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A New Differential Pressure Flow Meter for Measurement of Human Breath Flow: Simulation and Experimental Investigation

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Abstract

The development and performance characterization of a new differential pressure-based flow meter for human breath measurements is presented in this article. The device, called a “Confined Pitot Tube,” is comprised of a pipe with an elliptically shaped expansion cavity located in the pipe center, and an elliptical disk inside the expansion cavity. The elliptical disk, named Pitot Tube, is exchangeable, and has different diameters, which are smaller than the diameter of the elliptical cavity. The gap between the disk and the cavity allows the flow of human breath to pass through. The disk causes an obstruction in the flow inside the pipe, but the elliptical cavity provides an expansion for the flow to circulate around the disk, decreasing the overall flow resistance. We characterize the new sensor flow experimentally and theoretically, using Comsol Multiphysics[®] software with laminar and turbulent models. We also validate the sensor, using inhalation and exhalation tests and a reference method.

Keywords

transport; computational fluid dynamics; biomedical engineering; fluid mechanics; turbulence

Introduction

Breath analysis is an attractive noninvasive means of monitoring health. Human breath has a plethora of biomarkers,¹ and accurate measurement of breath flow rates is essential for the assessment of precise breath biomarker concentrations. In particular, clinical assessment of lung health relies on accurate measurements of breath flow rates, measured at either forced or relaxed breathing conditions. Forced exhalation flow measurements enable diagnosis of obstructive lung disorders via spirometry method,² which is highly accurate and low cost relative to other diagnostic methods. In contrast, relaxed exhalation flow measurements have wider clinical applications, including metabolic analysis, pulmonary assessment, and even cancer detection.^{3–7} To obtain good performance of relaxed flow measurements, the following requirements are necessary: (1) High sensitivity and accuracy to detect low flow rates. (2) Low flow resistance for enabling normal exhalation and inhalation measurements.⁸

(3) Capability to differentiate flow direction. (4) Capability to correct for humidity and temperature changes present during breathing.

There are many commercially available products currently for measuring relaxed breath flows,^{5,9} including turbine, hotwire, ultrasonic, and differential pressure flow meters. Each flow meter has its advantages and disadvantages. Turbine flow meters are ideal for applications with low flow-resistance; however, their accuracy is limited at low flow rates, and breathing maneuvers can cause errors due to “lag-before-start” and “spin-after-stop” effects.^{10,11} Hot-wire flow meters are fairly common in one-direction flow metering, but require warm-up, frequent calibration, and knowledge of gas composition.¹² Ultrasonic flow meters can measure flow direction and require infrequent calibration, but their sensitivity to factors altering the speed of sound (mainly gas composition, temperature, and humidity) limits their application in relaxed breath flow measurements.^{5,13}

Differential pressure flow meters are very widely used due to their simplicity, robustness, and adaptability (Figure 1A). Among them, Fleish-type meters, which rely on the pressure drop across a capillary screen, provide good linear response, but they are susceptible to clogs, difficult to clean, and require frequent calibration.¹⁴ Orifice flow meters are simple, robust, and inexpensive to implement, but they have high flow-resistance.⁹ Variable-orifice flow meters provide a reduced flow-resistance by having an orifice size proportional to flow; however, they are more expensive and prone to mechanical wear out over time.⁵ Alternatively, Pitot Tube flow meters have very low flow-resistance and can be inexpensive to implement (Figure 1B). In these meters, typically, pressure sampling is done in the center of fluid flow where the velocity is maximum. As the pressure sampling port is stationary, a stagnation pressure develops proportional to the flow rate. However, the sensitivity of a Pitot Tube is generally very low and limited to fairly steady flows.¹⁵ Thus, despite the field being very well developed for differential pressure flow meters, bidirectional flow measurement for relaxed breathing conditions is still nontrivial and undeveloped.

In the present work, a new flow meter called the confined pitot tube (CPT) is developed, which successfully overcomes the barriers of sensitivity, accuracy, resistance, bidirectionality, and cost of existing flow meters to measure human breath. The new device consists of an elliptical disk with a centered pressure sensing port, as shown in Figures 1C, D and 2A, B. The disk is located in an elliptical expansion cavity placed to reroute flow (Figure 2C). The design leads to increased stagnant pressure at the center of the elliptical disk on the side facing the incoming flow (Figure 2D). In addition, the elliptical disk is exchangeable and has different diameters. When the diameter is large, the gap between the disk and the expansion cavity is small, and high confinement flow conditions are developed (Figure 1C). On the contrary, when the disk diameter is small, the gap between the disk and the expansion cavity is high, and low confinement flow conditions are developed, and eddies are induced in the buffer regions (Figure 1D). In both, low and high CPTs, the viscous drag built from the confinement and inertial effects from bending the flow amplify the pressure at lower flow rates, allowing for greater accuracy. For conditions where the flow direction of the device is inverted (e.g., inhalation), the confinement of the flow meter yields a high negative pressure signal at the sampling pressure port (disk center), allowing for easy differentiation and measurement of flow direction.

Different configurations of the new CPT flow meter are presented here that use a constant inner radius of 10 mm for the elliptical expansion chamber, and variable inner disks of radii ranging from 4.25 to 9.25 mm (Figure 2E, F). The performance of these different configurations is investigated at constant flows of 0–20 L/min, as well as at simulated breathing cycles consisting of inhalation and exhalation conditions of [−60;60] L/min (negative referring to inhalation flows). An optimal geometry rendering maximum sensitivity for low flows, high accuracy, and low flow resistance for measurement of human breath flow is also identified.

Simulations

Simulations were performed using Comsol Multiphysics®. The CPT flow sensor was modeled axisymmetrically (see Figure 2D), neglecting the supports of the inner disk to reduce computation time. Two flow models, a laminar model and a $k-\omega$ Reynolds-Averaged Navier–Stokes (RANS) turbulent model, were applied using steady-state fluid flows and converged to a tolerance of 0.001.

Laminar model

Laminar fluid flow¹⁶ was simulated using the Navier–Stokes (1) and continuity (2) equations for a compressible flow

$$\rho(u \cdot \nabla)u = \nabla \cdot \left[-pI + \mu \left(\nabla u + (\nabla u)^T \right) - \frac{2}{3}\mu(\nabla \cdot u)I \right] + F \quad (1)$$

$$\nabla \cdot (\rho u) = 0 \quad (2)$$

where ρ is density, u is fluid velocity, p is pressure, and F is volume force, and ∇ , I , and T represent the gradient, identity matrix, and transpose operators, respectively. The laminar model was used to simulate flow profile and pressure distributions for constant flow through the flow meter.

$k - \omega$ RANS turbulence model

The $k - \omega$ RANS turbulence model was used because currently it is the best two-equation model for flows exhibiting high curvature. Being a two-equation model, it can predict turbulent behavior without previous knowledge of the turbulent structure. The first two-equation $k - \omega$ turbulence model was proposed by Kolmogorov.¹⁷ The model was later developed independently by Saffman, greatly improved by Wilcox and Alber, and then revisited and improved by Wilcox.¹⁸ The $k-\omega$ model accounts for the turbulent fluctuation (k) (Eq. 3), as well as the specific dissipation rate (ω) (Eq. 4)

$$\rho \frac{\partial k}{\partial t} + \rho u \cdot \nabla k = P_k - \rho \beta^* k \omega + \nabla \cdot ((\mu + \sigma^* \mu_T) \nabla k) \quad (3)$$

$$\rho \frac{\partial \omega}{\partial t} + \rho u \cdot \nabla \omega = \alpha \frac{\omega}{k} P_k - \rho \beta \omega^2 + \nabla \cdot ((\mu + \sigma \mu_T) \nabla \omega) \quad (4)$$

For further details on the $k - \omega$ RANS turbulence model see the Appendix.

Experimental

Flow meter design and fabrication

As mentioned before, the CPT was designed with internal confined pitot disks ranging from 4.25 to 9.25 mm in radius. This radius range was initially determined via simulation results. All the disks were interchangeable and placed in the common expansion cavity of 10 mm radius. The disks had a sample port located at the center as shown in Figure 2. The radii of the CPT disks allowed the classification of the resulting assembled flow meters as: (a) low confinement (4.25 mm), (b) moderate confinement (5.5–8 mm), and (c) high confinement (9.25 mm) (Figure 2F). The CPT flow meter device was fabricated using a Stratasys Object Photopolymer 3-D printer (Rocky Mountain Tech, LLC). Pressure readings were taken for different configurations, including: (a) flow-resistance (point ⑧ and ⑨ as shown in Figure 1C, D), (b) one-point sampling (point ① and ② as shown in Figure 1C, D), or (c) two-point sampling (point ① and ⑨ as shown in Figure 1C, D). A Freescale MP3V5004DP differential pressure sensor was used for the differential pressure measurements. The differential pressure sensor is internally temperature corrected from the range of 10–60°.

Steady state flow characterization setup

Flow experiments were performed using dry (24 ppm H₂O) breathing grade air (Praxair), passing through the CPT at a constant flow rate in the range of 0–20 L/min, which represents the typical average exhalation flow range of human subjects. The flow was measured by a reference flow meter: EM1 Sensirion Digital Mass Flow Meter, which was connected at the inlet of the CPT. In parallel, all pressure measurements were sampled by the differential pressure meter at 40 Hz and analyzed via boxcar averaging at a lower frequency of ~4 Hz.

Dynamic flow calibration setup

Cyclic-flow experiments were performed with a Vacumed® Lung Calibrator/Simulator, using a range of flow rates from –60 to 60 L/min. The setup worked as pictured in Figure 3. A rotating disk moves a piston, providing a sinusoidal flow pattern. The position of the piston at any point can be described as

$$x(\theta) = \sqrt{r \sin(\theta)^2 + a^2} + r \cos(\theta) \quad (5)$$

where x is the position of the piston, r is the radius of the rotating disk, a is the length of the bar attached to the wheel, and θ is the axial coordinate of the wheel. The change in position

of the piston can be described by equating the derivative of Eq. 17 to change in volume (V) with time (Eqs. 18 and 19)

$$\frac{dx}{d\theta} = \frac{r^2 \cos(\theta) \sin(\theta)}{\sqrt{a^2 + r^2 \sin^2(\theta)}} - r \sin(\theta) \quad (6)$$

$$\frac{dV}{dt} = \frac{dx}{d\theta} * \frac{d\theta}{dt} * A = \omega * A * \left[\frac{r^2 \cos(\theta) \sin(\theta)}{\sqrt{a^2 + r^2 \sin^2(\theta)}} - r \sin(\theta) \right] \quad (7)$$

As the Lung Calibrator/Simulator is designed for relatively low resistance applications, corrections of flow resistance introduced by the use of the CPT connected to the Lung Calibrator/Simulator were made at high flows using a reference turbine flow meter from Oxycon Mobile (Carefusion, Yorba Linda, CA). The correction factor was a linear function of the flow resistance, assessed by the differential pressure sensor from the CPT flow meter, and the reference turbine flow meter at the peaks and zeros of the sinusoidal cyclic flows.

Results and discussion

Steady state flow characterization results

Simulations were run for a CPT with five different inner radii disks ranging from 4.25 to 9.25 mm (Figure 2) in a flow rate range from 2 to 20 L/min of dry air. Both laminar and turbulent simulations were conducted as described in the experimental section, and considered smooth surfaces for the flow with no-slip conditions. Therefore, it is noteworthy that inaccuracies resulting from surface roughness were not accounted for, along with other minor features. For instance, neither model accounted for the disk supports or the actual pressure sampling ports, which were present in the actual flow chamber. Figure 4A shows an example of the pressure distribution profiles obtained for the CPT system under laminar flow. It can be observed that the optimal points for measuring differential pressure values are: (1) The pressure at the center of the CPT, which is a stagnant pressure, and (2) The pressure after the curvature of the CPT on the wall of the tube. These two pressure points were defined as sampling ports ① and ②, as shown in Figure 1C, D. The simulations verified the correct choice of the sampling ports for maximizing differential pressure measurements, which were correlated to flow rate. As mentioned before, the alternative choice for measuring differential pressure was sampling port ①, and a sampling point outside the system at atmospheric pressure, which was named atmospheric pressure port ④ (Figure 1C, D). Figure 4B shows the streamline profile of the CPT under turbulent simulation conditions. Note that the bulk of the profile is laminar. This indicates that the turbulent interactions in the simulation are only occurring on a very small length-scale.

Figure 5 shows plots for differential pressure experimental measurements vs. flow rate for different CPT flow meters of various radii with *one-point sampling* (left side) for assessment at sampling port ① vs. ④, and *two-point sampling* (right side) for assessment at sampling

port ① vs. ②. In addition, the simulation results for both models, laminar and turbulent, at each corresponding condition are shown for comparison. Two-point sampling provides a slightly higher signal, but not a significant difference in performance compared to one-point sampling.

Additional analysis of results represented in Figure 5 indicates a trend that could be classified as follows:

Low Confinement Flows—The smallest inner disk of 4.25 mm in radius (Figure 5A, B) was better described by the turbulent flow model. The relatively big gap between the external wall of the device and the inner wall of the disk yields a buffer area, which leads to turbulence at relatively low flow rates. Convergence for laminar flow could not be reached in Comsol at the lowest confinement conditions with a reasonable mesh.

High Confinement Flows—The largest inner disk of 9.25 mm in radius is well described by the laminar flow model (Figure 5I, J). In this case, a relatively high sensitivity of ~3 Pa at a low flow rate of 2 L/min flow using two-point sampling with a low flow-resistance was observed. The high confinement design provides a relatively small gap between the external wall of the device and the inner wall of disk, which yields mainly laminar flows. The analytical sensitivity and flow resistance of this specific design made it an inner disk dimension of choice for the design of a flow meter for breath analysis.

Moderate Confinement Flows—Intermediate inner disk radii of 5.5 to 8.0 mm were better described by the laminar model at flows less than 10 L/min and better described by the turbulent model flows greater than 12 L/min (Figure 5C–H).

Comparison of confined pitot tube flow meter and fixed orifice flow meter

All different CPT flowmeters were evaluated for breathing usability. The configuration with 9.25 mm was found to provide comfortable breathing for end-tidal conditions with the highest accuracy. Therefore, this configuration was the choice for further investigation of flow-resistance and comparison with existing flow meter technologies.

To benchmark the performance of the CPT flow meter, it was compared to a Fixed Orifice flow meter with a 2.5 mm orifice radius (Figure 1A) in both signal and flow-resistance. The dimension of the orifice was chosen to establish a comparison with state-of-the-art commercial flow meters for breath.¹⁹

Figures 6A, B show the differential pressure measured by a two-point sampling method vs. flow. At low flow rates of less than 3 L/min, the sensitivity given by the differential pressure of either device is similar (Figure 6B). At higher flow rates, higher differential pressure signal is observed for the fixed orifice flow meter. However, the higher sensitivity comes with a higher flow-resistance, as seen in Figure 6C. At 20 L/min, although the fixed orifice meter has roughly two times the signal, it also has nearly three times higher in flow-resistance. The resistance generated by the Fixed Orifice scales considerably higher than the resistance generated by the CPT. This point is of interest for breath flow meter measurements that require free-breathing or low flow-resistance such as capnography.⁵ The

CPT flow meter has considerably less flow-resistance than the Fixed Orifice meter, while still offering the same sensitivity to low flows (<3 L/min), which are of physiological interest in relaxed breathing conditions.⁵

Figure 6D shows the flow resistance for the CPT having different inner radii. Although the 9.25 mm disk yields a low flow resistance, the flow resistance could be further decreased using smaller inner disks if necessary for measuring other types of high flow rate breathing tests, such as spirometry.²

Dynamic flow calibration, and validation with real breath

Given the advantages of the 9.25-mm disk CPT flow meter as candidate of choice for breath flow meter, the meter was calibrated using a lung calibrator/simulator as reference flow using the method described in the experimental section. The use of the calibrator/simulator was essential to simulate intended use conditions of exhalation and inhalation. One-point sampling method was used (center of the disk against atmospheric pressure) in the CPT flow meter since a notable enough advantage was not seen with respect to two-point sampling method. Figure 7 shows the comparison between the measured differential pressure in the CPT and the reference flow. It is worth noticing that the square root of the measured differential pressure in the CPT flow meter scales linearly with the reference flow. However, there is no single square root relationship in the entire range of flow rates; therefore, a set of equations of differential pressure vs. flow rate were used to calibrate the system at different flow ranges. This practice of piecewise calibration is appropriate for most applications considering that the current computing power of smartphones and even microprocessors allow for very fast signal processing. For the present work, five linear equations relating the square root of the measured differential pressure to the flow rate were used. For commercial implementation, this could be further enhanced or simplified to a larger or smaller number of equations depending on the accuracy requirements of the flow metering application.

Once the CPT was calibrated, real breath samples were measured simultaneously using the CPT and a reference flow measurement (Oxycon Mobile). The accuracy was assessed by comparing the total volume of individual inhalation and exhalation cycles from the integration of the flow measured by both methods. Volume was used to assess compounded errors. In Figure 8A, a strong correlation is noted between the breath volumes from the CPT and reference flow meter for expired and inspired volumes. An offset between the two measurements is observed due to the aforementioned problems with the calibration method, such as leaks in the lung simulator system as the flow rate increased (see experimental). To combat this, a linear regression was computed and used to apply a linear correction. After applying a linear correction, the standard deviation between the two methods was about 50 mL, which happens to be the reported accuracy of the reference method; and the overall accuracy error is $<10\%$.

Summary of the operating principle of the confined pitot tube

The advantage of the CPT design is that at low flows (<10 L/min) there is high sensitivity, while at high flows (10–80 L/min) bears little flow resistance on the user. At low flows, a laminar regime is present, which increases the pressure by adding viscous drag to the

stagnant pressure from the constriction. Once higher flows occur, then the turbulent effects begin to show, but they do not reach the values expected for a fully developed turbulent flow regime. This is due to the high degree of confinement and high curvature of the sensor design. In addition, although laminar flow is generally sensitive to temperature and humidity, the effects of these on viscosity and density within the region of interest (15–35° and 0–100% relative humidity) are fairly low.²⁰ The design has no moving parts, and could easily be mass-produced by injection molding, providing a low-cost option.

Conclusions

This article describes the development of a new device called the CPT flow meter for measuring human breath. Based on experimental tests and simulations, the device performed on par with commercially available systems in terms of accuracy and low flow-resistance for both inhalation and exhalation measurements. Separate simulations of the system under laminar and turbulent flow models were compared with the experimental results, allowing the characterization of the new device, and assessment of flow regimes for different device configurations and flow rate operating conditions. It was found that the largest inner disk of the CPT flow meter design retained fairly laminar flow characteristics within the expected range of human breathing rates. In addition, under this condition, the highest sensitivity with relatively low flow-resistance was obtained. In addition, the CPT flow meter was able to measure bidirectional flows for inhalation and exhalations by means of an inexpensive single sensing element, a differential pressure sensor. With the addition of a temperature sensor, breath could also be further corrected to standard temperature, pressure, and dry conditions (known as STPD). Furthermore, calibration tests followed by validation tests with a reference flow sensor indicated that the CPT flow meter is accurate for measuring flow rates at relaxed breathing conditions in humans. Finally, given the features of simple design with no moving parts, the new CPT flow meter is an attractive option for inexpensive mass-production.

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Appendix

The $k - \omega$ RANS turbulence model utilizes the two equations below, where k is the turbulent fluctuation and ω is the specific dissipation rate

$$\rho \frac{\partial k}{\partial t} + \rho u \cdot \nabla k = P_k - \rho \beta^* k \omega + \nabla \cdot ((\mu + \sigma^* \mu_T) \nabla k)$$

$$\rho \frac{\partial \omega}{\partial t} + \rho u \cdot \nabla \omega = \alpha \frac{\omega}{k} P_k - \rho \beta \omega^2 + \nabla \cdot ((\mu + \sigma \mu_T) \nabla \omega)$$

P_k is the production term, which is defined as

$$P_k = \mu_T (\nabla u : (\nabla u + (\nabla u)^T - \frac{2}{3} (\nabla \cdot u)^2)) - \frac{2}{3} \rho k \nabla \cdot u$$

The closure coefficients (α , β , β^* , σ , σ^*) and eddy viscosity (μ_T) are defined as

$$\alpha = \frac{13}{25}, \beta = \frac{13}{125} f_\beta, \beta^* = \frac{9}{100} f_{\beta^*}, \sigma = \frac{1}{2}, \sigma^* = \frac{3}{5}, \mu_T = \rho \frac{k}{\omega}$$

The closing coefficients β and β^* are variant with factors f_β and f_{β^*} , which depend on nondimensional measures of vortex stretching χ_k and χ_ω , defined as

$$f_\beta = \frac{1+70\chi_\omega}{1+80\chi_\omega}, f_{\beta^*} = \begin{cases} 1 & \text{if } \chi_k \leq 0 \\ \frac{1+680\chi_k^2}{1+400\chi_k^2} & \text{if } \chi_k > 0 \end{cases}$$

The parameter χ_ω is based on Pope's nondimensional measure of vortex stretching parameter (χ_p), and is simply the absolute value of χ_p . χ_k is a nondimensional parameter defined by k and ω as shown below

$$\chi_\omega = |\chi_p| = \left| \frac{\Omega_{ij}\Omega_{jk}S_{ki}}{\left(\frac{9}{100}\omega\right)^3} \right|, \chi_k = \frac{1}{\omega^3} (\nabla k \cdot \nabla \omega)$$

where Ω_{ij} is the mean rotation tensor, defined by a difference of partial derivatives of the mean velocity ($\bar{u}_i$) and S_{ij} is the mean strain-rate tensor, defined by a sum of partial derivatives of the mean velocity ($\bar{u}_i$), as shown below

$$\Omega_{ij} = \frac{1}{2} \left(\frac{\partial \bar{u}_i}{\partial x_j} - \frac{\partial \bar{u}_j}{\partial x_i} \right), S_{ij} = \frac{1}{2} \left(\frac{\partial \bar{u}_i}{\partial x_j} + \frac{\partial \bar{u}_j}{\partial x_i} \right)$$

Figure 1. Streamline flows for: (A) an orifice flow meter, (B) a Pitot Tube flow meter, and (C, D) a CPT flow meter with high (C) and low (D) confinement, including a buffer region

The circles indicate sampling ports: ① and ②: at the center of the Pitot Tube (in B–D) and on the wall of the tube (in A–D), Ⓐ: at atmospheric pressure, which is outside the flow system (in A–D), Ⓡ: at the entrance of the flow to measure flow-resistance relative to atmospheric pressure Ⓐ (in A–D).

Figure 2. (A, B) Schematic of the CPT flow meter

(C) Photograph of the flow meter showing the interior design (view from inlet). (D) Flow pattern of the flow meter as modeled axisymmetrically (model is axisymmetric about the left boundary pictured). The arrows represent the fluid velocity and the tone represents the velocity magnitude (m/s). (E) Photograph showing the profile view of the CPT flow meter (inlet to the right) placed next to a quarter coin for scale. (F) Photograph showing the different sizes of inner disks used next to a quarter coin for scale. [Color figure can be viewed in the online issue, which is available at wileyonlinelibrary.com.]

Figure 3. Schematic of the Lung Calibrator/Simulator

The black disk rotates at a predefined rate, producing forward and backward movements of a piston, which creates positive and negative flows in the outlet/inlet of the system; then a breathing waveform is produced at the outlet with both inhalation and exhalation flows.

Figure 4. (A) Pressure distribution (Pa) in the CPT flow meter simulated with 9.25 mm inner radius disk at 20 L/min flow for the laminar model

Note that the direction of flow is from bottom to top; the greatest pressure is at the center of the face of the disk, where sampling port ① is located, and the lowest pressure is on the wall immediately after the disk, where sampling port ② is located (Figure 1C, D). (B) Streamline profile of the CPT flow simulated with 9.25 mm inner radius disk at 20 L/min for the turbulent model. [Color figure can be viewed in the online issue, which is available at wileyonlinelibrary.com.]

Figure 5. (A–J) Flow calibrations (differential pressure vs. flow rate) as compared to simulated results (laminar and turbulent model), either sampled at pressure port ① (center of the disk) and atmospheric pressure ④ (right side), or two pressure ports ① and ② (left side) (see Figures 1C, D and 2 for reference)

[Color figure can be viewed in the online issue, which is available at wileyonlinelibrary.com.]

Figure 6. Comparative differential pressure vs. flow rate plots for a CPT flow meter (9.25 mm) and a Fixed Orifice (2.5-mm radius)

(A) Comparison of pressure signal from 0 to 20 L/min. (B) Comparison of pressure signal from 0 to 5 L/min. (C) Comparison of flow-resistance from 0 to 20 L/min. (D) Comparison of flow-resistance between disk sizes from 0 to 20 L/min. [Color figure can be viewed in the online issue, which is available at wileyonlinelibrary.com.]

Figure 7. Comparison of the reference flow rate (dashed, left axis) and the lung calibrator/simulator, and the square root of the measured differential pressure (solid, right axis) from a CPT flow meter

The samples were taken at 40 Hz with no filtering. [Color figure can be viewed in the online issue, which is available at wileyonlinelibrary.com.]

Figure 8. (A) Comparison of volumes measured from real breath between the CPT flow meter (9.25 mm inner radius disk) and a reference method (Oxycon)

The dotted line represents the ideal 1:1 relationship, and the solid line represents a linear regression of the actual results. (B) Difference between measured volumes of the CPT and the Oxycon system. CPT measurements had a linear correction applied. The shaded area represents the standard deviation of the measurement difference, and the reported error of the Oxycon unit at low volumes (50 mL). The overall perceptual accuracy error is <10%.

[Color figure can be viewed in the online issue, which is available at wileyonlinelibrary.com.]

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