**High-Frequency Percussive Ventilation: Pneumotachograph Validation and Tidal Volume Analysis**

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**INTRODUCTION:** High-frequency percussive ventilation (HFPV) is an increasingly used mode of mechanical ventilation, for which there is no proven real-time means of measuring delivered tidal volume ($V_T$). 

OBJECTIVE: To validate a pneumotachograph for HFPV and then exploit flow-sensor data to describe the behavior of both low-frequency and high-frequency breaths.

METHODS: Sensor performance was gauged during changes in high-frequency (4–12 Hz) and low-frequency rate and ratio, mean airway pressure, oxygen concentration, heated or heated-humidified gas flow, and endotracheal tube diameter. Glass bottle (adiabatic $V_T$) and test lung (adiabatically derived low-frequency $V_T$) based adiabatic conditions provided both an initial source for analog-signal calibration and an accepted standard comparator to flow-sensor measurement of high-frequency and low-frequency breaths (flow-sensor-derived $V_T$), respectively.

RESULTS: Pneumotachography proved accurate and precise over an array of tested settings and conditions when analyzing both high-frequency ($\Delta V_T$ between mean ± SD high-frequency $V_T$ and adiabatic $V_T$ was $-0.2 \pm 1.8\%$, 95% confidence interval $-0.5$ to $0.9\%$) and low-frequency breaths (mean ± SD difference between flow-sensor-derived low-frequency $V_T$ and adiabatically derived low-frequency $V_T$ was $0.6 \pm 2.4\%$, 95% confidence interval $0.1$–$1.1\%$). High-frequency $V_T$ and frequency exhibited an exponential relationship. During HFPV, flow-sensor-derived low-frequency $V_T$ had a mean ± SD of 1,337 ± 700 mL, 95% confidence interval 1,175–1,499 mL.

CONCLUSIONS: Readily available pneumotachography provided accurate measurements of low-frequency and high-frequency $V_T$ during HFPV. In the setting of acute lung injury, typical HFPV settings may deliver injurious $V_T$. 

**Key words:** high-frequency percussive ventilation; HFPV; mechanical ventilation; tidal volume; $V_T$; pneumotachography. [Respir Care 2010;55(6):734–740]

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**Introduction**

Clinical application of high-frequency percussive ventilation (HFPV) in the setting of acute lung injury (ALI) and smoke inhalation has led to consistent improvements in gas exchange.$^1$–$^{11}$ HFPV employs a singular high-frequency flow to create a pressure-limited, time-cycled, “low-frequency” tidal volume ($V_T$) breath similar to that used in conventional mechanical ventilation (Fig. 1). However, neither the low-frequency nor the high-frequency volumes administered by HFPV are measured by the ventilator’s pressure transducers.

Importantly, contemporary mechanical ventilation strategies specific to ALI dictate that $V_T$ be confined to within 4–8 mL/kg ideal body weight.$^{12,13}$ It follows that our inability to ascertain $V_T$ magnitude could result in clinicians unknowingly inflicting a form of ventilator-induced lung injury known as volutrauma. We therefore developed an HFPV-specific flow sensor to quantify $V_T$ and reduce the risk of ventilator-induced lung injury.$^{14–21}$

For the purposes of this study, both pneumotachography and heated-wire flow sensors have consistently demonstrated the best accuracy and precision in measuring high-frequency $V_T$.$^{18–21}$ To the best of our knowledge, only...
Pneumotachograph sensors have received FDA approval for clinical use. Because of the projected appeal for such a device in other HFPV-equipped facilities, this exploration was limited to a readily accessible FDA-approved candidate.

Pneumotachograph readings were first assessed for accuracy and precision, using an in vitro validation construct. Once validated, the respective flow-sensor data were also exploited to describe high-frequency and low-frequency VT behavior across a series of ventilator settings and gas flow conditions. This study represents part of an ongoing series of in vitro investigations probing the merits and deficiencies of HFPV.

Methods

Ventilator and Glass Flask Setup: High-Frequency Tidal Volume Measurement

A high-frequency percussive ventilator (Volumetric Diffusive Respirator [VDR-4], Percussionaire, Sandpoint, Idaho) was connected in series with the flow sensor, an 8.0-mm cuffed endotracheal tube (ETT) (Hi-Lo, Mallinckrodt, Hazelwood, Missouri), and a 6-L glass flask. Total system compliance (ETT and glass flask) was 9 mL/cm H2O (Fig. 2).

The Fleisch-type, heating-optional pneumotachograph (3700, Hans Rudolph, Shawnee, Kansas) had a linear voltage signal over a range of flows (± 140 L/min) when assessed against a calibrating flow meter (VT Plus, Biotek, Winooski, Vermont). For each portion of the experiment the analog signal from the flow sensor was amplified, low-pass filtered (160 Hz), and digitally sampled at 500 Hz with a laptop computer and software (Labview 8.5, National Instruments, Austin, Texas) engineered to generate a raw flow/pressure signal (analog Vt). Each reported Vt reflected the mean of the inspiratory and expiratory Vt, or positive and negative deflections of the high-frequency analog signal, over a 2-second epoch. All data were continuously recorded and tabulated on a computer.
As an accepted standard comparator to analog V\textsubscript{T}, a pressure-integrated high-frequency volume measurement (or adiabatic V\textsubscript{T}) was derived from a transducer mounted within the glass-bottle apparatus. An implicit assumption of this protocol, and of preceding high-frequency flow-sensor validation studies, was that in vitro gas flow conformed to adiabatic properties.\textsuperscript{20} Following Boyle’s law, adiabatic V\textsubscript{T} was calculated by incorporating the high-frequency pressure amplitude into the adiabatic gas formula (see Equation 1). The (± 2 cm H\textsubscript{2}O) pressure transducer and amplifier (1110 series pneumotachometer amplifier, Hans Rudolph, Shawnee, Kansas) had a flat-frequency response of at least 25 Hz and a linear pressure response to ± 110 cm H\textsubscript{2}O. The pressure signal was similarly low-pass filtered (160 Hz) and sampled at 500 Hz.

\[
\text{Adiabatic } V_T = V \cdot \Delta P / \gamma \cdot P_o
\] (1)

in which V is the volume of the glass flask, \(\Delta P\) is the high-frequency pressure amplitude, \(\gamma\) is the heat-loss ratio (1.39 for oxygen, and 1.40 for dry air), and \(P_o\) is the barometric pressure.

**Development of a Corrective Flow-Integration Algorithm**

It has been consistently shown, regardless of flow-sensor construct, that raw analog V\textsubscript{T} signals must be digitally integrated or corrected to account for changes in frequency-dependent effects.\textsuperscript{20,21} Integration was performed by applying a real-time correction factor to the analog V\textsubscript{T} measurement at each frequency (1–12 Hz, in 1 Hz increments). The more accurate and precise calibrated V\textsubscript{T} (henceforth known as the corrected V\textsubscript{T} or high-frequency V\textsubscript{T}) was a derivative of the product of analog V\textsubscript{T} and the ratio of analog V\textsubscript{T} to adiabatic V\textsubscript{T} measured during baseline experiments (described below).

**High-Frequency Protocol**

Low-frequency breath cycling was turned off, resulting in a sustained high-frequency-only waveform. Calibrating baseline or default ventilator settings were performed at: 4–12 Hz, mean airway pressure (P\textsubscript{aw\textsubscript{aw}}) setting of 20 cm H\textsubscript{2}O, F\textsubscript{IO\textsubscript{2}} of 0.21, and a fixed inspiratory/expiratory ratio (high-frequency I:E) of 1:1, with an 8-mm ETT. As previously noted, analog V\textsubscript{T} and adiabatic V\textsubscript{T} results from the latter experiments were used to derive a corrected, frequency-specific V\textsubscript{T} or high-frequency V\textsubscript{T}. We then ascertained the magnitude of error in the corrected flow signal caused by changes in gas content, airway caliber, and/or physiologic conditions. The sensor was assessed, without further signal adjustment, across a range of frequencies (4–12 Hz), P\textsubscript{aw\textsubscript{aw}} (10, 20, 30 cm H\textsubscript{2}O), and F\textsubscript{IO\textsubscript{2}} (0.21, 0.50, and 1.0), as well as during active heating (gas temperature of 32°C), heating and humidification (gas temperature of 32°C, 70% relative humidity), with 6, 7, and 8-mm inner-diameter ETTs, and at a high-frequency I:E of 1:2. The high-frequency protocol included a total of 63 mean high-frequency V\textsubscript{T} measurements.

**Flow Sensor Dead-Space Effect**

In order to quantify flow-sensor-imposed dead space, adiabatic V\textsubscript{T} was measured during the same high-frequency protocol experiments, with and without the flow sensor in line.

**Ventilator and Mechanical Lung Setup: Low-Frequency Tidal Volume Measurement**

The previously described flow sensor and computer arrangement was used to measure and analyze low-frequency V\textsubscript{T}. Given the absence of a linear inspiratory flow, the low-frequency algorithm consisted of summing the difference between individual inspiratory and expiratory high-frequency V\textsubscript{T} over the time interval spanning end-expiration to end-inhalation, to produce the flow-sensor-derived low-frequency V\textsubscript{T}.

The low-frequency V\textsubscript{T} comparator (test-lung V\textsubscript{T}) was derived from a mechanical test lung (5600i, Michigan Instruments, Grand Rapids, Michigan) calibrated per the manufacturer’s instructions (see Fig. 2). A mechanical test lung was used for 2 reasons. First, it permitted one to examine whether glass-model-derived calibration could be extrapolated to other compliance/resistance conditions. Mechanical lung compliance and resistance could also be adjusted to more closely model physiologic conditions. Second, by studying HFPV under modeled physiologic conditions, one could comprehend potential test-lung V\textsubscript{T} behavior in the clinical setting. Test-lung compliance, validated at the airway pressures utilized for the study, was set to model ALI-like conditions (40 mL/cm H\textsubscript{2}O). Airway resistance was 5 cm H\textsubscript{2}O/L/s. The test lung included an embedded pressure transducer to obtain adiabatically-derived V\textsubscript{T} (manufacturer-recommended test-lung-specific software was used for measurement of test-lung V\textsubscript{T} [Pneuview 5600i software, Michigan Instruments, Grand Rapids, Michigan]). Notably, no test-lung-V\textsubscript{T}-based calibration adjustment of the low-frequency V\textsubscript{T} signal was required to enhance low-frequency flow sensor accuracy.

For the low-frequency protocol, conditions identical to the high-frequency protocol were explored with an additional array of low-frequency inspiratory time/respiratory rate combinations (10 breaths/min with an inspiratory time 1, 2, or 3 s, or 20 breaths/min with an inspiratory time of 1 or 2 s) and applied PEEP of 5 or 10 cm H\textsubscript{2}O). Each
experiment collected and averaged data over 10 consecutive low-frequency breaths. The low-frequency protocol included a total of 105 mean low-frequency VT measurements.

Statistical Analysis

For both the high-frequency and low-frequency experiments the comparison between high-frequency VT and adiabatic VT, and flow-sensor-derived low-frequency VT and adiabatically derived low-frequency VT, respectively, was performed with Bland-Altman analysis, with 95% confidence intervals (CIs). As error in VT may increase in proportion to VT size, the percent error (% error) (see Equation 2), as opposed to the absolute difference, was used.

\[
\text{% error} = 100\% \cdot \frac{(V_{TX} - \text{adiabatic } V_T)}{V_T \text{ mean}} \quad (2)
\]

in which x is either the high-frequency VT for the high-frequency protocol or the flow-sensor-derived low-frequency VT for the low-frequency protocol, and y is either the adiabatic VT for the high-frequency, or the adiabatically derived low-frequency VT for the low-frequency protocol.

The study also included a simple descriptive analysis (difference in the means at each frequency) of the effects of ventilator settings (eg, frequency, Paw, high-frequency I:E) and ETT diameter on adiabatically derived high-frequency VT. Baseline or default settings served as a comparison for each change in frequency, unless explicitly stated otherwise.

Results

As anticipated, the raw analog VT signals required integration or correction to account for changes in frequency-dependent effects. To illustrate, during initial high-frequency experiments, changes in frequency led to a systematic, linear, biasing effect on mean unadjusted analog VT relative to mean adiabatic VT (mean SD difference 3.1 ± 3.2%, 95% CI 1.9–4.3%).

Application of a real-time corrected VT (high-frequency VT) improved flow-sensor accuracy and precision, with a mean difference of –0.2 ± 1.8% (95% CI −0.5% to 0.9%) (Fig. 3). Changes in mean Paw, high-frequency I:E, and ETT size, or the application of heated and humidified gas did not diminish high-frequency VT accuracy or precision (Table 1). However, an FiO2 of 1.0 led to systematic error (mean error 9.2 ± 4.5%, 95% CI 6.3–12.1%) and required an additional correction factor. For oxygen calibration purposes, the previously mentioned baseline ventilator settings were reexamined with an FiO2 of 1.0. We then used the analog VT and adiabatic VT results from the latter experiments to generate an FiO2 = 1.0 specific corrected high-frequency VT. This adjustment further enhanced flow-sensor accuracy (see Table 1).

Mean high-frequency VT across the range of explored settings was 52.2 ± 27.1 mL (95% CI 41.9–62.5 mL). There was an exponential relationship between frequency and both mean adiabatic VT and mean high-frequency VT (see Equation 3 and Fig. 4).

\[
\text{Adiabatic } V_T = 119.9e^{-0.19\text{frequency}} \quad (3A)
\]

\[
\text{High-frequency } V_T = 118.27e^{-0.17\text{frequency}}(r^2 \geq 0.99) \quad (3B)
\]

Measurement of adiabatic VT with and without the flow sensor in place disclosed a reduction in mean VT of
Mean adiabatic VT magnitude varied with changes in frequency, high-frequency I:E, ETT dimension, and mean Paw (Table 2).

Flow sensor performance during low-frequency ventilation was accurate irrespective of inspiratory time and respiratory rate combination (mean difference 0.6 ± 2.4%, 95% CI 0.1–1.1%) (Fig. 5) or change in high-frequency I:E, heated and humidified gas flow, oxygen content, applied PEEP setting, or ETT diameter (Table 3). Extended experiments (at a fixed high-frequency rate of 6 Hz and mean Paw of 20 cm H2O) revealed the flow sensor remained accurate during broader changes in airway resistance (5–15 cm H2O/L/s) and compliance conditions (10–40 mL/cm H2O) (data not shown). Using the studied range of HFPV settings resulted in low-frequency VT extending from 607 mL to 3,452 mL (mean VT 1,337 ± 700 mL, 95% CI 1,175–1,499 mL) (see Fig. 5).

Discussion

This study represents the first effort to provide an accurate and precise measurement of 2 separate HFPV gas...
TIDAL VOLUME ANALYSIS DURING HIGH-FREQUENCY PERCUSSIVE VENTILATION

Table 3. Mean Difference in High-Frequency VT* Relative to Test-Lung VT†

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<tr>
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<th>Difference in High-Frequency VT (mean ± SD %)</th>
<th>95% CI</th>
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<tbody>
<tr>
<td>High-frequency I:E 1:1</td>
<td>–0.3 ± 0.6</td>
<td>–0.7 to 0.1</td>
</tr>
<tr>
<td>FIO2 1.0</td>
<td>0.4 ± 0.7</td>
<td>–0.1 to 0.9</td>
</tr>
<tr>
<td>Heated and humidified gas</td>
<td>–0.7 ± 2.0</td>
<td>–2.0 to 0.6</td>
</tr>
<tr>
<td>6-mm ETT</td>
<td>0.6 ± 2.1</td>
<td>–0.8 to 2.0</td>
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<tr>
<td>Applied PEEP (cm H2O)</td>
<td></td>
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<tr>
<td>5</td>
<td>–0.2 ± 0.9</td>
<td>–0.8 to 0.4</td>
</tr>
<tr>
<td>10</td>
<td>–0.8 ± 1.4</td>
<td>–1.7 to 0.1</td>
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* High-frequency VT = flow-sensor-derived high-frequency tidal volume.
† Test-lung VT = mechanical lung adiabatically derived low-frequency VT.
I:E = inspiratory/expiratory ratio
ETT = endotracheal tube
FIO2 = fraction of inspired oxygen

flow patterns in real time. A simple corrective algorithm for adjusting the analog flow signal proved exact across the studied range of frequencies, mean $P_{aw}$: high-frequency I:E ratios, $F_{IO2}$, and heated and humidified gas flows (see Tables 1 and 3 and Figs. 3 and 5). Moreover, flow-sensor accuracy was comparable to that achieved by flow sensors validated for use with other modes of high-frequency ventilation.18-21 Taken at face value, the use of pneumotachography for high-frequency flow analysis is a reliable and time-tested approach, and thus there is no novelty to our pneumotachograph validation. However, in the midst of performing a flow-sensor validation protocol specific to this dual-frequency mode of mechanical ventilation came several new discoveries and concerns regarding HFPV-administered high-frequency and low-frequency VT, respectively.

As with other modes of high-frequency ventilation, an increase in ETT diameter and I:E or a reduction in high-frequency rate amplified high-frequency VT (see Table 2).21,26,27 Surprisingly, there appeared to be a bell-shaped response to mean $P_{aw}$-specific VT responses. At mean $P_{aw}$ settings exceeding 20 cm H2O (going from 20 to 30 cm H2O) a reduction in mean high-frequency VT was noted (–9.2 ± 9.6%, 95% CI –2.9 to –15.5%). The origin of the latter finding is unclear but may be a consequence of high-$P_{aw}$-induced venting of gas flow through pressure-release valves adjacent to the HFPV assembly.22,23

Another original aspect to this study was the discovery that HFPV-administered high-frequency VT carried an exponential frequency-dependent relationship (see Equation 3 and Fig. 4). Scalfaro et al revealed that high-frequency oscillatory ventilation also follows an exponential VT reduction as frequency is increased.20 HFPV may therefore draw upon the same or similar principles invoked for high-frequency oscillatory ventilation to augment patient oxygenation or ventilation. However, this comparison must be examined in light of HFPV’s continued use of low-frequency ventilation.

Our aforementioned concerns regarding previously volumetrically unmeasured HFPV appear well founded. Applying a representative range of HFPV settings during ALI-modeled conditions led to a mean VT of 1,337 ± 700 mL (95% CI 1,175–1,499 mL). The latter results, extrapolated to a 70-kg patient, would correspond to a mean VT of 19.1 mL/kg, which is far in excess of the 4–8 mL/kg recommended for patients with ALI.12,13 This revelation suggests that HFPV must be used with caution in patients with ALI, or, at the least, should be guided by real-time flow-sensor measurements to avoid the delivery of inordinate VT. Continued exploration of VT-driven HFPV algorithms may eventually advance HFPV into the conceptual framework of lung-protective ventilation.

The flow sensor system suffers from limitations. Improving upon analog signal accuracy required high-frequency-rate and oxygen-concentration-specific corrections. However, frequency and oxygen dependent modification were anticipated and remediable confounding elements.14-17,20,21 Though the sensor attachment carries an additional small dead space burden, this impediment can be overcome by compensatory measures such as lowering the high-frequency rate or augmenting the low-frequency minute ventilation.21 Application of bench-top findings to clinically relevant bedside use can also be challenging. Nevertheless, as part of an ongoing follow-on protocol, preliminary experience has shown that the flow sensor is amenable to near-automated “plug-and-play” adaptability, permitting clinicians the opportunity to make real-time ventilator adjustments based on flow sensor measurements. Indeed, clinical application of the flow sensor has confirmed the existence of previously unrecognized large VT delivery during HFPV (unpublished observation). These findings have led to a marked change in how we approach HFPV programming.

Conclusions

A readily available pneumotachograph accurately and precisely gauged high-frequency and low-frequency VT during HFPV. This early experience suggests that the device may be an agreeable substitute or supplement to current HFPV transducers. Notably, flow-sensor measurements have pinpointed that, in the setting of ALI, typical HFPV settings may deliver injurious VT.

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REFERENCES


