

Effects of Imposed Resistance on Tidal Volume With 5 Neonatal Nasal Continuous Positive Airway Pressure Systems

Shannon E Cook, Katherine L Fedor CPFT RRT-NPS, and Robert L Chatburn RRT-NPS FAARC

BACKGROUND: Neonates with respiratory distress syndrome are often treated with nasal continuous positive airway pressure (CPAP). Nasal CPAP methods include electronic feedback control, underwater seal, flow opposition, and flow opposition with fluidic flow reversal on expiration. Few studies have compared those modes, and the results have been contradictory. **METHODS:** We compared the effect of resistive load on simulated tidal volume (V_T) with 5 neonatal nasal CPAP systems: Fisher and Paykel nasal CPAP tubing with Maquet Servo-i ventilator in NIV CPAP mode; Cardinal Health AirLife nasal CPAP system; Fisher and Paykel nasal CPAP tubing with water-seal pressure generator; AirLife infant nasal CPAP generator kit; and Hamilton Medical Arabella fluidic nasal CPAP generator. The lung simulator settings were: compliance 0.5 mL/cm H₂O, resistance 125 H₂O/L/s, sinusoidal patient-effort range 6.5–26 cm H₂O, rise 25%, hold 0%, release 25%, respiratory rate 65 breaths/min. We compared the mean values from 10 breaths. **RESULTS:** The mean inspiratory pressure drop and V_T difference (compared to the simulator alone, unloaded) increased with V_T , respectively, from 0.32 cm H₂O to 1.73 cm H₂O, and from –0.04 mL to –0.40 mL. Flow opposition had the smallest pressure drop (from 0.10 cm H₂O to 0.64 cm H₂O, $P < .001$). At V_T of ≤ 6 mL, the bubble nasal CPAP's pressure drop was largest ($P < .001$), whereas at V_T of ≥ 9 mL the electronic nasal CPAP's pressure drop was largest ($P < .001$). All systems except the ventilator did not have an average end-expiratory pressure of the targeted 5 cm H₂O. **CONCLUSIONS:** The differences in these nasal CPAP systems correlate with the differences in unassisted V_T due to loading effects. The ventilator imposed the least load, and the AirLife nasal CPAP system imposed the most. *Key words:* nasal continuous positive airway pressure; CPAP, bubble; work of breathing; lung simulator; neonate. [Respir Care 2010;55(5):544–548. © 2010 Daedalus Enterprises]

Introduction

Nasal continuous positive airway pressure (nasal CPAP) has long been used in neonatal intensive care units as an

adjunct in the treatment of respiratory distress syndrome and chronic lung disease.¹ There are several methods to deliver nasal CPAP and several available nasal CPAP systems. The 4 main methods to create nasal CPAP in neonates are:

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- Electronic feedback control valve (eg, ventilator or AirLife [Cardinal Health, Dublin, Ohio] nasal CPAP system)
- Water-seal (ie, bubble CPAP)²
- Flow opposition (ie, the patient's expiratory flow opposes a constant flow from nasal prongs; conventional neonatal CPAP)²
- Flow opposition with fluidic flow reversal during expiration (ie, a modification of conventional CPAP in which the patient's expiratory flow redirects the CPAP generator flow away from the patient via a fluidic "flip-flop" mechanism)²

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IMPOSED RESISTANCE AND TIDAL VOLUME WITH NEONATAL NASAL CPAP

Table 1. Nasal Continuous Positive Airway Pressure Systems Tested

System Components	Nasal Continuous Positive Airway Pressure System				
	Bubble/Water Seal	Flow Opposition	Flow Opposition Fluidic	Electronic	Electronic
Circuit	Fisher and Paykel	Cardinal Health AirLife	Hamilton Medical Arabella	Fisher and Paykel	Cardinal Health AirLife
Nasal prongs	Fisher and Paykel 4030	Cardinal Health AirLife Small	Hamilton Medical Arabella Small	Fisher and Paykel 4030	Cardinal Health AirLife Small
Flow generator	Flow meter	Flow meter	Flow meter	Servo-i ventilator	Cardinal Health

Little research has been conducted to determine the relative merits of these nasal CPAP systems. Few data are available regarding the resistive work of breathing (WOB) imposed by nasal CPAP systems. Air-flow resistance could be created by the nasal prongs and other system components, which could alter the neonate's ventilatory pattern. If the infant's ventilatory muscle reserve is low, the added resistance might reduce the tidal volume (V_T) and minute ventilation. On the other hand, the infant might simply increase the ventilatory effort to maintain adequate minute ventilation, at the expense of caloric consumption.

In their 2003 review, De Paoli et al posed the question: "What is the most effective source of pressure for CPAP?"¹ We sought to provide an answer. The purpose of this study was to evaluate the effects of 5 nasal CPAP systems on neonatal V_T , using a lung simulator. Specifically, we tested the hypothesis that different nasal CPAP systems have different load characteristics that affect V_T at constant effort.

Methods

Five nasal CPAP systems were evaluated, of 4 types: 2 electronic feed control valves, one water-seal, one flow opposition, and one flow opposition with fluidic reversal (Table 1). The electronic feedback control system drivers included the Servo-i ventilator (Maquet, Bridgewater, New Jersey) in nasal CPAP mode, and the stand-alone AirLife system, which is a device designed only to provide CPAP. It is controlled similarly to the ventilator, by an electronic feedback control. Both generate CPAP with a variable-resistance valve on the expiratory limb. The water-seal system was a traditional bubble CPAP system: the expi-

ratory limb of the circuit tubing is submerged in water, to the depth of desired CPAP (in this case, 5 cm H_2O). During flow opposition CPAP the patient breathes out against the constant flow provided by the flow driver. Fluidic flow opposition CPAP is similar to flow opposition CPAP, but during expiration the flow is diverted down a separate expiratory branch.

Each system was powered with dry air. All systems were set at nasal CPAP of 5 cm H_2O . Nares were simulated with holes drilled into a plastic tube, and the nasal prongs of each system were fitted into the holes, chosen according to the AirLife template for sizing. Whenever possible we used the same nasal prongs with the various tested systems. When that was not feasible, we used binasal prongs with similar nasal tip diameters, to keep a constant flow resistance with the different systems. Likewise, whenever possible we used the same circuits. The nasal prongs used with the bubble and electronic Servo-i system were neonatal prongs (model 4030, Fisher and Paykel, Auckland, New Zealand). With the flow opposition fluidic system we used the Arabella small prongs (Hamilton Medical, Reno, Nevada). With the flow opposition and electronic AirLife system we used AirLife small prongs.

We generated neonatal breathing patterns with a lung simulator (ASL 5000 IngMar Medical, Pittsburgh, Pennsylvania). We set the simulated compliance, resistance, and respiratory rate to represent a 1-kg neonate with respiratory distress syndrome.^{3,4} Because V_T and inspiratory flow rate affect the inspiratory resistive load (ie, the pressure drop due to flow resistance: the higher the V_T , the higher the peak inspiratory flow and the higher the resistance), we programmed the lung simulator to replicate a range of V_T values. We tested simulated muscle pressures in the range 6.5–26 cm H_2O , to generate V_T values in the range 3–12 mL (Table 2).⁵

The lung simulator calculated the mean and standard deviation of the inspiratory pressure drop below the set nasal CPAP level (ie, drop to P_{min} after start of effort, in cm H_2O) in all the experimental conditions except bubble nasal CPAP, which produced too much noise in the

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Table 2. Lung Simulator Settings

Uncompensated residual capacity (L)	0.200
Resistance (cm H ₂ O/L/s)	125
Compliance (mL/cm H ₂ O)	0.5
Filter type	5 Hz Butterworth
Breath detection setting (mL)	1.3
Muscle-pressure waveform	Sinusoidal
Simulated muscle pressure range (cm H ₂ O)	6.5–26
Respiratory rate (breaths/min)	60
Rise (% of cycle time)	25
Hold (% of cycle time)	0
Release (% of cycle time)	25

airway pressure signal to use the automated calculation. So, on the bubble nasal CPAP pressure-time graph we drew (by eye) a line representing a low-pass filtered signal (eliminating most of the pressure oscillations) and calculated the pressure drop from that. The pressure drop was defined as the end-expiratory pressure minus the minimum pressure during inspiration (drop to P_{min} from start of effort in the lung simulator’s post-run analysis window).

In the lung simulator’s data-analysis window we activated the 5-Hz Butterworth filter, which is a form of a low-pass digital filter to remove signal noise. We chose not to use the available moving-average filter, because it did not seem as effective as the Butterworth filter. The Butterworth filter was particularly useful when analyzing the bubble nasal CPAP data, in which there was a lot of noise. The breath-detection threshold was set at 1.3 mL for both the breath-start volume threshold and the expiratory-start volume threshold. All systems except for bubble nasal CPAP (which had a noisy volume signal) could have been analyzed with a breath-detection threshold of 0.7 mL, but for consistency and comparability we treated all the data with the 1.3 mL threshold.

The lung simulator calculated the mean V_T from at least 10 breaths in each experimental condition. We compared the mean pressure-drop and V_T values with 1-way analysis of variance. Differences associated with *P* < .05 were considered significant.

Results

Figures 1–5 show representative volume and pressure tracings. Resistive loading from the nasal CPAP systems is indicated by the pressure drop during inspiration.

The airway-pressure waveform from the ventilator requires further explanation. Figure 5 shows that after an initial pressure drop from about 5.25 cm H₂O to about 4.25 cm H₂O, airway pressure rose during inspiration, to about 5.75 cm H₂O. Inspiration is distinguished from expiration by the peak volume (see Fig. 5). The pressure rise during inspiration indicates that the ventilator provides a

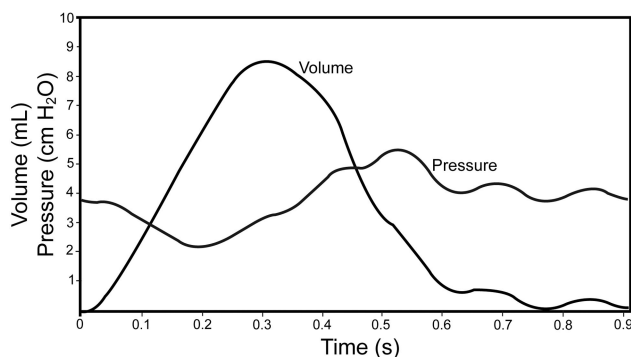


Fig. 1. Pressure and volume waveforms from bubble nasal continuous positive airway pressure.

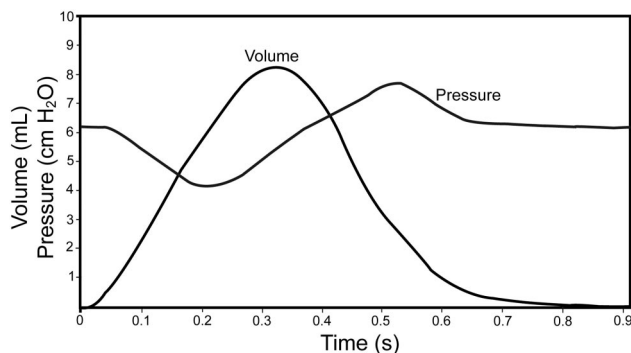


Fig. 2. Pressure and volume waveforms from electronic nasal continuous positive airway pressure.

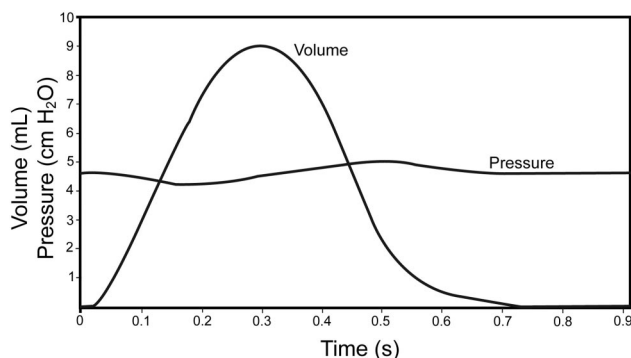


Fig. 3. Pressure and volume waveforms from flow opposition nasal continuous positive airway pressure.

small amount of pressure assist (ie, the ventilator does work on the patient), despite a pressure-support setting of zero. This is due to a slight overcompensation in the ventilator’s pressure-control algorithm.

Because the data were generated with a mechanical model, measurement variability was negligible. The standard deviations of the pressure-drop and V_T values were usually zero. Even in the noisy data from the bubble CPAP system the standard deviation was only 0.18 mL.

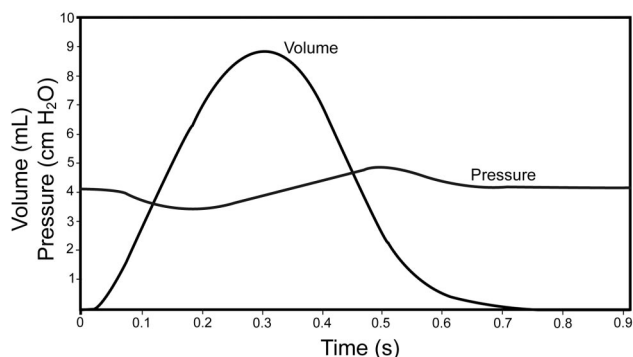


Fig. 4. Pressure and volume waveforms from flow opposition fluidic nasal continuous positive airway pressure.

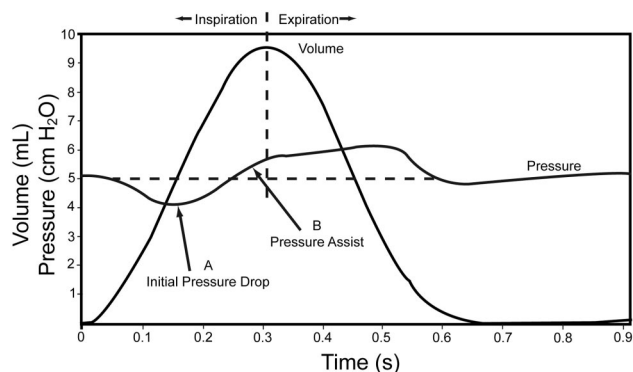


Fig. 5. Pressure and volume waveforms from ventilator nasal continuous positive airway pressure.

Table 3. Mean Inspiratory Pressure Drop

	Reference Tidal Volume (mL)			
	3	6	9	12
Inspiratory Pressure Drop (cm H ₂ O)				
Electronic	0.40	1.06	1.99	3.06
Bubble	0.62	1.22	1.70	2.47
Ventilator	0.28	0.57	0.95	1.45
Flow opposition fluidic	0.22	0.42	0.72	1.02
Flow opposition	0.10	0.21	0.42	0.64
<i>P</i>	< .001	< .001	< .001	< .001
Mean pressure drop	0.32	0.70	1.16	1.73

Table 3 shows the mean pressure-drop values, which, as expected, with all the systems, increased with V_T . The flow opposition system had the smallest inspiratory pressure drop (0.10 cm H₂O at V_T of 3 mL, up to 0.64 at V_T of 12 mL, $P < .001$): a difference that is probably not clinically important. The bubble CPAP system was the most difficult with which to accurately measure the pressure drop because of the signal noise generated by the vibrations from the bubbling water. At V_T of 3 mL and 6 mL, bubble CPAP produced the largest pressure drop

Table 4. Tidal Volume

	Reference Tidal Volume (mL)			
	3	6	9	12
Tidal Volume (mL)				
Ventilator	3.00	6.30	9.46	12.50
Flow opposition	3.00	6.00	8.90	11.90
Flow opposition fluidic	2.96	5.90	8.80	11.70
Bubble	2.94	5.54	8.34	11.08
Electronic	2.90	5.60	8.26	10.80
<i>P</i>	.12	< .001	< .001	< .001
Mean volume drop	-0.04	-0.13	-0.25	-0.40

($P < .001$). At V_T of 9 mL and 12 mL, the electronic CPAP system produced the largest pressure drop ($P < .001$).

Despite a CPAP setting of 5 cm H₂O, the CPAP with the Cardinal Health electronic CPAP system was 6 cm H₂O, with a V_T of 9 mL. This system was not the only one to give an inaccurate CPAP with a V_T of 9 mL. With the bubble nasal CPAP system, the actual CPAP was 4 cm H₂O rather than the set 5 cm H₂O. The flow opposition and flow opposition fluidic systems also both failed to create the exact CPAP of 5 cm H₂O. The flow opposition peaked at 5 cm H₂O and averaged approximately 4.6 cm H₂O. The flow opposition fluidic system averaged a CPAP of 4.25 cm H₂O, and peaked at 4.8 cm H₂O.

Table 4 shows the inspiratory V_T data. As with the pressure-drop data, the mean V_T difference, compared to unloaded breathing, with all the systems, except the ventilator, increased with V_T (difference range -0.04 to -1.2 mL). The V_T values were statistically different across the systems at all the simulated-muscle-pressure settings except at the 3 mL reference V_T (ie, that obtained with the unloaded system, that is, when the lung simulator was run with nothing attached). Because the ventilator provided a small amount of inspiratory pressure (0.65 cm H₂O), the V_T was slightly higher than during unloaded breathing (0.3 to 0.5 mL).

Discussion

CPAP has been used to support neonatal breathing since 1971.⁶ Determined benefits of CPAP include better oxygenation, reduced airway resistance, reduced apnea, and enhanced absolute lung volumes.¹ Predictably, practice and use of CPAP have changed over time. Various CPAP delivery methods have been used, but currently, the most popular is nasal CPAP, probably because of the easy access provided. The use of CPAP, too, has changed. In neonates the main uses of CPAP include preventing extubation failure and managing apnea.

This need is recognized by other respiratory professionals as well. In their review, DePaoli et al¹ summarized the current understanding of CPAP for neonates as of 2003,

and suggested topics for further research, including determining the most effective source of pressure for CPAP. Our study was designed to address this question.

Lee et al⁷ conducted a small study of bubble nasal CPAP and continuous flow opposition nasal CPAP in pre-term infants (mean weight $1,350 \pm 390$ g). Patients identified as ready for extubation remained intubated while being placed on alternating nasal CPAP delivery methods for predetermined periods of time. Lee et al concluded that bubble nasal CPAP reduced minute volume and respiratory rate, with a presumed decrease in the WOB. Accurate V_T measurements could not be obtained during bubble nasal CPAP, so resistance and compliance values could not be calculated; they could make no conclusions regarding the actual WOB. Pandit et al⁸ found that WOB was lower with flow opposition nasal CPAP than with electronic flow control nasal CPAP. Liptsen et al⁹ showed that resistive WOB and respiratory rate were greater with bubble nasal CPAP than with fluidic flow opposition nasal CPAP, and speculated that “the more labored and asynchronous breathing seen with bubble nasal CPAP may lead to higher failure rates over the long term.”⁹

Studies with human subjects often yield confusing results, due to differences between subjects and study designs. For example, subjects in Lipsten’s group⁹ were smaller ($1,048 \pm 241$ g) and younger (gestational age 28.1 ± 1.6 weeks) than those in Lee et al’s study ($1,350 \pm 390$ g, gestational age 30.7 ± 1.8 weeks).⁷ Differences in the equipment used and accuracy in measurement could also be a factor. Our data supports the previous human studies by Pandit et al⁸ and Lipsten et al.⁹ Our results suggest that bubble CPAP reduces V_T and minute volume, similar to the findings by Lee et al,⁷ except that they showed a much greater effect (39%) with no apparent effect on gas exchange, perhaps due to altered gas-exchange mechanisms or changes in lung compliance. By using a lung simulator we were able to apply a standardized simulated-muscle-pressure waveform, thus removing inter-subject and intra-subject ventilatory pattern variability. Our study is the first to compare all major CPAP systems simultaneously under controlled conditions to identify just the mechanical load effects on the breathing pattern.

Another relevant study, by Moa et al,¹⁰ showed that a new design for flow opposition CPAP reduced imposed WOB, compared to a threshold-resistor CPAP device. Those findings were confirmed by Klausner et al.¹¹

A 1-kg neonate with respiratory distress syndrome may have an unassisted spontaneous V_T of about 3 mL. At that V_T there was no difference among the systems we tested in the effect of resistive load on ventilation (see Table 4). However, larger infants, with V_T of ≥ 4 mL, may be affected by the nasal CPAP system. In vivo research is needed to test that hypothesis.

It is interesting to note that the only system that actually averaged the desired CPAP was the ventilator. Neverthe-

less, the pressure-targeting errors we recorded are probably not clinically important.

Conclusions

The studied nasal CPAP systems affected the simulated V_T when inspiratory effort was held constant. The clinical implications of that finding remain to be tested. The application of CPAP is challenging in itself, and when faced with determining a suitable nasal CPAP system, personal bias, interface issues, and simplicity often may take precedence. Our results may offer a logical way to select a system: for neonates weighing ≤ 1 kg, there is no difference between the systems. For neonates weighing > 1 kg, ventilator-derived nasal CPAP may offer an advantage. Rationally, clinical testing must occur to confirm this determination before putting it to use. However, one must consider the cost/benefit ratio and perhaps reserve the ventilator-derived method for the neonates who are at the greatest risk of nasal CPAP failure.

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