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Estimation of inspiratory muscle pressure in critically ill patients

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Abstract *Background:* Recently, a new technology has been introduced aiming to monitor and improve patient ventilator interaction (PVI monitor). With the PVI monitor, a signal representing an estimation of the patient's total inspiratory muscle pressure ($Pmus_{PVI}$) is calculated from the equation of motion, utilizing estimated values of resistance and elastance of the respiratory system. *Objective:* The aim of the study was to prospectively examine the accuracy of $Pmus_{PVI}$ to quantify inspiratory muscle pressure. *Methods and interventions:* Eleven critically ill patients mechanically ventilated on proportional assist ventilation with load-adjustable gain factors were studied at three levels of assist (30, 50 and 70%). Airway, esophageal, gastric and transdiaphragmatic (Pdi) pressures, volume and flow were measured breath by breath, whereas the total inspiratory muscle pressure (Pmus) was calculated using the Campbell diagram. *Results:* For a given assist, $Pmus_{PVI}$ throughout inspiration did not differ from the corresponding values calculated using the Pdi and Pmus signals. Inspiratory and expiratory time did not differ among the various methods of calculation. Inspiratory muscle pressure decreased with increasing assist, and the magnitude of this decrease did not differ among the various methods of pressure calculation. *Conclusions:* A signal generated from flow, volume and airway pressure may be used to provide breath-by-breath quantitative information of inspiratory muscle pressure.

Keywords Transdiaphragmatic pressure · Resistance · Elastance · Mechanical ventilation

Introduction

One of the main goals of mechanical ventilation is to unload the respiratory muscles [1]. Complete unloading of respiratory muscles may be achieved during controlled mechanical ventilation [2], which usually necessitates heavy sedation and occasionally administration of neuromuscular blocking agents. However, recent evidence in the literature indicates that active respiratory efforts during mechanical ventilation are beneficial, because they prevent respiratory muscle atrophy [3–6] and improve cardiovascular system function [7]. Moreover, switching mechanically ventilated patients to assisted modes of support reduces the complications associated with heavy sedation [8, 9] and thus is a priority in the intensive care unit [10, 11]. Nevertheless, during assisted mechanical ventilation, the patient interacts with the function of the ventilator, sometimes vigorously, and this interaction may influence the decision-making process and patient outcome [12, 13]. Patient-ventilator asynchrony largely prevents mechanical ventilation from achieving its goals and may impose significant harm to the patient [14].

It follows that assessment of respiratory output during assisted mechanical ventilation is crucial for proper titration of ventilator settings.

During mechanical ventilation, respiratory output can be monitored by the neural activity (i.e., NAVA technology) [15] and pressure output of respiratory muscles [16–18]. Notwithstanding that several factors may affect the transformation of neural activity to pressure output [19–22], the latter together with the ventilator pressure are the actual forces applied to respiratory system in order to execute the act of breathing. Thus, the respiratory muscle pressure may provide the caregiver important information regarding patient-ventilator interaction. This pressure output may be estimated either by transdiaphragmatic pressure or by calculating the instantaneous pressure output of the respiratory muscles [16, 17]. These approaches, however, necessitate the placement of esophageal and gastric catheters to record esophageal and gastric pressures, respectively. Other attempts to monitor respiratory muscle pressure are limited by the difficulties to estimate respiratory system mechanics or the dependency on the mode of support [18, 23, 24].

Recently, a new technology has been introduced aiming to monitor and improve patient ventilator interaction (PVI monitor, YRT, Winnipeg, Canada) [25]. With the PVI monitor, a signal representing an estimate of the patient's total inspiratory muscle pressure ($Pmus_{PVI}$) is calculated via the equation of motion, using estimated values of resistance and elastance of the respiratory system, obtained without additional interventions in mechanically ventilated patients [25]. The waveform of $Pmus_{PVI}$ is continuously displayed on-line on a breath-by-breath basis. The effectiveness of the PVI monitor in terms of identifying triggering delay, ineffective efforts and expiratory asynchrony has been retrospectively evaluated and recently reported by a study using pre-existing recording of flow, volume and airway pressure [25]. Since then, PVI software has been upgraded for the purpose of quantifying inspiratory muscle pressure. Thus, the aim of this study is to prospectively examine if inspiratory muscle pressure can be quantified during assisted mechanical ventilation using the method of PVI technology.

Methods (see also electronic supplementary material)

Patients

Thirteen patients admitted to the Intensive Care Unit for management of acute respiratory failure were studied. At the time of the study all patients were lightly sedated with propofol (Ramsay scale score 3), hemodynamically stable and ventilated on proportional assist ventilation (PAV) with the load-adjustable gain factors model with the ability to measure respiratory system mechanics (PAV+) semi-continuously (Nellcor Puritan Bennett LLC, Gosport, UK,). The study was approved by the hospital ethics committee, and informed consent was obtained from the patients or their families.

Measurements

Flow (V'), volume (V) and airway (Paw), esophageal (Pes) and gastric (Pga) pressures were measured as described previously [26, 27]. The proper position of the balloons was verified using standard tests [27, 28]. Transdiaphragmatic (Pdi) and transpulmonary (Ptp) pressures were derived by subtraction of Pes from Pga and Pes from Paw, respectively. Each signal was sampled at 200 Hz (WinDaq Instruments, Akron, OH) and stored on a computer disk for later analysis.

Study protocol

The patients were studied in semi-recumbent position (>45°) in order to obtain a Pes signal as accurately as possible. The patients were studied randomly at three levels of assist, 30, 50 and 70%. At each level the patients remained for 30 min.

Calculations and data analysis

Respiratory system and chest wall mechanics

Passive mechanical ventilation. At the end of the study, the patients were placed on volume-control, constant flow mode and ventilated passively with V_T of 10 ml/kg. End-inspiratory respiratory system elastance (Ers_p), maximum and minimum inspiratory system resistances (R_{max} and R_{min}), as well as chest wall elastance (Ecw_p) and resistance (Rcw) were measured by the technique of rapid airway occlusion using standard formulas [28].

Assisted mechanical ventilation. Mechanics during assisted ventilation were calculated at each level of assist using the unique feature of proportional assist ventilation, which is the tight link between neural inspiration and ventilator pressure [23, 24]. Initially at each level of assist the ventilator software calculated end-inspiratory respiratory system elastance (Ers_a), corrected for the presence of intrinsic PEEP [29]. Dynamic lung elastance (E_L) was calculated by dividing the difference in Ptp at zero flow in the beginning and end of inflation by the corresponding V_T [30]. E_L was subtracted from Ers_a to obtain chest wall elastance during active breathing (Ecw_a). Assuming that Rcw did not differ during active and passive respiration,

chest wall resistance was not calculated during active conditions.

Resistance of respiratory system during active respiration was calculated by (1) the ventilator software of PAV+ (Rrs_{PAV+}) and (2) the iso-volume technique (Rrs_{iso}) [30]. With the latter method lung resistance (R_L) was estimated, whereas Rrs_{iso} was obtained by adding Rcw to R_L [30].

Determination of inspiratory muscle pressure ($Pmus_{PVI}$)

$Pmus_{PVI}$ was calculated on a breath-by-breath basis by a research prototype (PVI Monitor, YRT Limited, Winnipeg, Canada) using a method described in detail previously [25]. Briefly, the inputs required by the monitor to calculate $Pmus_{PVI}$ were Paw and V', whereas V was obtained by V' integration. At least two points during expiration that satisfied passive conditions (i.e., flow was driven by the elastic recoil pressure) were automatically identified by the monitor. At these points, the equation of motion was applied, and elastance (Ers_{PVI}) and resistance (Rrs_{PVI}) of the respiratory system were calculated. Using these values and the equation of motion, $Pmus_{PVI}$ was calculated during each breath. Further fine-tuning of Ers_{PVI} and Rrs_{PVI} was performed to eliminate artifacts derived from $Pmus_{PVI}$ waveform. The monitor may calculate $Pmus_{PVI}$ during ineffective efforts (i.e., absence of ventilator triggering) using the estimated values of Ers_{PVI} and Rrs_{PVI} from the previous breaths and the change in V', V and Paw caused by the ineffective effort. The proper function of the monitor requires a peak inspiratory Paw at least 3 cmH_2O above PEEP. At lower Paw the calculation of respiratory system mechanics may not be reliable. However, under this circumstance the monitor continues to display the Pmus waveform, but its amplitude may not be accurate.

Pressure calculation generated by all respiratory muscles (Pmus)

Pmus was calculated from Pes taking into account the elastic and resistive properties of the chest wall. This calculation, which is based on the Campbell diagram (analysis of esophageal pressure–volume loops), has been previously described [17]. Pmus was calculated using both the values of Ecw_a ($Pmus_a$) and Ecw_p ($Pmus_p$).

Data analysis

The last 2 min of each 30-min period was analyzed and averaged to give the breath variables corresponding to each experimental condition. Patient mechanical inspiratory time was measured using the Pdi, $Pmus_a$, $Pmus_p$ and $Pmus_{PVI}$ signals as the interval between the beginning of the signal increase and the point at which these signals started to decline rapidly (T_IPdi, T_IPmus_a, T_IPmus_p and T_IPmus_{PVI}, respectively). Patient mechanical expiratory time was measured as the remainder of the respiratory cycle, determined from the corresponding waveforms (T_EPdi, T_EPmus_a, T_EPmus_p and T_EPmus_{PVI}, respectively). The rate of rise of these signals (dp/dt) was calculated as the difference between the signal peak and the value at the onset of signal increase divided by the corresponding time. Pressure time product (PTP) of these signals was also calculated as the area under the curve during the corresponding inspiratory time (PTPPdi, $PTPPmus_a$, $PTPPmus_p$ and $PTPPmus_{PVI}$). The level of PEEPi during the different experimental conditions was measured as the positive deflection of Pdi from the onset of neural inspiration to the point of zero flow [29].

Statistical analysis

Data were analyzed by multi-factor analysis of variance for repeated measurements (ANOVA), followed by Tukey's test for multiple comparison if the F value was significant. A p less than 0.05 was considered statistically significant. All values are expressed as mean ± SD.

Results (see also ESM)

Patients' characteristics are shown in Table 1. Two patients were excluded from the analysis (no. 12 and 13) for technical reasons (see ESM). In one patient (no. 3) Pes and Pga data were accidentally lost after Pdi calculation and thus only comparison between Pdi and $Pmus_{PVI}$ was made for this patient. Excluding this patient, the results did not change (see below). In one patient (no. 4) the PVI monitor did not estimate $Pmus_{PVI}$ at 30% of assist due to low peak Paw (<3 cmH_2O above PEEP) at this level of support.

Peak Paw increased significantly with increasing the level of assist, averaging 12.1 ± 2.1, 16.1 ± 3.6 and 19.8 ± 5.1 cmH_2O at 30, 50 and 70% of assist, respectively. V_T and breathing frequency (Fr) did not differ as a function of assist (V_T 0.40 ± 0.1, 0.43 ± 0.12, 0.43 ± 0.1 l; Fr 24.0 ± 6.3, 22.6 ± 6.8, 20.7 ± 5.5 br/min, at 30, 50 and 70%, respectively). Ineffective efforts were not identified during the study period.

Representative waveforms of Paw, flow, volume, Pdi and $Pmus_{PVI}$ are shown in Fig. 1. Independent of the method, all indices of inspiratory effort increased significantly with decreasing assist. For a given assist, peak $Pmus_{PVI}$, $dPmus_{PVI}/dt$ and $PTPPmus_{PVI}$ per breath did not differ from the corresponding values calculated using the Pdi, $Pmus_p$ and $Pmus_a$ signals (Table 2 and Figure S1). Although indices of inspiratory effort calculated from Pdi

Table 1 Patients' characteristics

No.	Sex	Age	Days on MV	Admission diagnosis
1	M	47	18	Lymphoma–encephalitis
2	M	63	10	Aspiration pneumonia
3	M	53	11	ARDS
4	M	70	6	CHF-pneumonia
5	F	75	26	AECOPD
6	M	70	26	CHF
7	M	72	15	Aspiration pneumonia
8	F	62	5	Polyneuromyopathy
9	M	77	12	ARDS
10	F	75	44	Pneumonia
11	M	74	3	Sepsis-ARDS
12	M	64	54	AECOPD-pneumonia
13	M	72	21	CHF-AECOPD

MV mechanical ventilation, *ARDS* acute respiratory distress syndrome, *CHF* congestive heart failure, *AECOPD* acute exacerbation of chronic obstructive pulmonary disease

were lower than those from the other three signals, the difference was not significant. Inspiratory and expiratory time did not differ among the various methods of calculation (Table 2). The onset of inspiratory effort identified by PVI monitor was delayed from that in Pdi signal by 0.06 ± 0.05, 0.05 ± 0.04 and 0.05 ± 0.03 s at 30, 50 and 70% assist, respectively.

Figure 2 shows mean Pdi, $Pmus_{PVI}$, $Pmus_p$ and $Pmus_a$ at 20, 40, 60, 80 and 100% of the corresponding inspiratory time, respectively. At any given % of inspiratory time $Pmus_{PVI}$, $Pmus_p$ and $Pmus_a$ were similar. At any given % of inspiratory time Pdi was slightly but non-significantly lower than $Pmus_{PVI}$, $Pmus_p$ and $Pmus_a$. In all patients $Pmus_{PVI}$, as a function of % of inspiratory time, was linearly related to Pdi, $Pmus_p$ and $Pmus_a$. The slope of the Pdi-$Pmus_{PVI}$ relationship was significantly higher than that of $Pmus_p$-$Pmus_{PVI}$ and $Pmus_a$-$Pmus_{PVI}$ (Table 3).

Figure S3 shows the mean increase in Pdi, $Pmus_{PVI}$, $Pmus_a$ and $Pmus_p$ as the level of assist decreased from 70 to 30%. For a given % of inspiratory time, the increase did not differ among the various methods of respiratory motor output calculation.

Respiratory system mechanics calculated during active respiration by the PVI monitor as well as by the PAV+ mode did not differ as a function of assist, and thus they were averaged for the simplicity of comparison. Although Ers_p (24.5 ± 8.2 cmH_2O/l) was higher than Ers_{PAV+} (20.3 ± 5.1) and Ers_{PVI} (22.3 ± 6.8), the difference was not significant. Rrs_{PVI} (9.9 ± 5.0 $cmH_2O/l/s$), Rrs_{PAV+} (8.5 ± 3.4) and Rrs_{iso} (10.6 ± 5.6) were significantly higher than Rmin (4.8 ± 3.2). Rmax, Rrs_{PVI}, Rrs_{PAV+} and Rrs_{iso} did not differ significantly. Figures S7

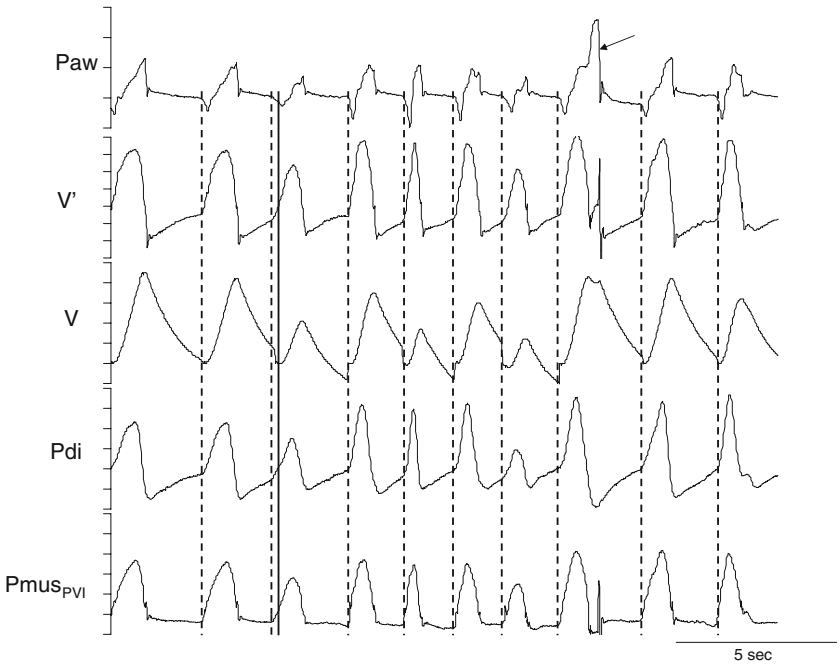

Fig. 1 Airway pressure (Paw 2 cmH_2O/division), flow (V' 0.2 l/s/division), volume (V 0.1 l/division), transdiaphragmatic pressure (Pdi 5 cmH_2O/divison) and inspiratory pressure calculated by the PVI monitor ($Pmus_{PVI}$ 5 cmH_2O/divison) in a representative patient ventilated with 30% of assist. *Arrow* indicates the breath in which occlusion at the end of inspiration was performed by the PAV+ software to calculate respiratory system mechanics. The *dashed vertical lines* indicate the beginning of patient's inspiration as defined by the Pdi waveform (rapid increase in Pdi from the value at the end of expiration). The *continuous vertical line* of the third breath indicates the point of zero flow. Notice that in all breaths $Pmus_{PVI}$ waveform tracks closely that of Pdi in terms of timing (see the *dashed lines*), amplitude (see the breath variability) and shape, even in the presence of dynamic hyperinflation [see the third breath in which intrinsic PEEP (measured as the change in Pdi to reverse flow from expiratory to inspiratory) was 3.12 cmH_2O]

Table 2 Breath characteristics as a function of assist level obtained by $Pmus_{PVI}$, Pdi, $Pmus_p$ and $Pmus_a$ signals

Assist	30%				50%				70%			
Signal	$Pmus_{PVI}$	Pdi	$Pmus_p$	$Pmus_a$	$Pmus_{PVI}$	Pdi	$Pmus_p$	$Pmus_a$	$Pmus_{PVI}$	Pdi	$Pmus_p$	$Pmus_a$
Peak	14.8 ± 5.6 (14.3 ± 5.7)	13.0 ± 7.9 (12.7 ± 8.3)	14.8 ± 5.7	14.6 ± 5.8	12.0 ± 5.2 (11.7 ± 5.4)	10.9 ± 7.2 (10.9 ± 7.6)	13.2 ± 5.5	12.8 ± 5.4	10.5 ± 5.0* (10.4 ± 5.7)	8.6 ± 6.1* (8.7 ± 6.5)	10.5 ± 4.5*	10.3 ± 4.7*
dp/dt	26.7 ± 15.7 (26.3 ± 16.5)	24.1 ± 16.9 (24.1 ± 17.1)	26.6 ± 14.2	27.3 ± 14.7	22.6 ± 13.0 (22.3 ± 13.7)	20.4 ± 14.7 (20.6 ± 15.4)	25.0 ± 12.5	24.7 ± 12.6	19.9 ± 12.6* (19.9 ± 13.3)	16.7 ± 14.4* (17.0 ± 13.1)	19.2 ± 11.0*	19.2 ± 11.8*
PTP/br	4.72 ± 1.7 (4.45 ± 1.5)	3.92 ± 2.7 (3.75 ± 2.8)	4.83 ± 2.4	4.57 ± 2.0	3.75 ± 1.3 (3.78 ± 1.4)	3.22 ± 2.2 (3.20 ± 2.3)	4.38 ± 1.9	4.42 ± 2.1	3.61 ± 1.5* (3.56 ± 1.6)	2.73 ± 1.7* (2.74 ± 1.8)	4.10 ± 1.9*	3.9 ± 1.8*
T_I	0.66 ± 0.3 (0.66 ± 0.3)	0.66 ± 0.3 (0.66 ± 0.3)	0.68 ± 0.3	0.68 ± 0.3	0.63 ± 0.2 (0.64 ± 0.2)	0.62 ± 0.2 (0.62 ± 0.2)	0.63 ± 0.2	0.63 ± 0.2	0.63 ± 0.2 (0.63 ± 0.2)	0.64 ± 0.2 (0.64 ± 0.2)	0.65 ± 0.2	0.67 ± 0.2
T_E	2.15 ± 0.85 (2.26 ± 0.8)	2.15 ± 0.87 (2.27 ± 0.8)	2.25 ± 0.82	2.24 ± 0.83	2.28 ± 0.82 (2.39 ± 0.8)	2.29 ± 0.83 (2.41 ± 0.8)	2.39 ± 0.78	2.40 ± 0.77	2.50 ± 0.83 (2.61 ± 0.8)	2.50 ± 0.84 (2.59 ± 0.8)	2.59 ± 0.81	2.56 ± 0.81

Values are mean ± SD, * $p < 0.05$. At 50 and 70% of assist, data based on $Pmus_p$ and $Pmus_a$ signals pertain to ten patients. At 30% of assist, data based on $Pmus_{PVI}$ and Pdi pertain to ten patients, whereas those based on $Pmus_p$ and $Pmus_a$ pertain to nine patients. The values in parentheses are obtained by excluding patient no. 3

$Pmus_{PVI}$ inspiratory muscle pressure calculated by the PVI monitor, *Pdi* transdiaphragmatic pressure. *$Pmus_p$*, *$Pmus_a$* inspiratory muscle pressure calculated using chest wall mechanics measured during passive and active respiration, respectively

and S8 show the relation and Bland-Altman analysis between Ers_{PVI} and Rrs_{PVI} and the corresponding respiratory system mechanics obtained during active (Ers_{PAV+}, Rrs_{PAV+}, Rrs_{iso}) and passive (Ers_p, Rmin, Rmax) conditions. There was a significant correlation in all cases, but there was considerable scatter. PVI mechanics were better correlated with those obtained during active breathing than those during passive mechanical ventilation.

Discussion

The main finding of this study is that a signal generated from flow, volume, and airway pressure can be used to provide breath-by-breath quantitative information of inspiratory muscle pressure in mechanically ventilated critically ill patients.

In our study inspiratory muscle pressure was assessed by (1) Pdi and (2) calculation of the pressure developed by all respiratory muscles (Pmus) using the Campbell diagram and chest wall mechanics during active ($Pmus_a$) and passive respiration ($Pmus_p$). We did not rely only on transdiaphragmatic pressure to estimate inspiratory muscle pressure because critically ill patients usually use other inspiratory muscles in addition to the diaphragm because of diaphragmatic dysfunction and increased workload [31, 32]. For this reason, Pmus waveform is a better reflection of respiratory muscle activity than transdiaphragmatic pressure. Indeed, Pdi-derived indices of inspiratory effort tended to be slightly lower, although not significantly, than those derived using $Pmus_a$ and $Pmus_p$, indicating that in these patients the diaphragm was not the only inspiratory muscle that determined total inspiratory muscle pressure.

We chose to study patients ventilated with PAV+ mode because with this modality it was possible to calculate chest wall elastance during active breathing and thus to eliminate the errors, if any, in calculation of Pmus using Pes waveform during active breathing and chest wall mechanics measured during passive mechanical ventilation (control mode) at different times. It is well known that both the type of breathing (active vs. passive) and time may affect, sometimes substantially, respiratory system mechanics [23, 24]. Nevertheless, there is no reason to believe that our results may not be applied in patients ventilated with other modes of assisted mechanical ventilation. The inputs required by the monitor to generate inspiratory muscle pressure waveform are airway pressure, flow and volume independent of their shape. Indeed, it has been shown that the PVI monitor can generate a reliable, at least in terms of timing, Pmus signal during pressure support ventilation, even at the absence of ventilator triggering (ineffective triggering) because of dynamic hyperinflation [25]. This is because the monitor calculates $Pmus_{PVI}$ based on a change of flow,

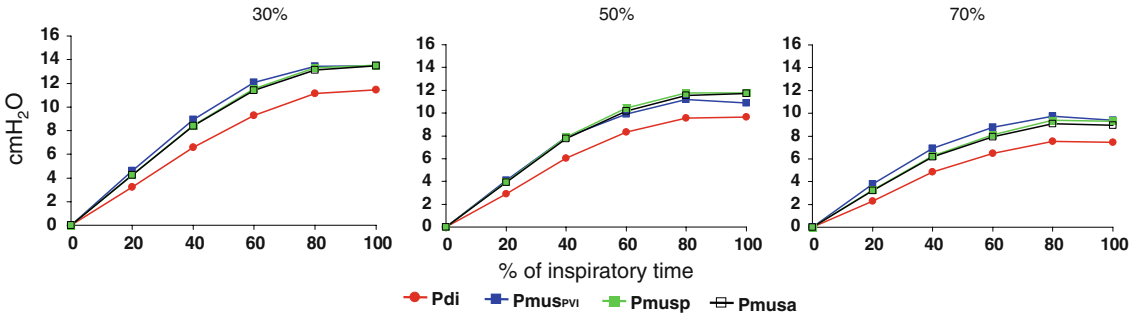

Fig. 2 Pdi, $Pmus_{PVI}$, $Pmus_p$ and $Pmus_a$ at 20, 40, 60, 80 and 100% of inspiratory time with 30, 50 and 70% of assist. $N = 10$ (patient no. 3 was excluded, see "Results"). SD was omitted for clarity of presentation

Table 3 Individual and mean values of slope (*S*), intercept (*I*) and *r* value of Pdi-$Pmus_{PVI}$, $Pmus_p$-$Pmus_{PVI}$ and $Pmus_a$-$Pmus_{PVI}$ relationships during inspiration

Pt. no.	Pdi-$Pmus_{PVI}$			$Pmus_p$-$Pmus_{PVI}$			$Pmus_a$-$Pmus_{PVI}$		
	S	*I*	*r*	*S*	*I*	*r*	*S*	*I*	*r*
1	1.17	−0.49	0.98	1.59	−0.14	0.92	1.49	−0.02	0.93
2	1.02	−0.14	0.996	0.96	−0.29	0.996	0.91	−0.15	0.995
3	1.25	0.46	0.95	NA	NA	NA	NA	NA	NA
4	1.20	1.64	0.91	1.01	1.20	0.95	1.06	1.37	0.92
5	2.03	1.38	0.94	1.05	−0.11	0.99	1.06	−0.20	0.99
6	1.06	3.00	0.83	1.04	2.60	0.85	1.05	2.65	0.85
7	0.76	1.41	0.99	0.92	1.45	0.99	0.91	1.47	0.99
8	1.66	0.22	0.95	0.59	0.59	0.99	0.65	0.07	0.98
9	1.74	0.90	0.94	0.90	−0.19	0.97	0.88	−0.13	0.97
10	1.27	0.48	0.83	0.72	0.31	0.80	0.80	0.17	0.83
11	1.01	0.21	0.98	0.89	0.22	0.97	0.99	0.007	0.98
Mean	1.29	0.82		0.97*	0.56		0.98*	0.52	
SD	0.37	0.99		0.26	0.93		0.22	0.97	

$Pmus_{PVI}$ inspiratory muscle pressure calculated by the PVI monitor, *Pdi* transdiaphragmatic pressure. $Pmus_p$, $Pmus_a$ inspiratory muscle pressure calculated using chest wall mechanics measured during passive and active respiration, respectively
* $p < 0.05$

volume and airway pressure caused by the application of inspiratory muscle pressure on the respiratory system. Thus, provided that peak airway pressure during inspiration is at least 3 cmH_2O above PEEP, the monitor can calculate inspiratory muscle pressure on a breath-by-breath basis independent of the mode of support. Nevertheless, the quantification of inspiratory muscle pressure during other than PAV+ assisted modes remains to be studied.

At all levels of assist inspiratory muscle pressure estimated with PVI was similar to that obtained using the Campbell diagram. The slope of $Pmus_a$-$Pmus_{PVI}$ and $Pmus_p$-$Pmus_{PVI}$ approaches unity, and is significantly lower than that of Pdi–$Pmus_{PVI}$, indicating that indeed Pdi in these patients underestimates inspiratory muscle pressure. Furthermore, $Pmus_{PVI}$ tracked with accuracy the load-induced changes in inspiratory muscle pressure; for a given % of inspiratory time, the increase in inspiratory muscle pressure when the assist decreased from 70 to 30% did not differ among the various methods of inspiratory muscle pressure calculation. It follows that, in critically ill patients, it is feasible to quantify the inspiratory muscle pressure without any intervention, using flow, volume and airway pressure, signals that may be easily obtained by the ventilator.

Our study showed that respiratory system mechanics estimated using PVI technology were similar to those estimated during active respiration using the ventilator software with PAV+ mode [23, 24] and the iso-volume technique of resistance measurement [30]. A good correlation was found for both elastance and resistance between PVI and PAV+ and iso-volume methods. Although good correlation was also observed between PVI-estimated and passive mechanics, resistance was higher and elastance lower when measured with PVI than when measured during passive mechanical ventilation. The difference between PVI and passive mechanics might not be entirely related to the methods of measurements for several reasons. Firstly, the measurements by the two methods were performed at least 1.5 h apart, and

respiratory system mechanics may have changed in the interval [23, 24]. Secondly, mechanics during passive mechanical ventilation may differ from those during active breathing [23, 24]. Thirdly, the deepening of sedation and the use of muscle relaxants in some patients to achieve passive conditions may have affected the mechanical properties of the respiratory system [33]. Fourthly, PVI estimates expiratory elastance and an average of inspiratory and expiratory resistances, whereas elastance and resistance measured during passive ventilation are inspiratory. Inspiratory resistance in critically ill patients may be several fold lower than expiratory [26, 34].

Younes et al. [25] showed that the PVI signal is able to identify ineffective efforts as well as the onsets and ends of patient mechanical inspiratory efforts with reasonable accuracy. Our results confirmed these findings. We demonstrated that inspiratory time, identified by the PVI signal, did not differ from that obtained using standard methods (Pdi or Pmus waveforms). In agreement with Younes et al. [25] the onset of inspiratory effort in PVI signal was very close to that in Pdi, the time delay between the events in the two signals averaging approximately 50 ms. Furthermore, our study showed that PVI technology may be used to estimate the respiratory system mechanics and thus quantify the inspiratory muscle pressure. Indeed, inspiratory muscle pressure waveform estimated by the PVI monitor did not differ from that calculated using Pes and Pga.

A final point worth mentioning is that the PVI monitor may not be able to calculate a reliable Pmus signal in patients with very severe flow limitation in whom inspiratory efforts fail to distort airway pressure and/or flow [25]. Therefore, in these patients phenomena recorded at ventilator line (pressure and flow) are dissociated from that produced by respiratory efforts and located at alveoli. However, in our study we did not observe such a dissociation, probably because only three patients had obstructive lung disease, and at the time of the study, none of them exhibited severe flow limitation as indicated by the low levels of PEEPi (range 0.2–1.9 cmH_2O).

The results of this study have important clinical implications. By identifying the onset and end of patient mechanical inspiratory time, the PVI signal may facilitate adjustments in ventilator settings to improve the triggering process and patient-ventilator synchrony [35, 36]. In addition, the current study showed that the PVI monitor can quantify inspiratory effort breath by breath and thus provide information about the degree of unloading of inspiratory muscles. Nevertheless, the optimal degree of unloading in mechanically ventilated patients is currently unknown, and thus setting the assist level based on various thresholds of inspiratory efforts [37, 38] may not be appropriate for an individual patient.

In conclusion, our study showed that in mechanically ventilated critically ill patients a signal generated from flow, volume and airway pressure may be used to provide breath-by-breath quantitative information of inspiratory muscle pressure.

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