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Evaluation of a new module in the continuous monitoring of respiratory mechanics

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Abstract *Objective*: Bedside monitoring of respiratory mechanics facilitates the use of lung protective ventilation in acute lung injury (ALI). We evaluated a new clinical monitor of respiratory mechanics.
Design: Prospective, in vitro and in vivo study.
Setting: University hospital.
Patients: Measurements were done using a lung model and in patients after cardiac surgery (n = 10) and in patients with ALI (n = 10).
Interventions and measurements: The monitor provides continuous monitoring of pressure, flow and volume waveform and loop data, and automatically collected variables of respiratory mechanics. Breath-by-breath respiratory mechanics data and the automated variables obtained with the new monitor were compared with flow and pressure reference data.
Results: Waveform data comparison showed errors of less than 5 % for most variables. Automatically recorded respiratory pressures and volumes showed good agreement within clinical standards when compared to reference (errors from 2.5 % to 6.2 %). Automatically recorded derived variables present poor agreement (errors from 8.1 % to 158.3 %).
Conclusions: The waveform data of the new monitor is accurate. The value of the automatically derived variables is limited by the fact that inspiratory plateau pressure and plateau compliance have no direct physiological meaning. Nevertheless, in clinical monitoring much information can be derived from the waveform signals alone and from pressure-volume and flow-volume loops. These facilitate monitoring changes in respiratory mechanics in the ALI patient.

Key words Open heart surgery · Acute respiratory failure · Side stream spirometry · Respiratory mechanics · Bedside monitoring

Introduction

High pressures and volumes during mechanical ventilation may injure previously healthy lung tissue [1, 2, 3, 4]. The use of small tidal volumes combined with higher levels of positive end-expiratory pressure (PEEP) has been advocated to limit the inspiratory pressures and volumes during the treatment of acute lung injury (ALI) patients. The application of respiratory mechanics in clinical practice has been limited due to the fact that these measurements are time-consuming and cumbersome. Online monitoring of respiratory mechanics may help to adjust ventilation in order to avoid lung overdistension and ventilator-induced lung damage.

Most available respiratory mechanics monitors are ventilator specific and the measurements are performed inside the ventilator. This sampling site does not necessarily reflect the changes at the airway, since the total compliance and resistance of the breathing circuit, gas compression, leak, and dead space in the system inter-

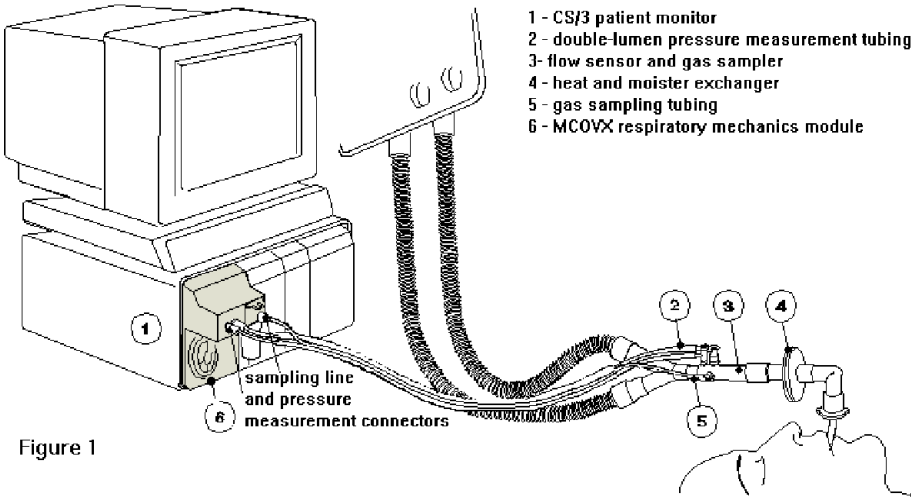


Fig. 1 Complete measurement system including patient monitor, respiratory mechanics module and flow sensor and gas sampler

fere. Furthermore, proper validation of the measurements is often not available.

In this study we aimed at evaluating the precision and bias of a new respiratory mechanics monitor that measures pressure and flow at the airway opening. The disposable sensor has little dead space and allows continuous monitoring and use of adequately humidified gases, without major problems with secretions, all features difficult to achieve with standard pneumotachometers. The waveform and derived mechanics variables of the device were compared to airway flow and pressure references, in order to establish its usefulness in the clinical monitoring of respiratory mechanics.

Material and methods

The respiratory mechanics module (MCOVX, Datex-Ohmeda, Finland) is integrated in a patient monitor (CS/3, Datex-Ohmeda, Finland). The module collects data at the airway opening through a flow sensor and gas sampler based on side stream spirometry (Patient Spirometry, Datex-Ohmeda, Finland) (Fig. 1) [5].

Flow sensor and gas sampler

The sensor consists of a flow-sensing element constructed into a normal side stream gas-sampling adapter. The device is a two-sided Pitot tube, which is essentially a pressure-based flowmeter with one fixed resistance interposed onto the airstream (Fig. 2) [5]. The construction is symmetrical to give the same response in both directions of gas flow (inspiration and expiration). Pressure difference caused by the gas flow over the restrictor is measured between the two pressure ports of the probe. Three channels in the middle of the sensor collect and average the airway pressure signals. A hole for sampling the gas for sidestream analysis penetrates the sensor in the middle of the pressure pick-up channels. The sensor is disposable and made of plastic molded as a single piece. It is 90 mm long (20 mm of which is just the endotracheal tube connector) and weighs 13 g. The dead space of the sensor is 9.5 ml. The flow resistance added by the flow sensor itself is 1 cmH_2O l s when measured at a flow of 0.5 l/s [5].

This sensor is not ventilator specific, being compatible with any kind of ventilatory support. It has proved linear with flows ranging from 0.04 to 1.67 l/s (a maximum error of 4% was found on the higher flows). Errors up to ± 5% were registered with tidal volumes ranging from 0.15 to 1.1 l [5]. These characteristics make the device potentially suitable for monitoring patients with ALI and high flows and small tidal volumes. The small weight and size of the sensor allow it to be placed and kept at the airway opening without interfering with patient nursing, being easily replaced if necessary. Adequate humidification of inspired gases can be kept and the small dead space avoids CO_2 retention.

Complete measurement system

The sensor head is interfaced with the monitor unit with three tubes, each 3 m in length (Fig. 1). Pressure difference and gas concentrations are used to calculate flow [5]. Flow and airway pressures are displayed as curves. Gas temperature is not measured, but a constant correction factor is applied for temperature difference between inspired and expired gas. Volume is integrated from flow [5].

Flow and airway pressure signals are sampled at a frequency of 100 Hz and converted from analog to digital. The digitized data are used to derive respiratory mechanics variables displayed on the screen for every breath and updated every 5 s for trend recording. Peak pressure (Ppeak), inspiratory plateau pressure (Pplat), dynamic total positive end-expiratory pressure (PEEPtot,dyn) [6], set positive end-expiratory pressure (PEEPset), intrinsic positive end-expiratory pressure (PEEPi), inspiratory tidal volume (Vt,insp), expiratory tidal volume (Vt,exp), plateau compliance (Cplat), and resistance (Table 1) are recorded.

The original digital signal (100 Hz) is sampled at 25 Hz in order to create the online flow and pressure waveform displays on the screen. This signal is then reconverted from digital to analog in order to provide the analog data output from the module.

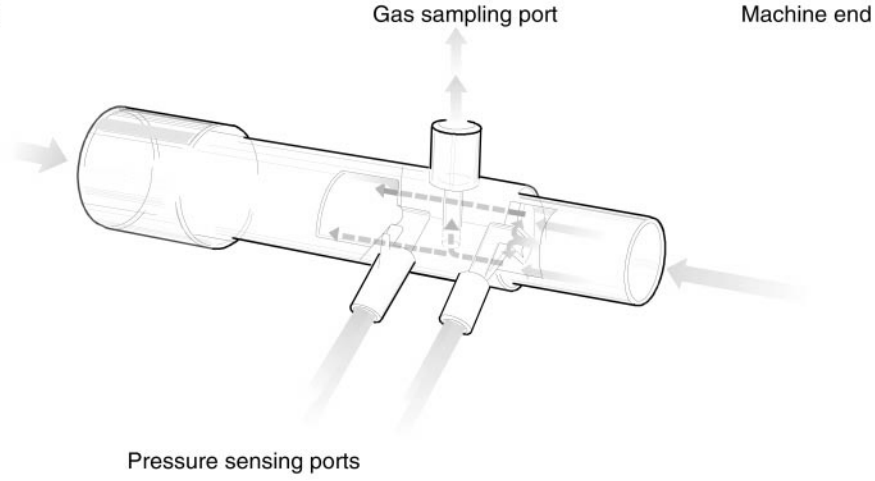


Fig. 2 Flow sensor and gas sampler

Reference system

A heated pneumotachometer (Fleisch No2, Lausanne, Switzerland) and a differential pressure transducer (Validyne, MP45, ± 2.0 cmH_2O; Validyne, Northridge, Calif., USA) were used as the flow reference. Volume was obtained by digital integration of the flow signal. The pneumotachometer system was linear over the range of flows used. Reference airway opening pressure was measured at the mouth via a side port proximal to the endotracheal tube by a pressure transducer (Validyne, MP45, ± 100.0 cmH_2O; Validyne, Northridge, Calif., USA). The sampling tubings used were 15 cm in length.

Comparison setup

The flow sensor and the pneumotachometer were connected in series between the endotracheal tube and the Y piece of the ventilator. The pneumotachometer was proximal to the patient. The gas sampling port in the new flow sensor was kept closed throughout the trial in order to avoid artifacts due to air sampling.

Calibration

Flow calibration of the new mechanics monitor and the pneumotachometer was done simultaneously before every patient (the pneumotachometer and the flow sensor attached together) using a 3 l calibration syringe (Model No 5530, Hans Rudolph, Kansas City, Mo., USA) and flows around 0.5 l/s. The airway pressure reference was calibrated against a water column. No pressure calibration was needed for the new monitor according to the specifications of the manufacturer.

Data collection

A physiologic recording system (Direc, Raytech Instruments, Vancouver, Canada) was used to record waveform signals from the pneumotachometer, differential pressure transducers, and from the analog output of the monitor at a sampling frequency of 100 Hz (Fig. 3). No sampling delays could be found among the multiple channels used. Automatically processed trend data from the mechanics monitor was collected using a laptop PC. Waveform data was analyzed using software programs (Windaq/EX, Dataq Instruments, Akron, Ohio, USA and Anadat, Version 5.2, RHT-InfoDat, Montreal, Quebec, Canada).

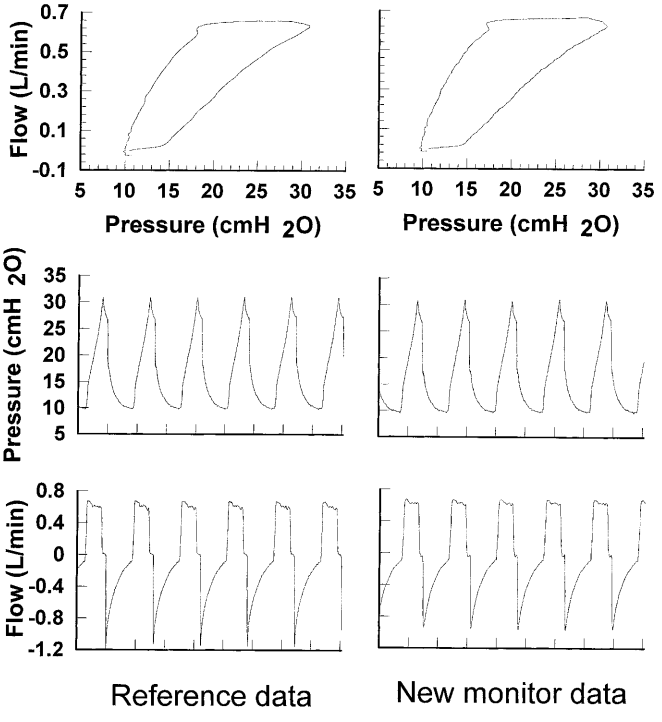


Fig. 3 Waveform and loop data from reference and new monitor from a representative ALI patient

In vitro

Signals from both the pneumotachometer and the mechanics monitor were recorded simultaneously while ventilating a water lung model (SSS-Tester, Datex-Ohmeda, Finland) with a ventilator (Servo 900, Siemens-Elema, Solna, Sweden). No humidification system was used. Fourteen different ventilator settings were stud-

Table 1

a) Automated data from the new module (data given automatically online in numeric format)
Ppeak (inspiratory peak pressure) = maximum pressure

Pplat (inspiratory plateau pressure) = pressure at the flow reversal point at the end of the inspiratory phase

PEEPtot,dyn (dynamic total positive end expiratory pressure) = total pressure in the lungs at the end of expiration, that is the pressure at the moment when flow changes direction between the expiratory and the inspiratory phase

PEEPset (set positive end expiratory pressure) = minimum pressure during expiration, that is the PEEP set from the ventilator

PEEPi (intrinsic positive end expiratory pressure) = PEEPtot,dyn-PEEPset

Vt,insp (inspiratory tidal volume)

VT,exp (expiratory tidal volume)

Cplat (plateau compliance) = Vt,exp/Pplat-PEEPtot,dyn)

Resistance (dynamic resistance at the ventilator frequency) = calculation by least squares fitting, using the equation
p(t) = Resistance*V'(t) + V(t)/Cplat + PEEPtot,dyn, where p(t), V'(t) and V(t) are the pressure, flow and volume measured at the sensor at a time t

b) Manually processed waveform data from both devices (values calculated manually and off-line from waveform data)
* calculated as previously described
Ppeak (inspiratory peak pressure)*

P1 = pressure at zero flow after inspiration, when the closure time of the occlusion valve has been taken into consideration (approximately 100 to 200 ms in Servo 900)

Pst,insp (static inspiratory pressure) = airway pressure at five seconds after airway occlusion at end-inspiration, when alveolar pressures have equilibrated with the ventilator system and a plateau has been reached

PEEPtot,dyn (dynamic total positive end expiratory pressure)*

PEEPtot,st (static total end expiratory pressure) = airway pressure at five seconds after airway occlusion at end-expiration, when alveolar pressures have equilibrated with the ventilator system and a plateau has been reached

Vt,insp (inspiratory tidal volume)

Vt,exp (expiratory tidal volume)

Cdyn (dynamic compliance) = Vt,insp/(P1-PEEPtot,dyn)

Cst (static compliance) = Vt,insp/(Pst,insp-PEEPtot,st)

R,aw (airway resistance) = (Ppeak-P1)/V'

Rtot (total airway resistance) = (Ppeak-Pst,insp)/V'

p(t) = Resistance*V'(t) + V(t)/Cplat + PEEPtot,dyn, where p(t), V'(t) and V(t) are the pressure, flow and volume measured at the sensor at a time t

ied (Table 2). Measurements were done using both heated and non-heated pneumotachometer in order to determine if the change in temperature due to the heating of the pneumotachometer influences the accuracy of the measurements.

The test lung is based on the use of two vertical water columns coupled together.

In vivo

Two groups of patients were included in the study: post cardiac surgery and ALI patients, in order to assess a wide range of respiratory mechanics and ventilator settings.

Ten patients undergoing CABG (coronary artery bypass graft) surgery, who were admitted to the Intensive Care Unit of Kuopio University Hospital, Finland, for postoperative care and ten patients with ALI were studied. One ALI patient was excluded from the data analysis due to spontaneous breathing efforts during the controlled ventilation. The study was approved by the local Ethics Committee. Written informed consent was obtained preoperatively from CABG patients. For ALI patients, the need for consent was waived since the measurements are part of the routine clinical management of ALI patients in this institution [7]. All patients were intubated and ventilated under volume controlled mode (Servo 900C or 900E, Siemens-Elema, Solna, Sweden). Humidification was provided by using active air humidification in ALI patients. No humidification system was used for CABG patients. Demographic and clinical data is presented in Table 3.

In all patients PV-curves were performed using different randomized ventilator settings (Table 2). Some measurements could not be performed due to unacceptably high inspiratory pressures (Ppeak pressures were limited to 45 cmH_2O for clinical safety reasons). The first ten settings recorded in each patient were used for analysis. During each different ventilator setting, static mechanics were recorded using the occlusion method [6, 8]. Periods of 60 s were recorded from the stabilization of signals until performing the airway occlusions in CABG patients, while in ALI patients the period had to be reduced to 30 s due to clinical instability (acute CO_2 retention and high pulmonary pressures caused by the added dead space of the pneumotachometer and the occlusion maneuvers).

Data comparison

The new monitor displays the flow and airway pressure waveform data on screen and respiratory mechanics values are calculated breath by breath. In contrast, data for trend display, including the derived variables, are not collected breath by breath, but sampled every 5 s. Hence, we first established a breath-by-breath comparison using the analog signal output data against reference. We then took the trend data and compared the mean values over a period of time to those manually calculated for the same time period from reference data.

In vitro waveform data

Three nonconsecutive breaths were analyzed per ventilator setting from both the new monitor and reference for Ppeak, PEEPtot,dyn, Vt,insp, peak inspiratory flow, peak expiratory flow, Cst, and Rtot.

In vivo waveform data

We analyzed waveform data from both the new monitor and the reference in all patients and each different ventilator setting for Ppeak, PEEPtot,dyn, Vt,insp, Cdyn [9], R,aw [6], Pst, insp [8, 10], PEEPtot, st [8, 10], Cst [6] and Rtot (Table 4) (190 breaths total). For PEEPtot, dyn calculations, flow was zeroed close to the measurements in order to avoid drift and the pressure values extrapolated to the zero flow point. For Cdyn and R,aw, P1 was corrected for the occlusion time of the ventilator valves.

Table 2 Ventilator settings for in vitro and in vivo measurements

In vitro		In vivo (PV-curve)		
# 1	baseline*	# 1	PEEPset = 9 cmH_2O	120 % Vt
The different settings were obtained by changing individually the following parameters from baseline:		# 2	PEEPset = 9 cmH_2O	100% Vt
		# 3	PEEPset = 9 cmH_2O	80% Vt
# 2	accelerating flow pattern	# 4	PEEPset = 12 cmH_2O	100 % Vt
# 3	I : E = 1 : 4	# 5	PEEPset = 12 cmH_2O	120 % Vt
# 4	I : E = 1 : 1	# 6	PEEPset = 12 cmH_2O	80 % Vt
# 5	I : E = 2 : 1	# 7	PEEPset = 3 cmH_2O	80 % Vt
# 6	PEEPset = 0	# 8	PEEPset = 3 cmH_2O	120 % Vt
# 7	PEEPset = PCV 10 cmH_2O	# 9	PEEPset = 3 cmH_2O	100 % Vt
# 8	PEEPset = PCV 20 cmH_2O	# 10	PEEPset = 6 cmH_2O	80 % Vt
# 9	Frequency = 25 min^{-1}	# 11	PEEPset = 6 cmH_2O	100 % Vt
# 10	Frequency = 35 min^{-1}	# 12	PEEPset = 6 cmH_2O	120 % Vt
# 11	Frequency = 15 min^{-1}	# 13	PEEPset = 15 cmH_2O	80 % Vt
# 12	Frequency = 8 min^{-1}	# 14	PEEPset = 15 cmH_2O	100 % Vt
# 13	mV = 1.65 l/min	# 15	PEEPset = 15 cmH_2O	120 % Vt
# 14	mV = 6.6 l/min	100 % Vt = Vt, insp set by the attending physician		
		120 % Vt = Vt, insp 20 % larger than the clinically set Vt		
		80 % Vt = Vt, insp 20 % smaller than the clinically set Vt		
		Minute ventilation was kept constant		

* Volume controlled mode; constant flow pattern; minute ventilation (mV) = 3.3 l/min; frequency = 15 min^{-1}; PEEPset = 10 cmH_2O; inspiration expiration ratio (I : E) = 1 : 2
I : E = inspiration expiration ratio
PCV = pressure controlled ventilation

In vivo automated data

Four ALI patients (patients numbers 6, 7, 9, and 10) were used to evaluate the automated data provided by the new monitor. Dynamic automated data variables (mean of six data time points over 30 s) were compared to the mean of three non-consecutive breaths of the same time period from reference (120 breaths total). The occlusion breaths were analyzed to assess static mechanics from reference (40 breaths total). No correction was used for the resistive pressure drop due to the endotracheal tube in airway resistance calculations, as this was not taken into consideration by the algorithm used in the new mechanics monitor.

Statistical analysis

Paired t-tests were done and percentage errors [100* (new module – reference)/reference] calculated for all variables in order to evaluate the differences between devices. Independent samples t-test was used to compare the differences between devices in the cardiac surgery patients with those of the ALI patients. $P < 0.05$ was considered significant. Values were given as mean ± SEM. Bland-Altman plots [11] were created for derived variables.

Results

In vitro

The percentage errors were less than 3.5 % for all variables, except for Vt,insp (4.3 ± 0.3 % for heated and 6.9 ± 0.4 % for non-heated pneumotachometer) and Rtot (–9.3 ± 1.3 % for heated pneumotachometer).

Table 3 Demographic and clinical data

Open heart surgery patients					
	Age (years)	Weight (kg)	CABG #	PaO_2/FiO_2	Cst (ml/cmH_2O)
# 1	68	70	5	272	78
# 2	75	74	6	163	48
# 3	73	95	5	208	58
# 4	67	60	3	221	47
# 5	58	73	3	411	89
# 6	62	89	5	225	95
# 7	67	70	3	203	67
# 8	69	82	6	311	57
# 9	72	70	6	234	79
# 10	64	64	3	210	33
Acute lung injury patients					
	Age (years)	Weight (kg)	Etiology	PaO_2/FiO_2	Cst (ml/cmH_2O)
# 1	61	65	Vasculitis	105	68
# 2	36	78	Pneumonia	76	32
# 3	42	92	Pancreatitis	60	40
# 4	31	68	Intoxication	131	14
# 5	28	80	Trauma	70	50
# 6	25	76	Pneumonia	131	48
# 7	49	68	Pneumonia	70	33
# 8	58	80	Pneumonia	80	65
# 9	58	56	LED	60	25
# 10	58	56	LED	55	33

CABG # – number of coronary artery bypass grafts
Cst – static compliance of the respiratory system (Vt/(Pst,insp-PEEPtot,st))
LED – Lupus eritematosus disseminatus

Table 4 Waveform data comparison between reference and the new respiratory module

		Dynamic mechanics				
		New monitor	Reference	% error	Differences	Paired T-test
Ppeak (cmH_2O)	CABG	21,2 ± 0,5	21,2 ± 0,5	0,2 ± 0,1	0,06 ± 0,03	0,035
	ALI	30,9 ± 0,7	30,8 ± 0,7	0,2 ± 0,1	0,10 ± 0,03	0,001
PEEPtot,dyn (cmH_2O)	CABG	7,7 ± 0,4	8,0 ± 0,4	–4,6 ± 1,1	–0,28 ± 0,04	0,000
	ALI	7,8 ± 0,4	8,1 ± 0,4	–4,6 ± 1,1	–0,30 ± 0,05	0,000
Vt,insp (ml)	CABG	463 ± 10	463 ± 11	0,2 ± 0,3	–0,51 ± 1,82	0,781
	ALI	502 ± 17	471 ± 16	7,3 ± 0,5	30,59 ± 1,98	0,000
Cdyn (ml/cmH_2O)	CABG	43,7 ± 1,0	45,0 ± 1,0	–2,7 ± 0,6	–1,29 ± 0,32	0,000
	ALI	29,7 ± 1,4	28,8 ± 1,4	3,9 ± 0,7	0,99 ± 0,22	0,000
Raw (cmH_2O/L/sec)	CABG	4,8 ± 0,1	4,8 ± 0,1	0,8 ± 1,8	–0,01 ± 0,08	0,889
	ALI	3,3 ± 0,2	4,0 ± 0,2	–17,4 ± 2,2	–0,75 ± 0,08	0,000

CABG – coronary artery bypass graft patients; ALI – acute lung injury patients

In vivo

The differences between devices in the waveform data were less than 5% except for the measurements of Vt,insp, R,aw, Cst, and Rtot, in the ALI population (–17.4 to 7.3% error) (Tables 5 and 6). Cst and Cdyn were underestimated by the new monitor in the CABG population, and overestimated in the ALI population (CABG vs ALI population: Cst and Cdyn; $P < 0.001$) (Fig. 4). Rtot was over and underestimated by the new monitor in the CABG and the ALI populations, respectively ($P < 0.001$).

Comparison of automated data from the new mechanics monitor to manually calculated variables from

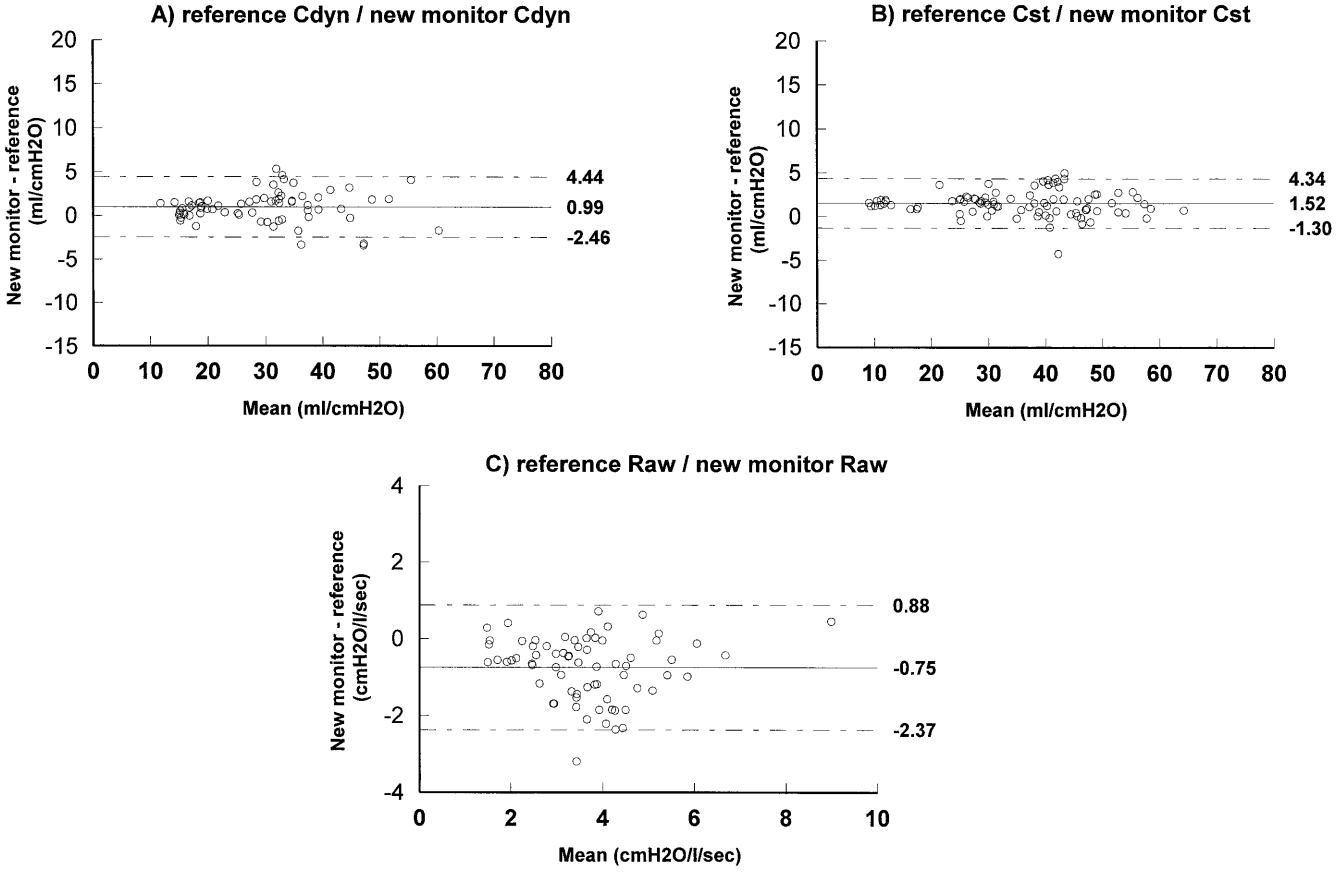


Fig. 4 Bland-Altman plots for comparison of waveform data from reference and from the new monitor in the ALI population

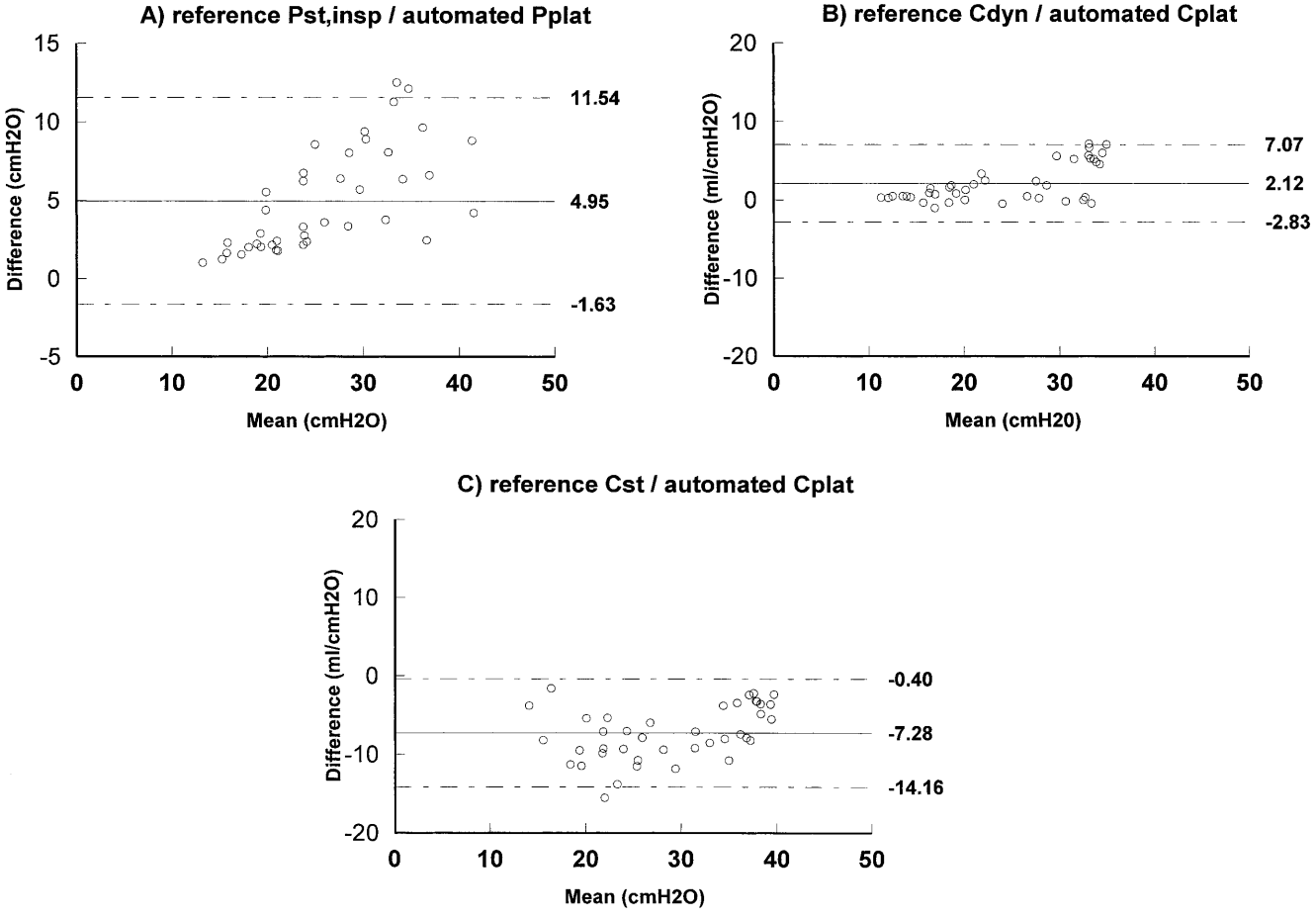
Table 5 Waveform data comparison between reference and the new mechanics monitor

		Occlusion method				
		New monitor	Reference	% error	Differences	Paired T-test
Pst,insp (cmH_2O)	CABG	16,0 ± 0,5	16,0 ± 0,5	–0,5 ± 0,1	–0,06 ± 0,02	0,002
	ALI	23,0 ± 0,6	22,9 ± 0,6	0,1 ± 0,1	0,03 ± 0,02	0,230
PEEPtot,st (cmH_2O)	CABG	7,5 ± 0,4	7,7 ± 0,4	–4,5 ± 1,0	–0,19 ± 0,02	0,000
	ALI	8,2 ± 0,4	8,4 ± 0,4	–2,5 ± 0,3	–0,16 ± 0,02	0,000
Cst (ml/cmH_2O)	CABG	57,4 ± 1,2	58,5 ± 1,2	–1,7 ± 0,3	–1,01 ± 0,21	0,000
	ALI	36,2 ± 1,4	34,6 ± 1,4	5,7 ± 0,6	1,52 ± 0,15	0,000
Rtot ($cmH_2O/L/sec$)	CABG	10,1 ± 0,3	10,0 ± 0,3	1,5 ± 0,5	0,16 ± 0,05	0,001
	ALI	15,0 ± 0,8	16,6 ± 0,9	–8,5 ± 0,7	–1,56 ± 0,16	0,000

CABG – coronary artery bypass graft patients; ALI – acute lung injury patients

reference waveform data shows errors of 0.2 ± 0.5 % ($P = 0.580$), –2.5 ± 1.8 % ($P = 0.002$), and 6.2 ± 0.7 % ($P < 0.001$) for Ppeak, PEEPtot,dyn, and Vt,insp, respectively (mean differences of 0.08 ± 0.15 cmH_2O, –0.21 ± 0.06 cmH_2O, and 31.36 ± 4.07 ml, respectively). Automated Pplat overestimates reference Pst,insp (20.4 % error) (Fig. 5 a). Automated Cplat presents an error of 8.1 ± 1.3 % when compared to reference Cdyn and of –23.9 ± 2.0 % when compared to reference Cst (Figs. 5 b and 5 c). Automated resistance values present errors of 158.3 ± 19.5 % and –32.9 ± 2.8 % when compared to R,aw and Rtot, respectively, (mean differences of 5.4 ± 0.3 and –6.4 ± 1.0 $cmH_2O/l/s$).

Fig. 5 Bland-Altman plots for comparison of waveform and automated data



Discussion

Accurate bedside measurements of respiratory mechanics help to optimize mechanical ventilation during a lung protective approach to ALI patients. Most of the existing devices have become available only recently and are ventilator specific. Furthermore, most of these devices have not been validated and perform the respiratory mechanics measurements in the ventilator, which does not necessarily reflect the changes at the airway. The inspired volume lost in the breathing circuit of some ventilators due to compliance of the system can reach 2–3 ml for each cmH_2O increase in peak pressure [12]. Any leakage in the system during inspiration reduces the tidal volume. Similar errors can be expected during expiration.

In general, the pressure, flow, and volume measurements as well as the variables derived manually from the waveform data were accurate. The small errors in pressure measurements may be in part related to the resistive pressure drop across the pneumotachometer. For the purpose of the experiment, the sampling line of the new monitor had to be occluded. In these conditions the monitor algorithm assumes a relative humidity of 50%. Since in the ALI population active humidification was used, a 1% overestimation of the flow by the new monitor is likely. Some error may be related to the sampling rates: the analog signal of the new monitor is regenerated from digitized data at 25 Hz, in contrast to the reference signal, sampled at 100 Hz. Visually, data from both devices appears identical (Fig. 3).

In contrast to the accurate waveform data, the automatically calculated data, which is also used for the trends, has both method-related and physiologic limitations. The automatically calculated data must be analyzed as mean values over a period of time, instead of breath-by-breath analysis; this contributes to some inaccuracy.

Pplat and Cplat commonly used as surrogates for static pressure and compliance do not have a clear physiologic meaning. Since these variables are intended to be used in the clinical setting, we compared them to Pst,insp, and Cdyn and Cst (Fig. 5). The poor agreement demonstrates that Pplat cannot be used as a surrogate of Pst,insp and that Cplat cannot be used as a surrogate of the dynamic or static compliance.

The new mechanics monitor calculates resistance by least square fitting of the equation of motion of the respiratory system [13] (see Methods). The different calculation method added to the flow overestimation error of the new monitor may explain, at least in part, the relevant differences found for this parameter. The resistances include the endotracheal tube resistance, which may be considered as a limitation of the monitor, but should not affect the accuracy of measurements. The higher inspiratory flows in the ALI patients (0.55 to 0.65 l/s) when compared to CABG patients (0.35 to 0.55 l/s) and the added resistance of the reference device, may have contributed to the even larger errors of R,aw and Rtot in the ALI patients.

Korst et al. [14] evaluated, using a test lung, the automated respiratory mechanics measurements from several microprocessor controlled ventilators and suggested that the respiratory mechanics software may be used to measure trends in the clinical setting during controlled ventilation. Cplat was used and resistance was not corrected for the endotracheal tube resistance. No information was provided on the algorithms of the microprocessors used and no direct waveform data comparison was done to the reference. Their conclusion is of limited value because the study was done in vitro and physiologically meaningful variables were not accessed.

We conclude that the waveform data of the new monitor is accurate. The value of the automatically derived variables is limited by the fact that Pplat and Cplat have no direct physiological meaning. On the other hand, for clinical monitoring, much information can be derived from the waveform signals alone and the display of pressure-volume and flow-volume loops. These facilitate monitoring changes in respiratory mechanics in the ALI patient.

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