applied with a surface coil of 2·5 cm inner diameter in close proximity to the electrodes. The stimulus was sufficient to cause large and extensive muscle contractions, but with the new preamplifier design did not disrupt the recording of the signal, with no saturation, distortion or baseline drift. A conventional amplifier would not have coped with these conditions and consequently part of the ECG trace would have been obscured.

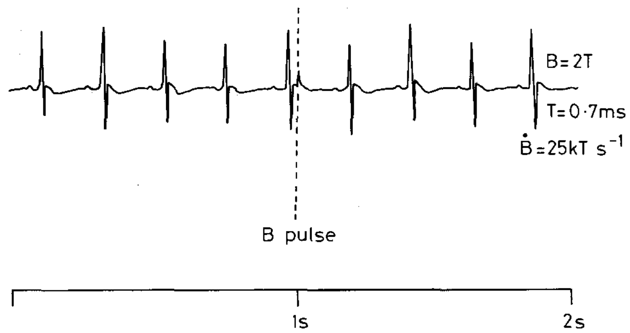

Fig. 4 *Rat ECG signals collected in the presence of a large magnetic stimulus using the new preamplifier and CMOS switching during the application of the magnetic field pulse*

The present design of the preamplifier does not comply with existing safety standards for medical electronic equipment (IEC 601–1, International Electrotechnical Commission, Geneva or BS5724, British Standards Institute, London). To achieve compliance and to be suitable for patient use, earthing of the patient should be avoided.

4 Conclusions

A novel principle for the preamplifier stage of a general-purpose electrophysiological waveform measurement has been described. The preamplifier compensates for DC offsets and drift without the use of capacitor coupling or a high-pass filter, using instead a DC rejection circuit. It recovers very rapidly from large unwanted input signals, as occurred in magnetic stimulation experiments. The preamplifier could be adapted for patient use by the inclusion of electrical isolation circuitry and by utilising the driven right leg configuration for the reduction of electrical interference.

References

McRobbie, D. W. (1985) Design and instrumentation of a magnetic nerve stimulator. *J. Phys. E: Sci. Instrumen.*, **18,** 74–78.

McRobbie, D. W. and Foster, M. A. (1985) Cardiac response to pulsed magnetic fields with regard to safety in NMR imaging. *Phys. in Med. & Biol.*, **30,** 695–702.

McRobbie, D. W. (1987) Relationships between changing B and induced electrical and physiological effects. In *Safety assessment of NMR clinical equipment.* Schmidt, K. (Ed.), Thieme, Stuttgart, New York, 64–70.

Technical note

Calibration of Fleisch and screen pneumotachographs for use with various oxygen concentrations

M. J. Turner I. M. MacLeod A. D. Rothberg

Departments of Electrical Engineering and Paediatrics & Child Health, University of the Witwatersrand and Johannesburg Hospital, P O Wits, 2050, South Africa

Keywords—*Artificial respiration, Calibration, Fleisch pneumotachograph, Intensive care unit, Mechanical ventilation, Oxygen concentration, Screen pneumotachograph*

Med. & Biol. Eng. & Comput., 1990, **28**, 200–204

1 Introduction

In an intensive care unit where patients are mechanically ventilated for respiratory insufficiency, inspired oxygen concentrations may range from 21 per cent to 100 per cent. Consequently, inspired gas viscosity may vary from that of humid air at approximately 187·4 μP to humid oxygen at approximately 207·3 μP (values calculated at 35°C using equations described elsewhere (Turner *et al.*, 1989)). Because the differential pressure developed by both Fleisch and screen pneumotachographs is proportional to gas viscosity (Johns *et al.*, 1982; Keller, 1963) recalibration is necessary each time the inspired oxygen content changes.

In the clinical situation patients might be studied repeatedly while being ventilated with increasing or decreasing oxygen concentrations. We therefore consider it important to be able to use a single calibration curve for any inspired oxygen fraction. Clinical measurements can then be made without having to perform a time-consuming calibration immediately before or after each measurement.

Correspondence should be addressed to M. J. Turner, Department of Electrical Engineering, University of the Witwatersrand, P O Wits, 2050, South Africa

First received 12th April and in final form 13th September 1989

In a capillary tube of a Fleisch pneumotachograph the Poiseuille equation describes the theoretical relationship between flow $\dot{V}$ and differential pressure P: $\dot{V} = P\pi r^4/8\,\mu l$ where μ is viscosity, r is capillary tube radius and l is length between pressure tappings. This equation can be rearranged to yield: $\mu\dot{V} = P\pi r^4/8l = KP$ where K is a constant dependent only on pneumotachograph characteristics. A similar equation can be derived for the screen pneumotachograph. If the pneumotachograph is perfectly linear then K can be determined from two or more calibration points in which $\dot{V}$, P and μ are known. In a measurement situation $\dot{V}$ can be calculated from measured P and calculated μ (TURNER *et al.*, 1989). However, it has been shown experimentally that the pressure developed across the Fleisch and screen pneumotachographs is in fact nonlinearly related to flow (FINUCANE *et al.*, 1972). Hence the Poiseuille and KELLER (1963) equations are not strictly valid and simple linear viscosity compensation for composition change may not yield adequate accuracy.

In this note we propose the use of a 'standard calibration polynomial' (SCP) to describe the nonlinear relationship between $\mu\dot{V}$ and P for Fleisch and screen pneumotachographs and determine the bounds of the error which may be incurred when an SCP is derived at one O_2 concentration and used at another. A similar (linear) technique has been implicitly suggested (TURNEY and BLUMENFELD, 1973), but has not been validated.

2 Methods

As an SCP is theoretically independent of gas properties, SCPs derived from calibration with different O_2 concentrations should be identical. If there is dependence on gas properties other than viscosity (e.g. density) then we would expect an SCP derived using an O_2 concentration midway between air and pure O_2 (60 per cent O_2) to be different from SCPs derived using either air or pure O_2. The difference (expressed in $\mathrm{ml\,s^{-1}}$) represents the maximum error in flow which would be obtained if calibration were carried out at 60 per cent O_2 and measurement carried out at any other O_2 concentration. We examined the differences between SCPs obtained from neonatal Fleisch and screen pneumotachographs using pure O_2, air and 60 per cent O_2.

We calibrated a Fleisch 00 pneumotachograph and a screen (Hans Rudolph 8300) pneumotachograph, each unheated and connected to a differential pressure transducer (Validyne MP45 $\pm$ 2 cm H_2O). The pneumotachographs were connected between a 3 mm endotracheal tube and a neonatal ventilator T-piece to imitate airway geometry in the intensive care unit. We defined flow into the lungs as positive. Our volume standard was a 7 litre wet-seal spirometer (Godart Statham 17330), which was calibrated against a 4 litre syringe (Cavitron PC3006) over the appropriate volume range. The 60 per cent oxygen mixture was obtained from the hospital oxygen supply through a blender (Sechrist 3500 HL). For each new gas mixture the spirometer was thoroughly equilibrated by filling and emptying several times, being allowed to stand for approximately an hour, then repeating the filling and emptying procedure immediately prior to calibration. Gas samples from the spirometer were analysed with a zirconium oxide analyser (Westinghouse model 211) accurate to ± 1 per cent. An uncertainty of 1 per cent in O_2 concentration leads to approximately 0·1 per cent uncertainty in viscosity.

HUGHES (1981) stated that neonatal pneumotachographs should be capable of measuring flows up to $400\,\mathrm{ml\,s^{-1}}$. We estimate a more reasonable maximum flow of $250\,\mathrm{ml\,s^{-1}}$ as follows. In our neonatal intensive care unit infants with severe lung disease have been ventilated with peak airway pressures up to 5 kPa. The time constant of the rising pressure waveform generated by neonatal ventilators has been reported as 10 ms (FIELD *et al.*, 1985). Thus 95 per cent of peak airway pressure appears across total respiratory resistance in 30 ms. Total respiratory resistance may be $2{\cdot}3\,\mathrm{kPa\,s\,litre^{-1}}$ (LESOUEF *et al.*, 1984), yielding a theoretical maximum flow of $220\,\mathrm{ml\,s^{-1}}$. We used 15 equally spaced gas flows of 0–$250\,\mathrm{ml\,s^{-1}}$. When setting up our apparatus we observed temperature and composition changes (humidification) in dry gas flowing into a water-sealed spirometer. We therefore generated constant flows by connecting the pneumotachographs to hospital wall suction (-60 kPa) through a needle valve. Flow variation during acquisition of a single data point was verified to be less than 0·5 per cent by analysis of differential pressure transducer output.

Gas temperature was measured immediately up- and downstream of the pneumotachographs with chromel-alumel thermocouples (model 300, Paul Beckman, Pennsylvania, USA). The temperature in the spirometer was measured with a thermistor (Siemens M85) whose resistance was indicated on a digital ohm-meter (Fluke 8060A). The thermocouples and thermistor and ohm-meter combination were calibrated with 0·1°C resolution in an agitated water bath against a mercury-in-glass thermometer accurate to 0·1°C. Pressure at the pneumotachogaphs was measured with a water manometer accurate to 0·01 kPa (D. Giles Co., Johannesburg, South Africa).

Gas in the spirometer was mixed and humidified prior to the measurement of each calibration point with a small brushless DC electric fan (Papst type 812 12 V 3 W, St Georgen, Federal Republic of Germany). The fan also dissipated sufficient energy to supply the latent heat of evaporation of the moisture which humidified each batch of gas, thereby keeping the spirometer temperature approximately constant from one calibration point to the next.

Voltages proportional to pneumotachograph differential pressure and pneumotachograph gas temperature were digitised at 100 samples $\mathrm{s^{-1}}$ using a 12-bit analogue-to-digital convertor (Dash-8, Metrabyte) plugged into an IBM-compatible computer (Olivetti M24). Data acquisition and analysis software was developed using Intel 8086 assembler language and the Forth-like language 'Asyst' (Macmillan Software Co., USA).

Gas flow from the spirometer was corrected for the temperature and pressure differences between the spirometer and pneumotachograph using standard equations. Gas in the spirometer was assumed to be at barometric pressure due to the low friction of the spirometer bearings and the large surface area of the top of the bell (TURNEY and BLUMENFELD, 1973).

The data were analysed by polynomial regression of the product of flow ($\mathrm{ml\,s^{-1}}$) and viscosity (μP) against differential pressure represented by the digital number generated by the analogue-to-digital convertor. Six separate polynomials for each pneumotachograph were fitted to the data sets for positive and negative flows of air, O_2 and 60 per cent O_2. The order of polynomials was determined by plotting the standard error of the regressions against order from first order to fifth order (DRAPER and SMITH 1981). The variance of each of the regression coefficients was calculated, as was the 95 per cent confidence interval of the regression lines (DRAPER and SMITH, 1981). The SCPs obtained using air, 60 per cent O_2 and pure O_2 were compared by plotting the difference between the polynomials for pure O_2 and 60 per cent O_2, and air and 60 per cent O_2.

Gas viscosities were calculated using equations which we have previously shown to be accurate to within 1 per cent (TURNER *et al.*, 1989). Gas flowing through the pneumotachograph was assumed to be saturated with water vapour at the lower of the two measured temperatures (pneumotachograph, spirometer). The partial pressure of water vapour was calculated according to BARENBRUG (1974).

3 Results

The spirometer volume calibration yielded a coefficient of variation of 0·3 per cent from 20 successive calibrations. Maximum pressure and temperature differences between spirometer and pneumotachographs were 2 kPa and 1°C, respectively.

Twelve SCPs were obtained from the two pneumotachographs: one in each direction for each of the three O_2 concentrations. Fig. 1 shows the standard errors (in ml s^{-1} air flow) of the 12 data sets as a function of polynomial order. Standard error decreases with increasing polynomial order at low order but little improvement is obtained through the use of high-order polynomials. Mean standard errors for the screen pneumotachograph with a second-order polynomial is 0·34 ml s^{-1}, and for the Fleisch pneumotachograph with a third-order polynomial is 0·32 ml s^{-1} equilvalent air flow. Based on this result, third-order polynomials were used for Fleisch calibration data and second-order for screen data.

Analysis of the regressions showed that all regression coefficients were significantly different from zero with $p < 0{\cdot}001$.

Fig. 2 shows the polynomial curves plotted against digital values, which are proportional to differential pressure. A digital value of 2048 represents zero differential pressure, and values above and below 2048 represent positive and negative differential pressures, respectively.

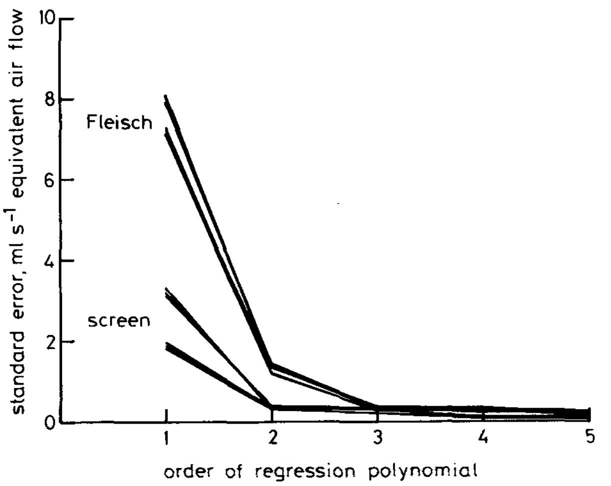

Fig. 1 *Standard error of 12 polynomial regressions of calibration data against order of polynomials. Equivalent air flow was obtained by dividing the flow-viscosity product by 187 μP*

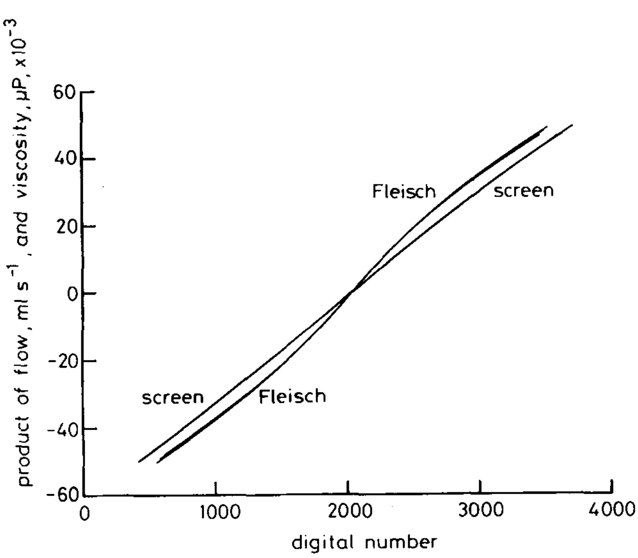

Fig. 2 *Calibration polynomials for Fleisch and screen pneumotachographs*

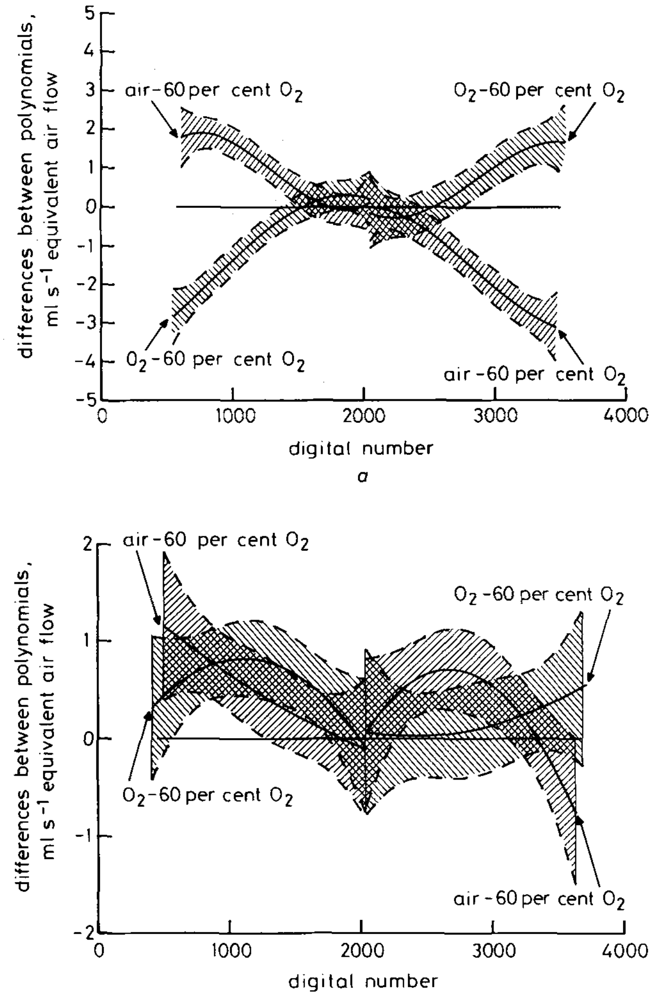

Fig. 3 *Difference between polynomials for O_2 and those for 60 per cent O_2 and polynomials for air and those for 60 per cent O_2 for (a) Fleisch and (b) screen pneumotachographs. Solid lines are mean differences, cross-hatched areas bounded by broken lines are 95 per cent confidence intervals of the differences. Equivalent air flow was obtained by dividing the flow-viscosity product by 187 μP. Note the different vertical scales in (a) and (b)*

The difference between the polynomials for O_2 and 60 per cent O_2, and air and 60 per cent O_2 are shown in Fig. 3*a* for the Fleisch pneumotachograph and Fig. 3*b* for the screen. The horizontal lines represent the (reference) curves for 60 per cent oxygen and the cross-hatched areas bounded by the broken lines are the 95 per cent confidence intervals for the differences (DRAPER and SMITH, 1981).

4 Discussion

The standard errors for the polynomials fitted to the calibration data are less than 0·4 ml s^{-1} or less than 0·2 per cent of maximum flow (250 ml s^{-1}), indicating the precision with which calibration can be carried out using a wet-seal spirometer. The characteristic of the screen pneumotachograph is more linear than that of the Fleisch (Fig. 2), allowing a lower-order polynomial to describe the data adequately.

An assumption of linear relationship between flow and differential pressure (TURNEY and BLUMENFELD, 1973) results in a standard error of approximately 8 ml s^{-1} (3 per cent of maximum flow) in the case of the Fleisch calibration curve and 4 ml s^{-1} (1·5 per cent of maximum flow) in the case of the screen (Fig. 1).

If the SCP is dependent on gas composition then calibration at 60 per cent O_2 would be expected to yield optimum error performance over the range 21–100 per cent O_2. The differences between SCPs plotted in Fig. 3 indicate the error which would be obtained if the pneumotachographs were calibrated at 60 per cent O_2 and used at 100 per cent or 21 per cent, O_2 i.e. the two worst cases. Although the SCPs for air and pure oxygen are significantly different from those for 60 per cent O_2 in the Fleisch pneumotachograph (Fig. 3*a*), one can be 95 per cent confident that the differences are less than approximately 4 ml s^{-1} equivalent air flow (1·6 per cent of maximum flow). In the case of the screen pneumotachograph the mean differences are smaller than 1 ml s^{-1} (0·4 per cent of maximum flow, the same order of magnitude as the uncertainty in spirometer calibration). The shaded confidence intervals (Fig. 3*b*) indicate that the differences are not significant at the 5 per cent level over substantial regions of flow. We postulate that the larger differences seen in the Fleisch pneumotachograph are associated with the higher Reynolds number in that transducer, related to the large capillary diameter relative to mesh dimensions in the screen pneumotachograph. The higher error obtained with the Fleisch pneumotachograph is consistent with its greater nonlinearity (Fig. 2)

Many different devices have been used as flow or volume standards against which pneumotachographs have been calibrated. Direct comparison calibration techniques have included the use of rotameters (JOHNS *et al.*, 1982; MULLER and ZAMEL, 1981), orifice plate flowmeter (BERTRAM, 1984), piston pumps (MILLER and PINCOCK, 1986; VAIDA *et al.*, 1983), turbine flowmeters (SODAL *et al.*, 1977) and the isothermal escape of compressed gas from a cylinder (PEDERSON *et al.*, 1983) as secondary standards. The manufacturer's specification of pneumotachograph resistance has also been used (FOND *et al.*, 1986), apparently without consideration of the effect of upstream geometry (FINUCANE *et al.*, 1972). Comparison of integrated or average indicated flow values with volume have included use of a spirometer (FINUCANE *et al.*, 1972), dry gas meter (HENDERSON *et al.*, 1983) and calibrated syringe (YEH *et al.*, 1982) as volume secondary standards.

In consideration of the effect of temperature and composition on gas viscosity (TURNER *et al.*, 1989), and the fact that respiratory gases are saturated with water vapour at temperatures well above room temperature, special devices for generating heated, humidified flows of air or gas mixtures have been constructed (MILLER and PINCOCK, 1986; VAIDA *et al.*, 1983; SODAL *et al.*, 1977).

Although most of the above calibration methods can be used with gas compositions other than air, the experimental validation of a composition-independent calibration curve has not been reported to date.

Our technique allows calibration to be performed at room temperature and with almost any flow or volume instrument as calibration standard. Nonhumidified gas at room temperature may be used because the difference in viscosity caused by the change in temperature and water vapour content is taken care of by the viscosity correction.

5 Conclusions

We have shown that second- and third-order polynomials yield a significantly better fit to calibration data than straight lines for neonatal screen and Fleisch pneumotachographs over the flow range 0–250 ml s^{-1}. We have also shown that a composition-independent calibration curve for Fleisch and screeen pneumotachographs may be obtained by regression of the product of flow and calculated gas viscosity against differential pressure. In the case of the Fleisch pneumotachograph, if the calibration is performed with 60 per cent O_2, then the calibration curve obtained by dividing the SCP by gas viscosity is likely to be in error by less than 1·4 per cent. In the case of the screen pneumotachograph the uncertainty caused by gas composition variation is of the same order of magnitude as the uncertainty in the calibration curve itself. We conclude that using this technique the neonatal screen pneumotachograph may be calibrated in any O_2 concentration and used with any inspired O_2 concentration with an expected error of approximately 0·4 per cent.

This study was performed using neonatal pneumotachographs and mixtures of air and oxygen which cover a small range of density and viscosity. We therefore caution against the use of this technique with gas mixtures whose properties are substantially different or cover a wider range, or with different pneumotachographs which may exhibit different characteristics, without prior experimental verification.

Acknowledgments—This study was supported in part by a grant from the South African Medical Research Council. The authors would like to thank the Johannesburg Hospital and the University of the Witwatersrand for the use of their facilities

References

BARENBRUG, A. W. T. (1974) *Psychrometry and psychrometric charts*, 3rd edn. Chamber of Mines of South Africa.

BERTRAM, C. D. (1984) Flowmeter for high frequency ventilation on the pneumotach principle. *Austral. Phys. & Eng. Sci. Med.*, **7**, (2), 85–90.

DRAPER, N. R. and SMITH, H. (1981) *Applied regression analysis*, 2nd edn. Wiley, New York.

FIELD, D., MILNER, A. D. and HOPKIN, I. E. (1985) Calculation of mean airway pressure during neonatal intermittent positive pressure ventilation and high frequency positive pressure ventilation. *Pediatr. Pulmonol.*, **1**, 141–144.

FINUCANE, K. E., EGAN, B. A. and DAWSON, S. V. (1972) Linearity and frequency response of pneumotachographs. *J. Appl. Physiol.*, **32**, (1) 121–126.

FOND, A., MORIZET-MAHOUDEAUX, P., KRIM, G. and RISBOURG, B. (1986) Online measurement of mechanical parameters in ventilated neonates with rejection of unrepresentative cycles. *Med. & Biol. Eng. & Comput.*, **24**, 305–310.

HENDERSON, A. M., MOSSE, C. A. and ARMSTRONG, R. F. (1983) Long term flow measurement: an assessment of the Siemens Servo 900B expiratory flow meter. *J. Med. Eng. & Technol.*, **7**, (3) 144–146.

HUGHES, T. J. (1981) Hot thermistor spirometry for the artificial ventilation of infants. Ph. D. Thesis, University of Cape Town, South Africa.

JOHNS, D. P., PRETTO, J. J. and STREETON, J. A. (1982) Measurement of gas viscosity with a Fleisch pneumotachograph. *J. Appl. Physiol.: Respirat. Environ. Exercise Physiol.*, **53**, (1), 290–293.

KELLER, J. B. (1963) Viscous flow through a grating or lattice of cylinders. *J. Fluid Mech.*, **18**, 94–96.

LESOUEF, P. N., ENGLAND, S. J. and BRYAN, A. C. (1984) Total resistance of the respiratory system in preterm infants with and without an endotracheal tube. *J. Pediatrics*, **104**, 108–111.

MILLER, M. R. and PINCOCK, A. C. (1986) Linearity and temperature control of the Fleisch pneumotachograph. *J. Appl. Physiol.*, **60**, (2), 710–715.

MULLER, N. L. and ZAMEL, N. (1981) Pneumotachograph calibration for inspiratory and expiratory flows during $He-O_2$ breathing. *J. Appl. Physiol.: Respirat. Environ. Exercise Physiol.*, **51,** (4), 1038–1041.

PEDERSON, O. F., NAERAA, N., LYAGER, S., HILBERG, C. and LARSEN, L. (1983) A device for evaluation of flow recording equipment. *Bull. Europ. Physiopath. Respirat.* **19,** 515–520.

SODAL, I. E., ADAMSON, H. P. and KIBLER, S. K. (1977) A new method for accurate calibration of respiratory flow meters. *Biomed. Sci. Instrum.*, **13,** 69–72.

TURNER, M. J., MACLEOD, I. M. and ROTHBERG, A. D. (1989) Effect of temperature and composition on the viscosity of respiratory gases. *J. Appl. Physiol.*, **67,** 472–477.

TURNEY, S. Z. and BLUMENFELD, W. (1973) Heated Fleisch pneumotachometer: a calibration procedure. *Ibid.*, **34,** (1), 117–121.

VAIDA, P., BARGETON, D. and GUENARD, H. (1983) BTPS calibration of heated Fleisch pneumotachometer. *Bull. Europ. Physiopath. Respirat.*, **19,** 635–640.

YEH, M. P., GARDNER, R. M., ADAMS, T. D. and YANOWITZ, F. G. (1982) Computerized determination of pneumotachometer characteristics using a calibrated syringe. *J. Appl. Physiol.: Respirat. Environ. Exercise Physiol.*, **53,** (1), 280–285.