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**Francesco Mojoli**
**Giorgio A. Iotti**
**Maddalena Gerletti**
**Carlo Lucarini**
**Antonio Braschi**

# Carbon dioxide rebreathing during non-invasive ventilation delivered by helmet: a bench study

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F. Mojoli (✉) · M. Gerletti · C. Lucarini · A. Braschi
Fondazione IRCCS Policlinico S. Matteo, Servizio di Anestesia e Rianimazione I, Piazzale Golgi 2, Pavia, Italy
e-mail: f.mojoli@smatteo.pv.it

F. Mojoli · A. Braschi
Università degli Studi di Pavia, Cattedra di Anestesiologia e Rianimazione, Pavia, Italy

G. A. Iotti
Fondazione IRCCS Policlinico S. Matteo, Servizio di Anestesia e Rianimazione II, Pavia, Italy

**Abstract** *Objective:* To define how to monitor and limit $CO_2$ rebreathing during helmet ventilation. *Design:* Physical model study. *Setting:* Laboratory in a university teaching hospital. *Interventions:* We applied pressure-control ventilation to a helmet mounted on a physical model. In series 1 we increased $CO_2$ production ($V'CO_2$) from 100 to 550 ml/min and compared mean inhaled $CO_2$ ($iCO_2$,mean) with end-inspiratory $CO_2$ at airway opening ($eiCO_2$), end-tidal $CO_2$ at Y-piece ($yCO_2$) and mean $CO_2$ inside the helmet ($hCO_2$). In series 2 we observed, at constant $V'CO_2$, effects on $CO_2$ rebreathing of inspiratory pressure, respiratory mechanics, the inflation of cushions inside the helmet and the addition of a flow-by. *Measurements and results:* In series 1, $iCO_2$,mean linearly related to $V'CO_2$. The best estimate of $CO_2$ rebreathing was provided by $hCO_2$: differences between $iCO_2$,mean and $hCO_2$, $yCO_2$ and $eiCO_2$ were $0.0 \pm 0.1$, $0.4 \pm 0.2$ and $-1.3 \pm 0.5\%$. In series 2, $hCO_2$ inversely related to the total ventilation (MVtotal) delivered to the helmet–patient unit. The increase in inspiratory pressure significantly increased MVtotal and lowered $hCO_2$. The low lung compliance halved the patient:helmet ventilation ratio but led to minor changes in MVtotal and $hCO_2$. Cushion inflation, although it decreased the helmet's internal volume by 33%, did not affect rebreathing. A 8-l/min flow-by effectively decreased $hCO_2$. *Conclusions:* During helmet ventilation, rebreathing can be assessed by measuring $hCO_2$ or $yCO_2$, but not $eiCO_2$. It is directly related to $V'CO_2$, inversely related to MVtotal and can be lowered by increasing inspiratory pressure or adding a flow-by.

**Keywords** Non-invasive ventilation · Helmet · Carbon dioxide · Rebreathing · Monitoring · Physical model

## Introduction

The head helmet is an alternative to the face mask for delivering both non-invasive continuous positive airway pressure (CPAP) [1–4] and non-invasive ventilation (NIV) [5–12]. This device is provided with a soft collar that ensures non-traumatic adhesion to the neck of the patient and good tightness, once connected to the ventilation system. In comparison to face masks, the helmet seems really advantageous in terms of patient comfort, tolerance and decreased skin lesions, thus improving feasibility, continuity and duration of non-invasive ventilatory assistance [3, 7, 8, 10–12].

On the other hand, some concern exists about carbon dioxide ($CO_2$) rebreathing during helmet application [9, 11, 13–16]: with every breath, the $CO_2$ expired by the patient does not completely leave the system, but partly dilutes within the internal volume of the helmet and is subsequently re-inhaled. It was recently demonstrated that, during CPAP delivered by helmet, inspired $CO_2$ is inversely related to the flow of fresh gas "washing" the internal volume of the device [13]. Therefore, it was suggested to use the helmet only with high continuous-flow CPAP systems and to monitor the inspiratory $CO_2$ concentration [13, 14].

Differently from CPAP, during NIV the fresh gas flow applied to the helmet is limited and typically intermittent, rather than continuous: this may worsen the $CO_2$ storage and rebreathing. On the other hand, the helmet responds to positive-pressure ventilation with a relatively high compliance, due to its large volume, soft collar and fixation with armpit straps. In this way only part of the volume delivered by the ventilator reaches the patient respiratory system, while another fraction intermittently distends the helmet. This "helmet ventilation" may improve the washout of $CO_2$ around the head of the patient, possibly limiting $CO_2$ rebreathing.

In a bench study, we analysed the mechanisms of $CO_2$ rebreathing during ventilation delivered by a low size helmet model specifically designed for NIV and provided with inflatable cushions to further decrease the internal volume. First, we looked for the most convenient method to monitor the inspiratory $CO_2$ concentration by sampling $CO_2$ at different sites. Then, we tested the effects on rebreathing of patient's $CO_2$ production and respiratory mechanics, ventilator setting, inflation of helmet cushions and application of a continuous "flow by" through the helmet. Finally, we verified the hypothesis that the inspiratory $CO_2$ concentration depends on just two factors: the sum of all flows passing through the helmet and the patient's $CO_2$ production.

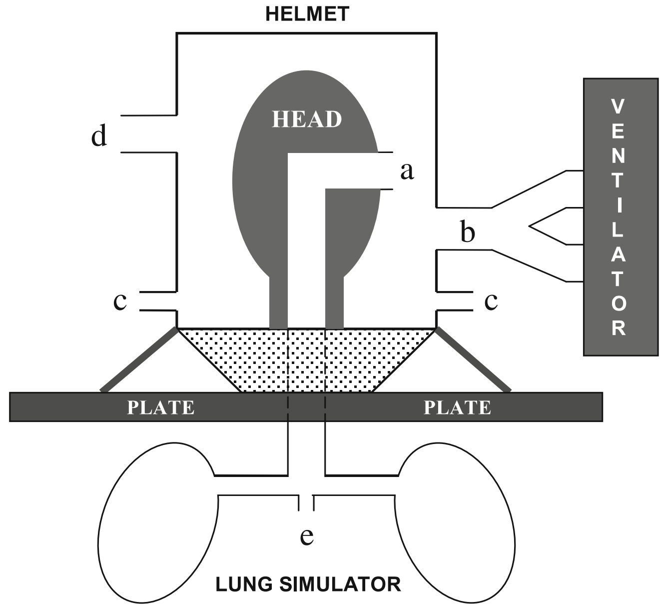


**Fig. 1** Physical lung model and experimental setup. The patient is simulated by an expanded polystyrene head provided with an internal proximal airway, connected to a mechanical lung simulator. The polystyrene head is connected to a plate. A NIV helmet is mounted on the head and fixed to the plate by four rigid straps. **a** Patient airway opening (site of measurement of $eiCO_2$, $etCO_2$ and V'insp). **b** 22-mm port for connection to the Y-piece of the ventilator circuit (site of measurement of $yCO_2$, $V_{t,i}$ and $V_{t,e}$). **c** Additional ports, used for flow-by inflow and outflow. **d** 22-mm port, used for helmet $CO_2$ ($hCO_2$) measurement. **e** $CO_2$ inflow for simulation of $CO_2$ production

## Materials and methods

A NIV helmet (CaStar R, Starmed, Mirandola, Italy) was connected to a physical model simulating a passive patient (Fig. 1). The model consisted of an expanded polystyrene head (about 3 l volume) connected to a plate that simulated the shoulders. The head was provided with a proximal airway that was connected, below the plate, to a passive mechanical lung simulator (Lungensimulator LS800; Dräger Medical, Lübeck, Germany) with adjustable respiratory system compliance and airway resistance. A bottle of pure $CO_2$ with a $CO_2$ flowmeter (RSP; Flowmeter, Wilson, OR, USA) was connected to a port in the lung model, for simulation of $CO_2$ production.

The helmet was placed around the head and secured to the plate by four straps. When the helmet was pressurized and ventilated, we could obtain a sealed connection between the helmet soft collar and the plate. As in the clinical setting, the helmet was able to slightly move up during inspiration and down during exhalation; moreover, the connection between the collar and the plate was not completely hermetic and leaks were allowed. At end-expiration, the internal volume of the helmet mounted on the head was 9 l, decreasing to 6 l once the cushions were inflated. Pressure-controlled ventilation was delivered by an Engström Carestation ventilator (GE Healthcare, Madison, WI, USA) by connecting the circuit Y-piece to the 22-mm port of the helmet.

Measurements of $CO_2$ concentration, airway pressure and flow were obtained by the ventilator sensors (calibrated according to manufacturer's instructions) and recorded on a personal computer by means of specialized software (RDPL Tool; GE Healthcare).

### Series 1

The effects of different levels of $CO_2$ production ($V'CO_2$) on the $CO_2$ concentration sampled at different sites were tested. We increased $V'CO_2$ from 100 to 550 ml/min in 50 ml/min steps. The respiratory pattern and the lung model mechanics were kept constant: PEEP 5 $cmH_2O$, inspiratory pressure above PEEP (Pinsp) 10 $cmH_2O$, respiratory rate (RR) 17/min, I:E ratio 1:2, airway resistance and compliance of lung model 8 cmH20/l/s and 50 ml/cmH20; no flow-by was applied and the cushions were left deflated.

### *Measurements and calculations*

Airway pressure and flow were recorded by connection of the ventilator sensors to the airway opening of the head

(Fig. 1 a). After a minimum equilibration time of 15 min and always when stable for at least 5 min, we recorded end-inspiratory ($eiCO_2$) and end-tidal ($etCO_2$) $CO_2$ at the airway opening, and then end-expiratory $CO_2$ at the Y-piece ($yCO_2$) and mean $CO_2$ inside the helmet ($hCO_2$) by sequential connection of the $CO_2$ sensor to three different sampling lines (Fig. 1 a, b, d). The $hCO_2$ was sampled far from both the ventilator connection port and the patient airway opening.

The $iCO_2$,mean was calculated from the instantaneous inspiratory $CO_2$ concentration ($CO_2$,insp) and flow rate (V'insp) sampled at 25 Hz at the airway opening:

$$\begin{aligned} iCO_{2,mean}(\%) = &\int[CO_{2,insp}(\%) \times V'insp(ml/s) \\ &\times 0.04(s)]/ \\ &\int[V'insp(ml/s) \times 0.04(s)] \end{aligned}$$

Series 2

During constant $V'CO_2$ (300 ml/min) the effects on $CO_2$ rebreathing of inspiratory pressure, patient passive mechanics, flow-by and inflation of helmet cushions were tested. We simulated a normal and a low compliance of 50 and 30 ml/cm$H_2O$. The cushion inflation was performed with 3 l of air.

For the flow-by, we used the helmet ports normally dedicated to nasogastric tubes (Fig. 1 c). The flow-by inflow and outflow were adjusted to the same value by two flowmeters (Dräger Medical, Lübeck, Germany) and obtained by delivering a constant flow of air from a source independent from the ventilator and by connection to a vacuum source, respectively.

Four levels (5, 10, 15 and 20 cm$H_2O$) of Pinsp and two values of flow-by (zero and 8 l/min) were used in any of three different settings: normal compliance, deflated cushions; low compliance, deflated cushions; and normal compliance, inflated cushions. Therefore, a total of 24 different conditions were studied. PEEP, RR and I:E ratio were kept constant as in series 1.

*Measurements and calculations*

We recorded inspiratory and expiratory volumes delivered by the ventilator (Vt,i and Vt,e) by connection of the ventilator sensors to the Y-piece (Fig. 1 b). End-tidal and end-inspiratory $CO_2$ at the airway opening and helmet $CO_2$ were recorded after a minimum equilibration time of 15 min and always when stable for at least 5 min by sequential connection of the $CO_2$ sensor to two different sampling lines (Fig. 1 a, d). The patient tidal volume (Vt,pat) was read directly on the lung simulator, on a scale precalibrated with a super-syringe.

The minute leakage (MVleak) and the minute ventilation of the patient (MVpatient), of the helmet (MVhelmet) and of the whole system (MVtotal) were calculated as follows:

$$MV_{leak} = (V_{t,i} - V_{t,e}) \times RR$$

$$MV_{patient} = V_{t,pat} \times RR$$

$$MV_{helmet} = (V_{t,e} - V_{t,pat})RR$$

$$MV_{total} = MV_{helmet} + MV_{patient} + MV_{leak} + \text{flow-by}$$

Paired *t*-test was used to compare conditions differing for just one parameter: Pinsp 5 versus 10, 15 and 20 cm$H_2O$, normal compliance versus low compliance (by discarding the eight conditions with the cushions inflated), deflated cushions versus inflated cushions (by discarding the eight conditions with the compliance low), no flow-by versus 8 l/min flow-by.

Calculated value of mean helmet $CO_2$ ($hCO_{2,calc}$) were obtained by application of the equation $hCO_{2,calc} = V'CO_2 / MVtotal$ and compared with measured values in series 1 and 2.

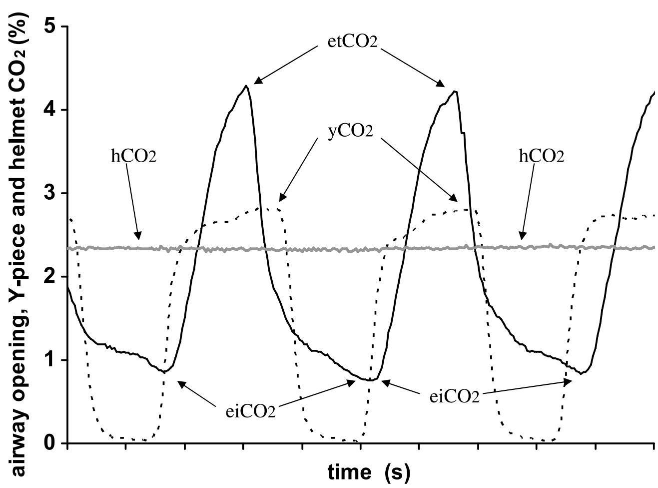


**Fig. 2** Continuous $CO_2$ concentration recordings at the airway opening (*black line*), at the Y-piece of the ventilator circuit (*dotted line*) and inside the helmet (*grey line*) during helmet ventilation, with 16.5 l/min of total minute ventilation and 350 ml/min of $CO_2$ production. At the airway opening the minimum value is the end-inspiratory $CO_2$ (*eiCO₂*) and the maximum value is the end-expiratory $CO_2$ (*etCO₂*). The $CO_2$ concentration measured at the Y-piece ranges from zero during inspiration (due to the fresh gas flow from the ventilator) and the end-expiratory value (*yCO₂*), which is lower than $etCO_2$ because the gas expired by the patient is diluted in the helmet internal volume. The $CO_2$ concentration measured in a quiet (not crossed by flows) point inside the helmet (*hCO₂*) is constant during the respiratory cycle and corresponds to the mean inhaled $CO_2$ concentration (see also Fig. 3)

## Results

### Monitoring of $CO_2$ rebreathing

In the 10 conditions of series 1, the mean inspired $CO_2$ ($iCO_2$,mean) was 2.2 ± 0.8% (range 1.0–3.3).

Figure 2 represents the instantaneous $CO_2$ concentrations (10-s record) at the airway opening of the patient, at the Y-piece of the ventilator circuit and inside the helmet. The $CO_2$ wave at the airway opening oscillated between an $eiCO_2$ level of 0.8 ± 0.3% (range 0.4–1.2) and an $etCO_2$ level of 3.9 ± 1.4% (range 1.7–5.9). The $CO_2$ wave at the Y-piece oscillated between zero during inspiration and an end-expiratory value ($yCO_2$) of 2.6 ± 1.0% (range 1.1–4.1). The $CO_2$ concentration inside the helmet ($hCO_2$) was extremely stable during the respiratory cycle, with a value of 2.1 ± 0.8% (range 0.9–3.3).

Figure 3 shows the relationships between $iCO_2$,mean and, respectively, $eiCO_2$, $yCO_2$ and $hCO_2$. All these three $CO_2$ measurements were highly and linearly related with $iCO_2$,mean ($r^2 > 0.97$). Differences between $iCO_2$,mean and, respectively, $hCO_2$, $yCO_2$ and $eiCO_2$ were 0.0 ± 0.1% (range –0.2 to +0.1), 0.4 ± 0.2% (0.1–0.8) and –1.3 ± 0.5% (–2.1 to –0.8).

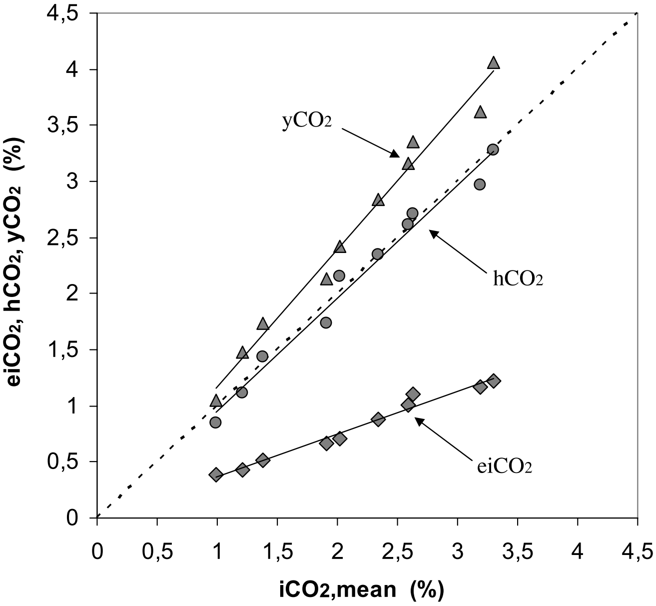


**Fig. 3** Relationships between mean inspired $CO_2$ (*$iCO_{2,mean}$*) and, respectively, end-inspiratory $CO_2$ at airway opening (*$eiCO_2$, rhombs*), helmet $CO_2$ (*$hCO_2$, circles*) and end-tidal $CO_2$ at the Y-piece (*$yCO_2$, triangles*). The identity line is traced as a *dotted line*

### Determinants of $CO_2$ rebreathing

In series 1, while MVtotal was constant at 16.5 l/min, $iCO_2$,mean was linearly and highly related to V'$CO_2$ ($iCO_2$,mean = V'$CO_2$ / 169.5+0.19; $r^2 = 0.98$).

The results obtained in series 2 are summarized in Table 1. As in series 1, $eiCO_2$ greatly underestimated mean helmet $CO_2$.

Increasing levels of Pinsp from 5 to 20 $cmH_2O$ were associated with significant increases of MVpatient, MVhel-

**Table 1** Changes of mean $CO_2$ concentration inside the helmet (*$hCO_2$*), end-inspiratory and end-expiratory $CO_2$ at the airway opening (*$eiCO_2$*, *$etCO_2$*) and minute ventilations (*MV*) due to different levels of inspiratory pressure (*Pinsp*) of 5, 10, 15 or 20 $cmH_2O$, respiratory system compliance (*Cpl*) of 50 or 30 ml/$cmH_2O$, deflated or inflated cushions and flow-by (*FB*) of 0 or 8 l/min during helmet ventilation

| | Pinsp 5 ($n=6$) | Pinsp 10 ($n=6$) | Pinsp 15 ($n=6$) | Pinsp 20 ($n=6$) | Cpl 50 ($n=8$) | Cpl 30 ($n=8$) | deflated ($n=8$) | inflated ($n=8$) | FB 0 ($n=12$) | FB 8 ($n=12$) |
|---|---|---|---|---|---|---|---|---|---|---|
| $hCO_2$ (%) | 2.3 ± 0.8 | 1.4 ± 0.3[b] | 1.1 ± 0.2[b] | 0.9 ± 0.2[b] | 1.4 ± 0.6 | 1.5 ± 0.8[a] | 1.4 ± 0.6 | 1.3 ± 0.7 | 1.7 ± 0.8 | 1.1 ± 0.3[b] |
| $eiCO_2$ (%) | 0.5 ± 0.1 | 0.3 ± 0.0[b] | 0.2 ± 0.0[b] | 0.2 ± 0.0[b] | 0.3 ± 0.1 | 0.3 ± 0.2 | 0.3 ± 0.1 | 0.3 ± 0.1 | 0.3 ± 0.2 | 0.2 ± 0.1[a] |
| $etCO_2$ (%) | 9.3 ± 2.8 | 5.9 ± 1.1[b] | 5.0 ± 0.7[b] | 4.7 ± 0.6[b] | 5.3 ± 1.2 | 7.7 ± 3.1[b] | 5.3 ± 1.2 | 5.7 ± 1.9 | 6.6 ± 2.8 | 5.9 ± 1.9[a] |
| MVpatient (l/min) | 4.7 ± 1.4 | 8.7 ± 1.5[c] | 11.8 ± 1.6[c] | 14.3 ± 1.6[c] | 11.2 ± 3.8 | 8.0 ± 3.7[c] | 11.2 ± 3.8 | 10.5 ± 4.0[c] | 9.9 ± 4.1 | 9.8 ± 3.9 |
| MVhelmet (l/min) | 3.1 ± 0.7 | 5.9 ± 1.1[c] | 8.8 ± 1.2[c] | 11.7 ± 1.6[c] | 6.4 ± 3.3 | 8.6 ± 3.9[c] | 6.4 ± 3.3 | 7.0 ± 3.1[b] | 7.7 ± 3.7 | 7.0 ± 3.3[b] |
| MVleak (l/min) | 2.4 ± 0.1 | 3.1 ± 0.2[c] | 3.5 ± 0.2[c] | 3.1 ± 0.3[b] | 3.2 ± 0.4 | 2.9 ± 0.5[a] | 3.2 ± 0.4 | 3.1 ± 0.5 | 3.0 ± 0.5 | 3.0 ± 0.5 |
| Flow-by (l/min) | 4.0 ± 4.4 | 4.0 ± 4.4 | 4.0 ± 4.4 | 4.0 ± 4.4 | 4.0 ± 4.3 | 4.0 ± 4.3 | 4.0 ± 4.3 | 4.0 ± 4.3 | 0.0 ± 0.0 | 8.0 ± 0.0 |
| MVtotal (l/min) | 14.3 ± 4.2 | 21.7 ± 4.1[c] | 28.1 ± 3.8[c] | 33.0 ± 3.9[c] | 24.8 ± 8.2 | 23.5 ± 8.7[c] | 24.8 ± 8.2 | 24.6 ± 8.5 | 20.7 ± 7.5 | 27.9 ± 7.3[c] |

Data are shown as mean ± SD. The $p$ values refer to the following comparisons (paired $t$-test): Pinsp 10, 15 and 20 versus 5 $cmH_20$, low versus normal Cpl, inflated versus deflated cushions, 8 l/min FB versus no FB.

[a] $p < 0.05$; [b] $p < 0.01$; [c] $p < 0.001$

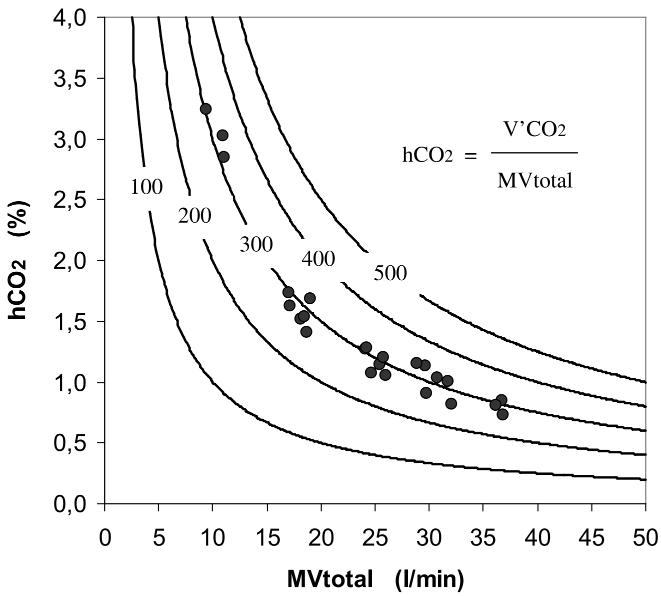


**Fig. 4** Relationship between mean $CO_2$ inside the helmet (*$hCO_2$*) and whole flow passing through the helmet (*$MV_{total}$*). Data obtained during the 24 conditions of series 2 (*circles*) and theoretical $hCO_2$–$MV_{total}$ curves (*continuous lines*) according to the equation of $CO_2$ steady state inside the helmet at five different values of $CO_2$ production (100, 200, 300, 400 and 500 ml/min)

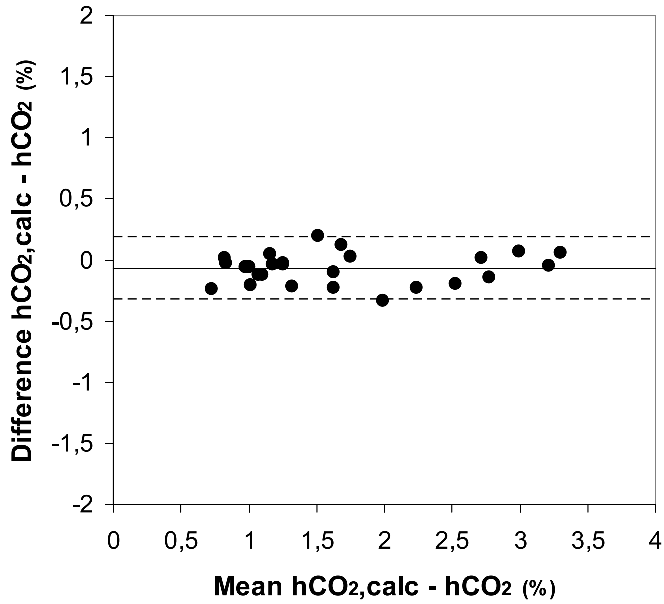


**Fig. 5** Difference between calculated (*$hCO_{2,calc}$*) and measured (*$hCO_2$*) values of mean helmet $CO_2$: Bland–Altman analysis. The calculated values were obtained by application of the equation $hCO_{2,calc} = V'CO_2/MV_{total}$ in the 10 conditions of series 1 and in the 24 conditions of series 2. *Continuous line* and *dotted lines*: mean value (−0.1%) and ± 2 standard deviations (± 0.2%) of the difference between $hCO_{2,calc}$ and $hCO_2$

met and MVleak, resulting in progressively increasing values of MVtotal and decreasing values of both $hCO_2$ and $etCO_2$.

The selection of a low Cpl on the lung simulator led to a major decrease in MVpatient, associated with an increase in MVhelmet: this halved the patient:helmet ventilation ratio. The low Cpl setting was associated with a slight decrease in MVtotal, a slight increase in $hCO_2$ and a large increase in $etCO_2$.

The 33% helmet volume reduction obtained by cushion inflation was associated with a minimal decrease in MVpatient, a minimal increase in MVhelmet and no significant changes in MVtotal, $hCO_2$ and $etCO_2$.

The addition of a 8 l/min flow-by through the helmet went along with an equal increase of MVtotal and effectively decreased $hCO_2$ and $etCO_2$, despite unchanged patient's ventilation.

During helmet ventilation at constant $CO_2$ production, we observed a very good inverse relationship between $hCO_2$ and MVtotal (Fig. 4). The equation $hCO_2 = 31.8$ / MVtotal–0.09 assured the best fit with the experimental data ($r^2 = 0.98$).

Conversely, the relationship between $hCO_2$ and MVhelmet was poor ($r^2 = 0.41$).

The calculated values of mean helmet $CO_2$ were nearly identical to the measured values in series 1 and 2 (Fig. 5): mean difference was –0.1 ± 0.1%, and the equation $hCO_{2,calc} = hCO_2 \times 1.0–0.1$ assured the best fit with the experimental data ($r^2 = 0.98$).

## Discussion

The main findings of this bench study are that, during helmet ventilation: (1) $CO_2$ rebreathing can be monitored measuring the $CO_2$ concentration at a "quiet" point inside the helmet or alternatively at the Y-piece, and (2) $CO_2$ rebreathing is inversely related to the total flow passing through the helmet and directly related to the patient's $CO_2$ production.

In our experience we observed that $CO_2$ concentration is not homogeneous inside the helmet. When the sampling was performed between patient and ventilator, at airway opening or Y-piece, $CO_2$ oscillated between low inspiratory and high expiratory levels. In contrast, $CO_2$ was very stable when measured at a quiet point, not affected by flows to and from the patient; moreover, this value was really equivalent to the mean inhaled $CO_2$. Therefore, when a "quiet" point inside the helmet can be found, this seems the best site for $CO_2$ sampling.

An interesting alternative, even if slightly less accurate, is represented by $yCO_2$, i.e. by reading the $etCO_2$ at the Y-piece. Of note, the value of $etCO_2$ at the Y-piece is much lower than the actual $etCO_2$ of the patient, because the expired $CO_2$ dilutes within the helmet internal volume, which is much greater than the patient's expiratory volume. Therefore, $yCO_2$ can be used to estimate $CO_2$ rebreathing but not to estimate the patient's arterial $CO_2$.

Usually $CO_2$ rebreathing had been evaluated by means of the end-inspiratory value at the airway opening [1, 13, 15, 16], but in our experience $eiCO_2$ grossly underestimated mean inspiratory $CO_2$, particularly when it was significant. The reason was the slow decrease in inhaled $CO_2$ concentration during the inspiratory phase of helmet ventilation.

When a steady state is reached and $CO_2$ is stable inside the helmet, the amount of $CO_2$ entering the helmet per minute must be equal to the amount of $CO_2$ leaving the device in the same time. Assuming a non-significant dead space at the Y-piece of the ventilator circuit, the only source of $CO_2$ entering the helmet is $CO_2$ production by the patient. Concerning the $CO_2$ leaving the system, that is a function of the mean $CO_2$ inside the helmet and the sum of all flows directed from inside the helmet to outside: namely the expired (by patient and helmet) ventilation through the Y-piece, plus the air leaks at the collar–neck interface, plus, if present, the outflow of an additional flow-by. Accordingly, the theoretical equation of $CO_2$ steady state inside the helmet is:

$$V'CO_2 = hCO_2 \times MV_{total}, \text{ or}$$
$$hCO_2 = V'CO_2/MV_{total}.$$

Therefore, the mean $CO_2$ concentration inside the helmet should be directly related to $V'CO_2$, inversely related to MVtotal and unaffected by any other factor. This hypothesis was fully confirmed by our experimental data (Fig. 4 and 5): of note, during phasic helmet ventilation we obtained results similar to those observed by Taccone et al. during continuous-flow CPAP [13].

As a matter of fact, all the manipulations tested in our second series affected $CO_2$ rebreathing insofar as they were able to change the total flow passing through the helmet.

The increase of inspiratory pressure above PEEP produced a progressive increase in MVtotal; accordingly, $hCO_2$ progressively decreased. In contrast with our findings, studies on healthy volunteers recently found no relationship between pressure support level and $CO_2$ rebreathing [15, 16]. A possible explanation is that, in order to keep their minute ventilation constant, volunteers may react to an increase in pressure support with a decrease in spontaneous inspiratory activity, thus limiting the effect on MVtotal. Most importantly, the theoretical relationship between MVtotal and $hCO_2$ (Fig. 4) tends to flatten for high values of MVtotal, and the flattening takes place for lower values of MVtotal when $V'CO_2$ is lower. In the study by Costa and coworkers [15], the mean MVtotal was close to the top of our second series, while $V'CO_2$ was much lower. Therefore, in these conditions the expected variations in helmet $CO_2$ due to changes in MVtotal were really low. Finally, rebreathing was assessed by the end-inspiratory $CO_2$ at the airway opening in these studies [15, 16], i.e. by a parameter we have found to have major limitations.

The relative distribution of MVtotal between patient, helmet and air leaks does not affect rebreathing. In confirmation of that, when we selected a low compliance in the lung model, this resulted in a large change in the patient:helmet ventilation ratio (from 2:1 to about 1:1) with only a small change in $hCO_2$, the latter fully explained by a proportionate change in MVtotal.

With regard to leaks, these can decrease $CO_2$ inside the helmet by an increase in MVtotal. In any case, a helmet provided with an "intentional" leakage port will decrease helmet pressurization and patient's inspiratory assistance, probably turning out counterproductive for $CO_2$ removal.

Concerning the role of helmet volume, inflation of the cushions resulted in a 33% volume reduction but in our experience was associated with no change in $hCO_2$. In clinical practice, the inflated cushions may facilitate the initial pressurization of the helmet, improve the comfort of the patient and stabilize the system, eventually limiting air leaks: the overall effect on rebreathing is difficult to predict, but probably of low significance.

The addition of a flow-by through the helmet was very effective in clearing $CO_2$ around the head of the patient. However, the flow-by system we used in our experimental setting is not very practical for clinical application. An interesting option might be the use of the bias flow of the ventilator as a flow-by. In order to force the bias flow to pass through the helmet, the ventilator must be connected to the helmet by two independent ports, for inspiration and expiration. Moreover, for a significant effect on $CO_2$ rebreathing the bias flow should be continuous or at least applied during the entire expiratory phase, and it should be adjustable at relatively high values, at least 10 l/min.

This study has some limitations. We performed a bench study, because the pure effects of changes of $CO_2$ production or minute ventilation on rebreathing would have been difficult to study in patients or volunteers. The lung model was passive and the ventilation controlled, in contrast to clinical practice. The use of an active physical model would have offered the opportunity, through modifications of the spontaneous activity, to change both the patient ventilation and the patient:helmet ventilation ratio. In our experimental setting, we obtained a similar effect by changing the lung model compliance. Finally, just one type of helmet was studied. Helmet models can differ in elastic properties, tightness or inner volume: these factors are related to rebreathing through their influence on helmet ventilation and leaks or, in the case of volume, not related at all.

## Conclusions

During NIV delivered by helmet, some $CO_2$ rebreathing is necessarily present and the inspiratory $CO_2$ must be monitored: gas sampling can be performed at a "quiet" point inside the helmet or, alternatively, at the Y-piece. The magnitude of rebreathing is directly related to the patient's $CO_2$ production and inversely related to the total flow passing through the helmet. Therefore, significant rebreathing can be expected when the patient's metabolic requirements are elevated and/or the ventilator volume delivery is low, discouraging the use of low levels of pressure support in this setting. To further decrease rebreathing when the pressure assistance to the patient's spontaneous activity is already optimal, the addiction of a flow-by can be considered.

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