Filters Alter the Performance of Noninvasive Ventilators

Julie Tolson, Rodney Hone, Christian Bandiera, Linda Rautela, Thomas J Churchward, Anna Ridgers, Mark E Howard, and Christopher J Worsnop

BACKGROUND: Noninvasive ventilation is recommended in hypercapnic respiratory failure secondary to ventilatory failure. Noninvasive ventilation may contribute to aerosol dispersion, which may increase the risk of transmission of COVID 2019. The addition of filters to the ventilator circuit has been recommended to reduce this risk. The aim of this benchtop study was to investigate the impact of adding filters to a ventilator circuit. METHODS: In this benchtop study, a breathing simulator was used with 4 commonly used ventilators. Ventilators were set to approximate the typical settings that are used for patients on long-term noninvasive ventilation. Ventilator performance was then evaluated with 3 circuit configurations in place: circuit A: no filter in situ; circuit B: 1 filter at the simulator end of the circuit; and circuit C: 1 filter at the simulator end of the circuit and a second filter at the ventilator end of the circuit. RESULTS: Ventilator variables were impacted by the addition of filters. Measurements of peak pressure ($P < .001$), tidal volume ($P < .001$), and peak flow ($P < .001$) decreased between circuit A and circuit C in all ventilators that were tested. Ventilator triggering was less sensitive in 3 of the 4 ventilators and the fourth ventilator did not trigger under the same simulator settings. CONCLUSIONS: This study demonstrated that ventilator settings established with filters in situ are not applicable if the ventilator is used without the filters. This is an important clinical consideration for patients who are hospitalized and require noninvasive ventilation in the COVID 2019 era. *Key words: COVID-19; infectious diseases; noninvasive ventilation; respiratory failure; benchtop study.* [Respir Care 2022;67(7):795–800. © 2022 Daedalus Enterprises]

Introduction

Noninvasive ventilation (NIV) is the standard of care for the management of acute respiratory failure due to exacerbations of COPD and of other respiratory and cardiac conditions. NIV is also used in the management of chronic respiratory failure due to various underlying respiratory conditions.[1] For patients with coronavirus disease 2019 (COVID-19), NIV therapy has been recommended in hypercapnic respiratory failure secondary to ventilatory failure.[2] NIV may be capable of generating aerosols, which may increase the transmission of SARS-CoV-2.[3,4] To reduce this risk, the addition of a filter to the patient end of the ventilator circuit has been recommended,[5] in conjunction with the appropriate use of personal protective equipment and other environmental modifications.[2,3]

In 2020, the Victorian Respiratory Support Service, a state-based chronic ventilation service with >1,000 patients in Victoria, Australia, adopted the recommended addition of a filter at the patient end of the ventilator circuit to minimize the potential spread of SARS-CoV-2 during hospital admission. Standard practice at the time was to add a separate filter to the ventilator end of the circuit, so each ventilator used for

Ms Tolson, and Mr Hone, Ms Rautela, Mr Churchward, Drs Ridgers, Howard, and Worsnop are affiliated with the Department of Respiratory and Sleep Medicine, Austin Health, Heidelberg, Victoria, Australia. Ms Tolson, and Mr Hone, Ms Rautela, Mr Churchward, Drs Ridgers, Howard, and Worsnop are affiliated with the Institute for Breathing and Sleep, Heidelberg, Victoria, Australia. Ms Tolson and Dr Ridgers are affiliated with the The University of Melbourne, Melbourne, Victoria, Australia. Mr Bandiera is affiliated with the Swinburne University of Technology, Hawthorn, Victoria, Australia.

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Correspondence: Julie Tolson BAppSc Hons, Department of Respiratory and Sleep Medicine, Austin Health, PO Box 5555, Heidelberg, Victoria, Australia, 3084. E-mail: julie.tolson@austin.org.au.

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NIV in hospitalized patients had 2 filters in its circuit. Our benchtop study (conducted at Austin Health, Department of Respiratory and Sleep Medicine, Heidelberg, Victoria, Australia) aimed to determine if ventilator performance was altered by adding filters to the ventilator circuit. Two experiments were undertaken to investigate this. The aim of experiment 1 was to determine the impact of filters on ventilator performance by measuring the following variables: peak pressure, PEEP, tidal volume, and peak flow. The aim of experiment 2 was to examine the impact of filters on ventilator triggering performance.

QUICK LOOK

Current knowledge

Noninvasive ventilation (NIV) may contribute to aerosol dispersion. NIV is the recommended treatment for hypercapnic respiratory failure secondary to ventilatory failure. The use of NIV in a health-care setting in patients with COVID-19 increases the risk of transmission of the virus. The addition of a filter to the ventilator circuit has been recommended to reduce this risk.

What this paper contributes to our knowledge

When using a bench model, 4 commonly used NIV devices were tested with 3 different circuit configurations: (1) no filter in the circuit, (2) a filter at the simulator end of the circuit, and (3) a filter at the simulator end and a filter at the ventilator end of the circuit. Ventilator variables of peak pressure, PEEP, tidal volume, and peak flow were impacted by the addition of filters. Triggering became less sensitive in 3 of the NIV devices and the fourth device failed to trigger.

Methods

This benchtop study was conducted in a laboratory setting and consisted of 2 separate experiments. The 2 experiments were designed to exclude patient-related factors that may impact ventilator performance such as mask leak and variations in effort. The equipment used in each experiment in an ASL 5000 Breathing Simulator (IngMar Medical, Pittsburgh, Pennsylvania) and 4 ventilators: 3 ResMed (San Diego, California) models: Stellar 150, Astral 100, and S9 VPAP, and the Philips Respironics (Pittsburgh, Pennsylvania) model (DreamStation AVAPS 30 AE). These ventilators were selected due to their common use in the chronic respiratory failure population. Three ventilator circuit configurations were evaluated: circuit A had no filter in the circuit; circuit B had a Pharma Mini port (bacterial/viral, heat and moisture exchange) filter (Pharma Systems, Knivsta, Sweden) at the simulator end of the circuit; and circuit C retained the Pharma Mini port filter at the simulator end and a Suregard bacterial/viral filter (Bird Healthcare, Melbourne, Victoria, Australia) was added to the ventilator end of the circuit (Fig. 1). The dead-space volume of the Pharma mini port filter was 26 mL, and the resistance was 1.5 cm H_2O/L/s at a flow of 20 L/min. The dead space volume of the Suregard filter was 50 mL and the resistance was 0.71 cm H_2O/L/s at a flow of 12 L/s.

In experiment 1, to approximate typical settings that are used for patients on long-term noninvasive ventilation, all ventilators were set to the following; spontaneous mode, inspiratory positive airway pressure, 20 cm H_2O; expiratory positive airway pressure, 10 cm H_2O; rise time, 150 ms; fall time, 200 ms; and minimum / maximum inspiratory time, 1.0 / 1.5 s. Trigger and cycle settings for the ResMed ventilators were set to medium and low, respectively. The Philips Respironics DreamStation ventilator has a proprietary automated triggering and cycling function that could not be adjusted. The Philips Respironics DreamStation has the option of average volume-assured pressure support (AVAPS) mode. This mode was not used in this study. The ASL 5000 was set to the following: rate, 12 breaths/min; resistance, 8 cm H_2O/L/s; compliance, 61 mL/cm H_2O; muscle pressure, 2 cm H_2O; rise time, 9%; release time, 2%; and inspiratory hold, 0%. In experiment 1, steady state was achieved by 30 s. The next 20 breaths were used to calculate the results for each simulation.

In experiment 2, the ventilator settings were the same as in experiment 1 except for trigger sensitivity. Four different trigger sensitivity settings were tested in the ResMed ventilators: low, medium, high, and very high. The DreamStation ventilator has a proprietary automated triggering and cycling function that could not be altered. To investigate the trigger function in experiment 2, some of the settings for the ASL 5000 had to be different from the settings used in experiment 1. The settings were as follows; rate = 12 breaths/min; resistance, 8 cm H_2O/L/s; compliance, 61 mL/cm H_2O; muscle pressure, 0.9 cm H_2O; rise time, 10%; release time, 10%; and inspiratory hold, 6%. The same 3 circuit configurations were used in experiment 2 as in experiment 1 (Fig. 1). Each simulation was run until triggering was clearly present or clearly absent, ~10–15 breaths.

The Compumedics Grael (Abbotsford, Victoria, Australia) polysomnography system (Fig. 1) was used as an interface to display selected results of experiment 2 (Fig. 2). The pressure signal as displayed in Figure 2 was taken from a port on the connector adjacent to the simulator (Fig. 1). The other signals displayed in Figure 2 (flow, tidal volume, and trigger/cycle) were computed by the Stellar 150 device. These signals were received by a ResMed connection module, which then performed an analog-to-digital conversion and the signals were then outputted to the Compumedics Grael polysomnography system. Before conducting experiment 2, the analog signals were calibrated by using the ResMed connection module and

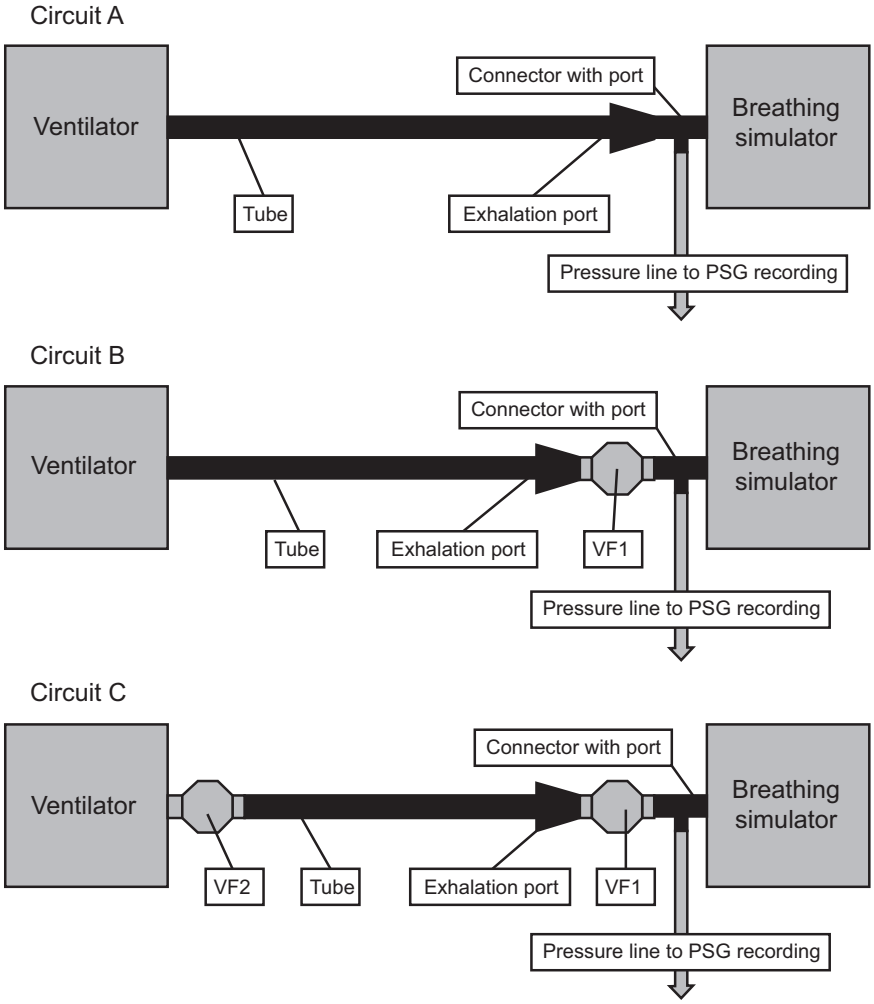

Fig. 1. Three benchtop simulator-ventilator circuit configurations. Circuit A: no filter in situ; circuit B: a VF1 filter in situ; circuit C: VF1 and VF2 filters in situ. VF1 = filter at the simulator end of the circuit; VF2 = filter at the ventilator end of the circuit; PSG = polysomnography.

the Compumedics Grael polysomnography system, and the pressure signal was calibrated with a fluid manometer.

Statistical Analysis

Statistical analyses were performed by using IBM SPSS Statistics software version 23 (Armonk, New York). Results are reported as median and interquartile range. For each ventilator model, when homogeneity of variances were met, one-way analyses of variance were conducted to determine if there were differences between circuit configurations and Bonferroni post hoc analyses were performed and, when homogeneity of variances were not met, Welch analyses of variance were conducted to determine differences among circuit configurations and Games-Howell post hoc analyses were performed. All statistical comparisons were 2-tailed, and the level of significance was set at $P < .05$.

Results

Results of experiment 1 are presented in Table 1. For each ventilator, there were significant differences in the ventilator variables of peak pressure, tidal volume, and peak flow between circuit configurations. There were no differences in PEEP among the circuit configurations for the Stellar 150, Astral 100, and DreamStation ventilators. Post hoc analyses showed that there was a reduction in most ventilator variables with the addition of filters to the circuit (Table 1). Results of experiment 2 are presented in Table 2. Triggering became less sensitive in all ResMed ventilators with the addition of filters. The DreamStation failed to trigger under the same simulated respiratory effort. Changes in pressure and flow profiles, along with changes in ventilator cycling in the Stellar 150 with the addition of filters to the circuit are shown in Figure 2. With no filter in situ (circuit A), ventilator triggering was achieved at the high setting; when 2 filters

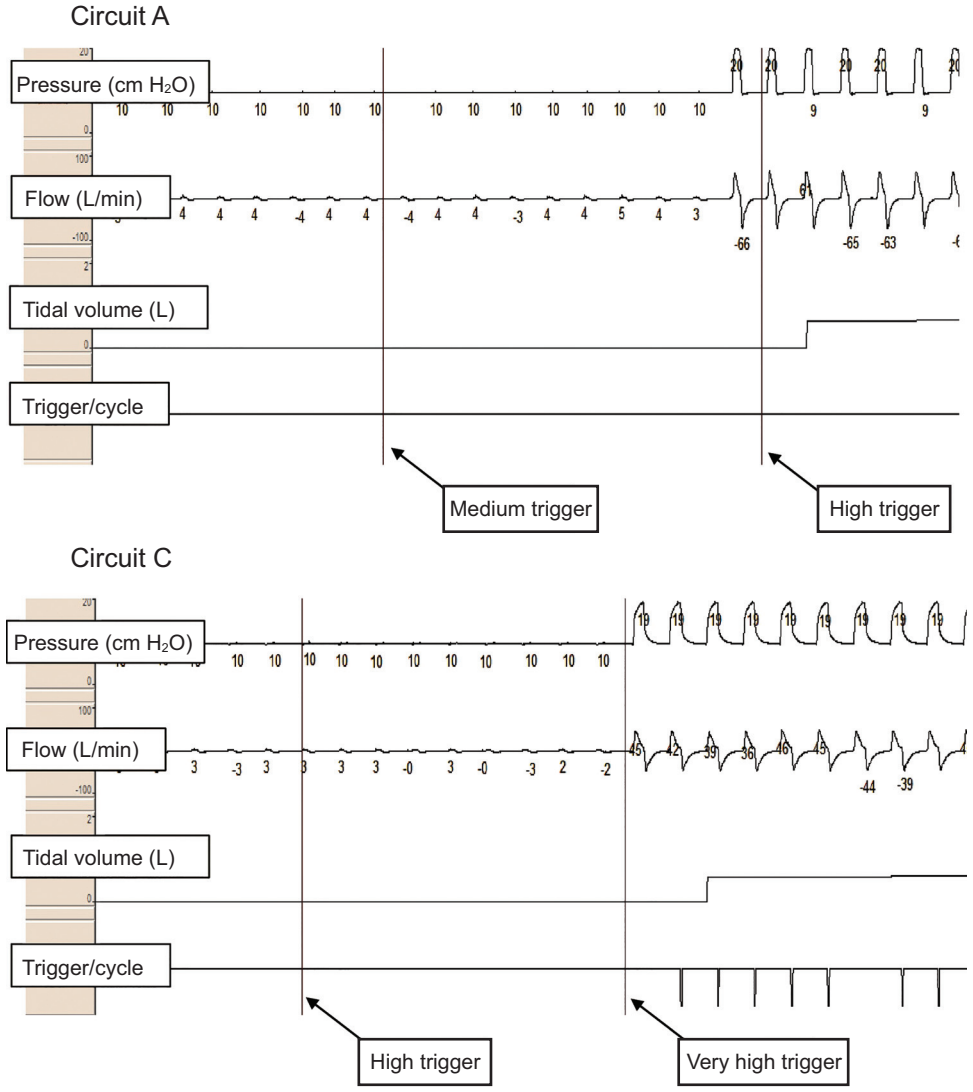

Fig. 2. Experiment 2: Compumedics Grael polysomnography (PSG) recording of signals produced by the ResMed Stellar 150 ventilator when connected to an ASL 5000 Breathing Simulator under 2 different circuit configurations. Circuit A (no filter in situ): ventilator triggering was achieved when the trigger sensitivity was changed from medium to high. Circuit C (2 filters in situ; 1 at the simulator end and 1 at the ventilator end of the circuit): ventilator triggering was achieved when the trigger sensitivity was changed from high to very high.

were in situ (circuit C), ventilator triggering was not achieved until the trigger setting was increased to very high (Fig. 2).

Discussion

Both experiments in this study demonstrated that ventilator performance was affected by modifying the circuit. Variables generated by the ventilator and the ventilator's ability to appropriately detect and trigger ventilation were affected when the circuit configuration was modified. The changes seen in ventilator performance were likely attributable to the extra resistance in the circuit with the addition of filters. In ventilators that use flow triggering, as the trigger sensitivity settings are increased, the required flow to achieve triggering is reduced. It is likely that the addition of filters reduces the circuit flow due to the extra resistance. Thus, a higher sensitivity setting will be required to achieve triggering for the same inspiratory effort. It is difficult to compare the trigger sensitivity settings that achieved triggering among the machines because the flows that correspond to these settings are different among the machines or are unknown. However, in the second experiment it was apparent that the ResMed ventilators required a higher sensitivity setting to achieve triggering when the filters were added and all other factors remained unchanged.

The Stellar 150 and Astral ventilators have a "learn circuit" function that measures circuit resistance and allows the device to compensate for this resistance. The instructions for this function are to not include any extra resistance between

Table 1. Experiment 1. Impact of Filters on the Performance of 4 Ventilators

Ventilator	Circuit Configurations			*P*	Post Hoc Comparison[†]
	Circuit A (1)	Circuit B (2)	Circuit C (3)		
S9 VPAP					
Peak pressure, cm H_2O	21.8 (21.7–21.8)	19.9 (19.9–19.9)	19.1 (19.1–19.1)	<.001*	1 vs 2[†]; 1 vs 3[†]; 2 vs 3[†]
PEEP, cm H_2O	10.1 (10.0–10.1)	10.1 (10.1–10.1)	9.7 (9.7–9.7)	<.001*	1 vs 3[†]; 2 vs 3[†]
Tidal volume, mL	589.5 (588.0– 592.0)	571.0 (571.0–571.0)	538.0 (538.0–538.0)	<.001*	1 vs 2[†]; 1 vs 3[†]; 2 vs 3[†]
Peak flow, L/min	73.7 (73.5–73.8)	50.0 (49.4–50.5)	43.9 (43.5–44.5)	<.001*	1 vs 2[†]; 1 vs 3[†]; 2 vs 3[†]
Stellar 150					
Peak pressure, cm H_2O	21.5 (21.4–21.5)	20.0 (20.0–20.0)	20.0 (19.9–20.0)	<.001*	1 vs 2[†]; 1 vs 3[†]
PEEP, cm H_2O	10.1 (10.1–10.1)	10.1 (10.1–10.1)	10.1 (10.1–10.1)	NA	
Tidal volume, mL	600.0 (598.3–602.8)	571.0 (571.0–572.0)	571.0 (571.0–571.0)	<.001*	1 vs 2[†]; 1 vs 3[†]
Peak flow, L/min	68.5 (68.3–68.8)	45.4 (44.6–46.3)	44.5 (44.0–44.9)	<.001*	1 vs 2[†]; 1 vs 3[†]
Astral 100					
Peak pressure, cm H_2O	21.2 (21.2–21.4)	20.0 (20.0–20.0)	20.1 (20.1–20.2)	<.001*	1 vs 2[†]; 1 vs 3[†]; 3 vs 2[†]
PEEP, cm H_2O	10.1 (10.1–10.1)	10.1 (10.0–10.1)	10.1 (10.0–10.1)	.36*	
Tidal volume, mL	578.5 (576.3–581.8)	560.0 (558.0–562.5)	571.5 (566.0–574.5)	<.001	1 vs 2[†]; 1 vs 3[†]; 3 vs 2[†]
Peak flow, L/min	59.7 (59.1–60.3)	41.9 (41.1–42.4)	39.1 (38.8–39.7)	<.001	1 vs 2[†]; 1 vs 3[†]; 2 vs 3[†]
DreamStation					
Peak pressure, cm H_2O	20.4 (20.4–20.4)	19.8 (19.8–19.8)	19.0 (19.0–19.0)	<.001	1 vs 2[†]; 1 vs 3[†]; 2 vs 3[†]
PEEP, cm H_2O	10.0 (10.0–10.0)	10.0 (10.0–10.0)	9.6 (9.6–9.6)	NA	
Tidal volume, mL	586.0 (585.0–588.0)	551.5 (549.0–555.8)	525.5 (521.3–528.8)	<.001*	1 vs 2[†]; 1 vs 3[†]; 2 vs 3[†]
Peak flow, L/min	42.1 (41.8–42.4)	28.8 (28.8–28.8)	26.0 (26.0–26.2)	<.001*	1 vs 2[†]; 1 vs 3[†]; 2 vs 3[†]

Data are presented as median (IQR) for peak pressure, PEEP, tidal volume, and peak flow for the 4 ventilators. Ventilator performance was evaluated with 3 different circuit configurations: Circuit A (1): no filter in situ; Circuit B (2): 1 filter at the simulator end of the circuit; Circuit C (3): 1 filter at the simulator end and a second filter at the ventilator end of the circuit.

*P from Welsh analysis of variance.

†$P < .001$.

NA = could not be calculated.

the patient interface and the exhalation port. For the filter to perform its role, it must be added to the circuit between the patient interface and the exhalation port; therefor, in this scenario, the filter will not be included in the resistance compensation feature. The second experiment also revealed that the DreamStation proprietary automated triggering function needed the simulator settings to be adjusted to achieve triggering. This proprietary automated triggering function uses an automated triggering, cycling, and leak compensation algorithm that adjusts ventilation to the patient's natural breathing patterns. It is not user adjustable, which may mean that it has limitations because it was not able to be triggered without changes being made to the simulator settings.

Patout et al[6] also extensively studied the effects of modifying noninvasive ventilator circuits to prevent aerosol dispersion in this COVID-19 era. Their studies were also benchtop, but they used a model head with an artificial lung. They assessed various strategies to minimize aerosol dispersion, by testing 8 circuit setups, a dual-limb circuit with either a helmet interface or an oronasal mask, a single-limb circuit with a passive exhalation valve, 3 single-limb circuits with custom-made introduced leaks and 2 single-limb circuits with active exhalation valves.[6] In addition, they used 1 of 2 types of filters: a heat-and-moisture exchange filter or a low-resistance bacterial filter. In their model, all the different types of circuits that they tested affected ventilator performance. Filters increased the inspiratory flow required to trigger the ventilator, with a greater inspiratory effort required, longer time to trigger with greater work of breathing, and more asynchrony.[6] Our study adds to their findings because firstly we used different filter brands, and we used the 2 filters (1 bacterial/viral, heat and moisture exchange filter and 1 bacterial/viral filter) in a series. We also tested not just 1 ventilator, but the 4 commonly used models of ventilators in Australia. Our results are consistent with those of Patout et al.[6]

The findings of our study have a number of clinical implications. Patient-ventilator asynchrony has been demonstrated to be associated with reduced tolerance of NIV therapy and a negative impact on sleep quality.[7-9] One of the key components of patient-ventilator asynchrony is trigger asynchrony. This occurs when there is an uncoupling between patient inspiratory effort and a triggering of a ventilator-supported breath. As demonstrated in experiment 2, modifying the circuit directly affected the trigger sensitivity, which may increase the incidence of patient-ventilator asynchrony. This is also clinically relevant if a patient with previously established chronic respiratory failure becomes acutely unwell and requires the addition of a filter to the circuit because his or her previous trigger sensitivity may no

Table 2. Experiment 2. Triggering in 4 Ventilator Models With 3 Circuit Configurations in Place

Trigger Setting per Ventilator Model	Circuit A	Circuit B	Circuit C
S9 VPAP			
Low	N	N	N
Medium	Y/N	N	N
High	Y	Y/N	Y/N
Very high	Y	Y	Y
Stellar 150			
Low	N	N	N
Medium	N	N	N
High	Y	N	N
Very high	Y	Y	Y
Astral 100			
Low	N	N	N
Medium	N	N	N
High	Y	Y/N	Y/N
Very high	Y	Y	Y
DreamStation			
Automated trigger function	N	N	N

Experiment 2: 10 to 15 breath simulations in the 4 ventilators. Trigger sensitivities were tested with 3 different circuit configurations in place: circuit A (no filter in situ), circuit B (1 filter at the simulator end of the circuit), and circuit C (1 filter at the simulator end and a second filter at the ventilator end of the circuit). Simulator setting: muscle pressure 0.9 cm H_2O.
N = ventilator not triggered
Y/N = inconsistent ventilator triggering
Y = ventilator triggered

longer be adequate. NIV titration, particularly in the acute hospital-based environment, involves assessment of clinical and ventilation variables, and arterial blood gas sampling. Review of ventilator variables such as tidal volume and minute ventilation are frequently used in the titration of NIV settings. Experiment 1 demonstrated that the ventilator-reported tidal volume was directly affected by the addition of the filters to the circuit. This is an important clinical consideration when performing bedside titration of NIV settings.

There were a few limitations of our study. Both experiments were benchtop simulations only and did not reflect a patient's respiratory mechanics. In patients, respiratory muscle effort, ventilatory drive, and respiratory compliance vary from breath to breath, thus the capture and associated triggering of a ventilator-delivered breath may improve. The 3 circuit configurations used were leak free because a breathing simulator was used in place of a patient. In clinical circumstances, it would be rare to observe a leak-free ventilator circuit in patients. Also, although we tested 4 ventilators and 3 circuit configurations, our results do not necessarily apply to other ventilators or filter configurations that may be used elsewhere.

Conclusions

With the use of a breathing simulator, this benchtop study demonstrated that modification of an NIV circuit configuration had direct effects on ventilator performance. This is an important consideration for patients with both acute and chronic respiratory failure who require NIV therapy when admitted to the hospital in the COVID-19 era. Further research is needed in other ventilator models.

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