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RECEIVED 02 December 2024

ACCEPTED 04 July 2025

PUBLISHED 23 July 2025

## CITATION

Li X, Dai B, Zhou C, Hou H, Wang S, Li X,  
Zhao H, Wang W and Tan W (2025) Effects of  
different exhalation valves on CO<sub>2</sub> rebreathing  
and ventilator performance during  
noninvasive ventilation.  
*Front. Med.* 12:1538280.  
doi: 10.3389/fmed.2025.1538280

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# Effects of different exhalation valves on CO<sub>2</sub> rebreathing and ventilator performance during noninvasive ventilation

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**Background:** Noninvasive ventilation (NIV) is widely used to improve oxygenation and reduce carbon dioxide (CO<sub>2</sub>) retention in patients with respiratory failure. However, it remains unclear whether different types of exhalation valves affect CO<sub>2</sub> rebreathing and ventilator performance during NIV.

**Methods:** Three noninvasive ventilators (V60, Flexo, and Stellar150) with single-limb circuits and four different exhalation valves (single-arch, whisper swivel, plateau exhalation, and vented mask valves) were separately connected in series to a lung simulator. CO<sub>2</sub> gas was injected from the simulated lung outlet, maintaining the end-expiratory CO<sub>2</sub> (PetCO<sub>2</sub>) at 80 mmHg. Both the CO<sub>2</sub> rebreathing volume (CO<sub>2REB</sub>) and the parameters displayed on the lung simulator and ventilator were recorded under each condition.

**Results:** The mean CO<sub>2REB</sub> values of the four aforementioned valves were  $18.51 \pm 2.87$ ,  $18.25 \pm 2.73$ ,  $17.78 \pm 2.98$ , and  $14.26 \pm 0.92$  mL/breath, respectively, with no significant differences among the first three types but all significantly higher than that of the mask valve (all  $p < 0.0001$ , rate of difference  $> 10\%$ ). Except with the V60 ventilator, some ventilator performance parameters (triggering and control performance) were significantly lower for the plateau valve than for the others, the rate of difference in tidal volume ( $V_T$ ) between the ventilator and the simulated lung exceeded 10% for all exhalation valves (all  $p < 0.01$ ).

**Conclusion:** Mask valves showed significantly lower CO<sub>2</sub> rebreathing than circuit-located valves (single-arch, whisper swivel, and plateau exhalation) in this NIV bench study. The different valves influenced ventilator performance differently, particularly the plateau valve. These findings necessitate further clinical validation *in vivo*.

## KEYWORDS

noninvasive ventilation, exhalation valves, CO<sub>2</sub> rebreathing, ventilator performance, plateau exhalation

## Introduction

Noninvasive ventilation (NIV) is extensively used among patients with acute and chronic respiratory failure, reducing the intubation rate, the risk of respiratory failure after extubation, hospital length of stay, and mortality (1–5). In certain patient populations, such as those experiencing an exacerbation of chronic obstructive pulmonary disease (COPD), NIV is considered first-line therapy and the standard of care (1–3).

NIV is delivered via a noninvasive ventilator with a single limb, utilizing an exhalation port that directs exhaled gasses from the breathing circuit to the atmosphere. Currently, four types of exhalation valve are available for clinical practice: the single-arch, plateau, whisper swivel, and vented mask (integrated leak port within the mask) valves (6). The characteristics of such valves including design features that allow controlled amounts of air to escape as “intentional leaks” vary greatly (7), influencing rebreathing risk and carbon dioxide retention, particularly in hypercapnic respiratory failure. Recent clinical data (8) suggested that insufficient CO<sub>2</sub> clearance may be associated with worse medium-term outcomes, highlighting the importance of CO<sub>2</sub> washout during inpatient NIV use. Besides, the characteristics of exhalation valves affecting the ability of a ventilator to generate pressure and thus its performance; this is particularly true of plateau exhalation valves (7). However, it remains unclear whether commercial exhalation valves differ in their effects on CO<sub>2</sub> rebreathing and ventilator performance when used with different ventilators, and the optimal standards for exhalation valve selection have yet to be identified. We therefore employed bench methodologies to systematically quantify valve performance mechanisms, establishing a mechanistic basis for subsequent clinical validation.

In the present bench study, we applied the simulated lung model to comprehensively compare CO<sub>2</sub> rebreathing among the four types of valve in combination with three noninvasive ventilators. We also evaluated the impact of valve type on ventilator performance, including triggering, control, and monitoring parameters.

## Materials and methods

### Bench model setup

The Active Servo Lung simulation system (ASL5000, Ingmar, Pittsburgh, PA) is a precise breathing simulator containing a piston that moves inside a cylinder, capable of simulating breathing patterns of patients with COPD. The following parameters adopted from previous literature were applied to the simulated COPD lung model: compliance of 60 mL/cmH<sub>2</sub>O, inspiratory resistance of 10 cmH<sub>2</sub>O/L/s, expiratory resistance of 15 cmH<sub>2</sub>O/L/s, and maximum inspiratory pressure drop of −8 cmH<sub>2</sub>O. To simulate the profile of the negative pressure produced by respiratory muscles, 5% of the respiratory cycle time was set to active inspiration, 3% to an end-inspiratory hold, and 15% to return pressure to baseline (9–13). Breathing frequency was set at 10, 15, and 25 breaths/min, respectively.

Three noninvasive ventilators were used: the V60 (Philips, Carlsbad, United States), Flexo (Curative Medical, Jiangsu, China), and Stellar150 (ResMed, Saint Priest, France) ventilators. Both the Flexo and Stellar150 ventilators were used in S (Spontaneous) mode with an inspiratory pressure airway pressure (IPAP) and expiratory positive airway pressure (EPAP) of 12/4 and 16/4 cmH<sub>2</sub>O, respectively. The other settings were as follows: inspiratory trigger sensitivity (Isens) of 2, expiratory trigger sensitivity (Esens) of 2, and inspiratory slope (Islop) of 2 for the Flexo ventilator; and a trigger of high, switch of high, rise of 150, and decline of 10 for the Stellar150. The V60 ventilator was used in S/T (Spontaneous/Timed) mode with the lowest respiratory rate (4 breaths/min), as there is no S mode. The pressure

settings were the same as above, with the following additional settings: inspiratory time (Ti) of 0.9, rise time of 1, and trigger of automatic. If miss-triggering occurred, the trigger parameter was readjusted to avoid this.

Two oronasal masks (first- and second-generation), with identical inner volumes and shape, were used from the same manufacturer (Bestfit™ I/II; Curative, Santa Clara, California). The first-generation mask, without an exhalation valve, was separately connected to a single-arch valve, whisper swivel valve, and plateau exhalation valve (all from Respirationics, Andover, MA). The second-generation mask has an exhalation port integrated into the mask itself, located above the nose bridge. To measure leak of the exhalation valve (Figure 1A), the VT Plus airflow analyzer (Fluke Biomedical, Everett, WA) was connected in series in the single-limb breathing circuit. The mask was fixed to the face of an adult-sized model head equipped with a mouth port and a headband. To avoid unintentional leaks, plasticine was used to seal any spaces around the interface between the interface and the head model. The leakage of each valve was measured at continuous positive airway pressures (CPAPs) of 4, 12, and 20 cmH<sub>2</sub>O.

The bench model is diagrammed in Figure 1B. The noninvasive ventilator was connected to the simulated lung through a single-limb breathing circuit behind the model head, which was connected one of the four valves. A micro gas flow controller (YJ-700CF, kongxin instrument company, Guangxi, China) was connected to a compressed CO<sub>2</sub> cylinder (0.3 kpa), and CO<sub>2</sub> was injected from the simulated lung outlet. The partial pressure of end-tidal carbon dioxide (PetCO<sub>2</sub>) was monitored using a vital sign monitor (NTID, newtech medical, Shenzheng, China), which was situated behind the model head to maintain PetCO<sub>2</sub> at 80 mmHg. A self-made gas analyzer, integrating flow rate and CO<sub>2</sub> sensors, was located between the model and the lung simulator.

### Data acquisition

Inspiratory flow was continuously monitored with a flow sensor (OOM109/OOM109-LF2, EnviteC, Wismar/Honeywell, Morristown, NJ) at a sampling frequency of 30 ms. The inspiratory phase was identified according to the flow waveform. We designed software that multiplied the CO<sub>2</sub> concentration by the inspiratory flow at each sampling point during the inspiratory phase. The delivered CO<sub>2</sub> volume was calculated using the equation shown in Figure 1C.

When the noninvasive ventilator and CO<sub>2</sub> ran stable for more than 5 min, ventilator parameters were recorded for 10 consecutive respiratory cycles (Figure 1D). The parameters included triggering performance, including time to trigger (T<sub>trig</sub>, in milliseconds [ms]), time from the beginning of the lung simulator's inspiratory effort to the lowest airway pressure needed to trigger the ventilator (T<sub>Pmin</sub>), and the magnitude of airway pressure drop needed to trigger (P<sub>trig</sub>); control performance, dividing into time to achieve 90% of the inspiratory target during inspiration (T<sub>90%</sub>, in ms), inspiratory time (Ti), peak inspiratory pressure (PIP), peak inspiratory flow (PIF), mean inspiratory flow (MIF), and positive end expiratory pressure (PEEP); and tidal volume (V<sub>T</sub>) parameters including actual V<sub>T</sub> (V<sub>T-L</sub>, that displayed on the simulated lung) and monitored V<sub>T</sub> (V<sub>T-V</sub>, that displayed on the ventilator).

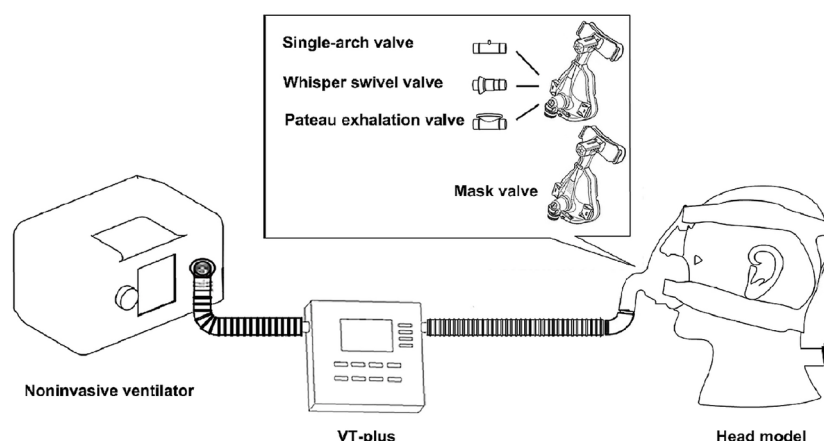
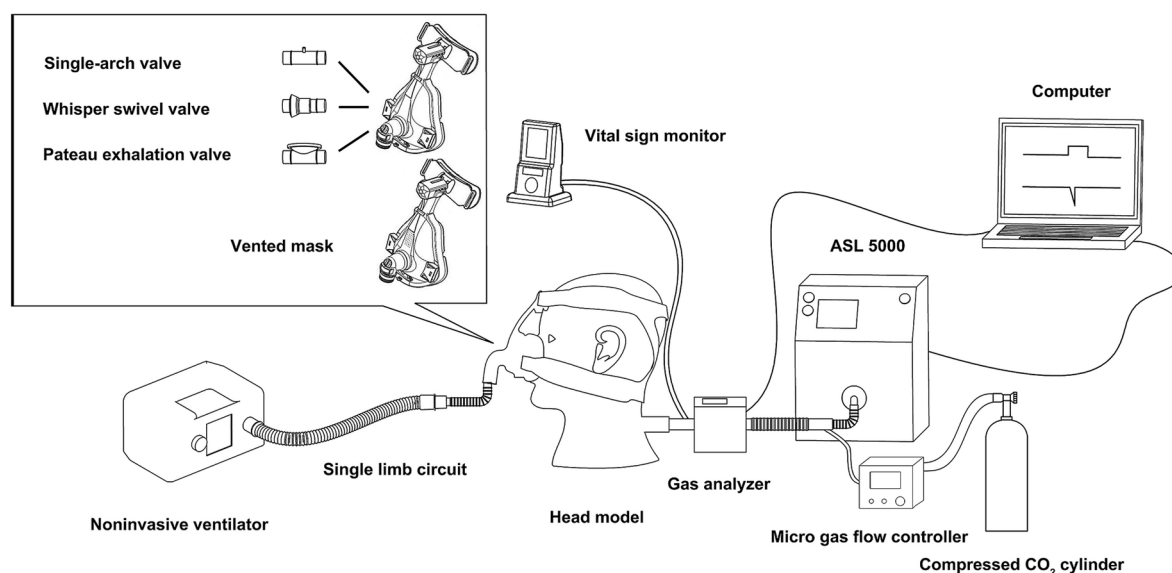
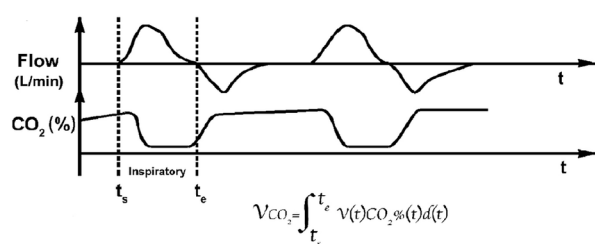
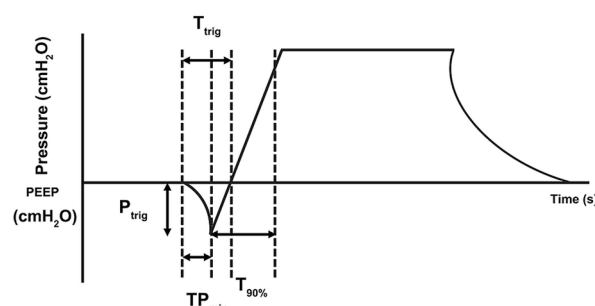
**A****B****C****D**

FIGURE 1

Illustration of the experiment. (A) Measuring the intentional leak of the exhalation valves, (B) schematic diagram of experimental setup, (C) time curve of flow and CO<sub>2</sub> concentration and the equation, (D) graphic explanation of the ventilator parameters.

## Statistical analysis

All statistical analyses were performed using GraphPad Prism (GraphPad Software, San Diego, CA). Data are expressed as means ± standard deviations. Differences between parameters displayed on the ventilator and lung simulator were calculated as follows: rate of

difference = (the parameters displayed on the ventilator – those on the lung simulator) / the parameters displayed on the lung simulator. A paired *t*-test was used to assess these differences. One-way analysis of variance (ANOVA) was used to compare the influence of different exhalation valves on CO<sub>2</sub> rebreathing and ventilator performance under the same condition. Changes were considered clinically

significant only when the rate of difference exceeded 10% and the differences were statistically significant at  $p < 0.05$  (9–13).

## Results

### Leakage

The leakages of different exhalation valves are shown in Figure 2. Those of the single-arch, whisper swivel, and mask valves increased linearly with increasing CPAP. With CPAP as the independent variable and leakage as the dependent variable, the general linear equations for the valves were  $y = 1.08x + 8.54$  ( $R^2 = 0.99$ ,  $p < 0.01$ ),  $y = 1.64x + 12.93$  ( $R^2 = 0.99$ ,  $p < 0.01$ ), and  $y = 2.28x + 10.40$  ( $R^2 = 0.99$ ,  $p < 0.01$ ), respectively. In contrast, the plateau exhalation valve exhibited a relatively constant leakage at different CPAP levels ( $24.44 \pm 0.9$  L/min) (Figure 2).

### CO<sub>2</sub> rebreathing

The mean CO<sub>2REB</sub> values under different ventilator settings for the single-arch, whisper swivel, plateau, and mask valves were  $18.51 \pm 2.87$ ,  $18.25 \pm 2.73$ ,  $17.78 \pm 2.98$ , and  $14.26 \pm 0.92$  mL/breath, respectively. There were no significant differences among the first three valve types, all of which were significantly higher than that of the mask valve (all  $p < 0.0001$ , rate of difference  $> 10\%$ ). Similar results were observed at different pressure settings (IPAP/EPAP = 12/4 and 16/4 cmH<sub>2</sub>O) (Figure 3A) and breathing frequencies (10 and 15, but not at 25, breaths/min) (Figure 3B), and for different types of ventilator (V60, Flexo, and Stellar150) (Figure 3C).

### Ventilator performance

The effects of each valve on ventilator performance varied greatly, particularly for the plateau valve. In regard to triggering performance

(Ttrig, T<sub>pmin</sub>, Ptrig), values were significantly lower for the plateau valve versus all others (all  $p < 0.01$ , rate of difference  $> 10\%$ ), except for the V60 ventilator (Figure 4A). Regarding control performance, T<sub>90%</sub> and T<sub>i</sub> were the lowest when using the plateau valve with the Flexo and Stellar150 ventilators (all  $p < 0.01$ , rate of difference  $> 10\%$ ); there were no significant differences in PIP, PIE, MIE, or PEEP among the valves (Figure 4B). Finally, in terms of V<sub>T-L</sub>, the value was significantly lower when using the plateau valve with the Stellar150 ventilator than any other combination (all  $p < 0.01$ , rate of difference  $> 10\%$ ) (Figure 4C).

Except for the V60 ventilator, the rate of difference in V<sub>T</sub> between the ventilator and the simulated lung exceeded 10% for all exhalation valve types. The rate of difference in PIP remained below 10% under all conditions (Table 1).

## Discussion

We comprehensively assessed the effects of four types of exhalation valve on CO<sub>2</sub> rebreathing and ventilator performance under different NIV settings and ventilators. Although no significant differences in CO<sub>2</sub> rebreathing were observed among the single-arch, whisper swivel, and plateau valves, all showed significantly higher rebreathing volumes than the mask valve. Additionally, except for the V60 ventilator, some ventilator performance parameters were significantly lower for the plateau valve than for the others, and the rate of difference in V<sub>T</sub> between the ventilator and the simulated lung with all exhalation valves exceeded 10%.

NIV can be delivered via a single-limb circuit, requiring a valve to enable exhalation. Clinically, four types of such valve are commonly used, depending on preference and equipment availability. However, their structural and functional characteristics differ. Previous study has indicated that leakage from the single-arch, whisper swivel, and mask valves increases with pressure, whereas the plateau valve maintains relatively constant leakage across different pressure levels due to its unique silicon diaphragm structure (7). Our study

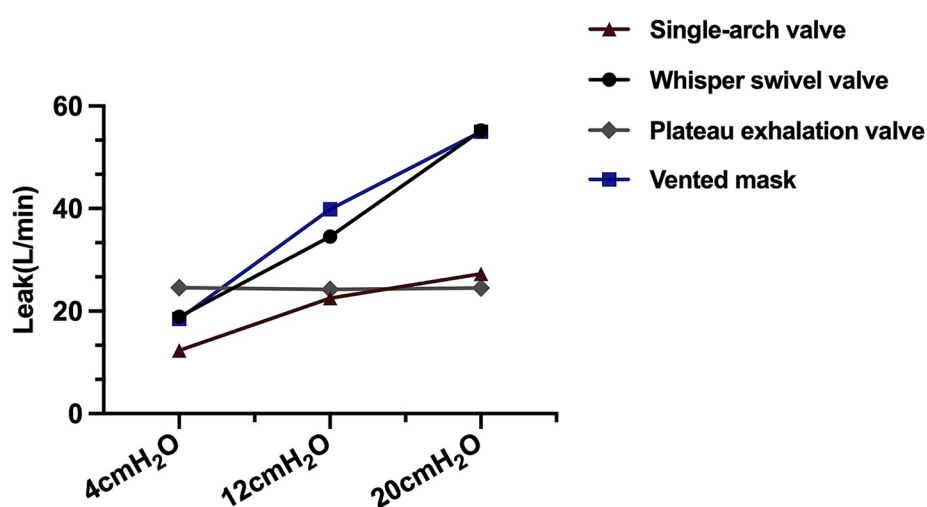


FIGURE 2  
The leakage of different exhalation valves.

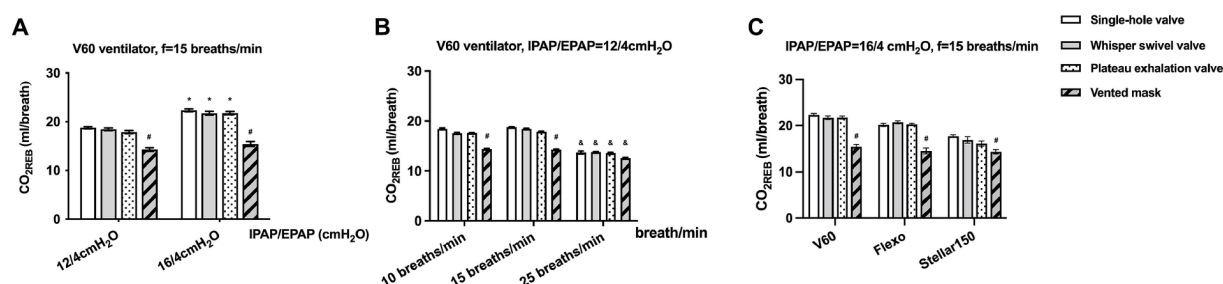


FIGURE 3

The  $\text{CO}_{2\text{REB}}$  of different exhalation valves under different conditions. (A) Comparison under different pressure settings (IPAP/EPAP = 12/4 and 16/4  $\text{cmH}_2\text{O}$ ); (B) Comparison at different breathing frequencies (10, 15, and 25 breaths/min); (C) Comparison across different ventilator types (V60, Flexo, and Stellar150). \* Significantly higher than IPAP = 12  $\text{cmH}_2\text{O}$  ( $p < 0.01$ ), # significantly lower than other exhalation valves ( $p < 0.01$ ), & significantly lower than  $f = 10/15$  breaths/min ( $p < 0.01$ ).

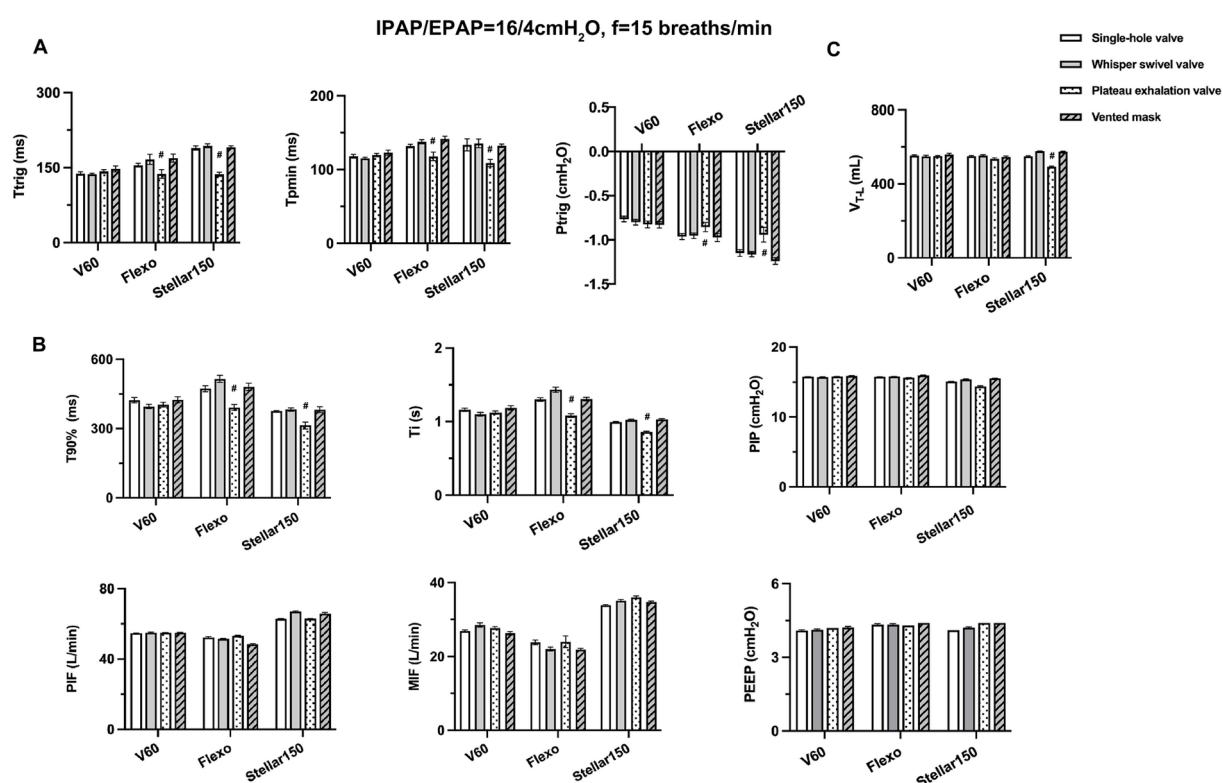


FIGURE 4

Effects of different exhalation valves on ventilator performance. (A) Triggering parameters (Ttrig, Tpm and Ptrig); (B) control parameters (T90%, Ti, PIP, MIF and PEEP); (C) tidal volume displayed on the lung simulator ( $V_{T-L}$ ). # significantly lower than other exhalation valves ( $p < 0.01$ ).

corroborates these findings, and further comprehensively analyzed their impact on  $\text{CO}_2$  rebreathing and ventilator performance.

Both bench and clinical studies (14–18) have demonstrated that  $\text{CO}_2$  may not be adequately cleared during NIV, for an additional rebreathing process occurred and exhaled gasses constituted a fraction of the delivered tidal volume. Various factors—including patient-associated characteristics (e.g., baseline  $\text{CO}_2$  levels, spontaneous breathing patterns), ventilator settings (e.g., IPAP, EPAP), and interface-related elements (e.g., exhalation valve type and position)—have been reported to affect  $\text{CO}_2$  rebreathing during NIV (15–18). The simulated COPD lung model, which replicates key pathophysiological features of hypercapnic COPD such as increased dead space, has been

widely used in previous bench studies (9, 11) to evaluate  $\text{CO}_2$  clearance efficacy. Similarly, maintaining a simulated  $\text{PetCO}_2$  of 80 mmHg—consistent with our previous approach (9)—provides a standardized hypercapnic condition for assessing the impact of different exhalation valves on rebreathing. As in prior methodology (15), the calculated  $\text{CO}_2$  rebreathing represented the total rebreathing volume from the ventilator circuit, mask, and upper airway dead space of the mannequin, and not solely the  $\text{CO}_2$  rebreathed from the circuit. Besides, previous studies also suggested that  $\text{CO}_2$  rebreathing can be minimized by using an EPAP of up to 8  $\text{cmH}_2\text{O}$  or an IPAP of up to 20  $\text{cmH}_2\text{O}$  (16). Therefore, to eliminate ventilator parameter influences while accounting for clinical constraints (specifically, the 4



TABLE 1 The difference rate of tidal volume ( $V_t$ , mL) and peak inspiratory pressure (PIP, cmH<sub>2</sub>O) between the ventilator and simulated lung with various exhalation valve types.

	V60			Flexo			Stellar150		
	$V_{t-L}$	$V_{t-V}$	the difference rate,%	$V_{t-L}$	$V_{t-V}$	the difference rate,%	$V_{t-L}$	$V_{t-V}$	the difference rate,%
The difference rate of $V_t$ (mL) between ventilator and simulated lung	553.53±4.34	595.40±2.84	7.57±0.79	549.70±4.43	819.01±14.86	49.0±2.87	548.58±4.01	611.15±5.35	11.41±0.44
	547.76±6.99	592.90±3.03	8.25±1.34	553.09±4.85	826.7±6.25	49.48±1.67	577.02±2.61	631.4±3.72	24.59±0.91
	550.40±3.40	599.10±2.73	8.85±0.86	535.27±5.18	829.1±10.51	54.91±2.47	492.14±3.45	394.63±103.5	-19.7±0.81
	558.56±6.49	600.50±4.06	7.52±1.36	546.06±4.83	932.9±4.04	70.86±1.70	573.56±2.66	702.29±0.8	22.44±0.5
	V60			Flexo			Stellar150		
	PIP <sub>-L</sub>	PIP <sub>-V</sub>	the difference rate,%	PIP <sub>-L</sub>	PIP <sub>-V</sub>	the difference rate,%	PIP <sub>-L</sub>	PIP <sub>-V</sub>	the difference rate,%
The difference rate of PIP (cmH <sub>2</sub> O) between ventilator and simulated lung	15.78±0.02	16.0	1.38±0.12	15.77±0.02	16.0	1.58±0.13	15.10±0.04	16.0	6.11±0.32
	15.73±0.03	16.0	1.69±0.22	15.82±0.02	16.0	1.30±0.14	15.39±0.06	16.0	4.00±0.40
	15.79±0.02	16.0	1.36±0.13	15.60±0.02	16.0	2.31±0.16	14.39±0.10	16.0	6.23±0.74
	15.88±0.04	16.0	0.73±0.29	15.96±0.02	16.0	0.21±0.15	15.50±0.03	16.0	3.20±0.20

cmH<sub>2</sub>O minimum EPAP setting on most devices), we applied IPAP levels of 12 or 16 cmH<sub>2</sub>O with a fixed EPAP of 4 cmH<sub>2</sub>O in this bench study. This design enables isolated investigation of exhalation valve effects on CO<sub>2</sub> rebreathing within an *in vitro* COPD model.

We found that CO<sub>2</sub> rebreathing with the mask valve was significantly lower than with the single-arch, whisper swivel, and plateau valves. This aligns with Schettino et al. (15), who demonstrated that exhalation port position influences CO<sub>2</sub> rebreathing. Compared to circuit-located leak ports, the mask-integrated exhalation port (as in the Facial-MEP design) enhances residual gas washout by fresh airflow, reducing dead space ventilation and rebreathing volume. Meanwhile, our previous work confirms substantial design-dependent variation in CO<sub>2</sub> rebreathing among oronasal masks with integrated ports, with mean values ranging 11.01–16.78 mL (9). Notably, all integrated-port masks exhibited lower rebreathing volumes than single-arch, whisper swivel, and plateau valves in this study.

Among the circuit-located exhalation ports, the plateau valve is generally believed to facilitate lower CO<sub>2</sub> rebreathing due to its unique controlled leakage mechanism (16). However, no significant differences in CO<sub>2</sub> rebreathing were observed among the single-arch, plateau, and whisper swivel valves in our study. This aligns with Hill et al.'s *in vivo* findings (19), where plateau valves showed no clinical advantage over whisper swivel valves in long-term NIV patients. This null effect likely relates to leakage characteristics. Except the minimum EPAP (4 cmH<sub>2</sub>O), plateau valves demonstrated equivalent leakage to whisper swivel and single-arch valves at both 12 and 20 cmH<sub>2</sub>O (Figure 2). Consequently, at our tested IPAP levels (12/16 cmH<sub>2</sub>O) with fixed EPAP (4 cmH<sub>2</sub>O), plateau valves conferred no additional rebreathing reduction. Notably, Ferguson and Gilmartin (16) observed superior performance of plateau valves at EPAP = 0 cmH<sub>2</sub>O, with valve differences disappearing at higher EPAP levels. Thus, at clinically relevant pressures (>4 cmH<sub>2</sub>O EPAP), rebreathing differences among these circuit-located valves become negligible. Further integrated bench-to-bedside studies are warranted to define valve selection criteria across therapeutic pressure ranges.

In terms of ventilator performance, we found that the plateau valve differed from the others. Triggering parameters (T<sub>trig</sub>, T<sub>pmin</sub>, P<sub>trig</sub>) and control parameters (T<sub>90%</sub>, T<sub>i</sub>) were significantly lower than for the others across the tested ventilators, except for the V60 ventilator. This may be due to the constant leakage of the plateau valve, and there is a greater leakage during the expiratory phase when EPAP is relatively low. This greater leakage may cause the ventilator to mistakenly perceive that the patient starting to inhale, thereby prematurely triggering the ventilator to deliver gas to the patient, resulting in a decrease in triggering parameters. Additionally, the relatively low leakage with the plateau valve at higher IPAP settings (16 cmH<sub>2</sub>O) may cause the ventilator to switch to expiration earlier, resulting in shorter inspiratory times and reduced tidal volumes. Therefore, when selecting exhalation valves in clinical practice, it is crucial to consider their impact on triggering and control performance during NIV.

Furthermore, we found that the rate of difference in  $V_T$  between the ventilator and the simulated lung exceeded 10% for all exhalation valve types when using the Flexo and Stellar150 ventilators but not the V60 ventilator. During NIV, the monitored tidal volume displayed on the ventilator is an estimated value. Luján et al. (20) reported that different valve types have different leakage, which can affect the accuracy of ventilator parameter monitoring during NIV. To be able to display accurate tidal and minute volumes, the ventilator must know the intentional leak characteristics of the specific mask/patient interface

and exhalation port. With V60 ventilator, select the desired exhalation port type or/and run an exhalation port test is needed before the ventilator starts working. We previously found that tidal volume monitoring was more accurate after leak tests when using certain industrial masks (13). These findings suggest that exhalation valves may not be interchangeable among ventilators with single-limb circuits. If interchangeable, leak tests of the valves from different manufacturers may be required.

Our *in vitro* findings yield some clinical and research implications. Consistent with prior studies, mask-integrated valve demonstrated significantly lower CO<sub>2</sub> rebreathing than circuit-located valves (single-arch, whisper swivel, plateau) during NIV, aligning with our previous work confirming superior CO<sub>2</sub> clearance across oronasal mask valves. Crucially, the plateau valve—despite its controlled-leak mechanism—showed no rebreathing reduction at clinically relevant settings, this provides actionable guidance for optimizing NIV trial protocols and clinical implementation. Furthermore, valve-dependent performance variations revealed some imperatives: rigorous monitoring of ventilator parameters (trigger sensitivity, control, and ventilation monitoring) during exhalation valves selection; development of ventilators with real-time intentional leak monitoring to optimize ventilator performance; and future studies also need to validate these findings across different patient populations (e.g., restrictive lung diseases) through bench-to-bedside investigations.

Several methodological limitations warrant consideration. This *in vitro* analysis employed a simulated COPD model with fixed hypercapnia (PetCO<sub>2</sub> 80 mmHg) and restricted ventilator pressure settings, thereby failing to capture physiological variations in respiratory patterns, metabolic CO<sub>2</sub> fluctuations, or compensatory mechanisms in chronic respiratory failure. The restriction to 12/4 and 16/4 cmH<sub>2</sub>O pressure settings—while clinically relevant—precludes evaluation across the full therapeutic spectrum, and the absence of human variables (e.g., airway humidity, spontaneous breathing variability) limits clinical extrapolation. Though providing rigorously controlled conditions for quantifying CO<sub>2</sub> rebreathing dynamics, these constraints necessitate: (1) expanded pathophysiological models (e.g., restrictive disease); (2) systematic valve evaluation across incremental pressures (e.g., IPAP 8–25 cmH<sub>2</sub>O; EPAP 4–12 cmH<sub>2</sub>O); and (3) prioritized clinical validation measuring ventilator performance parameters, and gas exchange efficacy in target populations.

## Conclusion

Our bench study demonstrated that different exhalation valves significantly affect CO<sub>2</sub> rebreathing and ventilator performance during NIV. CO<sub>2</sub> rebreathing was significantly lower with the mask valve than with the single-arch, whisper swivel, and plateau valves during NIV. Except for the V60 ventilator, ventilator performance parameters of the plateau valve differed from those of the other valves; additionally, the rate of difference in V<sub>T</sub> between the ventilator and the simulated lung exceeded 10% for all exhalation valves. Further clinical studies are needed to validate these results *in vivo*, taking into account patient variability in respiratory mechanics, ventilator settings, and CO<sub>2</sub> production.

## Data availability statement

The raw data supporting the conclusions of this article will be made available by the authors, without undue reservation.

## Author contributions

XinL: Conceptualization, Data curation, Formal analysis, Investigation, Methodology, Resources, Software, Writing – original draft. BD: Software, Supervision, Writing – review & editing. CZ: Investigation, Methodology, Writing – original draft. HH: Data curation, Formal analysis, Writing – original draft. SW: Software, Validation, Writing – original draft. XiaL: Validation, Visualization, Writing – original draft. HZ: Methodology, Supervision, Writing – review & editing. WW: Supervision, Validation, Writing – review & editing. WT: Formal analysis, Funding acquisition, Project administration, Software, Supervision, Visualization, Writing – review & editing.

## Funding

The author(s) declare that financial support was received for the research and/or publication of this article. This research was funded by the China Postdoctoral Science Foundation (Certificate number: 2023M733905).

## Acknowledgments

The laboratory of respiratory medicine of the first affiliated hospital of China medical university gave the necessary support.

## Conflict of interest

CZ was employed by Shenyang RMS Medical Tech Co., Ltd.

The remaining authors declare that the research was conducted in the absence of any commercial or financial relationships that could be construed as a potential conflict of interest.

## Generative AI statement

The authors declare that no Gen AI was used in the creation of this manuscript.

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