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RESEARCH ARTICLE

In-vitro Bench Evaluation of a Novel Adjus-Amplitude Bubble CPAP System for Neonatal Respiratory Support

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ABSTRACT

Respiratory distress syndrome (RDS) in preterm neonates remains a major contributor to neonatal mortality worldwide, with India bearing a disproportionate share of the global preterm birth burden, over 3 million births annually, more than 20% of the global total. Bubble continuous positive airway pressure (B-CPAP) is a widely used, low-cost mode of non-invasive respiratory support for these infants, but in every existing design, oscillatory amplitude is mechanically coupled to bias flow and therefore to the mean delivered pressure. Therefore, clinicians cannot raise one without raising the other. Humming BEE B-CPAP system decouples these variables through an independently controlled secondary extra-flow channel introduced into an enlarged bubble-generating chamber, conceptually importing an amplitude-control variable analogous to high-frequency oscillatory ventilation (HFOV) into the bubble CPAP platform.

In-vitro bench testing was performed using calibrated airway-pressure transducer and oscilloscope-based measurements across a range of bias-flow (5–8 L/min) and extra-flow (0–14 L/min) settings at a fixed CPAP set point of 5 cm H_2O. Introducing graded extra flow produced an approximately monotonic rise in oscillatory pressure amplitude of roughly 70 to 100% while delivered CPAP remained at its set point across the majority of conditions tested. Dominant oscillation frequency rose initially with extra flow before falling at higher settings, consistent with progressive bubble coalescence and the Minnaert relationship, yielding a favorable high-amplitude, reduced-frequency operating point at the upper end of the extra-flow range.

Preliminary observational clinical experience of Humming Bee B-CPAP in approximately 50 neonates with RDS has suggested that the device is feasible and well tolerated in routine neonatal practice.

These findings support a physiological rationale, grounded in bubble dynamics and HFOV principles, for why this capability may aid CO_2 clearance in neonates, particularly those with borderline rising $PaCO_2$ or in the acute low-compliance phase of RDS.

Keywords: Respiratory distress syndrome, B-CPAP, Oscillatory amplitude; High-frequency oscillatory ventilation; Non-invasive respiratory support; Minnaert relationship, $PaCO_2$

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INTRODUCTION

Preterm birth and respiratory complications arising due to it are the leading drivers of neonatal morbidity and mortality worldwide. According to WHO's estimates, around 13.4 million infants were born preterm in 2020, accounting for roughly one in ten live births.[1] This burden falls disproportionately in LMICs like India. In 2020, an estimated 3.02 million preterm infants were born in India, representing more than 20% of all

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preterm births worldwide.[2] The national preterm birth rate was approximately 13% of live births, substantially higher than the global average.[2]

One of the leading and most immediate complications of prematurity is respiratory distress syndrome (RDS), caused due to surfactant deficiency and structural lung immaturity. Therefore, the need to support safe exchange of gases in neonates at a low cost has given rise to decades of innovation in non-invasive respiratory technology. Continuous positive airway pressure (CPAP) has now become an important technology when it comes to non-invasive respiratory support in neonatal intensive care. By maintaining a constant distending pressure throughout the respiratory cycle, CPAP recruits and stabilizes collapsed alveoli, preserves functional residual capacity, conserves endogenous surfactants, improves ventilation-perfusion matching, and reduces the work of breathing.[3] This lowers the need for endotracheal intubation and invasive mechanical ventilation, thereby reducing the risk of complications associated with them. CPAP can be generated in several ways. Ventilator-derived CPAP partially occludes the expiratory limb of a mechanical ventilator circuit to hold a set pressure. Bubble CPAP, in contrast, generates pressure by submerging the distal end of the expiratory limb under a column of water. As gas escapes, it bubbles through the water, and the depth of submersion sets the mean distending pressure.[4] The clinically important difference between the two methods is that while ventilator CPAP delivers a comparatively smooth and constant pressure, bubble CPAP superimposes broadband, low-amplitude, high-frequency pressure oscillations generated by the bubbling process itself.[4-5] These oscillations are transmitted retrograde through the expiratory limb to the nasal interface and, from there, into the infant's airway, where they can be felt as fine vibration of the chest wall.[5]

REVIEW OF LITERATURE

In a randomized crossover trial of infants ready for extubation, Lee et al. found that bubble CPAP was associated with lower respiratory rate and lower minute ventilation than ventilator-derived CPAP at an equivalent mean pressure.[5] This study suggested that the oscillatory component itself contributes to more efficient gas exchange rather than simply reflecting a difference in distending pressure. In preterm lambs with induced respiratory distress, bubble CPAP produced superior lung volume recruitment and gas exchange compared with constant-pressure CPAP, with the authors proposing that the noisy oscillatory waveform promotes airway-opening events through a mechanism resembling stochastic resonance, particularly when lung compliance is low and atelectasis is present.[6] A 2023 Cochrane systematic review of 15 randomized trials similarly found that bubble CPAP, compared with ventilator- or infant flow driver-derived CPAP, was associated with a lower risk of CPAP treatment failure, albeit with an increased risk of nasal trauma.[7] Beyond this physiological advantage, bubble CPAP does not require an electrically powered oscillator or proprietary driver circuit. They only require a gas source, a humidifier, and a water column, making it markedly simpler, cheaper, and more robust to power and supply-chain interruptions than ventilator-derived or true high-frequency oscillatory systems.[7] This characteristic becomes particularly important in the resource-constrained settings that bear the greatest share of the global preterm-birth burden.

Bench studies have characterized what drives oscillatory amplitude and frequency of vibrations produced through the Bubble CPAP. Pillow and Travadi found bias flow and lung compliance both shape the magnitude and frequency of transmitted pressure oscillation, with higher bias flow increasing oscillation magnitude.[8] Wu et al. varied expiratory-limb diameter, submersion depth, and generator-bottle diameter, finding that a larger and more deeply submerged limb increased amplitude while lowering dominant frequency.[9] Separately, Iranpour et al.'s RCT of nasal HFOV versus nasal CPAP found shorter duration of non-invasive support, fewer intubations, and less intraventricular hemorrhage with nHFOV, supporting the value of superimposed high-frequency oscillation on a CPAP platform.[10] Outside CPAP specifically, Wyss Institute conducted a study in which they applied low-level stochastic vibration via a therapeutic mattress to 36 preterm infants and found approximately 50% reduction in apnea incidence and less severe bradycardia, independent of caffeine.[11]

Together, these studies demonstrate that the therapeutic performance of bubble CPAP is closely linked to the characteristics of the pressure oscillations generated by the device, highlighting the need to understand how device design influences oscillatory behavior

DEVICE DESCRIPTION

In every conventional bubble CPAP system studied to date, amplitude and bias flow are coupled. The only way to generate more vigorous bubbling, and therefore higher-amplitude oscillations, is to increase the bias gas flow, but doing so simultaneously raises the mean delivered pressure and the infant's resistive work of breathing, creating a therapeutic ceiling.[8] No commercially available bubble CPAP device allows the oscillatory amplitude to be titrated as an independent variable, separate from the preset CPAP level.

The novel Humming BEE bubble CPAP system is designed specifically to overcome this constraint. Developed as a CDSCO-approved Class C medical device under the Medical Devices Rules, 2017, Humming BEE is licensed for the provision of respiratory support in neonates and infants with conditions such as respiratory distress syndrome (RDS) and apnea of prematurity. By introducing a secondary, independently controlled extra-flow channel into an enlarged bubble-generating chamber, the device allows oscillation amplitude to be increased and dominant frequency to be lowered without increasing the bias flow or the preset CPAP. This is, in effect, an attempt to import a high frequency oscillatory ventilator (HFOV) like control variable into a bubble CPAP platform. Table 1 presents a comparison between

1: Comparison between conventional bubble CPAP & Humming BEE Bubble CPAP

Sr No	*Standard Features of B-CPAP*	*Features provided in Traditional B-CPAP*	*Humming BEE B-CPAP*
1	Air Oxygen Blender	YES	YES
2	Flow meter to set the blended gas flow.	YES Provided	YES
3	CPAP generation bottle.	YES	YES
4	Water capacity of CPAP generation bottle.	450 to 500 ml	1000 ml
4	CPAP generation Tube diameter of 10 mm.	YES	YES Provided, CPAP generation tube diameter of 12 mm.
5	Unit mounted on trolley.	YES	YES
6	Pressure gauge to indicate the delivered CPAP	Optional Depends on the manufacturer.	YES Digital Display Provided which shows frequency, amplitude and delivered CPAP
7	Approx Measured Frequency	20 to 40 Hertz Fixed.	10 to 20 Hertz, Varies automatically.
8	Adjus amplitude and frequency	NO.	YES The dominant frequency can be increased or decreased by adjusting the amplitude.

NOTE: For comparative purposes, internationally recognized brands, including Fisher & Paykel Healthcare (F&P) and Hamilton Medical, were considered. Products from Chinese manufacturers were excluded from the comparison.

the conventional Bubble CPAP (B-CPAP) system and the Humming BEE B-CPAP system.

THEORETICAL BACKGROUND

GENERATION OF MEAN DISTENDING PRESSURE

In a bubble CPAP circuit, the distal expiratory limb is submerged to a depth h beneath the surface of a water column. To escape, the bias gas flow must overcome the local hydrostatic pressure at that depth, which is approximately given by:

$$P = pgh$$

where,

ρ: density of water

g acceleration due to gravity

h: depth to which the expiratory tube is submerged

This hydrostatic pressure sets the mean component of the airway pressure i.e. the clinically prescribed CPAP level.

BUBBLE FORMATION

A bubble exists because the internal gas pressure at the end of the expiratory tube exceeds the local pressure within the surrounding liquid. This local pressure is determined primarily by the atmospheric pressure at the water surface and the hydrostatic pressure generated by the depth of submersion. In addition, the gas pressure must overcome the inward force exerted by surface tension of water at the gas-liquid interface. For a spherical bubble of radius R in a liquid of surface tension γ, the additional pressure required to balance the surface tension is described by the Young-Laplace relationship:

$$\Delta p = \frac{2y}{R}$$

where,

Δp: pressure difference between the interior of the bubble and the surrounding liquid.

γ: surface tension of water

R: Radius of the bubble

Therefore, bubble formation in a bubble CPAP system is governed by both hydrostatic pressure, which establishes the mean distending pressure, and surface-tension forces, which influence bubble size, formation, and detachment dynamics.

BUBBLE DYNAMICS

Once a bubble is formed at the submerged expiratory limb, it undergoes a dynamic sequence of growth, detachment, ascent through the water column, and eventual collapse at the liquid surface. Throughout this process, the bubble continuously exchanges energy with the surrounding liquid, generating transient pressure fluctuations that are transmitted through the CPAP circuit to the infant's airways. These pressure fluctuations constitute the oscillatory component of bubble CPAP. The motion of a spherical bubble in a liquid can be described by the Rayleigh–Plesset equatio.[12-13]

$$p\left[R\left(\frac{d^2R}{dt^2}\right) + \frac{3}{2}\left(\frac{dR}{dt}\right)^2\right] = P_B(t) - p_\infty - \left(\frac{2\sigma}{R}\right) - \frac{4n}{R}\left(\frac{dR}{dt}\right)$$

where,

p : density of the surrounding liquid

R : instantaneous bubble radius

$\frac{d^2R}{dt^2}$: radial acceleration

$\frac{dR}{dt}$: radial velocity

$P_B(t)$: pressure inside the bubble

P_∞ : pressure in the liquid far from the bubble

σ : surface tension

n : dynamic viscosity of the liquid

For small-amplitude oscillations about equilibrium, this equation reduces to a simple harmonic oscillator yielding the bubble's natural frequency. Minnaert demonstrated that the natural oscillation frequency of a spherical gas bubble is inversely proportional to its radius.[14]

$$f_0 = \frac{1}{2\Pi R}\sqrt{\left[\frac{3\kappa P_0}{\rho}\right]}$$

where,

f_0: natural frequency of the oscillation

R : radius of the bubble

κ : polytropic exponent of gas

P_0 : ambient pressure

ρ : density of the surrounding liquid

This relationship indicates that the natural oscillation frequency of a bubble is inversely proportional to its radius. Therefore, factors that increase the bubble size like increasing the diameter of the expiratory limb results in lower frequency oscillations and greater pressure amplitudes.[9]

The continuous formation, detachment, and oscillation of bubbles therefore generate a spectrum of pressure fluctuations superimposed on the mean CPAP pressure. These oscillations are believed to contribute to the unique physiological effects of bubble CPAP, including enhanced alveolar recruitment and improved gas mixing.

FORCED OSCILLATION AND OSCILLATORY AMPLITUDE AUGMENTATION

In conventional bubble CPAP, clinicians have limited ability to modify oscillatory amplitude without simultaneously altering the mean CPAP pressure or increasing bias flow. In the Humming BEE bubble CPAP, the introduction of an additional gas stream into the bubble generation chamber results in increased bubble interactions and larger pressure fluctuations without altering the mean CPAP pressure set by the clinicians or increasing the bias flow. When two bubble streams coexist within the same liquid medium, interactions between adjacent bubbles can result in coalescence, producing larger composite bubbles. According to the Minnaert relationship, larger bubbles possess lower natural oscillation frequencies than smaller bubbles. Simultaneously, larger bubbles displace a greater volume of liquid during formation and detachment, generating larger pressure fluctuations within the circuit.

The oscillatory component of the airway pressure waveform may therefore be represented as:

$$P(t) = P_{CPAP} + \Delta P(t)$$

where,

P_{CPAP} : mean distending pressure.

ΔP: time-varying oscillatory pressure component generated by bubble dynamics.

The magnitude of these oscillations can be quantified by the pressure excursion

$$\Delta P = P_{max} - P_{min}$$

where, P_{max} and P_{min} denote the maximum and minimum pressures measured during an oscillatory cycle.

Consequently, the introduction of a secondary bubble stream provides a mechanism for independently modifying the oscillatory component of the pressure waveform while preserving the mean distending pressure delivered to the patient.

PHYSIOLOGICAL BASIS FOR INCREASED AMPLITUDE AND REDUCED FREQUENCY

Bubble CPAP's oscillatory component is frequently compared, mechanistically, to HFOV, in which gas exchange is achieved through tidal volumes at or below the anatomical dead space delivered at supraphysiologic frequency.[15]

The physiological rationale for increasing oscillatory amplitude while reducing dominant frequency is derived from principles established in high-frequency oscillatory ventilation (HFOV). In HFOV, carbon dioxide elimination is influenced by both oscillatory frequency and the oscillatory tidal volume delivered to the lung.[16-17] This relationship is commonly described by the diffusion coefficient for carbon dioxide.

$$DCO_2 \propto f \times V_T^2$$

where,

f : oscillation frequency

V_T : oscillatory tidal volume.

Because tidal volume contributes as a squared term, relatively small increases in delivered oscillatory volume can substantially improve carbon dioxide clearance.

Consequently, when hypercapnia occurs during HFOV, clinicians typically increase oscillatory amplitude, decrease oscillatory frequency, or employ a combination of both interventions to increase oscillatory tidal volume and enhance carbon dioxide elimination. The same can now be done in a Humming BEE bubble CPAP which otherwise was not possible to do in a conventional CPAP.

MATERIALS AND METHODS: *IN-VITRO* BENCH TESTING

Configuration

The prototype circuit comprised a blended-gas inspiratory supply, a bubble-generating chamber filled with 1,000 mL of water, a bias-flow expiratory limb of 12 mm internal diameter, and an independent extra-flow delivery tube of 16 mm internal diameter immersed to a depth of 11 cm in the same chamber. The bias-flow expiratory limb was connected, via the inspiratory limb and nasal-interface analogue, to a fixed-compliance test lung of 50 mL capacity, used as a simplified, reproducible surrogate for the neonatal respiratory system. The nominal CPAP set point was held at 5 cm H_2O throughout testing by maintaining a constant bias-flow expiratory-limb submersion depth.

Instrumentation

Two independent measurement methods were used to characterize the oscillatory pressure waveform.

(i) A calibrated airway-pressure transducer recorded the P_{max} and P_{min} of the oscillatory waveform at the test-lung airway opening; the oscillatory amplitude was calculated as $\Delta P = P_{max} - P_{min}$, and the mean of the waveform was taken as the delivered CPAP.

(ii) The pressure signal was additionally coupled to a calibrated cathode-ray oscilloscope (CRO), a test instrument that graphically displays a varying voltage signal against a time base, allowing direct visualization of the waveform and measurement of its maximum voltage (V_{max}), minimum voltage (V_{min}), and peak-to-peak amplitude ($\Delta V = V_{max} - V_{min}$), used as an independent, electronically derived cross-check on the transducer-based ΔP values. Dominant oscillation frequency was determined from the CRO trace and corroborated by direct cycle counting, with a resolution of ±1 Hz.

Experimental Protocols

Protocol A

bias flow alone (no extra flow). With the extra-flow valve closed (0 L/min), bias flow was varied from 5 to 12 L/min in stepwise increments at a fixed CPAP set point of 5 cmH₂O, and dominant frequency, ΔP and delivered CPAP were recorded at each step. This protocol established the baseline behavior of the unmodified bias-flow circuit and the frequency/pressure ceiling achievable by bias flow alone.

Protocol B

combined bias-flow / extra-flow grid (manometric). For each of four fixed bias-flow settings (5, 6, 7, and 8 L/min), extra flow was increased from 0 to 14 L/min in 2 L/min increments (eight steps per bias-flow level; 32 conditions in total). At each of the 32 conditions, ΔP, delivered CPAP, and observed dominant frequency were recorded using the airway-pressure transducer.

Protocol C

combined bias-flow / extra-flow grid (oscilloscope cross-validation). The identical 4 × 8 grid of bias-flow and extra-flow settings used in Protocol B was repeated with the CRO as the measurement instrument, recording V_{max} , V_{min}, and ΔV, and the corresponding P_{max} , P_{min}, and ΔP derived from the calibrated voltage-to-pressure relationship of the transducer–oscilloscope chain. This protocol served as an independent, electronically instrumented replication of Protocol B.

2: Effect of bias flow on dominant frequency, oscillatory amplitude, and delivered CPAP with the extra-flow channel closed

Bias Flow (L/min)	*f (Hz ± 1)*	*ΔP (cmH₂O)*	*Delivered CPAP (cmH₂O)*
5	25–26	8	5
6	24–25	9	6
7	25–26	10	6–7
8	25–26	10–11	7–8
10	25	10–11	7–8
12	24	11–12	8–9

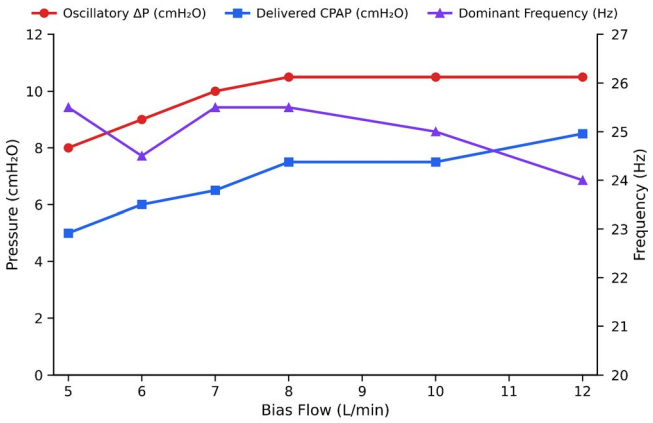

Figure 1: Baseline Characterization of Oscillatory Frequency, Amplitude, and Delivered CPAP as a Function of Bias Flow

EXPERIMENTAL RESULTS

Effect Of Bias Flow Alone On Frequency And Delivered Cpap (Protocol A)

With the extra-flow channel closed, increasing bias flow from 5 to 12 L/min produced only a modest, in dominant frequency and a small upward drift in delivered CPAP confirming that, in the absence of the extra-flow innovation, bias flow alone offers only a narrow and pressure-coupled range over which the oscillatory characteristics of the circuit can be altered. (Table 2, Figure 1)

Effect Of Graded Extra Flow Oscillatory Amplitude (Protocol B)

Across all four bias-flow conditions as shown in Table 3,

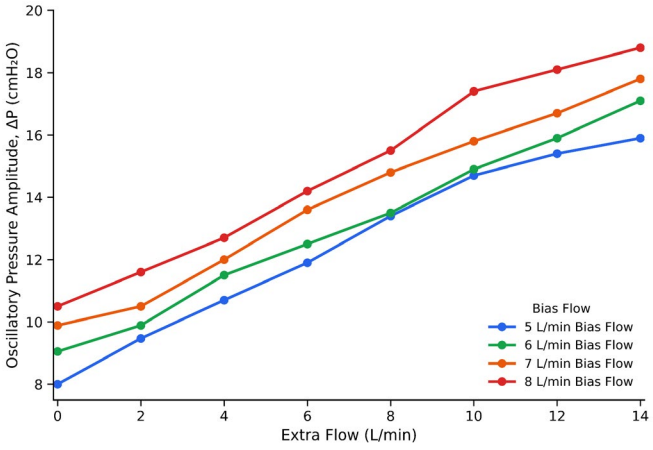

Figure 2: Modulation of Oscillatory Amplitude by Graded Extra Flow

3: Oscillatory pressure amplitude and dominant frequency as a function of bias flow and graded extra flow (set CPAP = 5 cmH₂O; CPAP-generation tube ID 12 mm, extra-flow tube ID 16 mm immersed to 11 cm).

Extra Flow (L/min)	*Bias flow 5 L/min*		*Bias flow 6 L/min*		*Bias flow 7 L/min*		*Bias flow 8 L/min*	
	ΔP	f	ΔP	f	ΔP	f	ΔP	f
0	8	10	9.05	10	9.89	10	10.5	14
2	9.47	15	9.89	15	10.5	15	11.6	16
4	10.7	18	11.5	18	12	18	12.7	18
6	11.9	20	12.5	20	13.6	20	14.2	20
8	13.4	19	13.5	19	14.8	19	15.5	19
10	14.7	20	14.9	20	15.8	20	17.4	20
12	15.4	14	15.9	14	16.7	14	18.1*	16
14	15.9	12	17.1	12	17.8	12	18.8*	14

NOTE: *Delivered CPAP began to drift above the 5 CmH₂O set point (to 6–8 cmH₂O) only at these two highest-bias / highest-extra-flow conditions.*

introducing and progressively increasing extra flow produced a consistent, approximately monotonic rise in oscillatory amplitude. At a bias flow of 7 L/min, for example, ΔP rose from 9.89 cmH₂O at an extra flow of 0 L/min to 17.8 CmH₂O at an extra flow of 14 L/min an increase of roughly 80% while the delivered CPAP remained at the 5 CmH₂O set point throughout the full extra-flow range. The same pattern was observed at bias flows of 5 and 6 L/min. At the highest bias-flow setting tested (8 L/min), delivered CPAP began to drift upward (6–8 cmH₂O) only once extra flow exceeded 12 L/min, identifying a practical upper boundary of the tested operating envelope (Table 3, Figure 2).

Unlike amplitude, the dominant oscillation frequency did not rise monotonically with extra flow. At a representative bias flow of 7 L/min, frequency rose from 10 Hz at zero extra flow to a peak of approximately 20 Hz at an extra flow of 6 to 10 L/min, before falling to 14 Hz at an extra flow of 12 to 14 L/min (Table 3, Figure 3). At low-to-moderate extra flow, the additional bubble stream adds high frequency oscillations and at higher extra flow, increasing bubble coalescence produces progressively larger composite bubbles whose natural frequency, by the Minnaert relationship, falls as bubble radius increases.

Effect Of Graded Extra Flow On Oscillatory Amplitude (Protocol C)

The complete CRO-based dataset, recorded across the same 4 × 8 bias-flow/extra-flow grid used in Protocol B, is reported in Tables 4 and Table 5.

At each condition, V_{max} and V_{min}, are the maximum and minimum oscilloscope voltages, ΔV their difference, and P_{max} , P_{min} and ΔP the corresponding pressures derived from the calibrated voltage-to-pressure transducer relationship. (Table 4, Table 5, Figure 4). Across all four bias-flow conditions, ΔP rose consistently with extra flow, from 8.0 to 10.5 cmH₂O at zero extra flow to 15.9 to 18.8 cmH₂O at 14 L/min, numerically matching the corresponding values in 3 to within rounding error and confirming the internal consistency of the two independent instrumentation methods. The data shows an overall reduction in P_{min} as extra flow increased (e.g., from 2.421 cmH₂O to 0.526 – 1.158 cmH₂O across the four bias-flow

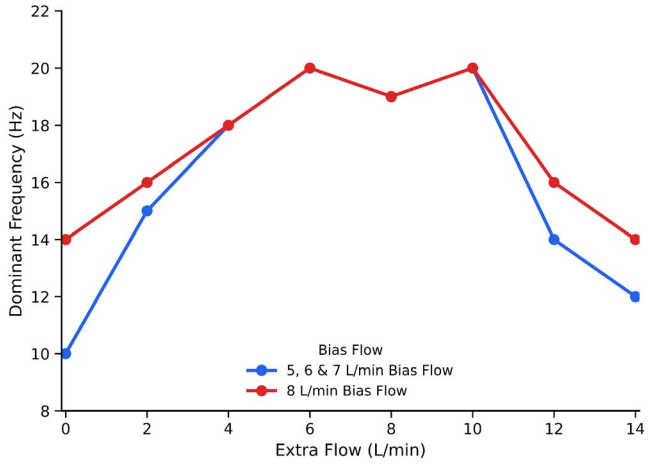

Figure 3: Change in Frequency due to Graded Extra Flow

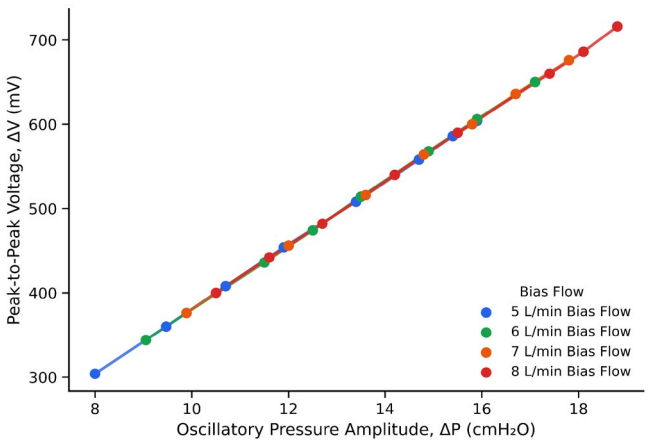

Figure 4: Concordance between transducer- and oscilloscope-derived pressure measurements

4: CRO-derived voltage and pressure parameters at bias flows of 5 L/min and 6 L/min across the full extra-flow range (Vmax, Vmin, ΔV in mV; Pmax, Pmin, ΔP in cmH_2O)

Extra Flow (L/min)	*Bias flow 5 L/min*						*Bias flow 6 L/min*					
	V_{max}	V_{min}	ΔV	P_{max}	P_{min}	ΔP	V_{max}	V_{min}	ΔV	P_{max}	P_{min}	**ΔP**
0	896	592	304	10.42	2.421	8	936	592	344	11.47	2.421	9.05
2	952	592	360	11.89	2.421	9.47	968	592	376	12.32	2.421	9.89
4	984	576	408	12.74	2	10.7	1020	584	436	13.68	2.221	11.5
6	1030	576	454	13.95	2	11.9	1050	576	474	14.47	2	12.5
8	1060	552	508	14.74	1.368	13.4	1090	576	514	15.53	2	13.5
10	1110	552	558	16.05	1.368	14.7	1120	552	568	16.32	1.368	14.9
12	1130	544	586	16.58	1.158	15.4	1150	544	606	17.11	1.158	15.9
14	1140	536	604	16.84	0.947	15.9	1170	520	650	17.63	0.526	17.1

5: CRO-derived voltage and pressure parameters at bias flows of 7 L/min and 8 L/min across the full extra-flow range (Vmax, Vmin, ΔV in mV; Pmax, Pmin, ΔP in cmH_2O).

Extra Flow (L/min)	*Bias flow 7 L/min*						*Bias flow 8 L/min*					
	V_{max}	V_{min}	ΔV	P_{max}	P_{min}	ΔP	V_{max}	V_{min}	ΔV	P_{max}	P_{min}	**ΔP**
0	968	592	376	12.32	2.421	9.89	984	584	400	12.74	2.211	10.5
2	992	592	400	12.95	2.421	10.5	1010	568	442	13.42	1.789	11.6
4	1040	584	456	14.21	2.211	12	1050	568	482	14.47	1.789	12.7
6	1100	584	516	15.79	2.211	13.6	1100	560	540	15.79	1.579	14.2
8	1140	576	564	16.84	2	14.8	1150	560	590	17.11	1.579	15.5
10	1160	560	600	17.37	1.579	15.8	1220	560	660	18.95	1.579	17.4
12	1180	544	636	17.89	1.158	16.7	1230	544	686	19.21	1.158	18.1
14	1220	544	676	18.95	1.158	17.8	1260	544	716	20	1.158	18.8

conditions), indicating that part of the amplitude gain reflects a deepening of the oscillatory trough alongside the rise in P_{max} . These findings suggest that both bias flow and extra flow contribute to the generation of oscillatory amplitude, with the effect of extra flow persisting across all bias-flow settings. Inspection of the pressure waveforms indicates that V_{min} falls modestly compared to the progressive rise in V_{max} . Thus the growth in ΔP is therefore driven predominantly by the rising peak of the oscillatory waveform rather than by an equivalent expansion of both upper and lower pressure limits.

DISCUSSION

These bench data supports the central engineering claim of this device that oscillatory amplitude can be raised substantially by roughly 70–100%, through an independently controlled extra-flow channel, without a proportionate increase in the mean, clinician-prescribed CPAP level. This is an important distinction from every previously characterized bubble CPAP configuration, in which amplitude and bias flow and therefore mean pressure are mechanically coupled.[8-9] The practical consequence is that the present device offers, for the first time within the bubble CPAP paradigm, an operator-controlled variable that maps conceptually onto the amplitude control of a clinical HFOV device, while the existing CPAP level control continues to map onto mean airway pressure. The non-monotonic frequency response is a more complex finding, but it is in tandem with the bubble dynamics defined by the Rayleigh–Plesset equation. The progressive introduction of a second bubble stream first adds frequency content, then, as coalescence comes to dominate at higher extra flow, produces fewer and larger bubbles whose natural frequency falls in accordance with the Minnaert relationship. From a design standpoint, this is a favorable result. High amplitude together with reduced frequency is reached at the same, upper end of the extra-flow range, rather than requiring amplitude and frequency to be manipulated by separate controls.

The central scientific question this paper addresses is not merely whether the amplitude can be increased, but when, physiologically, increasing amplitude while lowering frequency should be expected to benefit a spontaneously breathing neonate, and when it should not. Given the finding in the *in-vitro* bench tests, the most plausible clinical niches for an adjus-amplitude bubble CPAP device are narrow and specific rather than universal. They include infants on bubble CPAP with borderline or rising $PaCO_2$ who do not yet meet criteria

for intubation, in whom a trial of increased amplitude pressure oscillation titrated against transcutaneous or blood-gas CO_2 could plausibly reduce escalation to invasive ventilation; infants in the acute, low-compliance phase of moderate-to-severe RDS, in whom the lung-recruitment criterion is most applicable, with weaning of amplitude as compliance improves and resource-limited neonatal units that cannot access dedicated nHFOV equipment but already use bubble CPAP, for whom this device could offer a low-cost, incremental step along the same physiological spectrum.

Preliminary observational clinical experience with the Humming BEE B-CPAP system in approximately 50 neonates across multiple neonatal centers in India indicates that the device may provided effective non-invasive respiratory support in preterm and term infants with respiratory distress syndrome and related conditions. It was also noted that the independently adjustable oscillation amplitude facilitated individualized respiratory support while maintaining the prescribed CPAP level. Although these observations arise from observational experience rather than formal prospective clinical study, they provide encouraging results supporting the safety, feasibility, and clinical utility of the Humming BEE system in the future

LIMITATIONS

This current study represents a bench-top (*in-vitro*) characterization of the Humming Bee B-CPAP system. Accordingly, the findings primarily establish the device's engineering performance and its ability to independently modulate oscillatory amplitude and dominant frequency while maintaining the prescribed CPAP level. The experimental evaluation was performed using a fixed-compliance 50 mL test lung that does not replicate the heterogeneity of respiratory mechanics encountered in preterm infants. Neonatal lungs exhibit substantial variation in compliance, airway resistance, disease severity, and spontaneous respiratory effort, all of which may influence the transmission and attenuation of pressure oscillations.

In addition, the bench model did not incorporate clinically relevant factors such as nasal-interface leaks, upper-airway dynamics, secretion burden, or patient–device interactions, each of which may alter delivered airway pressure and oscillatory transmission during routine clinical use. Furthermore, the study did not directly measure tidal volume delivery, gas exchange, alveolar recruitment, carbon dioxide elimination, or work of breathing. Consequently, the proposed physiological advantages of increased oscillatory amplitude and reduced dominant frequency remain inferential and require validation in translational and clinical studies.

Preliminary clinical use of Humming Bee B-CPAP in approximately 50 neonates with RDS has suggested that the device might be feasible and well tolerated in routine neonatal practice. However, these observations were derived from routine clinical experience and therefore require adequately powered prospective randomized clinical trials to establish definitive conclusions regarding its clinical benefit.

CONCLUSION

This paper has presented Humming BEE B-CPAP system that, through an independent, secondary extra-flow channel introduced into an enlarged bubble-generating chamber, decouples oscillatory pressure amplitude from the preset CPAP level, a constraint present in every previously characterized bubble CPAP design. Grounded in the physics of bubble formation and oscillation in the established clinical practice of HFOV, we have argued that increasing amplitude while reducing dominant frequency should be expected to assist gas exchange specifically in infants on bubble CPAP with inadequate CO_2 clearance, in the acute low-compliance phase of moderate-to-severe respiratory distress syndrome, and in infants with immature but intact respiratory drive. Bench testing supports the core engineering claim that oscillatory amplitude increased by approximately 70-100% across a graded extra-flow range while delivered CPAP remained at its set point across the great majority of conditions tested, and the resulting high-amplitude, reduced-frequency operating point is consistent with the theoretical predictions developed here. Preliminary observational clinical experience in approximately 50 neonates shows positive signs that the device can be successfully integrated into routine neonatal respiratory care, supporting its practical feasibility in a real-world clinical environment. By combining the simplicity, affordability, and robustness of conventional bubble CPAP with the ability to independently regulate oscillatory characteristics, Humming BEE B-CPAP represents a promising advancement in non-invasive neonatal respiratory support and provides a strong foundation for future translational and clinical evaluation.

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