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A method for monitoring and improving patient: ventilator interaction

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Abstract *Objective:* To evaluate a new approach for monitoring and improving patient-ventilator interaction that utilizes a signal generated by the equation of motion, using improvised values for resistance and elastance obtained noninvasively. *Design and setting:* Observational study in intensive care units in five European centers. *Patients:* We studied 21 stable patients instrumented with esophageal/gastric catheters for a previous study and ventilated alternately with pressure support (PSV) and proportional assist (PAV) ventilation with a Tyco 840 ventilator. *Measurements and results:* Previously recorded digital files were analyzed in real-time by a prototype incorporating the new technology (PVI monitor, YRT, Winnipeg, Canada). Actual onsets ($P_{DI-T_{ONSET}}$) and ends ($P_{DI-T_{END}}$) of inspiratory efforts, ineffective efforts, and patient respiratory rate were

identified visually from transdiaphragmatic or calculated respiratory muscle pressure. Monitor-identified T_{ONSET} occurred 0.107 ± 0.074 s after $P_{DI-T_{ONSET}}$, substantially less than trigger delay observed with conventional triggering (0.326 ± 0.086 s). End of effort was identified 0.097 ± 0.096 s after $P_{DI-T_{END}}$, significantly less than actual cycling-off delay during PSV (0.486 ± 0.307 s) or PAV (0.277 ± 0.084 s). The monitor detected 80% of ineffective efforts. There was excellent agreement between monitor-estimated respiratory rate and actual patient rate over a wide range (17–59/min) of patient rates (mean (\pm SD) of difference -0.2 ± 1.9 /min for pressure support and 0.2 ± 0.9 /min for proportional assist) even when large discrepancies existed (> 35 /min) between patient and ventilator rates. *Conclusions:* The proposed approach should make it possible to improve patient-ventilator interaction and to obtain accurate estimates of true patient respiratory rate when there is nonsynchrony.

Keywords Mechanical ventilation · Trigger delay · Cycling-off delay · Ineffective efforts

Introduction

With conventional pressure or flow triggering, when dynamic hyperinflation is present, triggering does not occur until the patient makes enough effort to overcome the elastic recoil of the respiratory system present at the onset of effort [1–3]. Because dynamic hyperinflation is very frequent in ventilated patients [1, 2, 4], triggering is often substantially delayed. In many cases the effort terminates before the ventilator is triggered (ineffective efforts [1, 4, 5]). With pressure support and volume-cycled ventilation the ventilator cycle is not always constrained to cycle-off at the end of inspiratory effort [3]. As a result ventilator cycle may terminate before the end of effort or extend well into the patient's expiratory phase [3, 5]. There are reasons to believe that nonsynchrony may have clinical consequences [6, 7]. Asynchrony may result in increased need for sedation, impairment of sleep quality, or respiratory muscle injury (see [7] for review). Nonsynchrony may also lead to false diagnosis of weaning failure, with unnecessary continuation of mechanical ventilation [5, 7]. These consequences may underlie the recent demonstration that the duration of mechanical ventilation is longer in patients with nonsynchrony [6].

An entirely different approach to triggering and cycling-off has recently been proposed [8] in which the equation of motion [9, 10] is used to generate in real-time a signal that reflects respiratory muscle pressure output. However, instead of requiring real values of passive elastance and resistance, which are difficult to obtain during assisted ventilation, this approach utilizes improvised

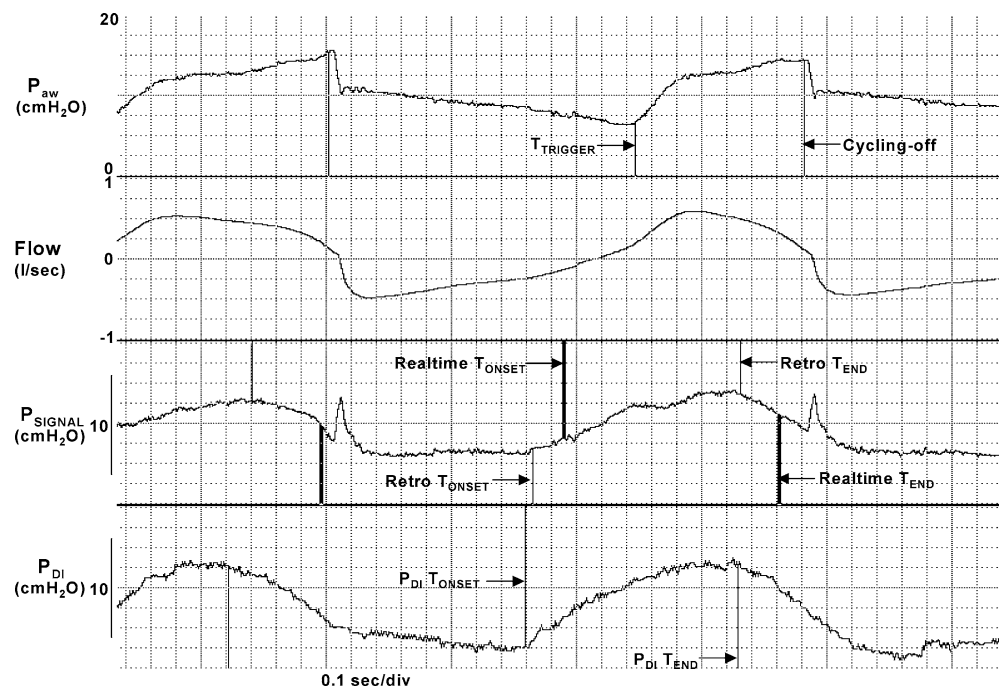
values that are selected empirically (see the Electronic Supplementary Material, ESM). This study compared the triggering and cycling-off times observed during conventional triggering with times determined from the improvised effort signal (P_{SIGNAL}).

Methods

A prototype (PVI Monitor, YRT, Winnipeg, Canada) was tested on 21 preexisting files that contained periods on pressure support (PSV, $n = 20$) and proportional assist ventilation (PAV; $n = 21$). These were generated during an earlier unrelated study involving patients with acute respiratory failure in stable condition [11]. The ethics committees of participating institutions approved the original clinical study. Patients or next-of-kin gave informed consent (see detailed methods in ESM).

Files contained airway pressure (P_{aw}), flow (\dot{V}), integrated flow (V), esophageal pressure (P_{ESO}), and, in 14 patients, gastric pressure (P_{GA}). Diaphragm pressure ($P_{\text{DI}} = P_{\text{GA}} - P_{\text{ESO}}$) or, when P_{GA} was unavailable, respiratory muscle pressure (P_{MUS} ; according to the Campbell method [12]) was calculated (see ESM). Approximately 1 hour from each file was analyzed. Markers were placed at the beginning ($P_{\text{DI-TONSET}}$) and end ($P_{\text{DI-TEND}}$) of each effort (Fig. 1). In four patients $P_{\text{DI-TEND}}$ could not be identified (see ESM). Respiratory mechanics were calculated using standard methods. Flow at $P_{\text{DI-TONSET}}$ was measured (flow at end-expiration). Negative values indicated the presence of intrinsic PEEP (PEEP_1). When P_{DI} was

Fig. 1 Example of monitor output (top three channels) and the corresponding diaphragm pressure (P_{DI}). P_{AW} Airway pressure; P_{SIGNAL} signal generated by monitor in real-time; T_{TRIGGER} time of ventilator triggering. Timing events identified by monitor are indicated in the P_{SIGNAL} tracing whereas the corresponding actual timing events are indicated in the P_{DI} tracing; $P_{\text{DI-TONSET}}$ and $P_{\text{DI-TEND}}$ are onset and end of effort as scored manually by an observer; *Real-time* T_{ONSET} and *real-time* T_{END} are onset and end of efforts identified by monitor in real-time; *Retro* T_{ONSET} and *Retro* T_{END} are onset and end of efforts identified by monitor retrospectively (i. e., after breath had elapsed). See text



available ($n = 14$), the increase in P_{DI} at zero flow crossing was measured to provide the component of $PEEP_1$ related to dynamic hyperinflation ($PEEP_1(DH)$ [13]). Monitor processed P_{aw} and flow data (at 200 Hz) as if received in real-time, generated a P_{SIGNAL} in real-time and made the following identifications (Fig. 1):

- Real-time T_{ONSET} was identified when P_{SIGNAL} increased by a threshold amount. T_{ONSET} threshold is automatically adjusted based on baseline noise in the 20 most recent breaths.
- Real-time T_{END} was identified when P_{SIGNAL} declined by 50% of the increase measured since real-time T_{ONSET} .
- Retrospective T_{ONSET} was determined, after breaths had elapsed, as the time when P_{SIGNAL} *began* rising (as opposed to real-time T_{ONSET} where P_{SIGNAL} must exceed a threshold to avoid autotriggering; Fig. 1). This is intended for noninvasive monitoring of trigger delay.
- Retrospective T_{END} was identified retrospectively as the point where P_{SIGNAL} *began* declining (cf. real-time T_{END} where P_{SIGNAL} must decline a threshold amount to avoid false cycling-off).
- Ineffective efforts were identified retrospectively when an increase in P_{SIGNAL} met minimum criteria for efforts but were not associated with a new ventilator breath. Ineffective efforts occurring during the inflation phase are referred to as extra-efforts.

- Patient's respiratory rate was the sum/minute of T_{ONSETS} that triggered ventilator breaths and ineffective/extra-efforts.

Sensitivity in detecting efforts was determined from the proportion P_{DI}/P_{MUS} efforts detected as efforts by monitor. False efforts were defined as efforts identified by the monitor that had no corresponding P_{DI}/P_{MUS} effort. Respiratory rates computed by monitor and from P_{DI}/P_{MUS} were compared by paired t -test. Actual trigger delay (trigger time minus $P_{DI}-T_{ONSET}$; Fig. 1) and cycling-off delay (cycling-off time minus $P_{DI}-T_{END}$) were calculated for each triggered breath. The corresponding values if monitor's real-time T_{ONSET} and T_{END} were used instead for triggering and cycling-off were also calculated. Pressure-time product required for conventional triggering (PTP_{TRIG}) and if real-time T_{ONSET} were used instead for triggering [$PTP_{TRIG}(\text{monitor})$] were calculated (see ESM). Ventilator T_I was expressed as a fraction of patient T_{TOT} (i. e., 60/patient respiratory rate). All reported values are mean \pm SD.

Results

Physiological characteristics of the patients

Respiratory mechanics varied widely (Table S1, ESM). Four patients had severe expiratory flow limitation.

Fig. 2 Impact of using monitor's real-time T_{ONSET} on trigger delay (a) and on pressure time product preceding triggering (PTP_{TRIG} ; b). *Conv. Trig* Conventional flow triggering on the 840 ventilator; *horizontal bars* averages for patients with (dashed lines) and without severe flow limitation; * $p < 0.02$; ** $p < 0.001$; *** $p < 0.0001$ lower than with conventional triggering

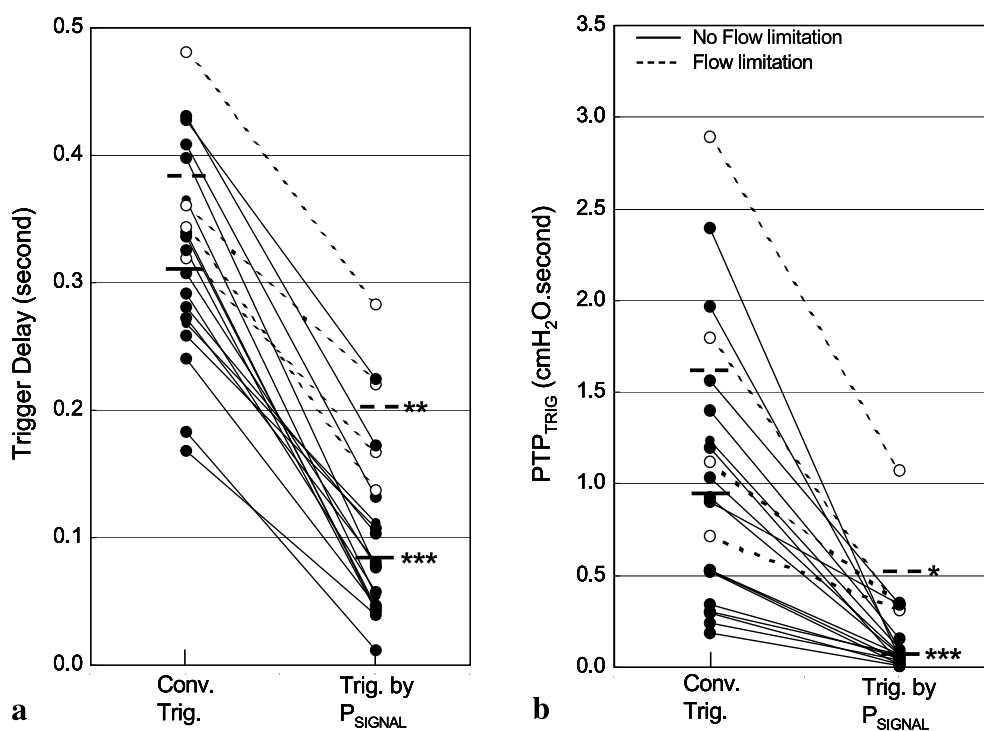


Table 1 Monitor identification of T_{ONSET} , T_{END} , and ineffective/extra-efforts (pooled PSV and PAV data) w(T_{TR} time of ventilator triggering; T_{ONSET} onset of effort as identified by monitor; T_{END} end of effort as identified by monitor; dT difference between time of event identified by monitor and time of corresponding event identified in diaphragm pressure or P_{MUS} ; *Sens.*: sensitivity; *VB*: ventilator breaths)

Case no.	VB	Retro- T_{ONSET} Identified ahead of T_{TR} (%)	dT (s)	Real-time T_{ONSET} Identified ahead of T_{TR} (%)	False T_{ONSET} (min^{-1})	dT_{ONSET} (s)	Retro T_{END} dT (s)	Real-time T_{END} dT_{END} (s)	Ineffective/extra-efforts Total no.	Sens. (%)	No. false (min^{-1})
1	1064	99.9	0.039	98.4	0.33	0.102	-0.168	-0.034	190	91.6	0.1
2 ^b	1720	100	0.011	98.7	0.03	0.042	-0.087	0.127	6	100.0	0.0
3 ^b	1059	99.2	0.004	90.0	2.31	0.019	-0.104	0.071	96	53.1	0.2
4	1490	100	0.007	95.0	1.91	0.050	NA	NA	15	100.0	2.1
5	1218	99.3	0.062	96.7	0.71	0.134	NA	NA	132	97.7	0.1
6	2104	100	0.042	97.7	0.07	0.052	-0.092	0.188	0	NA	0.0
7	963	99.9	0.052	99.0	0.31	0.099	-0.071	0.033	0	NA	0.0
8	1494	99.7	0.008	96.1	0.00	0.071	-0.033	0.121	126	87.3	0.1
9 ^b	1131	100	0.072	99.2	1.49	0.081	0.020	0.227	9	100.0	0.1
10 ^{a,b}	1080	99.4	0.099	93.8	1.91	0.167	0.023	0.197	18	72.2	0.6
11 ^b	2133	100	0.05	96.7	0.17	0.059	-0.004	0.166	0	NA	0.0
12 ^b	2305	100	-0.009	87.8	0.40	0.051	-0.019	0.160	107	97.2	0.7
13 ^a	1374	100	0.086	99.1	0.75	0.138	-0.258	-0.126	87	96.6	0.1
14 ^a	1164	100	0.221	99.1	0.56	0.290	-0.055	0.065	623	63.2	0.1
15	1556	100	0.039	99.6	0.26	0.073	-0.108	0.107	6	100.0	0.0
16	2823	100	0.024	76.1	0.52	0.062	0.019	0.115	948	92.2	0.0
17	1479	100	0.114	99.1	0.09	0.166	0.033	0.201	13	84.6	0.0
18	1208	100	0.064	98.7	0.10	0.085	NA	NA	0	NA	0.0
19 ^a	1364	100	0.125	95.7	1.26	0.233	-0.020	0.059	504	67.3	0.3
20 ^b	490	99.8	0.139	98.8	0.63	0.231	NA	NA	13	100.0	0.0
21	1418	99.9	-0.001	99.5	1.63	0.038	-0.146	-0.029	0	NA	0.3
Mean	1459	99.9	0.059	95.9	0.74	0.107	-0.063	0.097	138	87.7	0.2
SD	525	0.2	0.057	5.5	0.72	0.074	0.079	0.096	249	15.3	0.5

^a Patients with severe flow limitation; ^b Patients with no gastric (or PDI) pressure signal in whom P_{MUS} was calculated from esophageal pressure using the Campbell diagram

Most patients had intrinsic PEEP (flow at end-expiration -0.235 ± 0.176 l/s; Table S1, ESM). PEEP_I due to DH was 3.6 ± 2.6 cmH₂O. There were no significant differences on average between PSV and PAV in PEEP level (6.5 ± 4.1 vs. 6.7 ± 3.9 cmH₂O), pressure assist (peak P_{AW}-PEEP; 17.7 ± 4.0 vs. 16.6 ± 4.6 cmH₂O), or minute ventilation (11.4 ± 4.2 vs. 11.6 ± 3.8 l/min). Amplitude of P_{DI}/P_{MUS} excursions was relatively low (6.8 ± 4.1 vs. 8.8 ± 4.0 cmH₂O for PSV and PAV; $p < 0.01$). Diaphragm rate was high on average (30.6 ± 8.4 vs. 30.8 ± 9.5 /min for PSV and PAV, NS). Inspiratory duration of diaphragm efforts T₁(DIA) was brief in both modes (0.66 ± 0.17 vs. 0.69 ± 0.14 s; NS).

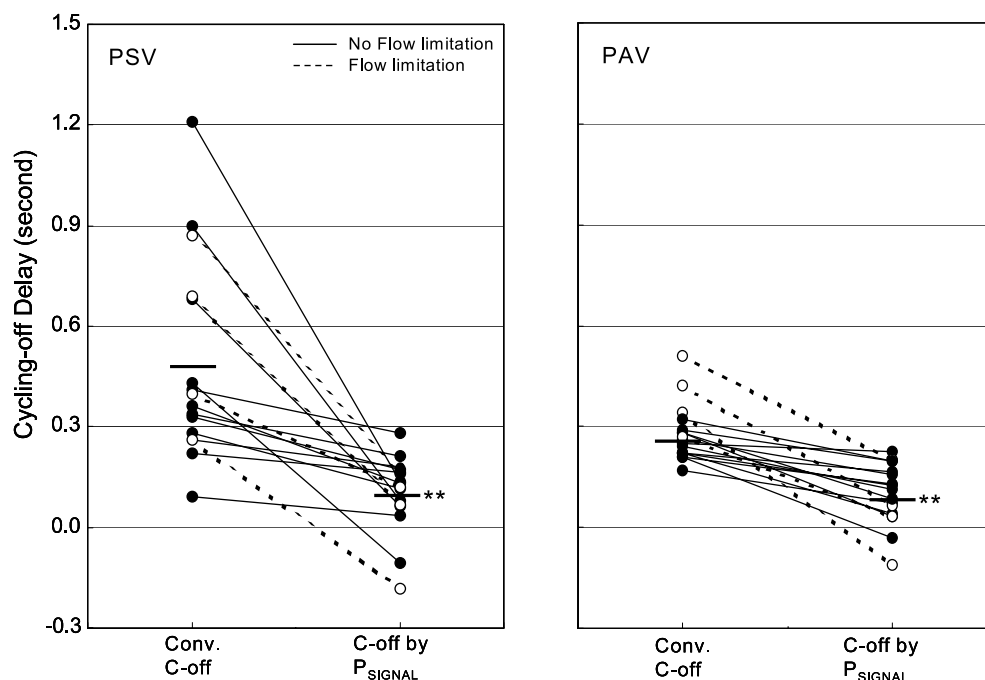
Trigger-delay

Trigger-delay was 0.33 ± 0.09 and 0.32 ± 0.08 s in PSV and PAV, representing $51 \pm 16\%$ and $48 \pm 12\%$ of T₁(DIA) (NS). PTP_{TRIG} was 0.96 ± 0.75 and 1.06 ± 0.74 cmH₂O·s for PSV and PAV (NS). Individual values are shown in Fig. 2 (PSV and PAV data were combined). Retrospective T_{ONSET} was identified ahead of triggering in $99.9 \pm 0.2\%$ of patient-triggered breaths (Table 1). It occurred slightly later than P_{DI}-T_{ONSET} ($dT = 0.059 \pm 0.057$ s; Table 1). dT was longer in patients with severe flow limitation in whom inspiratory effort often progresses for sometime before there is any external change (e.g., in flow) that can be measured by the monitor (e.g., Fig. S2, ESM). In patients without flow limitation dT was 0.042 ± 0.040 s.

Real-time T_{ONSET} occurred ahead of conventional triggering in 95.9% of ventilator breaths (Table 1). Of the missed events 81% were when inspiratory effort began very early in the ventilator's exhalation phase and these efforts were deliberately blanked to avoid premature triggering (see ESM). The reason for the remaining missed T_{ONSET} instances (approx. 1% of breaths) was usually a high threshold related to baseline noise (e.g., cardiac artifacts, secretions).

The time difference between P_{DI}-T_{ONSET} and real-time T_{ONSET} (DT_{ONSET}; what trigger-delay would be if real-time T_{ONSET} were used for triggering under identical conditions) averaged 0.107 ± 0.074 s (Fig. 2, Table 1). It was longer in patients with flow-limitation (0.21 ± 0.07 vs. 0.08 ± 0.05 s; Fig. 2). Concurrently PTP_{TRIG} decreased from 0.92 ± 0.65 to 0.09 ± 0.1 cmH₂O·s ($p < 0.0001$) and from 1.63 ± 0.95 to 0.51 ± 0.37 cmH₂O·s ($p < 0.02$) in patients without and with flow-limitation (Fig. 2B). False T_{ONSET} instances, which would result in autotriggering, averaged 0.72/min, not exceeding 2.5/min in any patient (Table 1). Actual autotriggering (ventilator breaths without a P_{DI} effort) averaged 0.23/min. The pattern of autotriggering was quite different in the two cases. Actual autotriggering occurred almost exclusively in three patients on PSV when the assist was excessive and efforts were indistinguishable from cardiac artifacts (e.g., Fig. S3, ESM). By contrast, monitor-generated false T_{ONSET} instances were sporadic and occurred in virtually all patients (Table 1).

Fig. 3 Impact of using monitor's real-time T_{END} on cycling-off delay during pressure support (left) and proportional assist (right); Conv. C-off Conventional cycling-off on the 840 ventilator; horizontal bars averages; ** $p < 0.0001$ lower than with conventional cycling-off



Cycling-off delay

In PSV the cycling-off delay varied widely from 0.09 to 1.21 s (Fig. 3), representing 18–186% ($47 \pm 46\%$) of patient T_E . With PAV the cycling-off delay ranged from 0.17 to 0.51 (Fig. 3) not exceeding 38% of T_E (DIA) in any patient. The ratio of ventilator T_I to patient T_{TOT} ranged from 0.20 to 1.27 in PSV while always remaining in the physiological range in PAV (0.20–0.44). Interestingly, ineffective efforts appeared in any patient in whom this ratio was greater than 0.42. In 13 of 17 patients in whom $P_{DI}T_{END}$ was available retrospective T_{END} was within ± 0.1 s of the corresponding $P_{DI}T_{END}$ (Table 1). In four patients it preceded $P_{DI}T_{END}$ by more than 0.1 s. Real-time T_{END} occurred 0.097 ± 0.096 s after $P_{DI}T_{END}$ and was identified before $P_{DI}T_{END}$ in three patients (Table 1). Figure 3 shows how cycling-off delay would be affected if real-time T_{END} were used for cycling-off the ventilator under identical conditions.

Impact of ventilator settings on monitor's identification of T_{ONSET} and T_{END}

Table S2 (ESM)) shows individual values of DT_{ONSET} and DT_{END} along with P_{AW} , V_T , and P_{DI}/P_{MUS} . Ventilator settings, V_T , and P_{DI} varied substantially between patients. Notably, P_{DI} ranged from 1.5 to 16.5 cmH₂O. However, there was no correlation between P_{AW} , tidal volume, or P_{DI} and either DT_{ONSET} or DT_{END} with either PSV or PAV. Likewise, although on average P_{AW} , V_T and P_{DI} did not differ substantially between PSV and PAV, within-patient differences were large in some cases (Table S2, ESM). The differences in DT_{ONSET} between PSV and PAV were small

(0.007 ± 0.025 s) and were not correlated with differences in P_{AW} , V_T or P_{DI} , indicating that they were not related to ventilator settings. Likewise, the differences in DT_{END} between PSV and PAV were small (0.001 ± 0.065 s) and not correlated with differences in P_{AW} , V_T or P_{DI} .

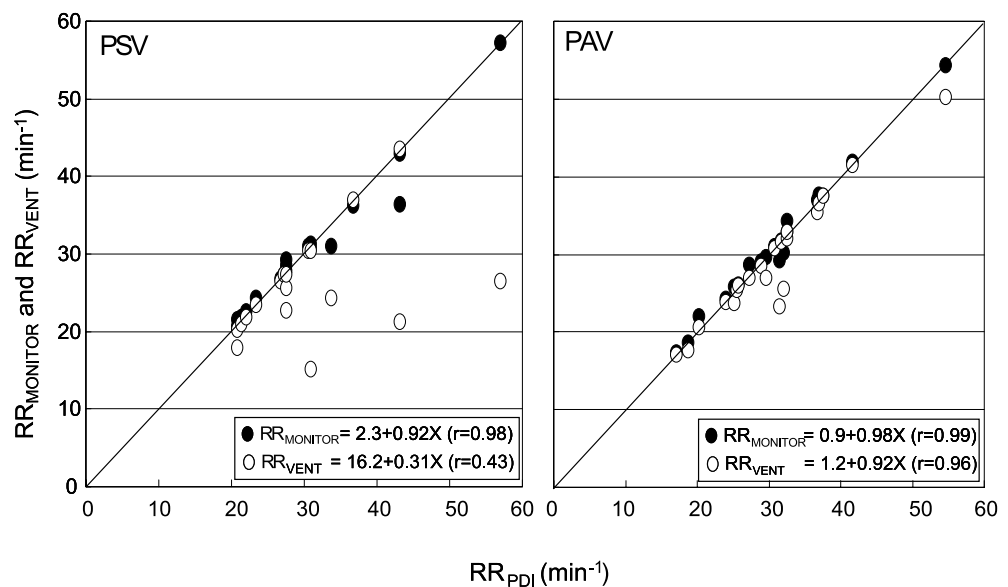
Ineffective/extra-efforts

In seven patients ineffective/extra-efforts were frequent ($> 10\%$ of RR) in PSV and resulted in an important difference between patient and ventilator rates (up to 32/min, open circles, Fig. 4 left). In the same patients on PAV ineffective efforts were either rare ($n = 3$) or considerably fewer (Fig. 4 right). A total of 2,893 ineffective/extra-efforts were identified in the P_{DI} tracing (Table 1). The monitor identified 2,319 of these (79.7%). Two-thirds of the unidentified efforts occurred in two patients with severe flow limitation (nos. 14, 19; Table 1) in whom these efforts had virtually no effect on flow (e. g., Fig. S2, ESM). In most other cases the unidentified events were associated with very weak P_{DI} efforts. False identification of ineffective efforts occurred infrequently (0.2 ± 0.5 /min), with the highest rate being 2.1/min (Table 1).

Patient's respiratory rate

Solid circles in Fig. 4 show the relationship between respiratory rate estimated by monitor and that from P_{DI} . The agreement was excellent both in PSV and PAV, with an average difference of -0.2 ± 1.9 /min in PSV and 0.2 ± 0.9 /min in PAV. By contrast, the relationship

Fig. 4 *Solid symbols* Relationship between respiratory rate identified by monitor ($RR_{MONITOR}$) and respiratory rate identified from diaphragm pressure (RR_{PDI}) during pressure support (PSV, $n = 20$ patients) and proportional assist ventilation (PAV, $n = 21$ patients). Note the excellent agreement between the two measurements in both modes. *Open symbols* Relationship between ventilator rate (RR_{VENT}) and RR_{PDI} . This relationship is quite poor in PSV



between ventilator rate and P_{DI} rate (open circles) was quite poor in PSV ($r = 0.43$, NS).

Improvised elastance (K_V) and resistance (K_F) values

Table S3 (ESM) shows individual values of K_V and K_F and their within-patient variability during PSV and PAV. Average K_V ranged from 12.2 to 43.6 in PSV (27.8 ± 9.1) and from 12.9 to 40.5 in PAV (25.2 ± 7.6 ; $p < 0.05$). K_F ranged from 2.5 to 24.1 in PSV (11.4 ± 6.3) and from 2.3 to 21.0 in PAV (10.2 ± 5.8) ($p < 0.005$). K_V and K_F changed from time to time in each patient but the changes were relatively small (Table S3, ESM). Figure S5 (ESM) shows the relation between K_V and K_F (average of PSV and PAV values) and the corresponding E_{RS} and R_{LUNG} measured by use of esophageal pressure. There was a significant correlation in both cases ($p < 0.0001$) but there was considerable scatter. Average difference (\pm SD) was -4.8 ± 6.1 cmH₂O/l for K_V vs. E_{RS} and 1.5 ± 4.1 cmH₂O/l per second for K_F vs. R_{LUNG} .

Discussion

The current study shows that a signal generated noninvasively from flow, volume, and P_{AW} using improvised resistance and elastance values can be used for monitoring patient-ventilator interaction and for providing information to optimize ventilator settings. The signal can also potentially be used to directly synchronize onsets and ends of ventilator cycles with patient efforts.

Critique of methods

We used preexisting files to test this approach. However, data were inputted in the monitor's computer in exactly the same fashion as would happen if they were acquired live. Because there were no interventions, results would have been the same if the patients were monitored live. It may be argued that using 15% of peak flow for cycling-off PSV may have unduly exaggerated the nonsynchrony in this mode. Using simulations, however, we found that the magnitude of nonsynchrony is only marginally greater with 15% peak than with the currently used 25%. Furthermore, extent of nonsynchrony found here is not substantially different from what was reported previously with PSV [1, 2, 4, 7].

This study showed that for identical ventilator outputs (P_{AW} and flow) and patient outputs (P_{DI} , patient RR) T_{ONSET} and T_{END} are closer to the true onsets and ends of the patient's efforts than conventional triggering algorithms (Figs. 2, 3). However, T_{ONSET} and T_{END} did not actively control ventilator cycles. It is possible that if this approach were used to directly control the ventilator's

timing a number of variables that affect synchrony may have changed (e. g., V_E , P_{MUS} , dynamic hyperinflation [1, 2, 5, 14]), so that the net effect on triggering and cycling-off delays may have differed from that reported here. However, T_{ONSET} and T_{END} were superior to conventional triggering in all patients (Figs. 2, 3), and in the same patients under two different modes, despite large differences in ventilator and patient outputs (Table S2, ESM). This suggests that the current approach remains superior to conventional triggering regardless of how the patient and ventilator respond to the different triggering method. The current approach was validated in intubated patients, in whom leaks are minimal, and externally measured flow may be assumed to reflect flow in and out of the patient with reasonable accuracy. The extent to which the accuracy of this method is influenced by leaks, such as may exist during noninvasive ventilation, is not known.

Use of P_{SIGNAL} for active triggering and cycling-off

Real-time T_{ONSET} occurred before conventional triggering in 95.9% of triggered breaths (Table 1). Most of the missed T_{ONSET} instances were deliberately blocked because the associated efforts occurred too early in the ventilator's expiratory phase (due to excessive cycling-off delay). Since the current system would preclude excessive cycling-off delay, instances where an effort would be missed for this reason would also be reduced. Thus this approach can reduce trigger delay in virtually every breath. Figure 3 shows that the potential reduction in trigger-delay is substantial.

In general, improved trigger sensitivity is associated with increased likelihood for autotriggering (e. g., [4]). This was not the case here; frequency of false T_{ONSET} instances remained quite low in all patients (Table 1). More importantly, the autotriggering that may occur would not result in overventilation, the main side effect of autotriggering [15, 16]. With the current approach autotriggering can only be sporadic. A false T_{ONSET} can be identified as false retrospectively because the change in P_{SIGNAL} would not meet respiratory effort criteria. T_{ONSET} threshold is then automatically increased thereby aborting further false triggers until the noise disappears. By contrast, with conventional triggering it is impossible to determine whether triggering was false in order to effect automatic changes in triggering threshold. Thus autotriggering may continue indefinitely [15], as was observed in three patients here (e. g., Fig. S3, ESM).

Real-time T_{END} occurred when P_{SIGNAL} decreased to 50% of its peak value (Fig. 1). Is this appropriate? During unassisted breathing the onset of expiratory flow is invariably delayed relative to end of neural inspiration [10, 17, 18]. This delay is a function of rate of decline in P_{MUS} and of respiratory mechanics (resistance \times compliance, RC) [17]. Because these vary greatly between individuals, there is no unique optimal cycling-off delay (see

Fig. S4, ESM). Ideally cycling-off should be linked to the rate of decline in P_{MUS} and to RC. In the current prototype real-time T_{END} was linked to the rate of decline in P_{SIGNAL} since with a faster declining rate the 50% point would be reached sooner. The choice of 50% was a compromise consistent with an average RC of approx. 0.6 s ($R = 15$, $C = 0.04$ [19, 20]). It would be possible to link the threshold proportional reduction to K_V and K_F , but at present the relationship between these values and actual respiratory mechanics is not sufficiently strong (Fig. S5, ESM).

Although the cycling-off delay that would result from using real-time T_{END} is appropriate on average (Fig. 3), it would not be prudent to rely entirely on real-time T_{END} . First, in three patients real-time T_{END} occurred before the end of patient T_I (Table 1). This is likely related to nonlinearities in the pressure-flow and/or pressure-volume relationships. Second, in some patients on PSV inspiratory effort ends immediately after triggering [21, 22, 23] so that cycling-off the ventilator at the end of effort would result in very short ventilator T_I . Accordingly, it would be necessary to establish a minimum duration for ventilator T_I .

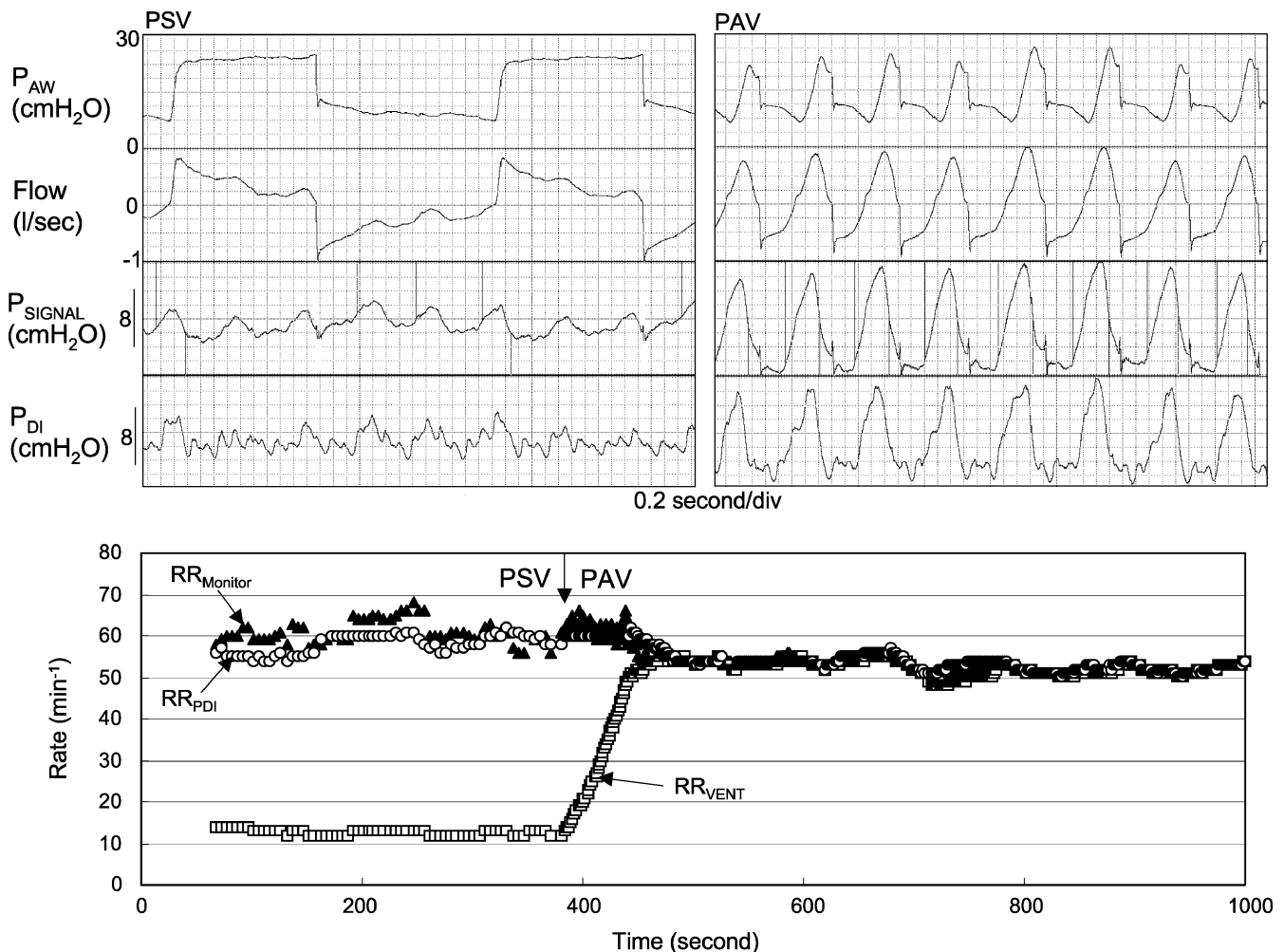


Fig. 5 Top Airway pressure (P_{AW}), flow, monitor-generated signal (P_{SIGNAL}) and diaphragm pressure (P_{DI}) in a patient while on pressure support (*PSV*) and later on proportional assist (*PAV*). Note the marked discrepancy between ventilator rate and patient's rate during *PSV*. Note also that efforts were very weak on *PSV*, indicating a high level of assist. Bottom Plot of different rates against time spanning the transition from *PSV* to *PAV*. Note that monitor's estimate of patient's rate ($RR_{MONITOR}$) accurately tracks respiratory rate determined from P_{DI} (RR_{PDI}). In this case the monitor clearly indicates

that the increase in ventilator rate (RR_{VENT}) following the transition was not related to an increase in patient's rate (in fact patient's rate decreased). All values are 1-min averages of rate. This accounts for the fact that RR_{VENT} appeared to increase gradually (over 1-min) following transition. In fact the change in ventilator rate was instantaneous, before there was any increase in P_{DI} . Event marks in P_{SIGNAL} are real-time T_{ONSET} instances (upward marks) and T_{ENDS} (downward marks) as identified by the monitor

Prinianakis et al. [4] evaluated a new triggering/cycling-off algorithm for PSV based on shape of the flow waveform. Relative to conventional flow-triggering, there was a modest reduction in trigger-delay (approx. 0.04 s) and in PTP_{TRIG} (approx. 0.1 cmH₂O·s; see Fig. 2). There were no changes in ventilator T_I , suggesting that cycling-off delay was not reduced. Ineffective efforts decreased marginally. For example, in the patient with the most ineffective efforts they decreased from 32 to 28/min. Although improvement in triggering was modest, autotriggering increased significantly [4].

Use of P_{SIGNAL} for passive monitoring of patient-ventilator interaction

When P_{SIGNAL} is displayed in real-time along with P_{AW} and flow, the user obtains a comprehensive visual feedback about the instant state of patient-ventilator interaction (e. g., Fig. 5, top). Excessive trigger delay, cycling-off delay and ineffective efforts would be immediately obvious. Such feedback can be readily used to optimize ventilator settings. For example, in the presence of excessive cycling-off delay the flow threshold for cycling-off in PSV may be gradually increased until reasonable synchrony is observed on the screen. By optimizing cycling-off it may be expected that ineffective efforts, if any, would be reduced and trigger delay and dynamic hyperinflation improved [24]. The onset of inspiratory effort while flow is still expiratory would clearly indicate the presence of intrinsic PEEP. The time difference between onset of effort and onset of inspiratory flow may be used to optimize external PEEP and reduce dynamic hyperinflation. Brief transients in P_{SIGNAL} that have the frequency of heart rate, along with absence of much larger transients that reflect efforts, would indicate autotriggering by cardiac artifacts resulting in overassist and apnea and would be countered by decreasing ventilator's trigger sensitivity. By freezing the screen at low speed the clinician would be able to count the actual respiratory rate of the patient.

Such visual feedback may be supplemented by numerical displays of the relevant variables. The current study shows that retrospective T_{ONSET} and T_{END} are fairly close to the actual onset and end of effort identified

from P_{DI}/P_{MUS} (Table 1). These retrospective values can therefore be used for automatic measurement of triggering and cycling-off delays. The study has also shown that the respiratory rate counted by the monitor faithfully reflects patient's respiratory rate (Fig. 4). When these values are stored, trends over time can be obtained. These may help troubleshoot respiratory distress. Furthermore, it would be possible to determine whether an increase in ventilator rate that occurs in the course of weaning trials or following a change in ventilator settings (e. g., a change in cycling-off threshold in PSV or a switch to PAV) is due simply to improved synchrony and hence may be ignored, or reflects a true change in patient's respiratory rate (i. e., distress). Since a large increase in ventilator rate during weaning trials is usually interpreted as weaning failure, such information may reduce instances of false weaning failure. Figure 5 shows a striking example of a large change in ventilator rate following a mode switch that was identified by the monitor as due to improved synchrony.

The current approach was not intended or designed to provide accurate estimates of respiratory mechanics or of the amplitude of P_{MUS} . It was of interest, however, to see the extent to which K_F and K_V agreed with actual resistance and elastance. The correlations were highly significant ($p < 0.0001$), but there was considerable scatter (Fig. S5, ESM). There are several possible reasons for the observed differences, and some of these are correctable. It remains to be seen whether the agreement between the improvised and actual values can be improved enough to make K_V and K_F useful surrogates for elastance and resistance. Because errors in estimated elastance and resistance result in errors in estimated P_{MUS} amplitude, until such improvement is achieved, the actual amplitude of P_{SIGNAL} should not be used to reflect actual P_{MUS} .

In summary, this study shows that a signal generated by applying the equation of motion, using improvised values of elastance and resistance, makes it possible non-invasively to identify the onsets and ends of inspiratory efforts with reasonable accuracy. Such signal may be used to synchronize the ventilator with the patient or to simply provide information on patient-ventilator interaction that may be helpful in making appropriate ventilator adjustments and in evaluating the patient, such as by providing actual patient's respiratory rate.

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