Effect of the Anatomic Reservoir on Low-Flow Oxygen Delivery Via Nasal Cannula: Constant Flow Versus Pulse Flow With Portable Oxygen Concentrator

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BACKGROUND: The F_{IO_2} for a nasal cannula with constant flow (CF) depends on the anatomic reservoir (AR), which is affected by changes in frequency and end-expiratory flow. Conversely, pulse flow (PF) devices do not require the AR. The purpose of this study was to compare the F_{IO_2} delivered by a nasal cannula supplied by CF via oxygen tank with that delivered by PF delivered via portable oxygen concentrator. Hypotheses were (1) a lung model of COPD with non-zero end-expiratory flow decreases F_{IO_2} for CF more than for PF, and (2) CF and PF perform differently in terms of F_{IO_2} delivery, despite having equivalent settings. METHODS: Normal and COPD lung models were simulated (IngMar Medical ASL 5000) using published human data: normal: breathing frequency = 15 breaths/min, R_{in} = 4 cm $H_2O \cdot s \cdot L^{-1}$, R_{out} = 4 cm $H_2O \cdot s \cdot L^{-1}$, C = 60 mL · cm H_2O^{-1}, tidal volume (V_T) = 685 mL, P_{max} = 11.95 cm H_2O, increase = 33%, and release = 28; COPD: breathing frequency = 20 breaths/min, R_{in} = 12 cm $H_2O \cdot s \cdot L^{-1}$, R_{out} = 25 cm $H_2O \cdot s \cdot L^{-1}$, C = 66 mL · cm H_2O^{-1}, V_T = 685 mL, P_{max} = 24.52 cm H_2O, increase = 35%, and release = 23%. CF was 1–5 L/min. Portable oxygen concentrators used were $Solo_2$ (Invacare), XPO2 (Invacare), FreeStyle (AirSep), Focus (AirSep), One G3 (Inogen), and LifeChoice ActivOx (Inova Labs). RESULTS: CF produced significantly higher F_{IO_2} at all settings for normal lungs but lower for COPD lungs compared with $Solo_2$. COPD reduced the F_{IO_2} for CF but had a smaller variable effect for PF. Data show there is no equivalency between PF setting and CF rates for the portable oxygen concentrators tested. CONCLUSIONS: CF oxygen delivery via a nasal cannula is significantly reduced by elimination of the AR in a model of COPD, yielding clinically important decreases in F_{IO_2}. PF (delivered with a portable oxygen concentrator) is relatively unaffected. This study supports the recommendation that clinicians and caretakers should titrate the PF setting to each patient's unique oxygen requirements. *Key words: lung simulator; nasal cannula; chronic obstructive pulmonary disease; oxygen therapy.* [Respir Care 2014;59(8):1199–1209. © 2014 Daedalus Enterprises]

Introduction

Long-term oxygen therapy in the home is often delivered using a nasal cannula and either compressed or liquid oxygen sources. More recently, portable oxygen concentrators have been developed and marketed for patients' home use on a prescriptive basis. For these oxygen sources, gas may be delivered either as a constant flow (CF) or pulse flow (PF) (eg, pulsed-dose oxygen-conserving systems). PF devices are intended to either conserve oxygen

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(for tanks) or conserve battery energy (for portable oxygen concentrators).

The F_{IO_2} during CF nasal cannula oxygen therapy is governed by the interaction of the patient's inspiratory flow pattern and the flow from the oxygen source (Fig. 1). The original mathematical model for this was described in a textbook by Shapiro[1] (Table 1, column A). As noted by Boatright and Ward,[2] this model is ". . . not clinically practical (for) calculating the . . . F_{IO_2} delivered by the nasal oxygen cannula (but) is a useful learning tool." Specifically, it is very useful for understanding the interactions of the variables that determine F_{IO_2}. Two important variables in this model are inspiratory time (T_I) and anatomic reservoir (AR). T_I is a function of breathing frequency and I-E ratio. As the frequency increases (due to either exercise or disease), T_I decreases, which in turn decreases F_{IO_2} (Table 1, column B). We will hereafter refer to this as the frequency effect.

The AR is the volume of oxygen stored in the upper airways (pharynx and nasal passage) due to flushing of the dead space by the cannula flow during the last portion of expiration. The AR is not a real physical space (ie, portion of the anatomic dead space) but rather an imaginary mathematical construct representing the volume of oxygen that may or may not exist in the anatomic dead space due to flushing by the cannula. If there is no flushing at end expiration (as with a pulse flow device) or if the flow from the cannula is opposed by the patient's expiratory flow, then the AR will not exist. Patients with COPD often have non-zero expiratory flows when the next inspiratory effort is made (ie, they have alveolar gas-trapping); that is, patients with COPD have long expiratory time constants and typically experience gas-trapping even during rest due to the fact that expiratory flow does not decay to zero before the next inspiratory effort. Thus, end-expiratory flow could prohibit CF devices from flushing away the upper airway dead space, prevent the AR of oxygen from accumulating, and thus decrease the F_{IO_2} (Fig. 2). As shown in Table 1 (column C), eliminating the AR has an even larger effect on F_{IO_2} than increasing frequency (in this case, decreasing F_{IO_2} from 0.34 to 0.26). We will hereafter refer to this as the reservoir effect.

PF devices built into portable oxygen concentrators are most frequently designed to maintain a constant minute volume of oxygen delivered by cannula as the frequency

QUICK LOOK

Current knowledge

The delivered F_{IO_2} during low-flow oxygen delivery via a nasal cannula is a function of the respiratory pattern and the presence of the anatomic reservoir (AR). During pulse dose oxygen delivery, the AR should not impact the delivered F_{IO_2}.

What this paper contributes to our knowledge

In a model system mimicking chronic obstructive lung disease, functional elimination of the AR due to the presence of flow at end expiration significantly reduced delivered oxygen via continuous low-flow oxygen by cannula. Pulse dose oxygen delivery was not impacted by the functional loss of the AR.

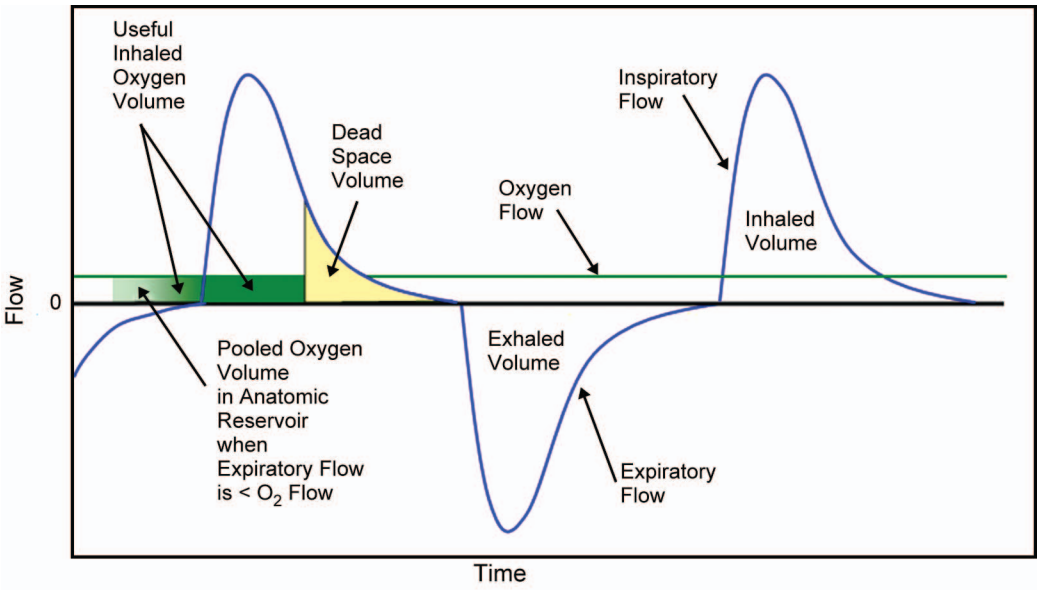

Fig. 1. Idealized waveforms for spontaneous breathing (blue) and constant flow nasal cannula (green) for patient with normal lungs. Areas between flow curves and time axis represent volumes.

Table 1. Simple Mathematical Model Showing the Variables Affecting F_{IO_2} for a Low-Flow Cannula

	A*	B†	C‡
Tidal volume (mL)	500	500	500
Dead-space volume (mL)	150	150	150
Alveolar volume (mL)	350	350	350
Cannula flow (L/min)	2	2	2
Cannula flow (mL/s)	33	33	33
Inspiratory time (s)	1	0.5	1
Inspired room air (mL)	417	433	467
Total inspired oxygen (mL)			
Anatomic reservoir	50	50	0
Cannula	33	17	33
From air	88	91	98
Total	171	158	131
F_{IO_2}	0.34	0.32	0.26

* Baseline F_{IO_2}
† F_{IO_2} with increased breathing frequency and decreased inspiratory time
‡ F_{IO_2} without the anatomic reservoir

Table 2. Math Models for Pulse Flow Oxygen-Conserving Device

	Pulsed Flow Pure Oxygen		Pulsed Flow POC (90%)	
	A	B	A	B
Tidal volume (mL)	500	500	500	500
Dead-space volume (mL)	150	150	150	150
Alveolar volume (mL)	350	350	350	350
Pulse volume setting 2 (mL)	24	65	24	65
Inspiratory time (s)	1	1	1	1
Inspired room air (mL)	476	435	478	442
Total inspired oxygen (mL)				
Anatomic reservoir	0	0	0	0
Pulsed flow	24	65	21.6	59
From air	100	91	100	93
Total	124	156	122	151
F_{IO_2}	0.25	0.31	0.24	0.30

Columns A and B show the change in F_{IO_2} over the range of pulse volumes measured in this study in different models of portable oxygen concentrators at setting 2.
POC = portable oxygen concentrator

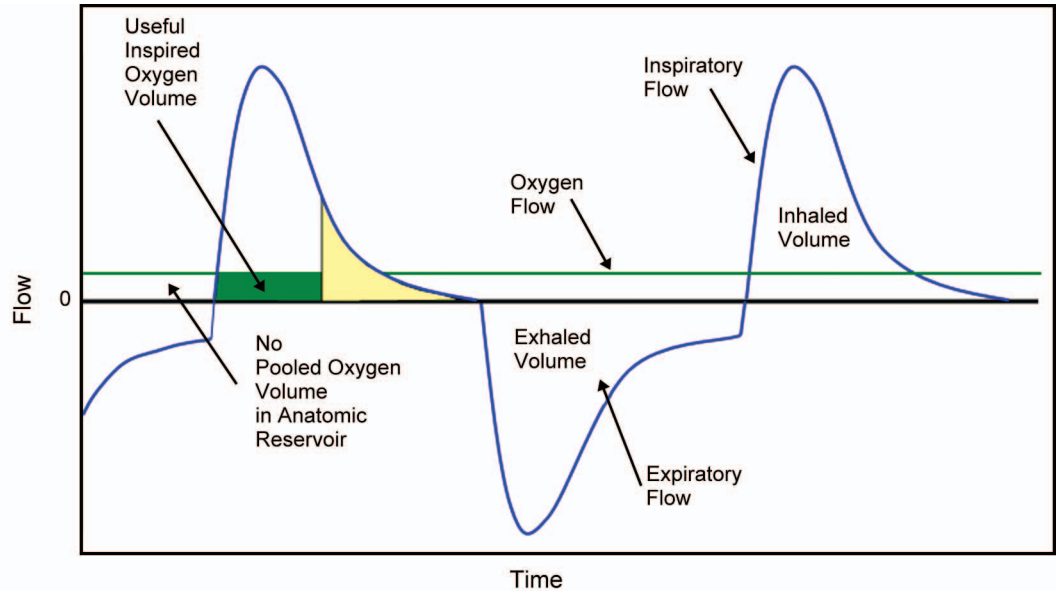

Fig. 2. Idealized waveforms for spontaneous breathing (blue) and constant flow nasal cannula (green) for patient with COPD. Areas between flow curves and time axis represent volumes. Note that end-expiratory flow does not decay to zero before the next breath, eliminating the atomic reservoir and decreasing the useful inspired oxygen volume.

increases (as opposed to some PF devices that maintain a constant pulse volume as frequency increases). Constant minute volume PF devices therefore should, like CF devices, show a decreasing F_{IO_2} with increasing breathing frequency. For PF devices, F_{IO_2} is largely determined by the pulse volume, and different portable oxygen concentrators deliver highly varied pulse volumes for the same setting (Table 2, compare columns A and B). Furthermore, portable oxygen concentrators do not deliver pure oxygen. Rather, the typical specification is 90 ± 3%. This has a small but noticeable effect on F_{IO_2} (Table 2).

One important difference between CF and PF devices is that no AR is established when using PF devices that are triggered by the patient's inspiratory effort. This may have clinical relevance when delivering oxygen to patients with lung disease. The implication is that a disease condition

Table 3. Specifications for Portable Oxygen Concentrators Used in This Study From Manuals

Device	Vendor	Weight (pounds)	Settings	Specification
$Solo_2$	Invacare	20	1–5	Minute volume range of 400–2,000 mL
XPO_2	Invacare	6	1–5	Minute volume range of 300–840 mL
FreeStyle	AirSep	4.4	1–3	Setting 3 is equivalent to CF at 3 L/min
Focus	AirSep	2	1	Equivalent to CF at 2 L/min
One G3	Inogen	4.9	1–4	No information available
LifeChoice ActivOx	Inova Labs	4.83	1–3	Settings are equivalent to CF (L/min) ± 20%

CF = continuous flow

such as COPD would decrease F_{IO_2} from CF devices but not from PF devices.

The purpose of this study was to compare the F_{IO_2} delivered with a nasal cannula attached either to CF of pure oxygen or to PF from a portable oxygen concentrator (constant minute volume design) generating ~90% purity. The primary hypothesis was that a lung model of COPD disease state with non-zero end-expiratory flow (ie, gas-trapping) decreases F_{IO_2} for CF more than for portable oxygen concentrators. A secondary hypothesis was that CF and portable oxygen concentrators perform differently in terms of F_{IO_2} delivery despite having equivalent settings.

Methods

Six portable oxygen concentrators were evaluated in this study (Table 3). As the largest device with a weight of ~20 pounds, the $Solo_2$ portable oxygen concentrator (Invacare, Elyria, Ohio) is often compared separately from the other small portable oxygen concentrators that weigh ~5 pounds. We decided to include it to illustrate the effects of the higher pulse volumes it is capable of delivering. Although the $Solo_2$ can deliver both CF and PF, we limited it to PF for this study.

As noted above, some manufacturers declare the portable oxygen concentrator settings to be equivalent to corresponding values of CF. Therefore, in this study, portable oxygen concentrator settings of 1–5 were compared with settings of CF values of 1–5 L/min from a compressed oxygen source. All portable oxygen concentrators were allowed to warm up for at least 5 min before beginning experimentation.

To determine whether these portable oxygen concentrators should be classified as constant minute volume devices (as opposed to constant pulse volume devices), we made measurements of pulse volume using an oxygen-conserving test system (Hans Rudolph) at two breathing frequencies (see lung model descriptions below). This device was also used to verify that the oxygen purity was within manufacturers' specifications.

Fig. 3. Photo of experimental setup showing model nares connected to the IngMar ASL 5000 lung simulator.

Lung Models

This experiment was conducted using a nasal cannula (AirLife, CareFusion, San Diego, California) and artificial nares connected to an ASL 5000 lung simulator (IngMar Medical, Pittsburgh, Pennsylvania) as shown in Figure 3. Two lung models were created with the ASL 3.3 software: normal and COPD (Table 4). Lung models were created in the ASL software by (1) specifying the mechanical properties of the respiratory system, that is, resistance and compliance (Fig. 4), and (2) specifying the parameters of the mathematical function that represents the force (pressure)

Table 4. Lung Model Parameters Used With ASL 5000 Lung Simulator

	Normal	COPD
Frequency (breaths/min)	15	20
R_{in} (cm H_2O/L/s)	4	12
R_{out} (cm H_2O/L/s)	4	25
C (mL/cm H_2O)	60	66
Tidal volume (mL)	685	685
P_{max} (cm H_2O)	11.95	24.52
Increase (%)	33.0	35.0
Hold (%)	0	0
Release (%)	28.0	23.0
Pause (%)	0	0

generated by the simulated respiratory muscle (Fig. 5). For these simulations, we used a sinusoidal muscle pressure function (P_{mus}) and specified the amplitude (P_{max}), rate of increase, hold, and decrease in P_{mus} as percentages of the time for one cycle of inspiration and expiration (T_{Tot}). Figure 4 is a screenshot of the ASL 5000 software that was used to set these parameters.

A normal patient was defined as being medically stable and devoid of ailments limiting pulmonary function. A COPD patient was defined as medically stable but diagnosed with COPD. Lung model parameters (eg, resistance, compliance, patient effort profile) were derived from previously published studies.

Normal Lung Model. For the normal model, breathing frequency was found from the average of the data obtained by Gallagher et al[3] and Im Hof et al,[4] ~13 breaths/min, which we arbitrarily rounded up to 15 breaths/min. Inspiratory resistance was taken from Younes and Riddle,[5] who reported passive resistance of 4 cm H_2O/L/s. Expiratory resistance was taken from Tobin[6] and was assumed to be the same as normal patients: 4 cm H_2O/L/s. Younes et al[7] report an average passive elastance of 17 cm H_2O/L, which was rounded to a compliance value of 60 mL/cm H_2O. Data taken from Gallagher et al[3] and Im Hof et al[4] yielded an average tidal volume (V_T) of 685 mL. P_{max} (maximum muscle pressure amplitude generated by the simulated respiratory muscles) parameters were as follows; percent increase was found by averaging percent inspiration of data obtained by Gallagher et al[3] and Im Hof et al[4] using the equation T_I/T_{tot}. T_I was averaged to be 1.6 s, and T_{tot} was averaged to be 4.935 s, yielding an increase of ~33%. Hold (delay between inspiration, expiration, and breaths) was based on analysis of proportional assist pressure waveforms, which showed a hold of 0% of T_{Tot}.[8] Percent release was taken from analysis of proportional assist pressure[8] waveforms, approximated to be 28% of T_{tot}. These model parameters are summarized in Table 4 and Figures 4 and 5.

For a constant P_{max} setting, as resistance increases, peak flow of the lung model decreases, and hence V_T decreases. Using the ASL 3.3 software, the P_{max} setting was manually adjusted during each experimental run to maintain the target V_T for the lung model (~685 ± 5 mL).

COPD Lung Model. For the COPD lung model frequency, O'Donnell et al[9] reported an average breath cycle time of 3.1 s, which corresponded with a frequency of ~19 breaths/min, rounded up to 20. Inspiratory and expiratory resistance values were taken from Mead et al.[10] Kondili et al[11] reported respiratory system compliance in slices (series of volume intervals for determining nonlinear dynamic respiratory system mechanics) ranging from 50 to 79 mL/cm H_2O. The average of 5 slices when PEEP = 0 is ~66 mL/cm H_2O. Data for V_T were taken from O'Donnell et al[9] and found to be 690 mL.[9] To maintain consistency between the two models, a common V_T of 685 mL was chosen. The P_{max} setting was manually adjusted during each experimental run to maintain the target V_T for the lung model (~685 ± 5 mL). P_{mus} waveform parameters were analyzed, yielding the following settings; O'Donnell et al[9] reported that for unassisted resting patients with COPD, the average T_I was 1.1 s, and the total cycle time was 3.1 s. Thus, the P_{mus} rise time was estimated to be 1.1/3.1 = 0.35 or 35%. Analysis of waveforms reported by Tassaux et al[12] indicated a P_{mus} hold setting of 0% and a release setting of 23% at a breathing frequency of 16 breaths/min (which was the closest to a frequency of 20 breaths/min that we could find in the literature). These model parameters are summarized in Table 4 and Figures 4 and 5.

Data Analysis

The main outcome variable of this study was F_{IO_2}, as measured by the ASL 5000 lung simulator. Prior to the beginning of experimentation, the lung simulator O_2 sensor was calibrated according to the manufacturer's instructions using compressed oxygen.

For each lung model (normal and COPD), data for 25 breaths were collected for each oxygen flow setting (after the oxygen concentration in the lung model stabilized, typically after 5 breaths). For CF, these included 1–5 L/min. For the portable oxygen concentrators, all numerical settings available were used. Measured F_{IO_2} values for the last 10 of the 25 breaths of each oxygen setting were averaged, and the SD was calculated using the ASL 3.3 simulator software. Mean F_{IO_2} values were compared using two-way analysis of variance (multiple comparisons were performed with the Holm-Sidak procedure) or Kruskal-Wallis analysis of variance on ranks (multiple-comparison procedures were performed using the Tukey test), with $P < .05$ considered significant. Mean F_{IO_2} for

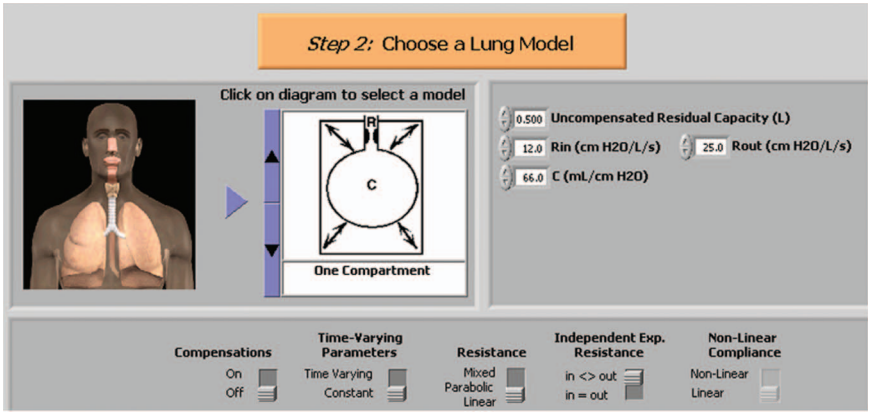

Fig. 4. Screen shot of the ASL 5000 lung simulator showing settings for lung mechanics.

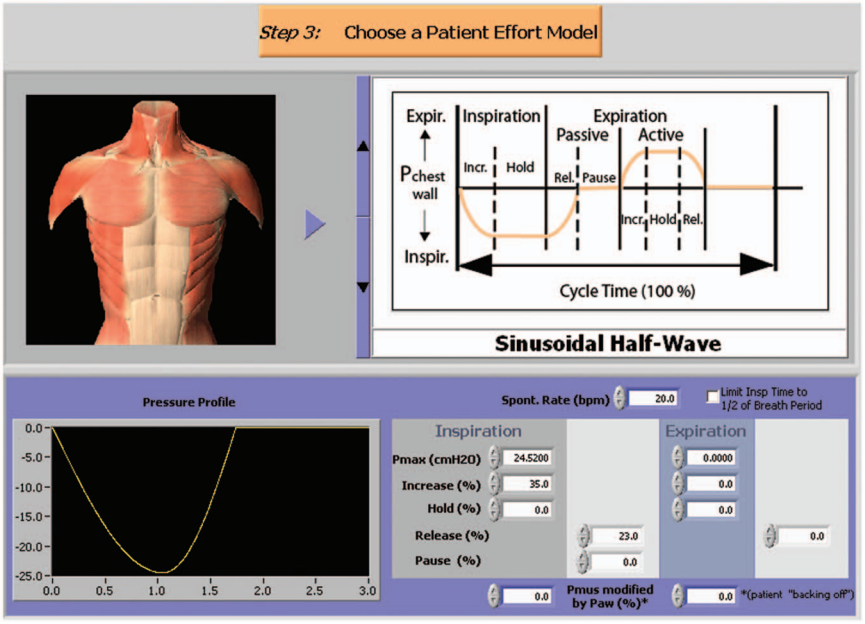

Fig. 5. Screen shot of ASL 5000 lung simulator software that allows setting of the parameters that define inspiratory effort during simulated breathing.

the AirSep Focus included only one portable oxygen concentrator setting, so a *t* test was used.

Results

Classification of Portable Oxygen Concentrators

Figure 6 shows that all portable oxygen concentrators maintained an essentially constant minute volume of oxygen delivery via the cannula as frequency increased from 15 to 20 breaths/min. This means that they all decreased their pulse volumes as the frequency increased. In contrast, devices that maintain a constant pulse volume show a dramatic increase in minute volume as frequency increases.[13]

Hypothesis 1: Effect of Disease State

Study data for both normal and COPD lung models (described in methods above) are summarized in Figure 7. For all experimental conditions, the resultant F_{IO_2} values were lower for the COPD lung model than for the normal lung model ($P < .001$). The F_{IO_2} difference increased as the CF or portable oxygen concentrator setting increased and was most pronounced at a flow of 5 L/min. However, if we define a clinically important difference as $\geq 2\%$, then the F_{IO_2} for CF was decreased by COPD for all oxygen flows greater than 1 L/min. The only clinically important difference for portable oxygen concentrators was at the highest setting for the $Solo_2$ (decrease of 4%).

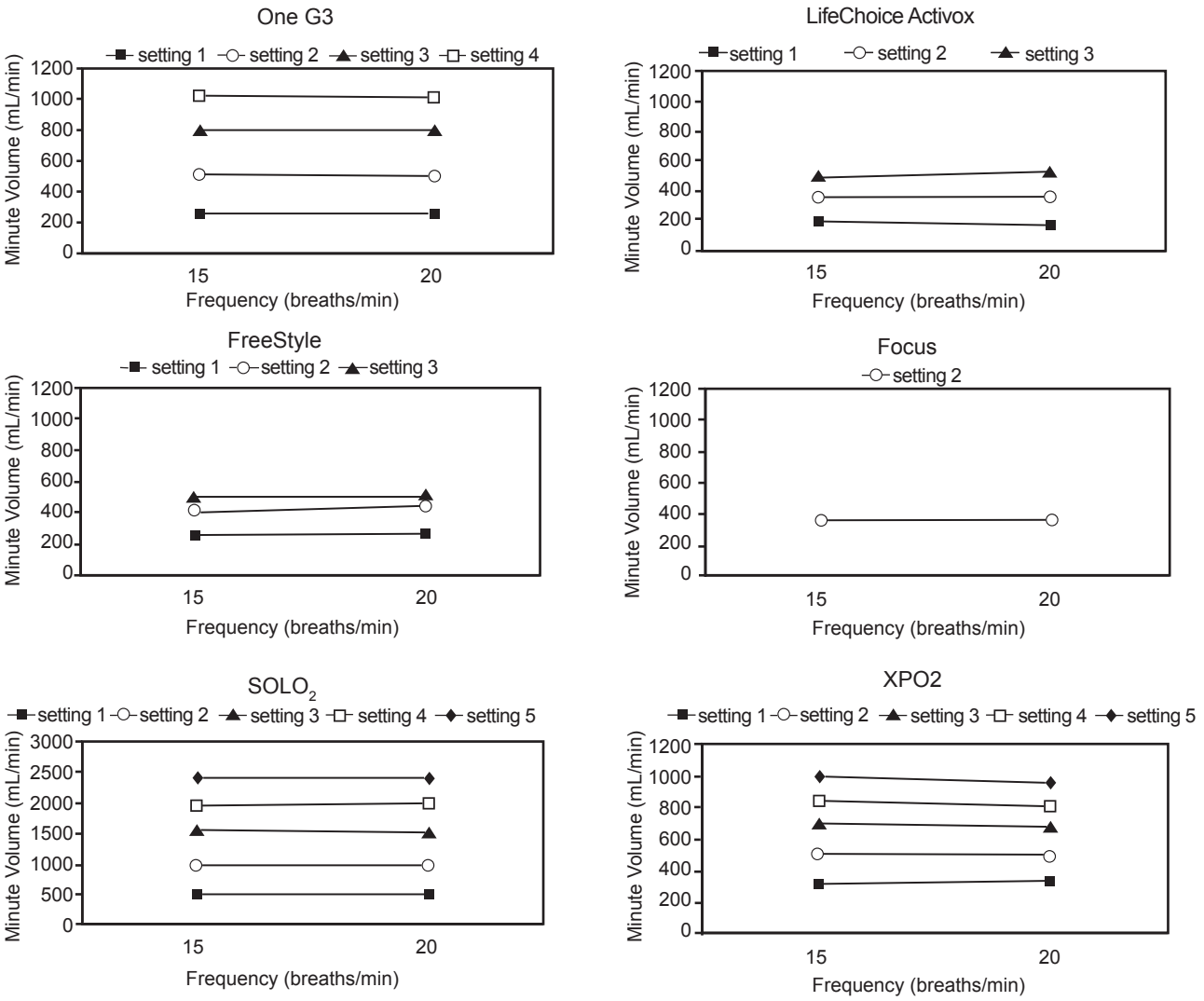

Fig. 6. Results of pulse volume measurement and calculation of minute volume of oxygen delivery. All devices maintained virtually constant minute volume as frequency was changed from 15 breaths/min (normal lung model) to 20 breaths/min (COPD lung model).

Hypothesis 2: Evaluation of CF and Portable Oxygen Concentrator Setting Equivalency

Our secondary hypothesis was that CF and portable oxygen concentrators perform differently in terms of F_{IO_2} delivery despite having equivalent settings. Study data are reproduced in Figure 8 and are grouped by normal and COPD lung models (lung parameters described in methods). Equivalent settings produced different F_{IO_2} values for both lung models ($P < .001$). Post hoc comparison results are shown in Table 5.

Discussion

To our knowledge, this is the first published study to compare the effects of eliminating the AR (using a model of COPD) on oxygen delivery via a nasal cannula using either CF or PF. The data support the hypothesis that F_{IO_2} in the COPD disease model is lower than in the normal model. Furthermore, F_{IO_2} with CF was decreased more in the disease model than PF using portable oxygen concentrators. We speculate that this is because F_{IO_2} with CF is reduced by two factors, the frequency effect and the reservoir effect, whereas PF is affected only by the frequency effect. We further speculate that PF oxygen-conserving devices that maintain a constant pulse volume as frequency increases (ie, increasing the minute volume delivery of oxygen) would be less affected or perhaps not affected at all by this COPD lung model.

This study also suggests that portable oxygen concentrator performance varies based on the minute volume of oxygen produced. In general, the higher the minute volume of oxygen, the higher the F_{IO_2} achieved. All of these

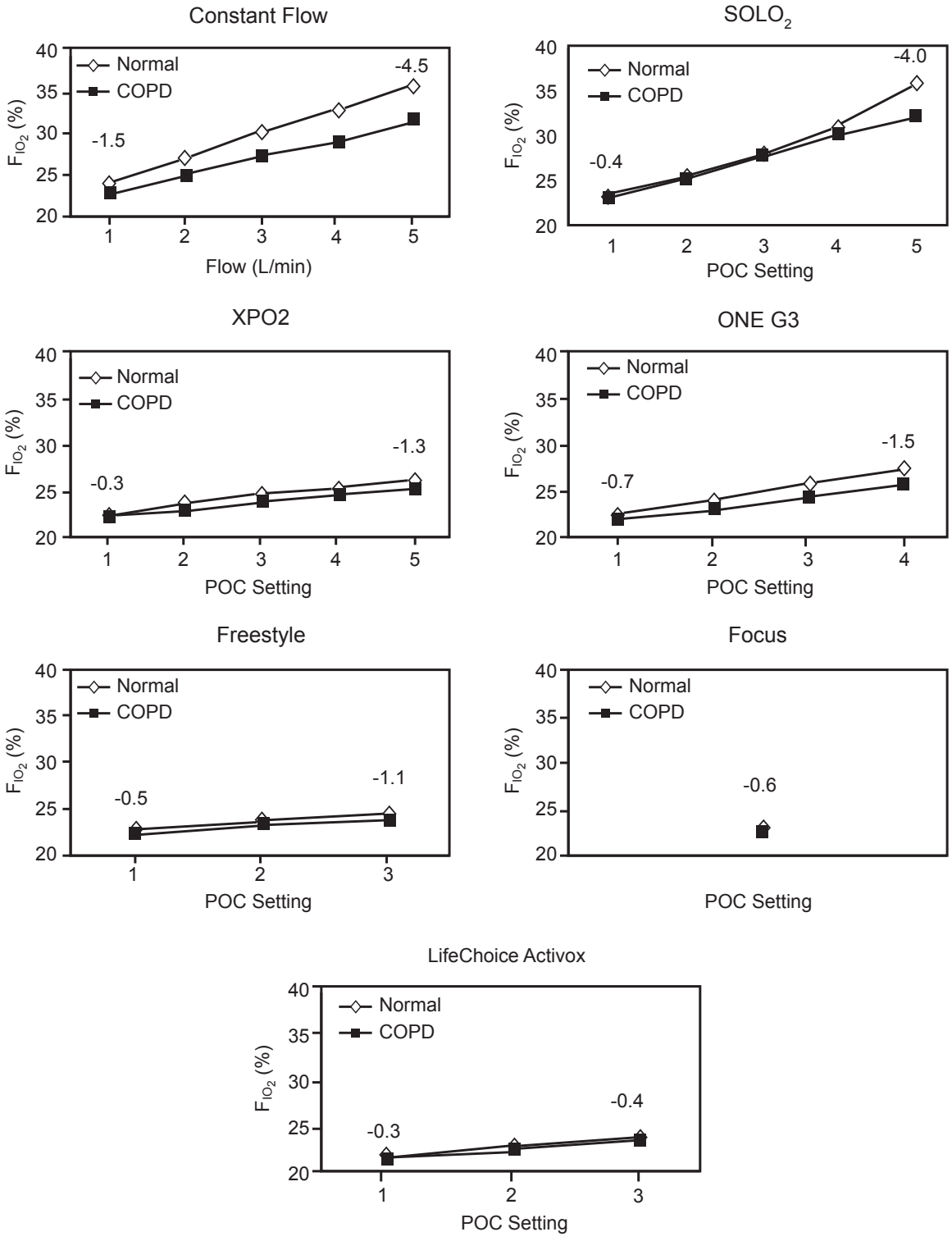

Fig. 7. Hypothesis 1: Graphs of measured F_{IO_2} (expressed as percentage) during simulated breathing. Negative numbers above curves represent the difference in F_{IO_2} (COPD lung model minus normal lung model) for the highest and lowest F_{IO_2} values. POC = portable oxygen concentrator.

devices delivered the pulse volume in less than ~ 0.6 s. Therefore, a reduction in T_I caused by a small increase in breathing frequency (from 15 to 20 breaths/min) for our study lung models is not expected to waste oxygen by

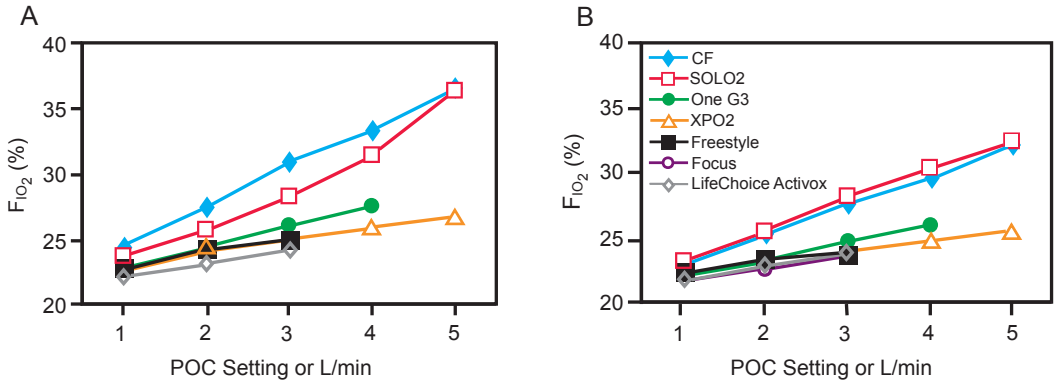

Fig. 8. Hypothesis 2: Comparison of constant flow (CF) and portable oxygen concentrator (POC) settings for normal (A) and COPD lung models (B).

Table 5. Results for Hypothesis 2. Data are F_{IO_2} for constant flow minus the F_{IO_2} for the indicated portable oxygen concentrator.

Setting	Continuous Flow					
	$Solo_2$	XPO_2	One G3	FreeStyle	Focus	LifeChoice
Normal						
1	**1**	**2**	2	2		2
2	2	3	3	3	4	4
3	**3**	6	**5**	6		7
4	**2**	7	6			
5	0	10				
COPD						
1	**0**	**0**	1	**1**	NA	1
2	0	2	2	2	3	2
3	−1	4	3	4		4
4	−1	5	4			
5	**0**	7				

Boldface values indicate nonsignificant results. Other values indicate $P < .05$.
NA = not available

delivering it into the dead-space volume at the end of a breath.

Finally, our data show that there is no such thing as equivalency between portable oxygen concentrator setting and CF rate for the devices tested. Therefore, the numerical settings on these portable oxygen concentrators should be interpreted as merely indicating increased relative minute volumes levels (ie, the higher the number, the higher minute volume of oxygen delivered via the cannula). These observations emphasize the importance of physicians or caretakers titrating the portable oxygen concentrator setting to each patient's unique oxygen requirements.

Our data are consistent with other studies in humans. LeBlanc et al[14] tested the ability of three portable oxygen concentrators to maintain $S_{pO_2} \geq 90\%$ during exercise in patients with COPD. The Eclipse 3 (chart SeQual Technologies, Ball Ground, Georgia), which delivers the largest oxygen bolus (pulse volume), yielded significantly higher S_{pO_2} during the walking exercise. In a similar fashion, our results confirm that the larger the pulse volume, the higher the F_{IO_2}.

A major limitation of this study was that our results were entirely dependent on the parameters of the simulations we designed. However, we believe that this is the first study using the IngMar ASL 5000 lung simulator that described every parameter of the model including not just mechanics but also the settings for the P_{mus} waveform. Furthermore, these parameters were based on an analysis of actual human data published in the literature. Therefore, we believe that our study design is particularly strong in its ability to model real-world clinical conditions.

Another limitation was that our simulated dead space (volume of nares and tube connecting the nose to the ASL 5000) was relatively small (40 mL) but still large enough to form an AR. However, the aim of this study was not to formulate a predictive model for patient F_{IO_2} but rather to offer insight into portable oxygen concentrator performance and the effects of normal and COPD disease states on F_{IO_2}. This study also offers insight into the relative effectiveness of CF and PF for treating COPD.

We speculate that the subject of our study has never been addressed because PF devices have historically been seen as valuable for their oxygen-conserving properties. Oxygen waste and conservation have played a significant role in the formulation of PF devices. In 1978, Auerbach et al[15] developed the first intermittent-demand nasal cannula, which delivered flow throughout inspiration. For this device, oxygen flow began when negative pressure was detected and when positive pressure ceased. Tiep et al[16] invented the first prototype for contemporary PF oxygen-conserving systems in 1985. This battery-operated device delivered 25 mL at the beginning of inspiration. Their study demonstrated that a PF device could provide the same oxygen saturation to patients using oxygen only at 230 mL/min compared with CF at 2 L/min.

Subsequent studies sought to examine the efficacy and effectiveness of these oxygen-conserving devices. Bower et al[17] treated hypoxemic patients with a constant minute volume demand oxygen-conserving system during rest, exercise, and sleep. This study found that the system produced arterial oxygenation equivalent to CF, but with oxygen savings of up to 60%. Another study by Tiep et al[18] found similar results. During treadmill exercise, hypoxemic patients achieved an equivalent arterial oxygen saturation (S_{aO_2}) of ~90% using oxygen at 1,600 mL/min for steady flow (CF) but only 210 mL/min using the oxygen-conserving device.

In another exercise study, Senn et al[19] established similar S_{pO_2} values for patients at 25–30% maximum work load using both CF and an oxygen-conserving device, with oxygen compared at equivalent flow settings. However, there was no explanation of how these settings were derived (a common trend). This relationship between PF settings and CF was further explored by Carter et al,[20] who compared COPD patient oxygenation using equivalent flow settings of CF and a demand flow device. At rest, a PF setting of 1 was found to be equivalent to CF at 1 L/min (S_{pO_2}), with an oxygen-conserving ratio of 4.6. During exercise, a PF setting of 4 resulted in higher S_{pO_2} than a CF of 4 L/min (S_{pO_2} 92% vs 89%), with an oxygen-conserving ratio of 7.2.

Braun et al[21] again compared COPD patient oxygenation using 5 different PF devices with a constant pulse volume or constant minute volume. At rest, PF was found to be equivalent to CF. During exercise, PF performed worse than CF (S_{pO_2} 89.6% vs 90.3%). Constant pulse volume also performed better than constant minute volume. This study concluded that oxygen-conserving devices vary in their ability to maintain S_{pO_2} during exercise. Roberts et al[22] found PF with a pulse volume of 35 mL with a pulse time of 200 ms to be equivalent to CF at 2 L/min. During the 6-min walk test, PF performed worse than CF (S_{pO_2} 78% vs 81%). PF was also found to be inferior in terms of patients' walking distance given supplemental oxygen. Fuhrman et al[23] found significant differences in demand oxygen delivery system performance for COPD patients. At lower oxygen flows, all four devices improved S_{pO_2} compared with CF. This study found that performance was better with PF at the start of inspiration. Palwai et al[24] showed significant differences in patient oxygenation at a PF setting of 2, with only one device equaling oxygenation with CF at 2 L/min. Kerby et al[25] examined the long-term mean S_{pO_2} values of hospitalized patients with diseases requiring oxygen therapy. This study found that PF and CF yielded equivalent mean S_{pO_2} values over the course of a day and a night.

In all of these studies, we see a consistent lack of information about how equivalent settings are derived. Perhaps it is not surprising then that the large variety of PF devices available today have a wide range of pulse volumes associated with the same setting values. Our study indicates one source of this variability, that is, patients with varying degrees of COPD-induced lung impairment (that variably affect the AR) will require different pulse volumes to attain the same oxygenation. Thus, studies using different kinds of patients will naturally derive different pulse volumes for the same settings.

In summary, when using a nasal cannula to deliver oxygen, the resultant F_{IO_2} could be reduced if the AR is reduced by COPD disease state, depending on whether CF or PF is used. Specifically, CF is subject to both the frequency effect and the reservoir effect, whereas PF (with a constant minute volume device) is subject only to the frequency effect. To further complicate this issue, differing PF characteristics (eg, pulse volume and pulse time) among different devices with the same settings would increase the variability of the results. Settings on PF devices are whole numbers that indicate increasing pulse volumes with increasing setting number (and presumably higher F_{IO_2} delivery). Some manufacturers go so far as to claim that the setting numbers are equivalent to CF values expressed in L/min (eg, a setting of 2 on a portable oxygen concentrator is supposed to deliver the same inspired oxygen as CF at 2 L/min). Although this notion has often been refuted in the scientific literature, it persists in marketing literature and promotional materials.[26] This can have dangerous repercussions on oxygen therapy, as clinicians and patients may believe that higher oxygen flows are delivered than in reality.

Conclusion

Our study shows that oxygen delivery via a nasal cannula is affected not only by the type of device used, that is, constant versus pulse flow, but also by the presence of end-expiratory flow (and hence absence of the AR), as in the case of COPD. As such, our data may help to explain differences in device performance found in human studies. Our data also show that there is no such thing as equivalency between portable oxygen concentrator setting and CF rate for the devices tested. Therefore, the numerical settings on these portable oxygen concentrators should be interpreted as merely indicating increased relative minute volume levels.

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