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Calibration of pneumotachographs using a calibrated syringe

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Tang, Yongquan, Martin J. Turner, Johnny S. Yem, and A. Barry Baker. Calibration of pneumotachographs using a calibrated syringe. *J Appl Physiol* 95: 571–576, 2003. First published April 18, 2003; 10.1152/japplphysiol.00196.2003.—Pneumotachographs require frequent calibration. Constant-flow methods allow polynomial calibration curves to be derived but are time consuming. The iterative syringe stroke technique is moderately efficient but results in discontinuous conductance arrays. This study investigated the derivation of first-, second-, and third-order polynomial calibration curves from 6 to 50 strokes of a calibration syringe. We used multiple linear regression to derive first-, second-, and third-order polynomial coefficients from two sets of 6–50 syringe strokes. In *part A*, peak flows did not exceed the specified linear range of the pneumotachograph, whereas flows in *part B* peaked at 160% of the maximum linear range. Conductance arrays were derived from the same data sets by using a published algorithm. Volume errors of the calibration strokes and of separate sets of 70 validation strokes (*part A*) and 140 validation strokes (*part B*) were calculated by using the polynomials and conductance arrays. Second- and third-order polynomials derived from 10 calibration strokes achieved volume variability equal to or better than conductance arrays derived from 50 strokes. We found that evaluation of conductance arrays using the calibration syringe strokes yields falsely low volume variances. We conclude that accurate polynomial curves can be derived from as few as 10 syringe strokes, and the new polynomial calibration method is substantially more time efficient than previously published conductance methods.

polynomial; linear regression; conductance; pneumotachometer

SINCE IT WAS INTRODUCED IN 1925 by Fleisch (3), the pneumotachograph (PT) has been used widely in the respiratory laboratory, intensive care unit, bioengineering laboratory, and in clinical anesthesia. Both Fleisch and screen PTs generate differential pressures approximately proportional to the volume flow and viscosity of the gas, but independent of the pressure. As viscosity varies with the gas composition, temperature, and humidity, flow-to-differential pressure conversion of a PT depends on all of these factors, as well as on the up- and downstream geometry of the tube (2, 5, 11). Therefore, PTs must be calibrated routinely under conditions that are as close as possible to those under which measurements are performed.

Several methods for describing a PT response curve have been reported, including first- (6), second-, and third-order polynomial (4, 9) and conductance arrays (8, 12). PTs are normally linear over a limited range of flows, and, if they are used outside that range, substantial errors may result. Polynomial methods have been shown to be practical and accurate over wider flow ranges (9), but the determination of the coefficients requires numerous data points at constant flows by using an accurate flow or volume standard as reference. Existing polynomial calibration procedures can be tedious and are not suitable for frequent routine calibration.

Yeh et al. (12) developed a simpler method for calculating PTs by using multiple strokes of a calibration syringe from which conductance arrays are calculated by a weighted-averaging algorithm. Yeh's iterative syringe stroke calibration method has the advantage that a complete set of conductance values can be determined easily and with moderate effort. A disadvantage of Yeh's conductance method is the fact that adjacent conductance values are independent and can differ substantially if no additional smoothing is applied, although physical considerations suggest that the curve should be continuous and smooth.

In this study, we report a new, efficient method for determining polynomial calibration curves from a small number of syringe strokes and compare this new method of calibration with the conductance array calibration method of Yeh et al. (12).

MATERIALS AND METHODS

Apparatus. The experimental apparatus included a 3-liter precision syringe (Pulmonary Data Service Instrumentation) connected to a screen PT (model 3700A, 0–160 l/min linear range, 5 $Pa \cdot l^{-1} \cdot min$ resistance; Hans Rudolph) and an airway pressure transducer (HCXM100D6 Sensortechnics, Puchheim, Germany). The PT was connected to standard anesthesia circuit components: biological filter and 8-mm endotracheal tube. The PT differential pressure taps were connected to a Validyne MP45 differential pressure transducer (Validyne) by 20 cm of 3-mm ID Tygon tubing. This configuration has been shown to exhibit a bandwidth of ~60 Hz (10).

A carrier demodulator (Validyne CD15, Validyne) converted the differential pressure to a voltage signal. Flow and airway pressure signals were filtered by matching sixth-

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order, linear-phase, low-pass filters (Telectronics, Sydney, Australia; −3 dB at 45 Hz), digitized at 320 Hz by a 12-bit analog-to-digital converter (ADC) (PCI-MIO-16-e, National Instruments) and acquired by an IBM-compatible computer. Barometric pressure was measured by using an electronic barometer (Vaisala PTB100A, Helsinki, Finland). Data acquisition was controlled by Matlab and Simulink software by using XPC Target and Realtime Workshop toolboxes (The Math Works, Natick, MA). The airway pressure transducer was calibrated by using a precision manometer (Airflow Developments).

Methods. The syringe was pumped manually with alternate low, medium, and high flows. All data points from all syringe strokes were checked for saturation of the measurement system. The mean value of the baseline zero-flow signal was calculated by using the first and last 1,000 samples of each sequence of syringe strokes and subtracted from the flow signal.

In *part A* of this study, peak flows did not exceed 120 l/min, which is within the linear range specified by the PT manufacturer. Fifty calibration strokes and 70 additional validation strokes were made in *part A*. In *part B*, peak flows up to 250 l/min were used, which exceeded the linear limit specified by the PT manufacturer by ~60%. The gain of the carrier demodulator was reduced to accommodate the higher differential pressures. Fifty calibration strokes and an additional 140 validation strokes were made in *part B*.

Theory. If the zero-flow offset is removed from the ADC output, a second-order polynomial for calculating the ith flow value ($\dot{V}_i$) as a function of the ith ADC output (n_i) can be expressed as

$$\dot{V}_i = b_1 n_i + b_2 n_i^2$$

where b_1 and b_2 are unknown constants. If the sample interval is T_s, the volume of gas (V) that flows through the PT can be calculated as follows

$$V = T_s \sum_i \dot{V}_i = T_s \sum_i (b_1 n_i + b_2 n_i^2) = b_1 T_s \sum_i n_i + b_2 T_s \sum_i n_i^2$$

The initial and final syringe pressures are atmospheric when the flow is zero, but the resistance of the airway or the connectors (in this study an endotracheal tube) causes gas exiting the syringe to flow through the PT at increased pressure, resulting in reduced volume flow in the PT. Resistance PTs are volume-flow transducers; therefore, comparisons with a volume standard require corrections for pressure changes. Each calculated flow sample requires correction for pressure changes in the PT as follows

$$V = T_s \sum_i k_i \dot{V}_i = T_s \sum_i k_i (b_1 n_i + b_2 n_i^2)$$

Hence

$$V = b_1 T_s \sum_i k_i n_i + b_2 T_s \sum_i k_i n_i^2 \qquad (1)$$

where k_i is given by

$$k_i = \frac{(P_B + P_i)}{P_B} \qquad (2)$$

where P_B is the barometric pressure and P_i is the airway pressure associated with the ith flow sample. In cases in which the airway geometry is such that the pressure in the PT remains at atmospheric pressure during calibration, then the pressure correction term k_i can be omitted.

Equation 1 is linear in the coefficients b_1 and b_2 and applies to any syringe stroke. Unique values of the coefficients b_1 and b_2 can be determined from any two different syringe strokes of known volumes. If more than two different strokes are made, then multiple linear regression (1) can be used to obtain better estimates of b_1 and b_2 (see APPENDIX A).

Data processing. We used multiple linear regression (see APPENDIX A) to estimate the coefficients of first-, second-, and third-order polynomials from sets of 6, 10, 20, 30, 40, and 50 syringe strokes. Regression software was written by using Matlab (The Math Works). Coefficients of n^2 in the second-order polynomials and n^3 in the third-order polynomials were tested for significance by examination of the 95% confidence intervals (with Bonferroni correction). With the use of each set of polynomial coefficients, the volume errors of the same set of calibration strokes and the additional sets of validation strokes were analyzed. Calibration data from *parts A* and *B* were processed separately. Conductance values for sets of 6, 10, 20, 30, 40, and 50 calibration syringe strokes were calculated by using the algorithm of Yeh et al. (12), modified to include airway pressure correction and iterated four times, as suggested by Stromberg and Gronkvist (8) (see APPENDIX B).

With the use of these conductance arrays, volume errors for the same set of calibration strokes and for the validation stokes were calculated.

Mean volume errors from all of the polynomial groups were compared with the mean volume errors of the conductance groups with corresponding numbers of syringe strokes, and with the 50-stroke conductance group. The variances of the volume errors in the second- and third-order polynomial groups were compared with the variances of the conductance groups with the corresponding number of syringe strokes, and with the 50-stroke conductance group. Volume errors in the 50-stroke conductance groups calculated by using the calibration strokes were compared with the corresponding errors obtained by using the evaluation strokes. Results are shown as means ± SD. One-way ANOVA with Tukey's post test was performed by using Prism V3.00 (GraphPad Software). The comparison of variances was conducted by using the two-tailed F-test with Bonferroni correction for multiple comparisons. $P < 0.05$ was considered to be statistically significant.

First-, second-, and third-order polynomials were derived from 50-stroke flow data of *part B* without correction for airway pressure and compared with polynomials derived with airway pressure correction to assess the error that would be incurred if pressure correction were omitted.

RESULTS

The second- and third-order polynomials and conductance arrays derived from 50 syringe strokes and normalized by subtraction of the first-order calibration curve are plotted in Fig. 1, *A* (*part A*) and *B* (*part B*). In Fig. 1*A*, the second- and third-order polynomial curves overlap over most of the flow range. The conductance values are randomly distributed around the higher order polynomial curves in Fig. 1. The 95% confidence intervals of the coefficients of n^2 in the second-order polynomials exclude zero in all groups of *parts A* and *B*, indicating significance. The 95% confidence intervals of the coefficients of n^3 include zero in all groups, indicating that these coefficients are not significant. The 95% confidence intervals of the third-order polynomial coefficients of both n^2 and n^3 in the six stroke groups of

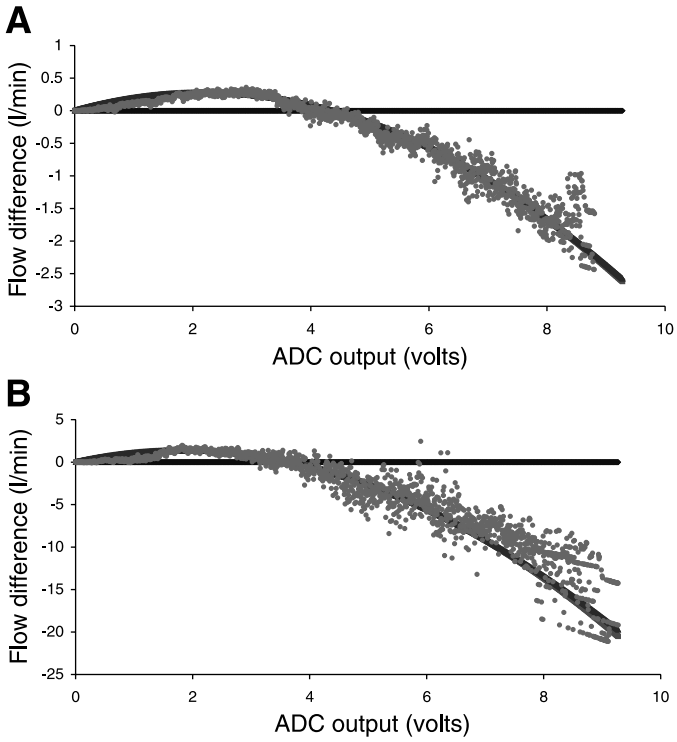

Fig. 1. Calibration curves obtained from 50 syringe strokes normalized by subtraction of the first-order curve [*part A* low flow (*A*); *part B* high flow (*B*)]. The thick horizontal line at zero difference represents the first-order polynomial. The second- and third-order polynomial curves are almost exactly superimposed and are represented by the thick curved lines. The conductance array is represented by the dotted points surrounding the polynomial curves. ADC, analog-to-digital converter.

both *parts A* and *B* included zero, indicating lack of statistical significance.

Figure 2 shows means ± SDs of the volume errors evaluated by using the same set of 6–50 syringe strokes from which the calibration curves were derived. There are no statistically significant differences in mean errors in *part A* or *B*. The variances of the first-order polynomial groups in *parts A* and *B* are all significantly larger than the variances of the corresponding conductance groups in *parts A* and *B*. The variances of the 20-stroke second- and third-order polynomial groups in *part A* and the six-stroke second- and third-order polynomial groups in *part B* are significantly different from the variances of the corresponding conductance groups in *part B*. In *part B*, the variances of the six-stroke third-order polynomial group and the 6-, 10-, and 20-stroke conductance groups are smaller than the variance of the 50-stroke conductance group.

Figure 3 shows means ± SDs of volume errors evaluated by using the separate set of 70 validation syringe strokes in *part A* and 140 validation syringe strokes in *part B*. In *part A*, none of the variances of the second- and third-order polynomial groups differed from the variance of the 50-stroke conductance group. In *part B*, all of the second- and third-order polynomial groups, with the exception of the six-stroke third-order high-

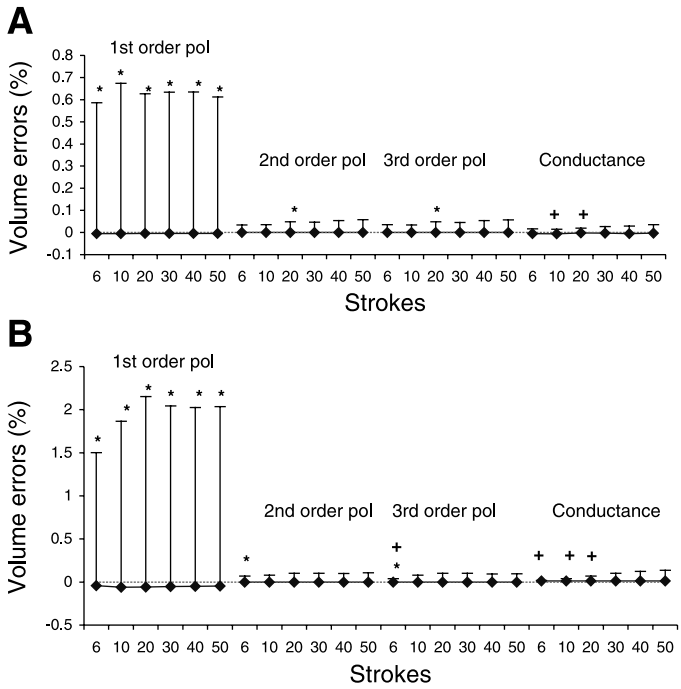

Fig. 2. Volume errors obtained by analysis of the same syringe strokes used to derive the calibration curves [*part A* (*A*); *part B* (*B*)]. Values are means ± SD. Variance differs significantly from *corresponding conductance group and from $^{+}$50-stroke conductance group: $P < 0.05$. pol, Polynomial.

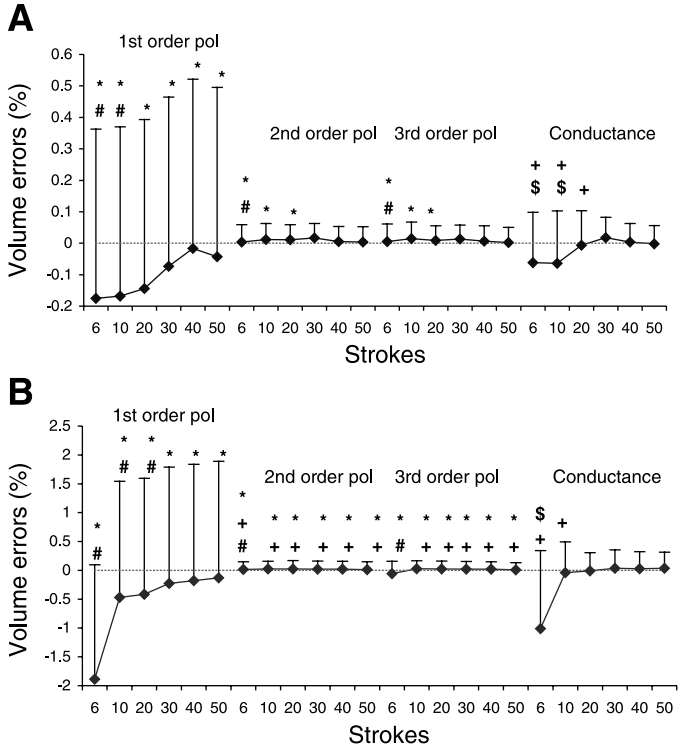

Fig. 3. Volume errors obtained by analysis of 70 validation strokes (*A*) and by analysis of 140 validation strokes in the nonlinear range (*B*). Values are means ± SD. Mean differs significantly from $^{\#}$corresponding conductance group and from $^{\$}$50-stroke conductance group; variance differs significantly from *corresponding conductance group and from $^{+}$50-stroke conductance group: $P < 0.05$.

flow group, exhibit variances that are significantly smaller than both the corresponding conductance groups and the 50-stroke conductance group. The variances in the first-order polynomial groups are all significantly bigger than corresponding variances in the conductance groups. The mean volume errors in all of the six-stroke second- and third-order polynomial groups are significantly smaller than the mean volume errors in the corresponding conductance groups.

Hence, in *parts A* and *B* both, both second- and third-order polynomials yield volume errors with variability that is either not different from or is significantly smaller than the volume variability delivered by the conductance method at 50 strokes.

The variances of the 50-stroke conductance volume errors calculated by using the calibration strokes (Fig. 2) were significantly smaller than the corresponding variances calculated by using the evaluation strokes (Fig. 3).

Airway pressures measured in *part B* ranged between 0 and 10.2 kPa. The differences between first-, second-, and third-order polynomials derived without and with airway pressure correction, as well as the rise in airway pressure expressed as a percentage of barometric pressure, are shown in Fig. 4. Second- and third-order calibration polynomials derived without airway pressure correction overestimate flow by approximately the same factor as the airway pressure increase, whereas the first-order curve overestimates flow by a constant 2.2%.

DISCUSSION

This study describes a calibration method in which polynomial calibration curves for a screen PT can be derived from as few as 10 different strokes of a calibration syringe. Our results show that second- and third-order polynomial calibration curves derived from 10 syringe strokes yield volume errors with means and variances that are not greater than those obtained from conductance arrays derived from 50 syringe strokes in a set of independent validation data. When the flow is extended 60% beyond the linear range of the PT, a 10-stroke polynomial calibration yields volume variances that are smaller than those obtained by the conductance method with 50 strokes. Our polynomial technique is substantially more efficient than the conductance method for flows up to 60% greater than the linear range of the PT.

Yeh et al. (12) recommended 50–100 calibration strokes for the determination of conductance arrays, which is time consuming and tedious, especially if frequent calibration is required. Our results (Fig. 3*A*) suggest that calibration with the use of the conductance method needs at least 30 syringe strokes to achieve stable results. If fewer calibration strokes are used to determine conductance arrays, the accuracy decreases, and individual conductance values vary as fewer data points are obtained at each differential pressure.

The weighted averaging process of the conductance method can result in adjacent conductance values that differ substantially, leading to discontinuous conductance curves (Fig. 1), if additional smoothing is not used. In some cases, a large number of syringe strokes is required to generate an acceptably smooth curve, particularly at low and high flows, which are difficult to generate with a syringe by hand. The continuous, smooth nature of polynomials matches the expected physical characteristics of the PT and facilitates the determination of smooth calibration curves from a small number of strokes.

Figure 2 shows that falsely accurate results can be obtained if calibration curves are evaluated by using the same data set from which the curves are derived. The conductance method appears to be particularly susceptible to this effect. Our study suggests that calibration curves should be evaluated by using a separate data set.

We found that an assumption of a linear relationship between flow and differential pressure results in volume errors that are significantly higher than those obtained by using polynomial relationships, even when the maximum flow is well within the manufacturer-specified linear range. The volume errors are, however, small physiologically (SD $< 0.6\%$, Fig. 3*A*), if the PT used in this study is used within its linear range. Linear resistance PTs are available from many manufacturers and cover a wide range of flows. PTs not identical to the device calibrated in this study might exhibit stronger nonlinearity inside the manufacturer-specified linear region. Appropriately selected higher order polynomial calibration methods should allow most nonlinearity errors at any flow range to be reduced. A limitation of polynomial calibration curves is that a high-order polynomial might be required for a PT with a strong nonlinearity, and it is more difficult to estimate the coefficients of high-order polynomials. In our case, the minor nonlinearity within the manufacturer-specified linear region linear range can be corrected (Fig. 3*A*), and the dynamic range of the PT can be extended

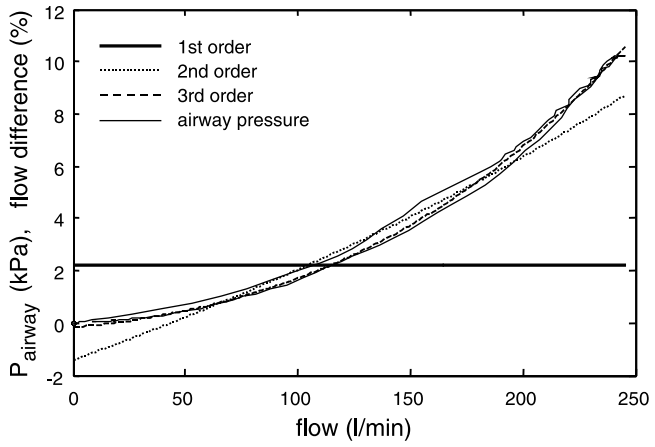

Fig. 4. Percent difference between 50-stroke calibration polynomials derived without airway pressure (P_{airway}) correction and polynomials derived with pressure correction. The P_{airway} of the highest flow calibration stroke, expressed as a percentage of barometric pressure, is superimposed.

well beyond the specified range, with acceptable accuracy with a second-order polynomial (Fig. 3*B*).

The volume errors obtained in this in vitro study are substantially smaller than might be expected from clinical measurements in which temperature, humidity, gas composition, and airway geometry often vary. The emphasis of this study, however, is on the comparison of two calibration methods and not the low absolute errors that were obtained under ideal laboratory conditions.

Fleisch and screen PTs are volume flow transducers that generate differential pressures proportional to the product of volume flow and gas viscosity (9). Viscosity is independent of pressure at pressures encountered in physiological measurements (7); therefore, at constant temperature, the differential pressure is proportional to volume flow. It is standard practice to correct calibration data for pressure and, if necessary, temperature differences between a volume standard and the PT during calibration (9). In our study, the filter and 8-mm endotracheal tube, which were connected downstream of the PT, caused the pressure in the PT to increase at high flow. At sea level, the volume of an ideal gas flowing through a PT at increased pressure is reduced by ~1%/kPa. Consequently, we expect calibration curves forced to fit data that are not corrected for airway pressure to overestimate volume flow by ~1% for every kilopascal that airway pressure rises above barometric pressure. Our results reveal that polynomials derived from uncorrected data cause systematic overestimation of volume flow by a factor approximately equal to the increase in airway pressure over barometric pressure (Fig. 4). In applications in which airway pressure remains close to atmospheric pressure (e.g., spirometry), pressure correction may not be necessary during calibration or use. When airway pressure varies by >1–2 kPa, for example, during calibration with high-resistance airway connectors or during mechanical ventilation of subjects with stiff lungs, pressure compensation is necessary to avoid systematic errors. In this study, we analyzed only inspiratory flow in which gas flows from the syringe through the PT, filter, and endotracheal tube. If a syringe is used to derive calibration curves for expiratory flow in a similar airway geometry, then the airway pressure will be reduced as the syringe plunger is withdrawn, and calibration curves derived from uncorrected data will underestimate volume flow.

Conclusions. This study shows that polynomial calibration curves for PTs can be obtained by using a linear regression technique based on as few as 10 strokes of an appropriately sized syringe. Our new polynomial calibration method is more time efficient and produces results equivalent to, or better than, previously published conductance methods.

APPENDIX A: MULTIPLE LINEAR REGRESSION

Equation 1 in the text describes a second-order relationship between the ADC output and the integrated flow in the PT. If N strokes of a syringe of volume V_s are made and *Eq. 1* is generalized to a polynomial of order p, we can write a set of N simultaneous equations as follows

$$\begin{aligned}
V_s &= b_1 T_s \left(\sum_i k_i n_i\right)_1 + b_2 T_s \left(\sum_i k_i n_i^2\right)_1 + \cdots + b_p T_s \left(\sum_i k_i n_i^p\right)_1 + \epsilon_1 \\
V_s &= b_1 T_s \left(\sum_i k_i n_i\right)_2 + b_2 T_s \left(\sum_i k_i n_i^2\right)_2 + \cdots + b_p T_s \left(\sum_i k_i n_i^p\right)_2 + \epsilon_2 \\
&\vdots \qquad\quad \vdots \qquad\qquad \vdots \qquad\qquad \vdots \qquad\quad \vdots \\
V_s &= b_1 T_s \left(\sum_i k_i n_i\right)_N + b_2 T_s \left(\sum_i k_i n_i^2\right)_N + \cdots + b_p T_s \left(\sum_i k_i n_i^p\right)_N + \epsilon_N
\end{aligned}$$

where $\epsilon_{1 \ldots N}$ are the errors in the calculated volumes of each stroke. These equations can be written in matrix form

$$Y = X\beta + E$$

where Y is a column vector of N syringe volumes; β is a column of p unknown polynomial coefficients; E is a column vector of N volume errors; and X is an $(N \times p)$ matrix

$$X = \begin{bmatrix} b_1 T_s \left(\sum_i k_i n_i^1\right)_1 & \cdots & b_p T_s \left(\sum_i k_i n_i^p\right)_1 \\ \vdots & \ddots & \vdots \\ b_1 T_s \left(\sum_i k_i n_i^1\right)_N & \cdots & b_p T_s \left(\sum_i k_i n_i^p\right)_N \end{bmatrix}$$

A least squares regression estimate B of the polynomial coefficient vector is obtained by evaluating the matrix (1)

$$B = (X'X)^{-1}X'Y$$

If the data set contains too many similar syringe strokes, then the matrix $(X'X)$ might be ill-conditioned. Syringe strokes should be made with random variations in peak flow and flow profile. Estimates of the individual confidence intervals (ci) for the coefficient are given by

$$\mathrm{ci}_j = b_j \pm t(\upsilon, 1 - \alpha/2)s_j \quad j = 1 \cdots p$$

where s_j is the estimated standard deviation of the jth coefficient given by the jth element of the diagonal of the matrix $(X'X)^{-1}\, e_s$, and e_s is the standard error of the regression given by

$$e_s = \frac{\sum_i (\hat{Y}_i - Y_i)^2}{N - p}$$

APPENDIX B: DETERMINATION OF CONDUCTANCE ARRAY

The conductance array C_r of a PT (12) is defined by the equation

$$\dot{V} = C_r n_r \quad r = 0 \ldots 2{,}047$$

where r refers to the possible values that n can take (0–2,047 for a binary 12-bit ADC and bidirectional flow), and C_r are 2,047 conductance values associated with the 2,047 possible ADC outputs.

Initial estimates C_q of the conductance values of the PT are obtained from N syringe strokes by using the equation

$$C_q = \left(\frac{V_s}{T_s \sum_i k_i n_i} \right)_q \quad q = 1 \ldots N$$

where i refers to the ADC samples covering each syringe stroke and k_i are the pressure correction factors defined in *Eq. 2* in the text.

The initial conductance array is calculated by using the equation

$$C_r = \frac{\sum_q (C w_r)_q}{\sum_q (w_r)_q} \quad r = 1 \ldots 2{,}047$$

where w_r is the number of times the rth ADC value occurs in the qth syringe stroke. Conductances associated with ADC values that are not covered in a set of syringe strokes are set to the average of the 11 neighboring conductances.

To improve an existing set of conductance estimates, the following procedure is used.

First, using the current conductance array, the volumes (V_q) of a set of N syringe strokes are calculated

$$V_q = T_s \sum_i (C_i k_i n_i)_q \quad q = 1 \ldots N$$

Second, a set of volume correction factors g_q are calculated for each stroke

$$g_q = \frac{V_s}{V_q} \quad q = 1 \ldots N$$

Third, conductance correction factors e_r are calculated as weighted averages of the volume correction factors

$$e_r = \frac{\sum_q (g w_r)_q}{\sum_q (w_r)_q} \quad r = 1 \ldots 2{,}047$$

Fourth, conductance values are corrected

$$C_{r\,\text{corrected}} = e_r C_r \quad r = 0 \ldots 2{,}047$$

Conductances associated with ADC values that are not covered in a set of syringe strokes are set to the average of the 11 neighboring conductances.

This study has been accepted for presentation in part at the 77th International Anesthesia Research Society Clinical and Scientific Congress, New Orleans, LA, 2003.

DISCLOSURES

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REFERENCES

1. **Draper N and Smith H.** *Applied Regression Analysis*. New York: Wiley, 1981.
2. **Finucane KE, Egan BA, and Dawson SV.** Linearity and frequency response of pneumotachographs. *J Appl Physiol* 32: 121–126, 1972.
3. **Fleisch A.** Der pneumotachograph: ein apparat zur geschwindigkeisregistrerung der atemluft. *Pflügers Arch* 209: 713–722, 1925.
4. **Giannella-Neto A, Bellido C, Barbosa RB, and Melo MF.** Design and calibration of unicapillary pneumotachographs. *J Appl Physiol* 84: 335–343, 1998.
5. **Kreit JW and Sciurba FC.** The accuracy of pneumotachograph measurements during mechanical ventilation. *Am J Respir Crit Care Med* 154: 913–917, 1996.
6. **Pearce DH, Milhorn HT Jr, Holloman GH, and Reynolds WJ.** Computer-based system for analysis of respiratory responses to exercise. *J Appl Physiol* 42: 968–975, 1977.
7. **Reid RC, Prausnitz JM, and Poling BE.** *The Properties of Gases and Liquids*. New York: McGraw-Hill, 1987.
8. **Stromberg NO and Gronkvist MJ.** Improved accuracy and extended flow range for a Fleisch pneumotachograph. *Med Biol Eng Comput* 37: 456–460, 1999.
9. **Turner MJ, MacLeod IM, and Rothberg AD.** Calibration of Fleisch and screen pneumotachographs for use with various oxygen concentrations. *Med Biol Eng Comput* 28: 200–204, 1990.
10. **Turner MJ, Macleod IM, and Rothberg AD.** Measurement of the frequency response and common-mode gain of neonatal respiratory pressure and flow measurement systems. Part 2: Results. *Clin Phys Physiol Meas* 10: 231–240, 1989.
11. **Turney SZ and Blumenfeld W.** Heated Fleisch pneumotachometer: a calibration procedure. *J Appl Physiol* 34: 117–121, 1973.
12. **Yeh MP, Gardner RM, Adams TD, and Yanowitz FG.** Computerized determination of pneumotachometer characteristics using a calibrated syringe. *J Appl Physiol* 53: 280–285, 1982.

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Pages 571–576: Yongquan Tang, Martin J. Turner, Johnny S. Yem, and A. Barry Baker. "Calibration of pneumotachographs using a calibrated syringe" (http://jap.physiology.org/cgi/content/full/95/2/571). The equation on page 575 defining the matrix X contains a typographical error. The parameters $b_1 \ldots b_p$ should not appear in this equation. The new equation appears below.

$$X = \begin{bmatrix} T_s\left(\sum_i k_i n_i^l\right)_1 & \cdots & T_s\left(\sum_i k_i n_i^p\right)_1 \\ \vdots & \ddots & \vdots \\ T_s\left(\sum_i k_i n_i^1\right)_N & \cdots & T_s\left(\sum_i k_i n_i^p\right)_N \end{bmatrix}$$