RESPIRATORY CARE Paper in Press. Published on May 21, 2013 as DOI: 10.4187/respcare.02466

A comparison of leak compensation in acute care ventilators during non-invasive and invasive ventilation; a lung model study

Jun Oto, M.D., PhD, Research fellow Respiratory Care Services, Massachusetts General Hospital

E-mail address: JOTO@PARTNERS.ORG

Christopher T. Chenelle, B.A., Respiratory Care Services, Massachusetts General Hospital

E-mail address: CCHENELLE@PARTNERS.ORG

Andrew D. Marchese, B.S., Respiratory Care Services, Massachusetts General Hospital

E-mail address: ANDY@CSAIL.MIT.EDU

Robert M Kacmarek PhD, RRT, Respiratory Care Services, Massachusetts General Hospital

E-mail address: RKACMAREK@PARTNERS.ORG

Address correspondence to:

Robert M Kacmarek PhD, RRT, Respiratory Care Services, Massachusetts General Hospital

E-mail address: RKACMAREK@PARTNERS.ORG

55 Fruit street, Boston, MA, 02114

Tel: 617-724-4490, Fax: 617-724-4495

RESPIRATORY CARE Paper in Press. Published on May 21, 2013 as DOI: 10.4187/respcare.02466

Financial support: Funded in part by COVIDIEN, Inc., 6135 Gunbarrel Avenue, Boulder, CO 80301

Keywords: Leak compensation, Mechanical ventilation, Acute care ventilator

Conflict of interest: Robert M Kacmarek: Maquet Inc received honorarium for lecturing, research grant from COVIDIEN, Consultant for COVIDIEN. All other authors have no conflict of interest.

Word Count: 3665

Word Count Abstract: 275

RESPIRATORY CARE Paper in Press. Published on May 21, 2013 as DOI: 10.4187/respcare.02466

Abstract

***Background*:** Although leak compensation has been widely introduced to acute care ventilators to improve patient-ventilator synchronization in the presence of system leaks, there is no data on these ventilators ability to prevent triggering and cycling asynchrony. The goal of this study was to evaluate the ability of leak compensation in acute care ventilators during invasive and non-invasive ventilation (NIV).

***Method*:** Using a lung simulator, the impact of system leaks was compared on 7 ICU ventilators and 1 dedicated NIV ventilator during triggering and cycling at two respiratory mechanics (COPD and ARDS models) settings, various modes of ventilation (NIV mode: pressure support ventilation; and invasive mode: pressure support and pressure assist/control), and two PEEP levels (5, and 10 cmH_2O). Leak levels used were up to 35–36 L/min in NIV mode and 26–27 L/min in invasive mode.

Results: Although all of the ventilators were able to synchronize with the simulator at baseline, only 4 of the 8 ventilators synchronized to all leaks in NIV and 2 of 8 ventilators in invasive ventilation. The number of breaths to synchronization was higher in increasing than decreasing leak. In the COPD model, miss-triggering occurred more frequently and required a longer time to stabilized tidal volumes than in the ARDS model. The PB840 required fewer breaths to synchronize in both invasive and NIV modes compared with the other ventilators ($p < 0.001$).

Conclusions: Leak compensation in invasive and NIV modes has wide variations between ventilators. The PB840 and the V60 were the only ventilators to acclimate to all leaks but there

RESPIRATORY CARE Paper in Press. Published on May 21, 2013 as DOI: 10.4187/respcare.02466

were differences in performance between these two ventilators. It is not clear if these differences have clinical significance.

RESPIRATORY CARE Paper in Press. Published on May 21, 2013 as DOI: 10.4187/respcare.02466

Introduction

During mechanical ventilation, system leak is a major cause of patient-ventilator asynchrony[1]. Leaks may be due to the endotracheal tube cuff, ventilator circuit, or chest drain during invasive ventilation. The incidence of endotracheal tube cuff leaks has been reported at ranges from 11 to 24%[2, 3]. With non-invasive ventilation (NIV), leaks around the face or nasal mask are common cause of system leak[1]. Patient-ventilator asynchronies have been reported to occur with a high incidence during both invasive[4, 5] and non-invasive ventilation (NIV)[6, 7]. Thille AW et al. reported that 24% of patients showed patient-ventilator asynchrony in greater than 10% of their total ventilatory rate during invasive ventilation[5]. Vignaux et al. demonstrated that auto-triggering was present in 13% of patients and delayed cycling in 23% of patients during NIV[6]. Patient-ventilator asynchrony can significantly increase the work of breathing[8, 9] and a high incidence of patient-ventilator asynchrony is associated with a longer duration of mechanical ventilation[4, 5].

Although ICU ventilators were initially built to function without leaks, leak compensation has been added to ICU ventilators to improve patient-ventilator synchronization in the presence of system leaks. Ideally, leak compensation should automatically adjust during triggering and cycling to ensure the ventilator rapidly responds to changes in leak without affecting patient-ventilator synchrony and maintains pressurization capacities. Ferreira et al. evaluated the ability of 9 critical-care ventilators to function in the presence of leaks[10]. The BiPAP Vision and the Servo-i were the only ventilators able to adapt to the leaks. Carteaux et al.

RESPIRATORY CARE Paper in Press. Published on May 21, 2013 as DOI: 10.4187/respcare.02466

reported that dedicated NIV ventilators promoted better patient-ventilator synchronization than critical care and transport ventilators, even when the NIV mode was used[11].

Technological improvements by ventilator companies are introduced rapidly. Currently, there is no assessment of the performance of many ICU ventilators or the recent upgrades of ICU ventilators on their ability to prevent triggering and cycling asynchrony in both invasive and NIV. The aim of our study was to evaluate the ability of current acute care ventilators to prevent triggering and cycling asynchrony caused by increasing and decreasing leaks during both invasive and NIV.

RESPIRATORY CARE Paper in Press. Published on May 21, 2013 as DOI: 10.4187/respcare.02466

Methods

Seven ICU ventilators (Maquet Servo-*i*, Covidien PB840, Hamilton C3, Hamilton G5, General Electric Carestation, Drager V500, and CareFusion Avea) and one dedicated NIV ventilator (Respironics V60) (Table 1) were compared using a lung simulator (ASL5000 lung model, IngMar Medical; Pittsburgh, PA)[12] during increasing and decreasing system leaks.

Study setup (ASL5000 interface and Mannequin)

The simulator was adjusted to simulate 1) the COPD model and 2) the ARDS model. In the COPD model, compliance was 60 ml/cm H_2O, inspiratory resistance 10 cm H_2O/L/s and expiratory resistance 20 cm H_2O/L/s. In the ARDS model, compliance was 20 ml/cm H_2O, inspiratory and expiratory resistance 5 cm H_2O/L/s. The inspiratory time of the simulator was 0.92 s, the maximum inspiratory pressure drop was –5 cm H_2O, the pressure drop generated 100 ms after the onset of an occluded inspiratory effort was –3.6 cm H_2O, and the respiratory rate was 15 breaths/min. For the profile of the negative pressure created by the respiratory muscles, 5% of the respiratory cycle time was active inspiration, 3% was an end-inspiratory hold, and 15% was for return of pressure to baseline. The ASL5000 incorporates a series of three user-controlled leaks with a simulator bypass and leak valve module (SBLVM) [IngMar Medical][12].

In the NIV mode, a mannequin head was used to simulate the patient-mask interface (Suppl. Figure. 1). An oronasal facemask (PerformaTrack SE; Respironics Inc; Murrysville, PA) was affixed to the head of the mannequin with standard straps. A baseline leak of 3 to 4 L/min

RESPIRATORY CARE Paper in Press. Published on May 21, 2013 as DOI: 10.4187/respcare.02466

(baseline leak: B) at a mean airway pressure of 7.5 cmH_2O was established. The SBLVM was set to leaks of 9 to 10 L/min (leak level 1: L1), 26 to 27 L/min (L2) and 35 to 36 L/min (L3) at a mean airway pressure of 7.5 cmH_2O. We chose these levels to represent the range of leak flows that are likely to be experienced clinically[13]. All combinations of increasing (n=6) and decreasing (n=6) leaks were evaluated.

During invasive ventilation, the ventilators were affixed to the ASL5000 with an 8 mm internal diameter endotracheal tube (Suppl. Figure. 1). The SBLVM was set to leaks of 3 to 4 L/min (L1), 9 to 10 L/min (L2) and 26 to 27 L/min (L3) at a mean airway pressure of 7.5 cmH_2O. All combinations of increasing (n=6) and decreasing (n=6) leaks were evaluated.

Ventilator setup

During the non-invasive ventilation assessment, all of the ventilators were set in NIV mode as follows: pressure support ventilation (PSV); inspiratory pressure, 12 cm H_2O; PEEP 5 and 10 cmH_2O; respiratory rate, 10 breaths/min; and leak compensation activated if available. Trigger sensitivity was set to 3 L/min if available; inspiratory rise time, when adjustable, was set to the most rapid setting while avoiding overshooting of the set peak pressure. In the COPD model, termination criteria, when adjustable, was set to insure that the lung model ending of inspiration and the ventilator ending of inspiration did not differ by more than ± 5% at baseline leak. In the ARDS model, termination criteria were set at 25% of the peak flow. The maximum duration of inspiration was set to 1.5 sec.

During invasive ventilation, all ventilators were set in PSV and pressure assist/control

RESPIRATORY CARE Paper in Press. Published on May 21, 2013 as DOI: 10.4187/respcare.02466

ventilation (PAC): pressure level, 12 cmH_2O; PEEP, 5 and 10 cmH_2O: respiratory rate, 10 breaths/min; leak compensation activated if available. Trigger sensitivity and cycling criteria were set the same as during NIV. In PAC mode, inspiratory time was set at 0.90 sec, approximately equal to the inspiratory time of the lung model (0.92 sec).

Variables and evaluation

For the evaluation of synchronization, we recorded the number of breaths to synchronization after leak change (B to synch). The following variables were also recorded: auto-triggering, the number of cycles not triggered after a change in leak until synchronization; miss-triggering, the number of efforts not recognized by the ventilator prior to synchronization; time to settle, the number of breaths from the moment leak was increased or decreased until the tidal volume was within 2 standard deviations of the mean tidal volume for each leak level. In addition, the following variables were evaluated: time to baseline pressure, the time from the beginning of an inspiratory effort to the return of airway pressure to baseline during triggering; triggering pressure, the airway pressure change needed to trigger; delivered tidal volume, and cycling delay time, time from the end of inspiratory effort to the moment the ventilator cycled to expiration. Each specific evaluation scenario was repeated three times.

Data collection and analysis

After each change in leak level, we waited up to 1 min for the ventilator to synchronize with the simulator. If synchronization was not achieved, the ventilator was considered unable to

RESPIRATORY CARE Paper in Press. Published on May 21, 2013 as DOI: 10.4187/respcare.02466

compensate at the specific leak setting and data was not collected. If synchronization was achieved within 1 minute, a total of 2 minutes of data after each change in leak level was collected and analyzed. Offline analysis of each breath was performed by the ASL5000 software (Labview; National Instruments; Austin, TX). A $p < 0.05$ was considered significant. Data are presented as the mean ± standard deviations or median (inter quartile range: IQR) depending on the parametric or non-parametric nature of the data distribution. Regarding the time to baseline pressure, triggering pressure, delayed cycling time, and delivered tidal volume, we report only differences that were both statistically significant ($p < 0.05$) and clinically important (>10%).

RESPIRATORY CARE Paper in Press. Published on May 21, 2013 as DOI: 10.4187/respcare.02466

Results

Synchronization

In non-invasive mode, the Servo-*i*, the PB840, the C3 and the V60 synchronized to all increasing and decreasing leaks in both the COPD model and the ARDS model (Table 2).

In invasive mode, only the PB840 and the V60 synchronized to all increasing and decreasing leaks in both PSV and PAC (Table 2).

Increasing leak vs. decreasing leak: (Figure 1)

B to synch, miss-triggering, auto-triggering and time to settle were higher for increasing than decreasing leaks (Suppl. Table 1A, B). During increasing leak, auto-triggering occurred more frequently than miss-triggering, and during decreasing leak, miss-triggering occurred more frequently than auto-triggering, in both non-invasive and invasive ventilation (Figure 1). As the magnitude of the change in leak increased, B to synch and miss-triggering increased, but not auto-triggering ($p < 0.001$).

COPD vs. ARDS model:

In non-invasive mode, miss-triggering and time to settle were higher with the COPD model than the ARDS model. Auto-triggering was higher in the ARDS model than in the COPD model (Suppl. Table 1A).

In invasive mode, miss-triggering (PSV) and time to settle (PSV and PAC) were higher

RESPIRATORY CARE Paper in Press. Published on May 21, 2013 as DOI: 10.4187/respcare.02466

in the COPD model than the ARDS model. Auto-triggering was higher in the ARDS model than the COPD model in both PSV and PAC (Suppl. Table 1B).

PEEP 5 vs. PEEP 10 cmH_2O:

Breaths to synchronization, miss-triggering, auto-triggering and time to settle were higher with PEEP 10 than 5 cmH_2O in both non-invasive and invasive ventilation (Suppl. Table 1A, 1B).

Comparison among ventilators:

We compared synchronization only among ventilators that could synchronize to all leak scenarios. In non-invasive ventilation, the PB840 significantly outperformed the other 3 ventilators (the Servo-*i*, the C3 and the V60) in three categories (B to synch, miss-triggering and time to settle) (Suppl. Table 2A).

In invasive mode, the PB840 significantly outperformed the V60 in B to synch, miss-triggering and time to settle in both PSV and PAC. (Suppl. Table 2B).

PSV vs. PAC:

There were significant differences in B to synch, miss-triggering and time to settle between PSV and PAC, with PAC outperforming PSV in all three categories (Suppl. Table 3).

Triggering delay, delayed cycling and delivered tidal volume

RESPIRATORY CARE Paper in Press. Published on May 21, 2013 as DOI: 10.4187/respcare.02466

Triggering delay time: (Figure 2)

In non-invasive mode, the time to baseline pressure was longer in the COPD model (148±22 ms) than the ARDS model (128±26 ms) ($p < 0.001$). The time to baseline pressure was longer in the C3 and the V60 than the Servo-*i* ($p < 0.001$). All ventilators except for the G5 showed a time to baseline pressure < 150 ms at baseline leak. However, the C3 and the V60 showed time to baseline pressures over 150 ms in L2 and L3 (Figure 2).

In invasive mode, the time to baseline pressure was longer in COPD model (139±25 ms and139±23 ms) than ARDS model (112±15 ms and 113±15 ms) in both PSV and PAC ($p < 0.001$) (Figure 3). There were no differences in mean triggering delay time between PSV and PAC in both the COPD model and the ARDS model (COPD model: PSV vs. PAC, 139 ms vs. 139 ms, P=0.89, respectively) (ARDS model: PSV vs. PAC, 112 ms vs. 113 ms, p=0.23, respectively). Comparing the PB840 and the V60, there was no significant difference between the two ventilators.

Delayed cycling time: (Figure 3)

In non-invasive mode, all ventilators except for the G5 showed delayed cycling time within 50 ms in the COPD model. In the ARDS model, all ventilators except the C3 and the G5 showed a delivered inspiratory time more than 100 ms shorter than the lung simulator inspiratory time in all leak scenarios (Figure 3). The C3 and the G5 showed prolonged cycling time in the ARDS model (more than 2 times the lung simulator inspiratory time) during L2 and L3. In PAC,

RESPIRATORY CARE Paper in Press. Published on May 21, 2013 as DOI: 10.4187/respcare.02466

all ventilators showed a cycling delay time of approximately 50–100 ms in both the COPD and the ARDS models (Suppl. Figure 5).

Triggering pressure and delivered tidal volume are described in supplemental materials (Suppl. Figure 6 and 7).

RESPIRATORY CARE Paper in Press. Published on May 21, 2013 as DOI: 10.4187/respcare.02466

Discussion

The main findings of this study are as follows: 1) at baseline, all ventilators were able to synchronize without miss- or auto-triggering, but there were wide variations in synchronization capability; 2) ventilators performed better during decreasing than increasing leak; 3) ventilators performed better with lower than with higher PEEP; 4) miss-triggering occurred more frequently and longer times were required to stabilize tidal volumes in the COPD model than the ARDS model; auto-triggering occurred more frequently in the ARDS model than the COPD model; 5) ventilators were better able to avoid miss-triggering and achieve synchronization and stabilization of tidal volume in PAC than PSV; 6) the PB840 and the V60 were the only ventilators to maintain synchrony in all leak scenarios without adjustment of sensitivity or inspiratory termination criteria, but there were differences in performance between these two ventilators. While previous studies have evaluated leak compensation in NIV alone[10, 11], to the best of our knowledge this is the first assessment of leak compensation during invasive and NIV.

Ventilators performed better during decreasing than increasing leak scenarios. These findings were consistent with a previous report[14]. As system leak increases, ventilators misinterpret the resulting changes in flow as inspiratory efforts, leading to frequent auto-triggering. If leak flow reaches the trigger threshold, auto-triggering occurs. Because of this, the frequency of auto-triggering does not depend on the magnitude of the increase in leak[6]. On the other hand, if the leak is large enough, ventilators may not detect respiratory efforts, leading to miss-triggering. Auto-triggering also may induce miss-triggering if inspiratory time is

RESPIRATORY CARE Paper in Press. Published on May 21, 2013 as DOI: 10.4187/respcare.02466

prolonged due to auto-triggering overlapping the patient's next inspiratory effort. In other words, cycle asynchrony can produce trigger asynchrony. Thus, in order for ventilators to maintain synchrony in the presence of leaks, ventilators have to automatically adjust trigger sensitivity and/or cycling time.

The ventilators may automatically decrease trigger sensitivity according to the level of leaks to avoid auto-triggering in the presence of leak. However, as the leak decreases, the trigger sensitivity increases. This can lead to miss-triggering, particularly if the change is larger than the inspiratory effort. If the change in leak is smaller than the inspiratory effort, miss-triggering is unlikely, though higher patient effort is required to reach this threshold. Because all ventilators measure one or several cycles and adjust trigger/cycling for the subsequent cycles following the leak level changes it is not possible to synchronize on the exact breath that the leak changes. Due to this technical constraint, leak compensation on current acute care ventilators is limited in its ability to provide synchrony.

Ventilators performed better at PEEP 5 cmH_2O than 10 cmH_2O. Gas leak was created at the airway opening of the lung simulator and the extent of the gas leak is nonlinearly related to pressure and flow. At higher PEEP, the gas leak is increased due to the higher baseline pressure and this may lead to an inability to distinguish the trigger signal from the leak.

In the COPD model, miss-triggering was more frequent and required a longer time to stabilize tidal volumes than in the ARDS model. These results are consistent with those of previous reports[4, 5]. Compared to the ARDS model, the COPD model increased time to baseline, triggering pressure, and delayed cycling time. This may be explained by the presence of higher

RESPIRATORY CARE Paper in Press. Published on May 21, 2013 as DOI: 10.4187/respcare.02466

airway resistance and lung compliance in the COPD model, which impede the transmission of respiratory effort to the ventilator and require larger efforts to reach the trigger threshold. On the other hand, auto-triggering was more frequent in the ARDS model than the COPD model. Expiratory leaks can mimic an inspiratory effort, leading to auto-triggering. Premature cycling in the ARDS model prolongs expiratory time and may facilitate this expiratory leak effect. Another possible reason is that pressure signal noise induced by the low compliance of the simulated lung may also facilitate auto-triggering[15]. In the clinical setting, low respiratory drive, respiratory frequency and absence of hyperinflation are associated with auto-triggering[16-18].

In this study, all ventilators except for the G5 showed acceptable cycling off capability (approximately ± 50 ms) even in the presence of system leak in the COPD model. However, Ferreira et al. reported that the Servo-*i* and the BiPAP Vision showed cycling delay times of approximately 200 ms and 800 ms respectively under similar lung mechanics and leak conditions[10]. One possible reason is that we set the termination criteria to insure that the ending of inspiration of the lung model and the ventilator did not differ by more than ± 5% at baseline leak in the COPD model. In this study, we set the cycling termination criteria around 40–45 % of peak inspiratory flow. In the ARDS model, most ventilators showed premature cycling at 25% of peak inspiratory flow due to the low compliance. In general, setting the expiratory trigger at a low percentage of the peak inspiratory flow may attenuate premature cycling. However, in the preliminary evaluation, all ventilators could not adjust the ending of inspiration to be within ± 5% of the lung model ending of inspiration at baseline leak even when termination criteria were set to the lowest possible values in the ARDS model; therefore we adjusted the expiratory trigger

RESPIRATORY CARE Paper in Press. Published on May 21, 2013 as DOI: 10.4187/respcare.02466

sensitivity to 25% of peak inspiratory flow. The C3 showed a prolonged inspiratory time (>600 ms) at L2 and L3 in the ARDS model (termination criteria was set at 25% of peak inspiratory flow) but not in the COPD model (termination criteria was set at 45% of peak inspiratory flow). In these cases, we believe that system leak may prevent airflow from reaching the preset expiratory flow, leading to a prolonged inspiratory time at lower values of peak inspiratory flow. Tokioka et al. reported that delayed termination with a duty cycle of > 0.5 sometimes occurred with the lowest values of termination criteria (1% peak inspiratory flow) in 2 of 8 patients with ARDS or acute lung injury[9].

In PAC, miss-triggering occurred less frequently and ventilators required fewer breaths to synchronization compared with PSV. As shown in Supplemental Figure 3, just after the system leak was increased, cycling delay due to inspiratory system leak induced miss-triggering, resulting in a longer time to synchronization during PSV. However, in PAC, inspiratory time was fixed and cycling delay did not induce miss-triggering. This is one reason why PAC provided better synchronization than PSV. Calderini et al. reported that in the presence of leaks, PAC provided better synchrony and patient comfort than PSV during NIV[19].

According to our study, the leak compensation in acute care ventilators can correct partially or completely for system leak interferences, but there were wide variation among ventilators. Vignaux et al. performed a bench study of 8 ICU ventilators featuring an NIV mode[20]. For most of the tested ventilators, leaks led to an increase in trigger delay and ~~work load,~~ a decrease in ability to reach the pressure target and delayed cycling. Similar to our findings, they found that NIV mode partially or completely corrected triggering and cycling delay, and there

RESPIRATORY CARE Paper in Press. Published on May 21, 2013 as DOI: 10.4187/respcare.02466

were marked variations between ventilators. Contrary to our study, some investigators have reported that dedicated NIV ventilators could produce better performance and synchronization than ICU ventilators in the presence of leak [7,8,21]. Miyoshi E et al. evaluated the effects of gas leak on triggering function during NIV with dedicated NIV and ICU ventilators using a lung simulator[21]. They found that the dedicated NIV ventilators triggered properly at several levels of leak (up to 44.2 L/min at 5 cmH_2O) and triggering was more effective than with the ICU ventilators. However, the NIV mode was not tested in the ICU ventilators. Carteaux G et al. compared the operation of 8 ICU ventilators, 5 transport ventilators and 6 NIV ventilators in NIV mode in a lung model and clinical study[11]. In the lung model study, they found that even though there were wide variations in synchronization capabilities among ICU and transport ventilators, the Servo-*i*, the PB840 and the V500 could avoid auto-triggering completely when using NIV mode. In the clinical study, they found the NIV ventilators allowed better patient-ventilator synchrony than the ICU ventilators, but they did not test the Servo-*i*, the PB840 or the V500 in the clinical study. Ferreira et al. evaluated the ability of 9 ICU and 1 NIV ventilators to function in the presence of leaks in a COPD model[10]. As leak increased, all ventilators except for the Servo-*i* and the BiPAP Vision needed adjustments of triggering or cycling criteria to synchronize appropriately with the lung simulator. They concluded that the Vision had slightly better synchrony with triggering and the Servo-*i* with cycling. However, some ICU ventilators tested had not incorporated leak compensation. Since the manufacturers have not revealed the exact triggering and cycling algorithms used during system leak, it is difficult to explain the discrepancies among the different studies. However, considering the rapid growth of technology,

RESPIRATORY CARE Paper in Press. Published on May 21, 2013 as DOI: 10.4187/respcare.02466

it is critical to regularly repeat evaluations of ventilators to determine their ability to prevent triggering and cycling asynchrony caused by system leaks.

There were some limitations in this study. First, this study was not conducted on patients, raising the question of clinical relevance. However, lung simulator studies assure that experimental conditions are the same for each ventilator evaluated. It is impossible to control the level of the leak or maintain stable baseline conditions in clinical settings. Another limitation is that we tested only a limited range of leaks and ventilator settings. However, we chose these to represent the range of leak flows that are likely to be encountered in clinical settings. In addition, we were interested in evaluating the maximum capabilities of the ventilators tested. Third, in some patients with acute respiratory failure, ventilatory efforts may be higher than that of our simulated respiratory efforts and these higher efforts may affect our results.

In conclusion, all ventilators synchronized at baseline leak and some synchronized at low-level leaks, but there are wide variations of ventilator performance. The PB840 and the V60 were the only ventilators to synchronize with simulated respiratory efforts in all leak scenarios both in non-invasive and invasive ventilation modes. Ventilator performance is strongly influenced by leak, lung mechanics and PEEP settings. In clinical practice the ventilator is applied to patients with different lung mechanics, different ventilator settings, and rapid leak variations. Considering the above, the performance of leak compensation algorisms is a crucial issue regarding patient-ventilator interactions during mechanical ventilation. Although it is unclear how different ventilators may affect clinical outcome, clinicians should be aware of these differences when applying leak compensation during invasive and non-invasive ventilation.

RESPIRATORY CARE Paper in Press. Published on May 21, 2013 as DOI: 10.4187/respcare.02466

Further studies are needed to determine the impact of different ventilators on outcome during non-invasive and invasive ventilation.

RESPIRATORY CARE Paper in Press. Published on May 21, 2013 as DOI: 10.4187/respcare.02466

References

1. Hess DR. Patient-ventilator interaction during noninvasive ventilation. Respir Care 2011;56(2):153-165.
2. Stauffer JL, Olson DE, Petty TL. Complications and consequences of endotracheal intubation and tracheotomy. A prospective study of 150 critically ill patients. Am J Med 1981;70(1):65-76.
3. Rashkin MC, Davis T. Acute complications of endotracheal intubation. Relationship to reintubation, route, urgency, and duration. Chest 1986;89(2):165-167.
4. Chao DC, Scheinborn DJ, Stearn-Hassenpflug M. Patient-trigger asynchrony in prolonged mechanical ventilation. Chest 1997;112(6):1592-1599.
5. Thille AW, Rodriguez P, Cabello B, Lellouche F, Brochard L. Patient-ventilator asynchrony during assisted mechanical ventilation. Intensive Care Med 2006;32(10):1515-1522.
6. Vignaux L, Vargas F, Roeseler J, Tassaux D, Thille AW, Kossowsky MP, Brochard L, Jolliet P. Patient-ventilator asynchrony during non-invasive ventilation for acute respiratory failure: a multicenter study. Intensive Care Med 2009;35(5):840-846.
7. Vignaux L, Tassaux D, Carteaux G, Roeseler J, Piquilloud L, Brochard L, Jolliet P. Performance of noninvasive ventilation algorisms on ICU ventilators during pressure support: a clinical study. Intensive Care Med 2010;36(12):2053-2059.
8. Nava S, Bruschi C, Fracchia C, Braschi A, Rubini F. Patient-ventilator interaction and inspiratory effort during pressure support ventilation in patients with different pathologies.

RESPIRATORY CARE Paper in Press. Published on May 21, 2013 as DOI: 10.4187/respcare.02466

Eur Respir J 1997; 10(1):177-183.

9. Tokioka H, Tanaka T, Ishizu T, Fukushima T, Iwaki T, Nakamura Y, Kosogabe Y. The effect of breath termination criterion on breathing patterns and the work of breathing during pressure support ventilation. Anesth Analg 2001;92(1):161-165.
10. Ferreira JC, Chipman DW, Hill NS, Kacmarek RM. Bilevel vs. ICU ventilators providing noninvasive ventilation: effect of system leaks: a COPD lung model comparison. Chest 2009;136(2):448-456.
11. Carteaux G, Lyazidi A, Cordoba-Izquierdo A, Vignaux L, Jolliet P, Thille AW, Richard JCM, Brochard L. Patient-ventilator asynchrony during noninvasive ventilation: A bench and clinical study Chest 2012;142(2):367-376.
12. IngMar Medical. ASL5000 Active Servo Lung Computerized Breathing Simulator and Ventilator Test Instrument user`s manual. Pittsburgh, PA: IngMar Medical, 2006
13. Meyer TJ, Pressman MR, Benditt J, McCool FD, Millman RP, Natarajan R, Hill NS. Air leaking through the mouth during nocturnal nasal ventilation: effect on sleep quality. Sleep 1997; 20(7):561-569.
14. Sulemanji D, Marches A, Kacmarek RM. Noninvasive ventilation, synchronization following increasing and decreasing leaks: A lung model study (Abstract). Crit Care Med 2009; 37: no12 (Suppl) A469
15. Louis B, Leroux K, Isabey D, Fauroux B, Lofaso F. Effect of manufacturer-inserted mask leaks on ventilator performance. Eur Respir J 2010;35(3):627-636.
16. Imanaka H, Nishimura M, Takeuchi M, Kimball WR, Yahagi N, Kumon K. Autotriggering

RESPIRATORY CARE Paper in Press. Published on May 21, 2013 as DOI: 10.4187/respcare.02466

caused by cardiogenic oscillation during flow-triggered mechanical ventilation. Crit Care Med 2000;28(2):402-407.

17. Prinianakis G, Delmastro M, Carlucci A, Ceriana P, Nava S. Effect of varying the pressurization rate during noninvasive pressure support ventilation. Eur Respir J 2004;23(2):314-320
18. Hill LL, Pearl RG. Flow triggering, pressure triggering, and autotriggering during mechanical ventilation. Crit Care Med 2000;28(2):579-581.
19. Calderini E, Confalonieri M, Puccio PG, Francavilla N, Stella L, Gregoretti C. Patient-ventilator asynchrony during noninvasive ventilation: the role of expiratory trigger. Intensive Care Med 1999;25(7):662-667.
20. Vignaux L, Tassaux D, Jolliet P. Performance of noninvasive ventilation modes on ICU ventilators during pressure support: a bench model study. Intensive Care Med 2007;33(8):1444-1451.
21. Miyoshi E, Fujino Y, Uchiyama A, Mashimo T, Nishimura M. Effects of gas leak on triggering function, humidification, and inspiratory oxygen fraction during noninvasive positive pressure ventilation. Chest 2005;128(5):3691-3698.

RESPIRATORY CARE Paper in Press. Published on May 21, 2013 as DOI: 10.4187/respcare.02466

Figure legends

Figure 1

Auto- and miss-triggering under increasing and decreasing leak

Top: Auto- and miss-triggering under increasing and decreasing leak during non-invasive ventilation.

Miss-triggering:

†, Increasing leak: B→L3 > B→L1 ($p < 0.01$), B→L3 > L2→L3 ($p < 0.01$)

†, Decreasing leak: L3→L1 > L1→B ($p < 0.001$), L3→B > L1→B ($p < 0.001$),

L2→B > L1→B ($p < 0.01$), L3→B > L3→L2 ($p < 0.05$)

Bottom: Auto- and miss-triggering under increasing and decreasing leak during pressure support ventilation.

There was no significant difference between each leak scenarios.

Each box represents the inter-quartile range between the 25th and the 75th percentiles, with the median value. Vertical bars represent the maximum and minimum values except for outliers. Outliers identified with a circle are any data values which lie greater than 1.5 times the box length beyond the lower or higher edges of the box and within 3.0 times the box length.

RESPIRATORY CARE Paper in Press. Published on May 21, 2013 as DOI: 10.4187/respcare.02466

Extreme outliers identified with a star are any data which lie more than 3.0 times the box length beyond the lower or higher edges of the box.

PSV, pressure support ventilation; B to synch, number of breaths to synchronization

Figure 2

Time to baseline pressure under leak scenarios in COPD and ARDS models

Top- non-invasive mode and bottom- invasive mode, pressure support ventilation.

Left showed COPD model and right of ARDS model.

All data included the values of PEEP 5 cmH_2O and 10 cmH_2O. Absent bars indicate failure to synchronize during the leak scenario.

*G5 shows only the values of PEEP 5 cmH_2O in L2 and L3 categories because of non-synchronization with PEEP 10 cmH_2O under L2 and L3.

The histogram bars showed mean value. PSV, pressure support ventilation.

Figure 3

Cycling delay time under leak scenarios in COPD and ARDS models

Top- non-invasive mode and bottom- invasive mode, pressure support ventilation.

Left showed COPD model and right of ARDS model.

RESPIRATORY CARE Paper in Press. Published on May 21, 2013 as DOI: 10.4187/respcare.02466

Positive values represent delayed cycling and negative values represent premature cycling. All data included the values of PEEP 5 cmH_2O and 10 cmH_2O. Absent bars indicate failure to synchronize during the leak scenario.

*G5 shows only the values of PEEP 5 cmH_2O in L2 and L3 categories because of non-synchronization with PEEP 10 cmH_2O under L2 and L3. The histogram bars showed mean value. PSV, pressure support ventilation.

RESPIRATORY CARE Paper in Press. Published on May 21, 2013 as DOI: 10.4187/respcare.02466

Table 1. Ventilator specification

Ventilator	Software	Leak Compensation	IT Range	ET Range
Servo i	V5.00.00	NIV:50 L/min	0-100%, -20–0 cmH_2O	1–40%
		IV: None		
PB840	4-070212-85-AG	NIV: 65 L/min	0.2–20 L/min	1– 80%
		IV: 65 L/min		
C3	1.0.0	No information	0.5–15 L/min	5–70%
G5	2.1X	No information	0.5–15 L/min	5–70%
V500	2.23	180 L/min	0.2–15 L/min	5–70%
CareStation	5.0	No information	1–9 L/min, -10–0 cmH_2O	5–50%
Avea	4.4	No information	0.1–20 L/min	5–45%
V60	PN 1076723 Auto-Track+	60 L/min	Auto-Trak	Auto-Trak

IT Range, Inspiratory trigger range and ET Range, Expiratory trigger range, expressed as a percentage of inspiratory flow.

NIV, non-invasive ventilation mode; IV, invasive ventilation mode

RESPIRATORY CARE Paper in Press. Published on May 21, 2013 as DOI: 10.4187/respcare.02466

Table 2A. Synchronization capability under leak scenarios during non-invasive ventilation

	B to synch			Auto-triggering			Miss-triggering			Time to settle		
Ventilators	L1	L2	L3	L1	L2	L3	L1	L2	L3	L1	L2	L3
Servo-*i*	1.5	3.7	4	0.5	4.7	5.8	0	1.8	2.2	1.5	4	4
PB840	0.2	1	1.2	0	1	1	0.2	0	0.7	1	1	1
C3	1.8	1.8	2.3	0	2.8	3.2	1.3	1.8	2.3	3	2	2.5
G5*	3.2	3.2*	3.2*	0	5.7*	6.5*	4	3.2*	3.2*	4	3.2*	3.2*
CareStation	3.2	NS	NS	1.8	NS	NS	3.2	NS	NS	7.3	NS	NS
V500	3	NS	NS	0	NS	NS	3	NS	NS	5	NS	NS
Avea	NS	NS	NS	NS	NS	NS	NS	NS	NS	NS	NS	NS
V60	0.2	2.5	4.2	0	1.8	2	0.2	1.7	3.7	1.2	2.7	4.7

All data expressed as median value

B to synch, Number of breaths to synchronization; NS, cannot synchronize

L1: 9 to 10 L/min, L2: 26 to 27 L/min and L3: 35 to 36 L/min at a mean airway pressure of 7.5 cmH_2O.

L1 include the values of B→L1, L2→L1,and L3→L1; L2 include the values of B→L2, L1→L2, and L3→L2; L3 include the values of B→L3, L1→L3, and L2→L3

*G5 shows only the values of PEEP 5 cmH_2O in L2 and L3 categories because of non-synchronization with PEEP 10 cmH_2O under L2 and L3.

RESPIRATORY CARE Paper in Press. Published on May 21, 2013 as DOI: 10.4187/respcare.02466

Table 2B. Synchronization capability under leak scenarios during invasive ventilation

PSV mode	B to synch			Auto-triggering			Miss-triggering			Time to settle		
Ventilators	L1	L2	L3	L1	L2	L3	L1	L2	L3	L1	L2	L3
Servo-*i*	NS	NS	NS	NS	NS	NS	NS	NS	NS	NS	NS	NS
PB840	0	1	1	0	1.7	1.8	0	0.7	0.7	1	1	1.3
C3	1.8	2.3	NS	2.3	3.2	NS	0.8	2	NS	1.5	2.3	NS
G5	0	2.2	NS	0	3.2	NS	0	1	NS	0.3	1.2	NS
CareStation	0.3	NS	NS	0	NS	NS	0	NS	NS	1.2	NS	NS
V500	2.3	4.7	NS	0	5	NS	1.2	4.5	NS	2.8	7	NS
Avea	NS	NS	NS	NS	NS	NS	NS	NS	NS	NS	NS	NS
V60	0	0.3	4.3	0	0	2	0	0.3	3.8	0.5	1.5	4.3
PAC mode	B to synch			Auto-triggering			Miss-triggering			Time to settle		
Ventilators	L1	L2	L3	L1	L2	L3	L1	L2	L3	L1	L2	L3
Servo-*i*	NS	NS	NS	NS	NS	NS	NS	NS	NS	NS	NS	NS
PB840	0	1	1	0	1.7	2	0	0.7	1	0.3	1	1
C3	1.8	2	NS	2.3	3	NS	1.3	1.7	NS	1.3	2	NS
G5	0	1.7	NS	0	2	NS	0	0.7	NS	0.5	1.2	NS
CareStation	0	NS	NS	0.2	NS	NS	0.2	NS	NS	0.8	NS	NS
V500	2.3	4	NS	1.3	6.8	NS	1.7	3.8	NS	2.3	5	NS
Avea	NS	NS	NS	NS	NS	NS	NS	NS	NS	NS	NS	NS
V60	0	0.7	4.7	0	0	2	0	0.2	3.2	0	0.8	4.7

All data expressed as median value

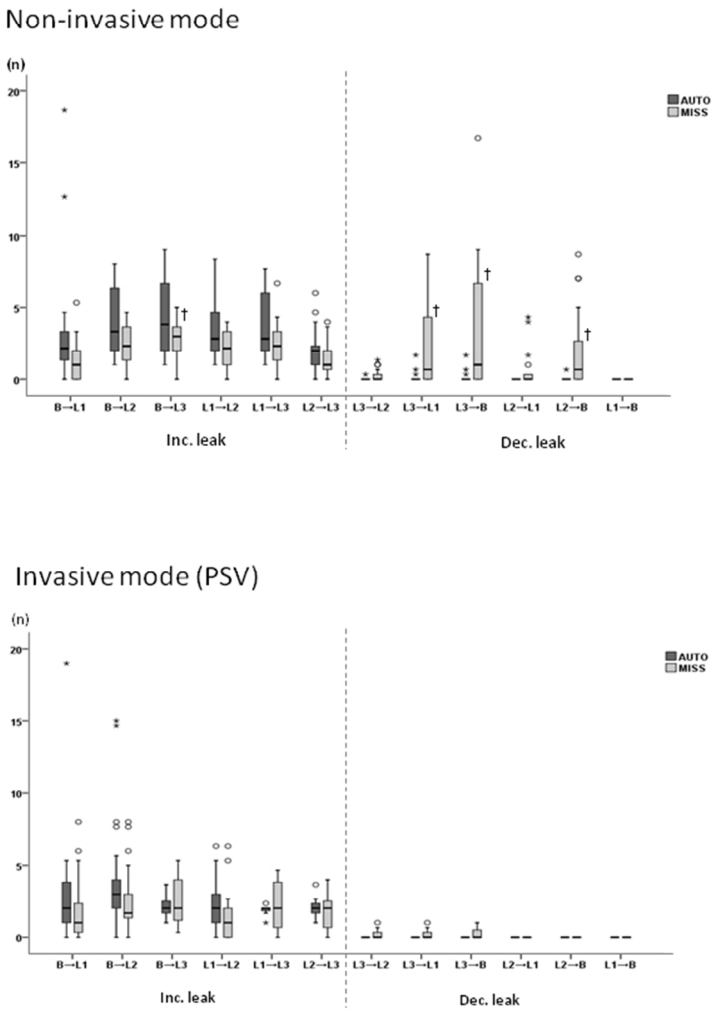
B to synch, Number of breaths to synchronization; NS, cannot synchronize

L1 include the values of B→L1, L2→L1,and L3→L1; L2 include the values of B→L2, L1→L2, and L3→L2; L3 include the values of B→L3, L1→L3, and L2→L3

L1: 3 to 4 L/min, L2: 9 to 10 L/min and L3: 26 to 27 L/min at a mean airway pressure of 7.5 cmH_2O.

RESPIRATORY CARE Paper in Press. Published on May 21, 2013 as DOI: 10.4187/respcare.02466

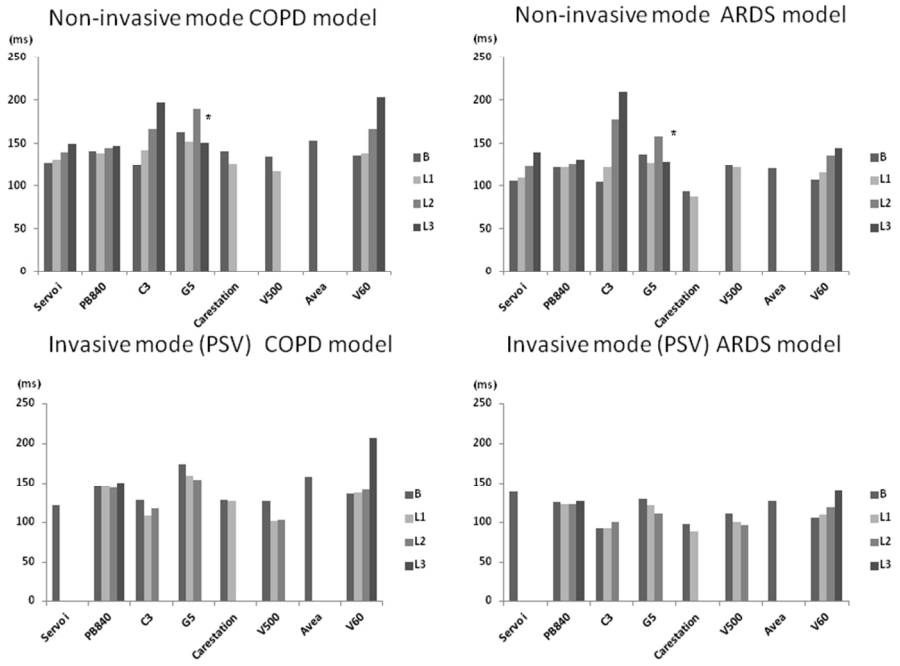
Figure 1



190x254mm (96 x 96 DPI)

RESPIRATORY CARE Paper in Press. Published on May 21, 2013 as DOI: 10.4187/respcare.02466

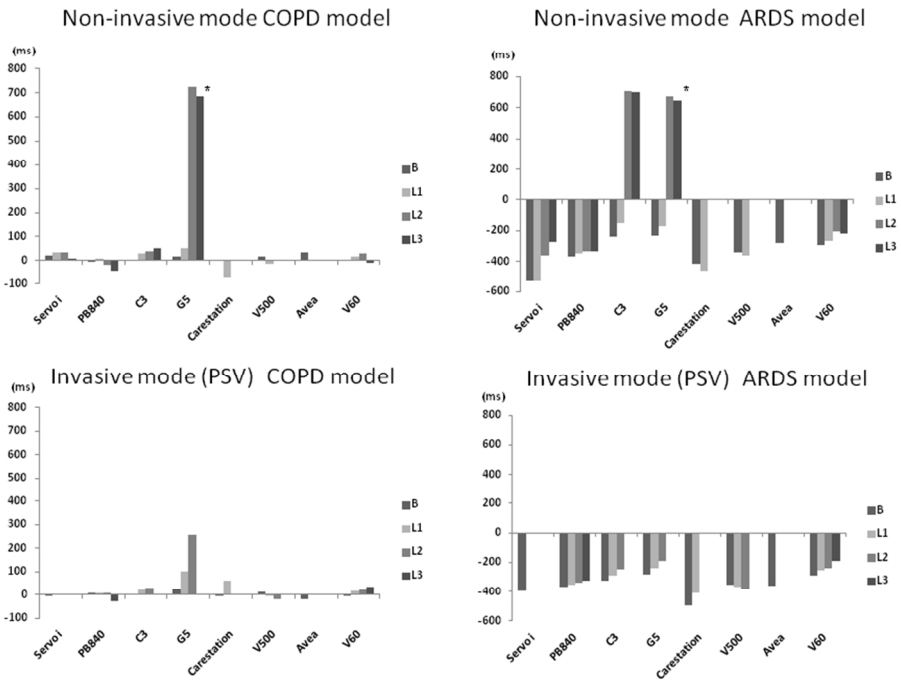
Figure 2



254x190mm (96 x 96 DPI)

RESPIRATORY CARE Paper in Press. Published on May 21, 2013 as DOI: 10.4187/respcare.02466

Figure 3



254x190mm (96 x 96 DPI)