The Effect of Filters on CPAP Delivery by Helmet

Daniele Privitera, Nicolò Capsoni, Francesco Zadek, Paolo Vailati, Chiara Airoldi, Mattia Cozzi, Federico Pierotti, Roberto Fumagalli, Andrea Bellone, and Thomas Langer

BACKGROUND: When helmet CPAP is performed using a Venturi system, filters are frequently interposed in the respiratory circuit to reduce noise within the helmet. The effect of the interposition of these filters on delivered fresh gas flow and the resulting F_{IO_2} is currently unknown. METHODS: In a bench study, 2 different Venturi systems (WhisperFlow and Harol) were used to generate 3 different gas flow/F_{IO_2} combinations (80 L/min-F_{IO_2} 0.6, 100 L/min-F_{IO_2} 0.5, 120 L/min-F_{IO_2} 0.4). Different combinations of filters were applied at the flow generator input line and/or at the helmet inlet port. Two types of filters were used for this purpose: a heat and moisture exchanger filter and an electrostatic filter. The setup without filters was used as baseline. Gas flow and F_{IO_2} were measured for each setup. RESULTS: Compared to baseline, the interposition of filters reduced the gas flow between 1–13% ($P < .001$). The application of a filter at the Venturi system or at the helmet generated a comparable flow reduction ($-3 \pm 2\%$ vs $-4 \pm 2\%$, $P = .12$), whereas a greater flow reduction ($-7 \pm 4\%$) was observed when filters were applied at both sites ($P < .001$). An increase in F_{IO_2} up to 5% was observed with filters applied. A strong inverse linear relationship ($P < .001$) was observed between the resulting gas flow and F_{IO_2}. CONCLUSIONS: The use of filters during helmet CPAP reduced the flow delivered to the helmet and, consequently, modified F_{IO_2}. If filters are applied, an adequate gas flow should be administered to guarantee a constant CPAP during the entire respiratory cycle and avoid rebreathing. Moreover, it might be important to measure the effective F_{IO_2} delivered to the patient to guarantee a precise assessment of oxygenation. *Key words: CPAP; noninvasive ventilation; respiratory insufficiency; noise; emergency department*. [Respir Care 2022;67(8):995–1001.

Introduction

CPAP is widely used in the acute care setting for the treatment of hypoxemic respiratory failure due to acute cardiogenic pulmonary edema[1-3] and pneumonia, including viral pneumonia due to COVID-19.[4-6] Frequently, CPAP is delivered through a helmet.[7] The helmet is broadly used in southern Europe and particularly in Italy, mainly for the treatment of hypoxemic respiratory failure and acute cardiogenic pulmonary edema.[7] To perform helmet CPAP, a continuous flow generator is used to provide a fresh gas flow.[8] This gas flow is delivered inside the helmet and generates a PEEP as it flows through an expiratory valve.[9] Three main types of flow generators are commonly used: multiple columns, gas blenders, and Venturi system flow generators.[9] All these systems are able to deliver continuous gas flows at high rates. These high gas flows are required for 2 main reasons: first, to exceed the patient's peak inspiratory flow, ensuring a stable CPAP throughout the entire respiratory cycle; second, to avoid CO_2 rebreathing.[10] In most cases, 60 L/min is considered an adequate gas flow to perform helmet CPAP.[8,11] However, higher flows may be required in case of severe respiratory distress with tachypnea, large tidal volumes (V_T), and high minute ventilation.[12]

Different kinds of PEEP valves are commercially available: water-sealed valves, precalibrated fixed PEEP valves, and adjustable PEEP valves.[9] Among these, adjustable valves have shown a variable degree of flow dependence, potentially leading to higher-than-expected continuous airway pressures.[9] On the other hand, precalibrated and water-sealed valves exhibit a better performance: Precalibrated valves have a flow-independent behavior, whereas water-sealed valves have a low degree of flow dependence.[9]

In studies comparing the effectiveness of face masks and helmets, pressure injuries are the main complications of noninvasive ventilation delivered with the former,[13] whereas the latter interface can avoid this problem, as there is no direct contact between the helmet and the patient's face. However, the use of helmets is characterized by higher noise levels, potentially limiting patient's comfort.[14-16] In a study on healthy volunteers, Lucchini et al demonstrated that applying a heat and moisture exchanger filter (HMEF) on the inlet gas port of helmet CPAP significantly reduced the noise level recorded inside the helmet, improving subjects' comfort.[16] However, while reducing the noise within the helmet, HMEFs add resistance to the respiratory circuits, possibly affecting the delivered fresh gas flow.[17] Data about flow variations due to the use of a heat and moisture exchanger and electrostatic filters positioned either at the flow generator input line (as indicated by the manufacturer) and/or at the inlet port of the helmet are currently lacking.

Therefore, the present bench study aimed to assess the impact of the application of filters during helmet CPAP performed using Venturi systems on (1) delivered fresh gas flow, (2) F_{IO_2}, and (3) the noise level both inside and outside the helmet.

QUICK LOOK

Current knowledge

Helmet CPAP is usually performed using a fresh gas flow (> 60 L/min) provided through a continuous flow generator. Respiratory filters (usually positioned at the inlet gas port of the helmet) are frequently employed to reduce the noise level inside the helmet, thus improving patient comfort.

What this paper contributes to our knowledge

When helmet CPAP was provided using a Venturi flow generator, the application of filters within the respiratory circuit reduced the delivered fresh gas flow. This was associated with an increased F_{IO_2}. In this setting, if filters are applied, an adequate gas flow should be guaranteed. Moreover, it might be important to measure the effective F_{IO_2} to assess oxygenation precisely.

Methods

The present bench study was performed at the ASST Grande Ospedale Metropolitano Niguarda, Milan, Italy. A half-body manikin (size medium) was used for the study (Fig. 1). A certified, medium-sized CPAP helmet (DIMAR, Mirandola, Italy) was sealed with standard armpit straps. A fixed precalibrated 10 cm H_2O PEEP valve (Harol, San Donato Milanese [Milan], Italy) was applied to the expiratory port of the helmet. A 150-cm respiratory circuit for adults with a smooth inner surface was used (285/5063, DAR, Medtronic, Mansfield, Massachusetts). Two different Venturi systems were used for flow generation (9293/D, Harol; and WhisperFlow, Philips Respironics, Murrysville, Pennsylvania). The flow generators underwent testing and calibration according to the manufacturers' specifications. These devices require a single source of oxygen to generate a fresh gas flow. They use the Venturi effect to drag room air into the circuit and generate a high gas output flow. A supplementary oxygen source located downstream from the air-entrainment valve can be connected to increase F_{IO_2}. According to the selected oxygen flow on the main flow meter, the generator takes a variable amount of air from the environment, which, added to the oxygen flow, determines the total flow of fresh gas delivered to the patient (between 0−180 L/min). By acting on the second oxygen flow meter, F_{IO_2} can be adjusted between 0.3−1.0.

An electronic flow meter (TSI 4000 Series, TSI, Shoreview, Minnesota), able to measure flows between 0–200 L/min, and a rapid-response oxygen analyzer (Handi+, Maxtec, Salt Lake City, Utah) were positioned between the circuit and the CPAP inlet port. Two sound level meters (MK09, 30–130 dB, 31.5–8 KHz, Meterk, Shenzhen, China) were used: The first was placed on the right ear of the manikin, close to the helmet inlet port to measure the noise level within the helmet; the second was positioned 1 m away from the flow generator to simultaneously measure the environmental noise. The oxygen pipeline supply pressure to the flow meter was > 4 bar, that is, adequate for the correct functioning of the flow generators.

Experimental Setup

For each Venturi system, 3 combinations of fresh gas flow and F_{IO_2} were tested: 80 L/min with F_{IO_2} 0.6, 100 L/min with F_{IO_2} 0.5, and 120 L/min with F_{IO_2} 0.4. Flows of the main and supplementary oxygen sources were adjusted to reach the desired fresh gas flows and F_{IO_2}, measured with the electronic flow meter and the oxygen analyzer, respectively.

To test the impact on flow, F_{IO_2} and noise, 2 different types of filters were used: an HMEF (labeled H) (V_T 150–1,000 mL) and an electrostatic filter (labeled E) (V_T 300–1,500 mL), both from the same manufacturer (DAR,

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The authors have disclosed no conflicts of interest.

Supplementary material related to this paper is available at http://www.rcjournal.com.

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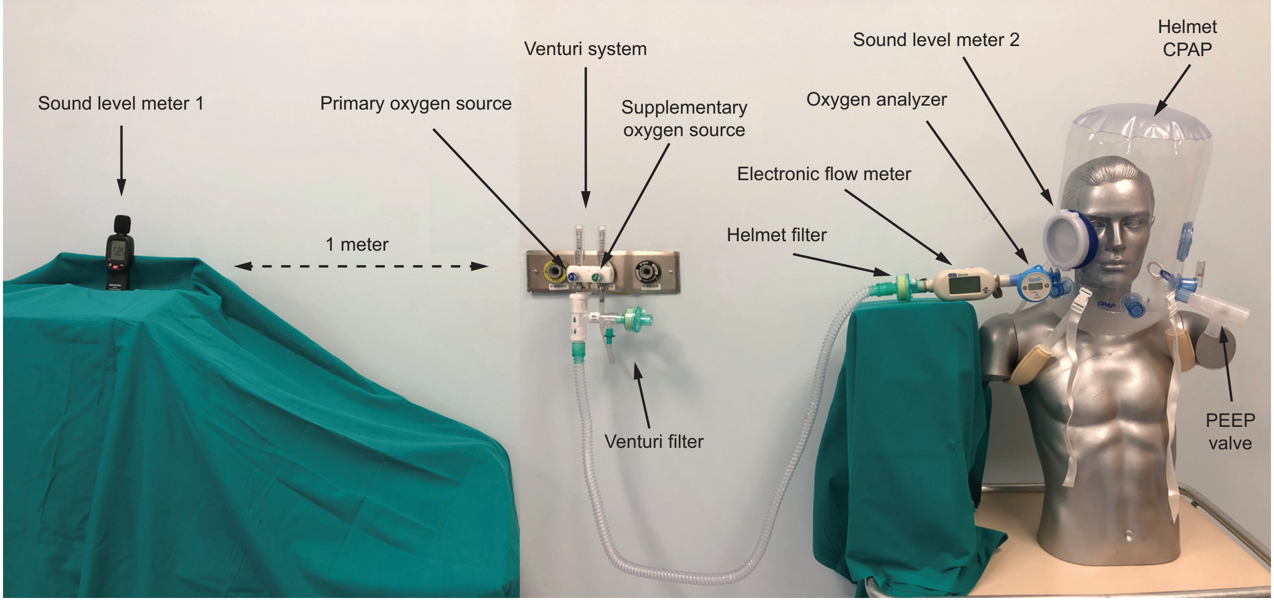

Fig. 1. Experimental setup.

Medtronic). The filters were applied at the flow generator input line (labeled Venturi), at the helmet inlet port (labeled helmet), or at both sites (labeled both), obtaining the different setups reported in Table 1. The setups without the interposition of filters served as reference and were labeled as baseline.[18]

Measurements

Before initiation of the experiments, environmental noise was measured using the sound meter placed 1 m away from the flow generator. For each Venturi system, gas flow/ F_{IO_2} couple, and filter setup, the fresh gas flow was recorded in triplicates. The average value of 3 measurements performed 30 s apart was used for analysis. The F_{IO_2} was acquired after the measured values were stable for 30 s. The maximum noise level was recorded simultaneously inside and outside the helmet. The highest value measured was collected.

Table 1. Experimental Setups

Venturi System	Flow/F_{IO_2}	Filter Combinations
Harol	80 L/min - F_{IO_2} 0.6	0/0 (Baseline)
WhisperFlow	100 L/min - F_{IO_2} 0.5	E/0 (Setup 1)
	120 L/min - F_{IO_2} 0.4	E/E (Setup 2)
		E/H (Setup 3)
		0/E (Setup 4)
		0/H (Setup 5)
		H/0 (Setup 6)
		H/E (Setup 7)
		H/H (Setup 8)

Table 1 summarizes the 2 Venturi systems, the 3 flow/F_{IO_2} couples, and the 9 different filter combinations tested. The different positions of the filters were expressed as flow generator input filter/helmet filter, coding the filter type as follows: 0 = no filter; H = heat and moisture exchanger filter; E = electrostatic filter.

Statistical Analysis

Data from the different gas flow/F_{IO_2} combinations were pooled to analyze factors associated with the reduction in fresh gas flow. Continuous variables are expressed as mean ± SD or as median (interquartile range) according to their distribution. For each setup, the mean of repeated triplicate of flow measurements was used for analysis. The repeatability coefficient of measured flows was calculated by multiplying the within-subject SD by 2.77 as previously reported.[19,20] Percentage variations from baseline values of fresh gas flow and noise were calculated. F_{IO_2} variations are indicated as an absolute percentage change. The difference in flow reduction according to the applied filter (H or E) and employed Venturi system (WhisperFlow or Harol) was assessed using the pairwise *t* test or Wilcoxon signed-rank test as appropriate. Pearson correlation coefficient was employed to assess the strength of association between flow and resulting F_{IO_2}. A 2-way analysis of variance (ANOVA) was performed considering as outcome variable flow variation or sound level and as factors both the site of filter interposition (Baseline, Venturi, Helmet, Both) and the type of Venturi system (WhisperFlow or Harol). Bonferroni correction was used for post hoc pairwise comparisons. Statistical significance was defined as $P < .05$. Analysis was performed with SigmaPlot v.12.0 (Systat Software, San Jose, California).

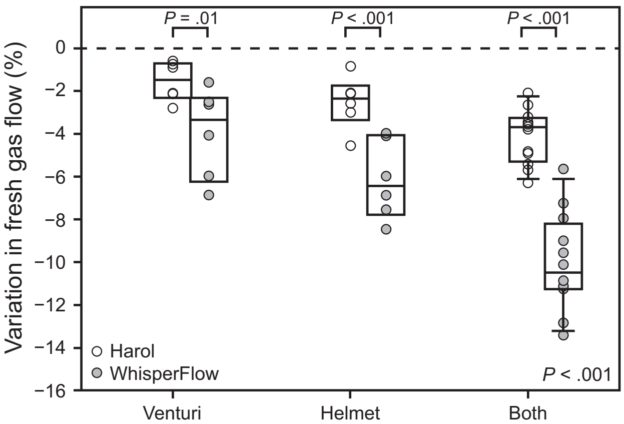

Fig. 2. Box and dot plot representation of percentage variations in gas flow resulting from the interposition of respiratory filters at the Venturi system before the helmet or at both sites (no. = 54; 48 points are presented + 6 baseline values). The horizontal dashed line represents 0% variation, that is, the baseline reference value without the interposition of filters. $P < .001$ refers to the P value of the 2-way analysis of variance.

Results

The interposition of filter(s) within the circuit determined a reduction of the total flow of fresh gas. The entity of this reduction ranged between 1–13% of baseline fresh gas flow. The repeatability coefficient of measured flows was ± 0.49 L/min.

Data from the different couples of gas flow/F_{IO_2} were pooled to analyze the role of presence/site of filter interposition (Baseline, Venturi, Helmet, or Both), type of used Venturi system (Harol or WhisperFlow), and their interaction on the reduction of gas flow (Fig. 2). As compared to baseline in all studied conditions (Venturi, Helmet, and Both), a significant reduction in fresh gas flow was recorded ($P < .001$). The application of a filter at the Venturi system or at the helmet generated a comparable flow reduction (−3 ± 2% vs −4 ± 2%, respectively, $P = .12$). However, the interposition of filters at both sites caused a significantly greater reduction in delivered fresh gas flow as compared to both the application of a filter at the Venturi system (−7 ± 4% vs −3 ± 2%, $P < .001$) and at the helmet (−7 ± 4% vs −4 ± 2%, $P < .001$). The flow reduction observed with Harol was significantly lower the one recorded when using the WhisperFlow both overall (−3 ± 2% vs −7 ± 3%, $P < .001$) and for each site of filter interposition (Venturi $P = .013$, Helmet $P < .001$, Both $P < .001$) (Fig. 2). Overall, the effect of the interposition of filters was more pronounced on the WhisperFlow system ($P < .001$, 2-way ANOVA interaction). To evaluate if the type of filter had a role in flow reduction, we compared flow reductions in the settings where either H (setups 1, 2, and 4) or E (setups 5, 6, and 8) filters were used. We excluded from this analysis setups 3 and 7, in which a combination of H and E filters was used. The observed difference regarding the reduction of fresh gas flow was −6 ± 3% versus −4 ± 3%, $P = .060$ for H and E, respectively.

The interposition of filters had a significant effect also on the resulting F_{IO_2}. In particular, for all 3 setups, a strong negative linear relationship was observed between the resulting fresh gas flow and the increase in F_{IO_2} (Fig. 3).

Regarding noise levels inside the helmet, Harol generated, overall, a significantly higher noise than the WhisperFlow system (82 ± 4 dB vs 76 ± 3 dB, $P < .001$). The application of a filter at the inlet line of the flow generator (Venturi) had no effect on the noise within the helmet, as compared to baseline (Fig. 4). The interposition of a filter at the helmet significantly reduced the noise within it compared to baseline (78 ± 4 dB vs 82 ± 5 dB, $P = .008$). Finally, the interposition of filters at both sites resulted in a similar noise level as the single filter at the helmet inlet but was significantly lower as compared to both baseline and the single filter applied at the flow generator (Venturi) ($P < .001$ for both). Overall, the effect of the interposition of filters on noise within the helmet

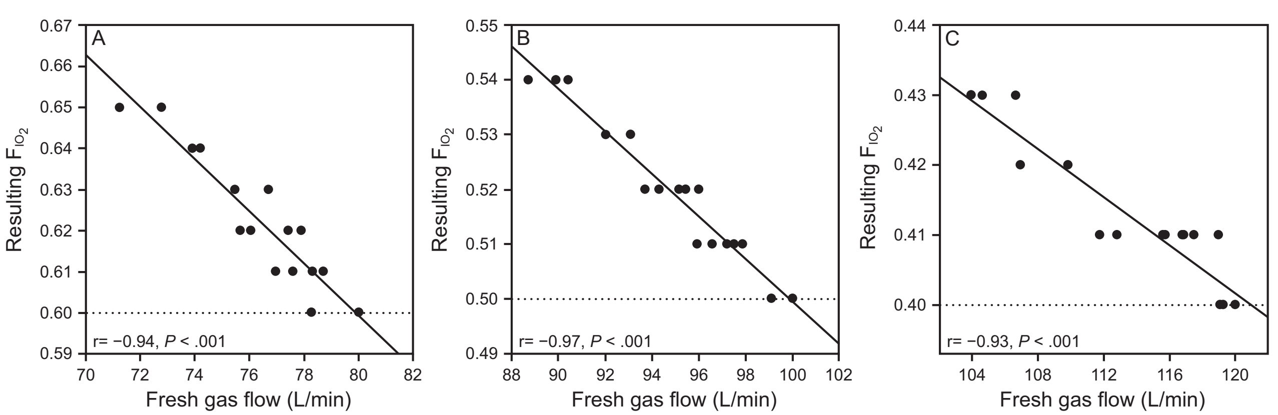

Fig. 3. Fresh gas flow to resulting F_{IO_2} relationship in the 3 experimental settings. A: Gas flow 80 L/min-F_{IO_2} 0.6; B: gas flow 100 L/min-F_{IO_2} 0.5; and C: gas flow 120 L/min-F_{IO_2} 0.4. Horizontal dashed lines represent baseline F_{IO_2}. No. = 18 for each experimental setting.

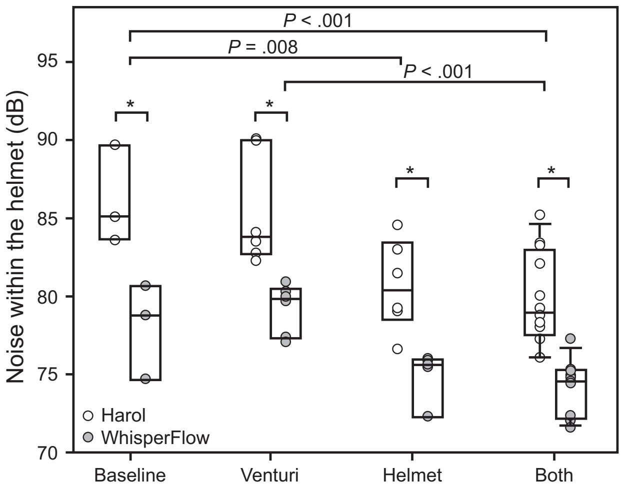

Fig. 4. Box and dot plot representation of absolute noise levels within the helmet at baseline (without the interposition of filters) and after the interposition of filters at the flow generation system (Venturi), at the inlet port of the helmet, and at both sites (no. = 54). Data are presented separately according to the used Venturi system. A 2-way analysis of variance was performed ($P = .80$)* $P < .001$ versus WhisperFlow.

was similar for the 2 flow generation systems ($P = .80$, 2-way ANOVA interaction).

Before starting the test, the environmental noise inside the room ranged between 39.5–41.0 dB. During the test, baseline values, ie, values without the interposition of filters, were 74 ± 2 dB and 61 ± 1 dB for Harol and WhisperFlow, respectively, ($P < .001$) (Figure E1, see related supplementary materials at http://www.rcjournal.com). The application of a filter at the Venturi system significantly reduced the environmental noise as compared to baseline (61 ± 7 dB vs 67 ± 7 dB, $P < .001$), whereas no effect was recorded when the filter was applied to the helmet (67 ± 7 dB vs 67 ± 7 dB, $P > .99$). Finally, the application of filters at both sites had no relevant additional muffling effect.

Values of fresh gas flow, F_{IO_2}, and noise resulting from the interposition of the filters for the 3 gas flow/F_{IO_2} couples are reported in Tables E1, E2, and E3 (See related supplementary materials at http://www.rcjournal.com) of the supplementary materials.

Discussion

Our main finding is that the interposition of filters, which are frequently applied in clinical practice, significantly reduced the delivered fresh gas flow. The most relevant flow reduction was observed when 2 filters were applied at the same time. Due to the application of filters, F_{IO_2} increased, and the increase was linear with flow reduction. Finally, although applying an additional filter to the Venturi system did not significantly reduce the noise within the helmet, it did decrease the environmental noise.

HMEFs are usually interposed in the respiratory circuit to heat and humidify air entering the airways when the upper airways are bypassed, such as during invasive mechanical ventilation. In addition, electrostatic filters are frequently interposed within the breathing circuit with the aim of reducing the transmission of microbes.[21] In the context of helmet CPAP performed with a Venturi system, filters are employed to muffle the noise within the helmet to increase the patient's comfort.[16] However, the effect of these filters on delivered fresh gas flow was unknown.

We observed a consistent reduction in delivered fresh gas flow applying filters within the respiratory circuit. The largest flow reduction was observed when filters were applied both at the Venturi system and at the helmet inlet (Fig. 2). Of note, the reduction differed according to the flow generation system used, with greater reductions observed when the WhisperFlow was employed. Furthermore, the type of filter seems to have a role in the flow reduction since larger drops in the flow were observed when HMEFs were used. It is conceivable that HMEFs generate greater resistance due to their hygroscopic membrane (absent in electrostatic filters).[17,22,23]

In addition to the effect on delivered gas flow, we observed that the application of filters within the respiratory circuit caused a slight increase in delivered F_{IO_2}. The variations were not particularly marked, as they reached a maximum increase in F_{IO_2} of 5%. Interestingly, in all settings, F_{IO_2} increased linearly with the reduction in the delivered fresh gas flow (Fig. 3). This finding might be explained by a modification of the gas mixture. For instance, both Harol and WhisperFlow systems use the negative pressure developed by the Venturi effect to drag room air into the system. The use of filters has the net effect of increasing the resistance to flow, reducing the negative pressure generated by the Venturi effect, and, ultimately, limiting the entrance of room air (F_{IO_2} of 0.21) in the circuit. On the contrary, the amount of oxygen delivered to the system is constant, justifying the increase in F_{IO_2}.

Finally, we studied noise within the helmet and in the environment. Noise is often a cause of patients' intolerance to CPAP and of therapeutic failure.[16] The 2 Venturi systems employed in our study generated significantly different noise levels (Fig. 4), with the Harol system characterized by higher decibels. In accordance with the literature,[14,16] we observed that the application of a filter at the helmet inlet port determined a marked reduction of the noise inside it. On the contrary, the application of a filter only on the flow generator input line did not have this effect. Of note, the muffling intensity of respiratory filters was similar for the 2 studied flow generators ($P = .80$).

Environmental noise in the emergency department often exceeds the maximum of 40 dB recommended by the World Health Organization and is thus a serious problem for both patients and medical/nursing staff.[24-26] A noisy environment favors errors and is considered a risk factor for provider burnout and negative outcomes for patients.[27] We observed significant differences between the environmental noise generated by the 2 Venturi systems used in our bench study. Also in this case, the Harol system generated higher noise levels. The application of a filter at the flow generator input line significantly reduced the environmental noise but, as discussed above, contributed to the reduction in fresh gas flow.

Clinical Implications

In patients requiring respiratory support with helmet CPAP, two aspects need to be considered. On the one hand, an adequate flow inside the helmet is fundamental to exceed the patient's peak inspiratory flow, maintain a constant CPAP, avoid rebreathing, and thus optimize the respiratory support.[13,28,29] On the other hand, noise needs to be reduced to increase patients' comfort and tolerance of the respiratory support.[14,16] Our study shows that the 2 applied Venturi systems generated different noise levels. Moreover, it confirms that the application of filters within the respiratory circuit was effective in reducing noise within the helmet. However, when using Venturi flow generators, filters reduced the delivered fresh gas flow and thus caused a slight increase in F_{IO_2}. In this context, it is, therefore, important to be aware of these implications. Flow can easily be increased to ensure adequate values; F_{IO_2} can be measured to guarantee a precise assessment of oxygenation.

Limitations

The bench nature of our study has advantages and disadvantages. It allowed us to assess and compare precisely the effects of the interposition of filters on several outcome variables. However, we could not evaluate patient-centered outcomes, such as comfort and respiratory variables. Moreover, our model does not account for the potential impact on flow and F_{IO_2} of the patient's inspiratory effort.

Conclusions

The interposition of filters within the breathing circuit reduced the fresh gas flow delivered to the helmet. In addition, the interposition of filters slightly increased the effective F_{IO_2} linearly with gas flow reduction. When filters are interposed within the circuit to reduce the noise level, attention should be paid to guarantee an adequate fresh gas flow. Finally, resulting F_{IO_2} should be confirmed to avoid underestimations of the severity of hypoxemia.

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