.J. Tobin, Assessment of neural inspiratory times in ventilator-supported patients, *Am. J. Resp. Crit. Care Med* 162 (2000) 546–552.
- [14] P.F. Baconnier, A. Eberhard, P.Y. Carry, J.P. Perdrix, J.M. Fagnoli, A computer program for automatic measurement of respiratory mechanics in artificially ventilated patients, *Comput. Meth. Progr. Biomed.* 47 (1995) 205–220.