Measurement of Pulsatile Tidal Volume, Pressure Amplitude, and Gas Flow During High-Frequency Percussive Ventilation, With and Without Partial Cuff Deflation

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OBJECTIVE: With a high-frequency percussive ventilator and a mechanical lung model, to measure tidal volume (VT), pulsatile pressure amplitude (difference between peak and nadir pulsatile pressure [\(\Delta P\)]) and mean airway pressure (Paw) at various pulsatile frequencies, pulsatile inspiratory-expiratory ratios (I:Ep), and pressures (measured at the interface between the pulse-generator and the endotracheal tube [Pvent]). METHODS: With the endotracheal tube inside an artificial trachea, we manipulated the high-frequency percussive ventilation settings and adjuncts, including pulsatile frequency, I:Ep, and Pvent by manipulating pulsatile flow. We also studied the effects of partially deflating the endotracheal tube cuff. We measured Paw, pulsatile pressure amplitude at the carina (\(\Delta P_c\)), and pulsatile VT at the carina. With the cuff partly deflated, we measured the fraction of inspired oxygen (FIO2) in the gas efflux above and below the cuff. RESULTS: Increasing the pulsatile frequency from 300 cycles/min to 600 cycles/min and changing the I:Ep from 1:3 to 1:1 significantly reduced VT (p < 0.001). Paw and \(\Delta P_c\) were unaffected by the change in pulsatile frequency or I:Ep, except when we did not preserve the pulsatile flow. The measured VT range was from 19.1 mL (at 600 cycles/min) to 47.3 mL (at 300 cycles/min). Partial cuff deflation did not significantly reduce Paw or \(\Delta P_c\), but it did significantly reduce VT and FIO2. CONCLUSION: During high-frequency percussive ventilation, the pulsatile frequency is inversely related to VT. Partial cuff deflation reduces the delivered FIO2.

Key words: volumetric diffusive ventilation, volumetric diffusive respiration, high-frequency percussive ventilation, high-frequency ventilation, tracheal tube, cuff leak.

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Introduction

High-frequency percussive ventilation (HFPV) is used extensively in burn units for severe inhalation injury and as an advanced ventilator mode for patients with acute respiratory distress syndrome who fail conventional ventilation.1–11 HFPV can be viewed as a hybrid ventilation mode in which a high-frequency (100–900 cycles/min) pulsatile waveform is integrated with a lower-frequency (5–30 cycles/min), conventional, time-cycled, pressure-limited breath with a phasic waveform (Fig. 1). Superimposition of the pulsatile waveform at higher frequencies (300–900 cycles/min) on a conventional pressure-cycled breath is thought to promote additional oxygen diffusion, while at lower frequency (100–300 cycles/min) convective alveolar ventilation and airway mucociliary clearance are enhanced.11–13 In addition, with high-frequency breath-stacking, HFPV may maintain airway pressure in the presence of large air leaks. With that in mind, partial deflation...
of the endotracheal tube (ETT) cuff has been recommended as an adjunct to assist in ventilation and secretion clearance during HFPV. However, there has been no detailed study of the effects of partial cuff deflation during HFPV on the characteristics of the high-frequency waveform and gas flow.

In HFPV, the air pulses/percussions are created by a device called a phasitron, which has a component that slides back and forth to generate a high-frequency pulsatile flow, in which the frequency and amplitude are joined or coupled (Fig. 2). Increasing the pulsatile frequency or changing the pulsatile inspiratory-expiratory ratio (I:Ep) from 1:3 to 1:1 may decrease the pulsatile pressure amplitude or pulsatile flow, and vice versa. The additional tidal volume (VT) generated by the pulsatile waveform is thought to be less than the anatomical dead space. However, to our knowledge no study has directly isolated and measured the pulsatile volume or investigated the interaction between HFPV settings and pulsatile VT, mean airway pressure (Paw), and pulsatile pressure amplitude (difference between peak and nadir pulsatile pressure [ΔP]).

To determine the relationship between the various aspects of the HFPV waveform, we adjusted the HFPV settings of pulsatile frequency, I:Ep, and the internal pressure measured by the HFPV at the interface of the ETT and the HFPV circuit (Pvent) to study the behavior of VT, Paw, and ΔP. We also examined the effects of partial cuff deflation on distal airway pressure, Vt, and the fraction of inspired oxygen (FiO2) distal and proximal to the cuff in the artificial trachea.

Methods

An HFPV device (VDR-4 [volumetric diffusive respirator], Percussionaire, Sandpoint, Idaho) was attached to a test lung (5600i, Michigan Instruments, Grand Rapids, Michigan) through an artificial trachea intubated with an 8.0-mm cuffed ETT (see Fig. 2). The artificial trachea is composed of 0.5-cm thick noncompliant plastic tubing, 22 cm in length and 9 cm inner diameter. The test lung was set to a compliance of 0.03 L/cm H2O. For all the experiments, the HFPV settings were maintained at a constant Pvent of either 20 cm H2O or 40 cm H2O, by adjusting pulsatile flow. Similar separate trials were also performed without adjusting the pulsatile flow. The phasic I:E was set with full “counterclockwise” inspiratory time while the expiratory time was turned to off. The HFPV system includes an aneroid manometer that measures the pressure within the HFPV circuit between the phasitron and the attachment to the ETT (Fig. 3).

The phasic I:E represents the conventional inspiratory and expiratory times upon which the pulsatile frequency is superimposed (see Fig. 1). To isolate and focus on the interaction of only the high-frequency aspects of the pulsatile volume, Paw, and ΔP, we turned off the convective pressure rise, oscillatory/demand continuous positive airway pressure, and the ventilator’s nebulizer function.

The artificial trachea has 2 gas sampling ports, 5 cm proximal and distal to the ETT cuff (see Fig. 2). We measured FiO2 and gas balance (ie, nitrogen and carbon dioxide) with a gas analyzer (Datex-Ohmeda S/5, GE Healthcare, Waukesha, Wisconsin) at both the sampling ports. FiO2 on the HFPV was set to 1.0, which was externally verified with the gas analyzer.

The cuff was either kept fully inflated or set to partial cuff deflation, which was implemented by decreasing Pvent by 5 cm H2O from the 40-cm H2O inflated-cuff baseline. We gradually deflated the cuff and then restored Pvent by increasing pulsatile flow.
To determine the origin of entrained air, we sequentially occluded the expiratory and inspiratory fail-safe valves and the circulation tubing to prevent ambient air from being entrained into the circuit, while retaining the 1.0 \text{FIO}_{2} gas flow directly delivered by the high-frequency flow-interrupter to the phasitron (see Fig. 3). Occluding these circuit apertures allowed us to sequentially analyze ambient-air entrainment and delivered gas.

We measured exhaled $V_T$, $P_{aw}$, and the high-frequency pressure amplitude with a neonatal high-frequency heated-wire sensor (Florian, Acutronic Medical Systems, Hirzel, Switzerland) placed in-line, distal to the tip of the ETT (location equivalent to the carina). Airway pressure amplitude and $P_{aw}$ were measured via the co-housed Florian manometer, which measured airway pressure via an in-line connection at the carina ($\Delta P_c$). We studied pulsatile frequencies of 300 cycles/min and 600 cycles/min, and I:E$_p$ settings of 1:1 and 1:3. Changing the I:E$_p$ changes the registered pulsatile frequency, so when we changed the I:E$_p$, we also changed the pulsatile frequency setting to maintain the original pulsatile frequency.

Each experiment was performed in triplicate. The data are expressed as mean ± standard deviation values. We used analysis of variance testing for repeated measures to assess the differences at 300 cycles/min and 600 cycles/min. Differences were considered significant when $p < 0.05$.

**Results**

**Effects of Pulsatile Frequency and I:E$_p$ on $V_T$, $P_{aw}$, and $\Delta P$**

At a set I:E$_p$ of 1:3 and a $P_{vent}$ of 20 cm H$_2$O, increasing the pulsatile frequency from 300 cycles/min to 600 cycles/min significantly reduced $V_T$, from 47.3 ± 0.76 mL at 300 cycles/min to 19.1 ± 1.94 mL at 600 cycles/min ($p < 0.001$). Increasing the pulsatile frequency from 300 cycles/min to 600 cycles/min caused no statistically significant change in $\Delta P_c$ or $P_{aw}$. Adjusting the I:E$_p$ to 1:1 while maintaining the same $P_{vent}$ significantly reduced $V_T$, compared to the I:E$_p$ 1:3 baseline at the same respective frequencies; the measured mean volumes were 43.2 ± 0.35 mL at 300 cycles/min ($p < 0.001$) and 17.8 ± 0.05 mL at 600 cycles/min ($p = 0.027$). Mean $\Delta P_c$ and $P_{aw}$ were not significantly different at I:E$_p$ 1:1 than at I:E$_p$ 1:3.

We reasoned that maintaining an artificially constant $P_{vent}$ by manipulating the pulsatile flow might confound the interpretation of pulsatile pressure and volume, so in a separate trial we evaluated the impact of increasing the pulsatile frequency without adjusting the pulsatile flow. In this trial we found a significant reduction in mean $V_T$ at 300 cycles/min: 47.4 ± 0.61 mL with pulsatile flow maintenance, versus 42.9 ± 0.40 mL without pulsatile flow maintenance ($p < 0.001$), whereas there was no significant difference from the pulsatile-flow-preserved $P_{vent}$ trials in the observed $\Delta P_c$ at 300 cycles/min or 600 cycles/min or in $V_T$ at 600 cycles/min. However, $P_{aw}$ did significantly decrease with increasing frequency at both I:E ratios (3.25 ± 0.75 cm H$_2$O, $p < 0.001$) relative to the pulsatile-flow-preserved test.

**Manipulation of $P_{vent}$**

Increasing $P_{vent}$ from 20 cm H$_2$O to 40 cm H$_2$O at a pulsatile frequency of 300 cycles/min increased $P_{aw}$ by 20 ± 0.0 cm H$_2$O and increased $\Delta P_c$ by 20.3 ± 5.5 cm H$_2$O ($p < 0.001$). $V_T$ was significantly reduced at $P_{vent}$ of 40 cm H$_2$O, at both 300 cycles/min ($V_T = 39.1 ± 0.06$ mL, $p < 0.001$) and 600 cycles/min ($V_T = 15.0 ± 0.91$ mL), $p < 0.001$ (Fig. 4). At the same high $P_{vent}$, adjusting the I:E$_p$ to 1:1 did not significantly affect $\Delta P_c$ or $P_{aw}$ relative to the I:E$_p$ 1:3 setting. However, $V_T$ at I:E$_p$ of 1:1 was significantly reduced at 300 cycles/min ($V_T = 34.2 ± 0.404$ mL, $p < 0.001$) and 600 cycles/min ($V_T = 13.8 ± 0.70$ mL, $p = 0.007$). At the I:E$_p$ 1:3 setting, increasing the pulsatile frequency from 300 cycles/min to 600 cycles/min significantly decreased $V_T$, from 38.4 ± 1.5 mL to 15.3 ± 2.2 mL ($p < 0.001$) but did not significantly change $P_{aw}$ or $\Delta P_c$.

In a separate trial we evaluated the impact of increasing the pulsatile frequency without adjusting the pulsatile flow. We found no significant difference in $\Delta P_c$ or $V_T$, compared to the $P_{vent}$ trials in which we preserved the pulsatile flow. As in the other trial, $P_{aw}$ significantly decreased (4.0 ± 0.0 cm H$_2$O) when we increased the frequency, compared to when the pulsatile flow was preserved.
entrainment into the circuit during partial cuff deflation (see Fig. 3). Occluding the inspiratory fail-safe valve decreased \( P_{\text{vent}} \) by 15 cm \( \text{H}_2\text{O} \). The \( F_{\text{IO}_2} \) measured at the distal and proximal ports after occluding the inspiratory fail-safe valve was 0.99 \pm 0.0. Occluding the expiratory fail-safe valve did not change airway pressure or oxygen concentration. The circuit was then restored, with the exception of a disconnection (not an occlusion) of the inspiratory circulation tubing, and exposure of the same port to \( F_{\text{IO}_2} \) of 0.21 from ambient air. The proximal and distal port \( F_{\text{IO}_2} \) decreased to 0.48 \pm 0.02 and 0.56 \pm 0.01, respectively.

With occlusion of the inspiratory circulation tubing and the inspiratory fail-safe valve, the proximal and distal port \( F_{\text{IO}_2} \) increased to 0.81 \pm 0.08 and 0.93 \pm 0.04, respectively. These findings are consistent with gas sampling through the oxygen reservoir contained within the expiratory circulation tubing (see Fig. 3). In summary, the “open circuit” effect of the inspiratory fail-safe valve appeared to be responsible for ambient-air entrainment intended to maintain aneroid manometer pressure during partial cuff deflation and, in so doing, compromised the delivered \( F_{\text{IO}_2} \).

**Discussion**

Given the mechanical properties of the phasitron, the pulsatile frequency and amplitude are thought to be coupled.\(^{11-12}\) However, a significant increase in \( P_{\text{aw}} \) and \( \Delta P_{\text{c}} \) implemented by increasing the \( P_{\text{vent}} \) significantly reduced \( V_T \) (see Fig. 4), which suggests an uncoupling of pulsatile \( V_T \) and any resultant change in \( P_{\text{aw}} \) or \( \Delta P \). Similarly, we found dissociation between \( \Delta P \) and \( V_T \) as a function of increasing pulsatile frequency (see Fig. 4). This is probably because of the frequency and flow-dependent constituents of impedance—namely, flow-dependent resistance, frequency-dependent inertance, and lung compliance. For instance, by increasing pulsatile flow to sustain the elevated \( P_{\text{vent}} \) goal, \( V_T \) is compromised by the resultant increase in flow-dependent resistance. Conversely, \( \Delta P \) is relatively preserved because of a distal decrease in lung compliance caused by the increase in \( P_{\text{aw}} \). On the other hand, with an increase in pulsatile frequency inertance effects then attenuate distal \( V_T \) and, possibly through an increase in airflow and thus flow-dependent resistance, prevent a similarly significant decrease in distal \( \Delta P \). Inertance represents the limits imposed by high-frequency volume accelerations; it is expressed as a proportionality constant of the difference in pressure across the mass of the airway and surrounding parenchyma and the mass of the administered gas. It also appears that an increase in pulsatile frequency may significantly compromise \( P_{\text{aw}} \), which implies that the bias flow and breath-stacking mechanisms
employed by HFPV may be insufficient to maintain $P_{aw}$ in the absence of augmented pulsatile flow.

Increasing the pulsatile flow compensates for the loss of pressure associated with partial cuff deflation. Given that pulsatile flow is drawn from the high-frequency flow-interrupter in the HFPV device, the reduced delivered $F_{I02}$ was an unexpected and profound result. More revealing was that the HFPV circuit appeared to be the main, if not the only, source of ambient-air entrainment. The ambient-air entrainment may be a consequence of the loss of reflected $P_{aw}$ against the phasitron. With the reduction in back-pressure, the HFPV circuit may mechanistically respond by permitting ambient-air supplementation to restore or compensate for this pressure change. As a caveat, the reduction in $F_{I02}$ may also reflect the extent to which our ETT/artificial-trachea model permitted peri-cuff airflow, which would be anticipated to vary depending on the ratio of ETT or ETT cuff to trachea diameter and luminal airflow properties. The potential deleterious clinical effects of compromised $F_{I02}$ suggest reconsideration of partial cuff deflation in patients who are difficult to oxygenate with a fully inflated cuff. We are currently exploring corrective options to restore distal $F_{I02}$ delivery.

Limitations to our study include the lack of multipoint analog-to-digital-conversion sampling technology, which may improve pulsatile $V_T$ and pressure amplitude resolution. Lung modeling may also (through its intrinsic mechanical properties) magnify or attenuate pulsatile flow changes, limiting extrapolation of our results. Our trials also did not explore the effects of HFPV-unique and advantageous programming to include expiratory pressure and oscillation and convective pressure rise, among others, as this experiment focused only on the high-frequency waveform characteristics of HFPV.

To the best of our knowledge this is the first study to directly measure high-frequency pulsatile $V_T$ during HFPV, to establish the fundamental effects of HFPV settings adjustments on high-frequency waveform features, and to analyze those waveform features and gas flow during partial cuff deflation. Understanding the means of measuring and manipulating high-frequency pulsatile $V_T$ sets a foundation for studying the optimal method of employing HFPV for lung-protective (low-$V_T$) ventilation in acute respiratory distress syndrome.

Conclusion

During HFPV, pulsatile frequency is inversely related to $V_T$. Partial cuff deflation caused an unanticipated reduction of delivered $F_{I02}$; this requires clinical study.

REFERENCES