Ventilation Measurement

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77.1 Ventilation

This section deals mainly with **spirometry**, i.e., with the measurement of volumes and flows associated with respiration. Spirometric tests (which embody useful information about parameters related to pulmonary function) are often used for diagnostic purposes in conjunction with other measurements; Figure 77.1 and Table 77.1 show spirometric quantities of clinical interest [1].

The classification of spirometric instruments may be based on different criteria. For example, one may consider *open-circuit* devices (the subject takes a full inspiration or expiration before connecting to the meter) and *closed-circuit* devices (the subject remains connected to the apparatus during one or several respiratory cycles). Another distinction concerns whether the instrument is a *portable* one and is mainly used for monitoring purposes or is a *diagnostic* one whose purpose is to provide an accurate value that may be compared with a reference value. As for the nature of the physical functions more directly investigated, a subdivision into *volume measurements* and *flow measurements* is important, even though, using some care, it is sometimes possible to shift from one class to the other by time-differentiation or time-integration procedures. Another technique deals with the direct evaluation (as a function of frequency) of the complex impedance of the respiratory system viewed as a suitable network of resistors, capacitors, and inductors. This method has been the subject of considerable investigation in the last two decades, thanks to the advent of computers and to their use in the spectral analysis of signals. This topic will be discussed in the final section.

As a general comment about measurements associated with ventilation, it may be said that there are few other fields in applied science where a correct *operation procedure* is virtually as important as the reliability of the measuring instruments themselves. See, for example a study conducted on nearly 6000 patients affected by some airflow obstruction (Reference 2 and references therein, especially n. 26).

Volume Measurements

In this case, gas volumes associated with the respiratory process are the main target of investigation; the principal instruments that have been used so far in the clinical routine and in research activity are [3]

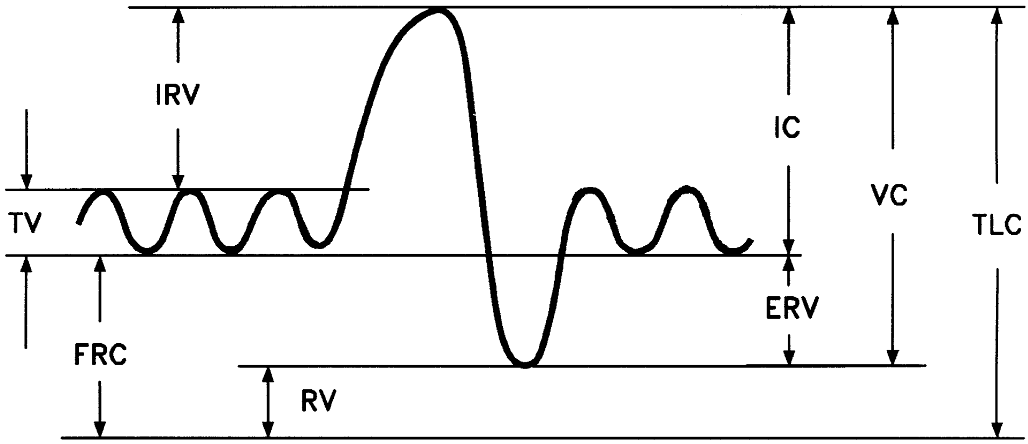

FIGURE 77.1 Spirometric trace illustrating the definitions of some significant quantities commonly evaluated in spirometry (see also Table 77.1).

TABLE 77.1 Common Spirometric Quantities

Vital capacity	VC
Inspiratory capacity	IC
Inspiratory reserve volume	IRV
Expiratory reserve volume	ERV
Tidal volume	TV
Functional residual capacity	FRC
Residual volume	RV
Total lung capacity	TLC

1. *Spirometer:* An expandable chamber whose volume is monitored during inspiration or expiration. The subject is instructed to blow into a conduit communicating with the chamber; the latter may consist of a bell, a piston, or more often a bellows (as in the once ubiquitous Vitalometer®).
2. *Turbine meter:* Based on the principle that air blown through the inlet produces the rotation of a turbine connected to a revolution counter.
3. *Impedance plethysmograph:* Based on the measurement of resistance (*strain gage plethysmograph*) or of inductance (*inductance plethysmograph*); in both cases, the impedances of an elastic coil wrapped around the subject's chest and one wrapped around the subject's abdomen are monitored during respiration.
4. *Total body plethysmograph:* A kind of sealed telephone booth inside which the subject sits; the pressure inside the box is sensed and converted to volume values.

Flow Measurements (Pneumotachometry)

In this case the airflow through the upper airways is directly measured; in principle, the flow could be evaluated by time differentiating the spirometer records, but this method would suffer from limitations due to nonlinearity effects and to errors related both to poor frequency response and to some hysteresis of the volume meter. The reverse procedure is in fact more common: since they are normally connected to an electronic integrating device, flowmeters can be employed as volume-measuring instruments as well. Calibration is normally performed by discharging a syringe (of known capacity) through the flowmeter; the response of the latter is time integrated to check whether it corresponds to the calibration volume. It is convenient to repeat the calibration procedure a few times, with the syringe discharged each time at a different speed, in order to detect possible effects due to deviations from linearity of the flow sensor.

A useful summary of standard specifications governing the performance of several types of pneumotachometers (PTM) together with a concise description of the more common PTM types can be found in Reference 4. Conventional devices for measuring respiratory flow are

1. *Linear Resistance* PTM (LRPTM): Evaluates the pressure difference generated by the (laminar) airflow across a fixed hydrodynamic resistance. This procedure is based on the Poiseuille equation and is analogous to evaluating current by measuring voltage across a known resistor and using Ohm's law. The resistive element may consist either of a bundle of tubelets (Fleisch-type) or of a wire mesh screen (Lilly-type).
2. *Hot Wire* PTM (HWPTM): A very thin wire heated by an electric current, cooled by the flowing gas; the rate at which heat is conveyed away from the wire depends on the fluid flow; a variant of this instrument uses a heated film in lieu of the heated wire.
3. *Ultrasonic* PTM (UPTM): Based on the principle that the speed of a beam of ultrasounds exchanged between a pair of transducers is increased or decreased as it propagates through the moving air; the variations of the time of flight of the ultrasonic beam is related to the average speed of the flowing air.
4. *Vortex-Shedding* PTM (VSPTM): Basically measures the frequency at which vortices are generated in the wake of a suitably shaped obstacle (the "bluff body") exposed to the flow.

Respiratory Impedance Measurements

In recent years, considerable interest has been devoted to the impedance of the respiratory system, a parameter that contains information about the morphology of our breathing apparatus and plays a role analogous to the impedance of an electrical circuit. **Respiratory impedance** may be obtained by connecting the patient's upper airways to an alternating pressure generator and by evaluating the ratio of the measured pressure to the measured flow; this can be done at several excitation frequencies, using a method that is referred to as the forced oscillation technique.

77.2 Instrumentation: Principles and Description

Measurements of Volume

Spirometer

A widely used spirometer is the bell-type (see Figure 77.2): the volume of the lungs is monitored by the position of a light cylindrical bell (possibly equipped with counterweights) connected to the patient's mouth. The bell spirometer is rather cumbersome but, thanks to its great reliability, is an ideal tool for calibration purposes as well as for comparing data recorded at different centers. In its turn, the bell spirometer needs periodic calibration at least every 3 months; this may be performed by means of a 3-L syringe equipped with electronic volume readout [5]. The syringe must be accurate to within 25 mL and the spirometer should be able to measure volumes of at least 8 L with an accuracy of at least ±3% or 0.050 L, whichever is greater, with flows betweeen zero and 14 L/s [6].

Turbine Meter

Its operating principle is quite straightforward. The revolutions of the turbine wheel, whose speed is proportional to the flow, are counted by electronic sensors (optoelectronics or Hall effect based) and processed to give volume or flow values. The main limitation of the turbine meter is that low flow values are underestimated since a greater fraction of the air slips past the wheel as the flow rate decreases; it also displays poor frequency response and can be used only in unidirectional flows. Although these features make the device unsuitable for accurate laboratory measurements, it is used extensively to monitor the ventilation of patients in intensive care and in portable instruments, thanks to its overall (sensor and pulse processing) simplicity and low cost.

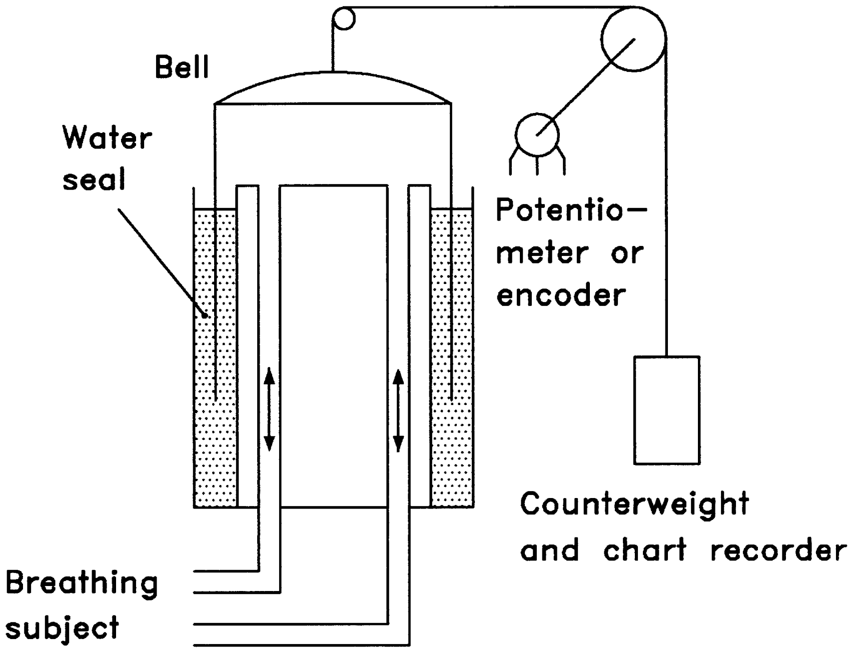

FIGURE 77.2 Schematic view of the classical bell spirometer.

Inductance Plethysmograph

Like the spirometer, the pneumotachometer, and the total-body plethysmograph, this technique allows for noninvasive monitoring of human ventilation but it does not require a connection to the airways. Two elastic belts with serpentine inductors encircle the torso, the first around the rib cage, the other around the abdomen. Each inductor is part of the resonant tank circuit of a free-running oscillator (resonant frequency from 0.2 to 1 MHz) whose output is optically coupled to the demodulator circuit (for the purpose of electrical isolation of the subject from the equipment, usually operated by the ac power line). Ventilation causes the inductor cross section areas to vary changing their inductances and thus varying the frequency of the two oscillators (FM modulation).

Strain Gage Plethysmograph

Its principle of operation is similar to that of the inductance plethysmograph; in this case, the sensing element consists of a strain gage whose resistance is varied as the gage is stretched and released during respiration.

Total-Body Plethysmograph (Constant-Volume Box)

This is a sealed chamber that allows the measurement of thoracic gas volume (TGV) by exploiting the pressure–volume relation of a fixed quantity of ideal gas. A schematic illustration of the method for a constant-volume plethysmograph is shown in Figure 77.3. The balloon-shaped object represents the subject's respiratory system, the volume of which varies during breathing. In the course of the measurement operation (which lasts only a few seconds), the shutter appearing in Figure 77.3 must remain closed in order that the alveolar pressure may be identified with the pressure measured at the mouth. In the following, the subscript A (for alveolar) denotes the gas in the lungs and the subscript C denotes the gas in the chamber.

For moderate respiratory acts, the air contained in the lungs at a given time can be treated as an approximately isothermal gas (temperature 37°C, water vapor saturated) while the air in the chamber is more appropriately treated [1] as an adiabatic gas. By differentiating the ideal gas equation for the isothermal air in the lungs,

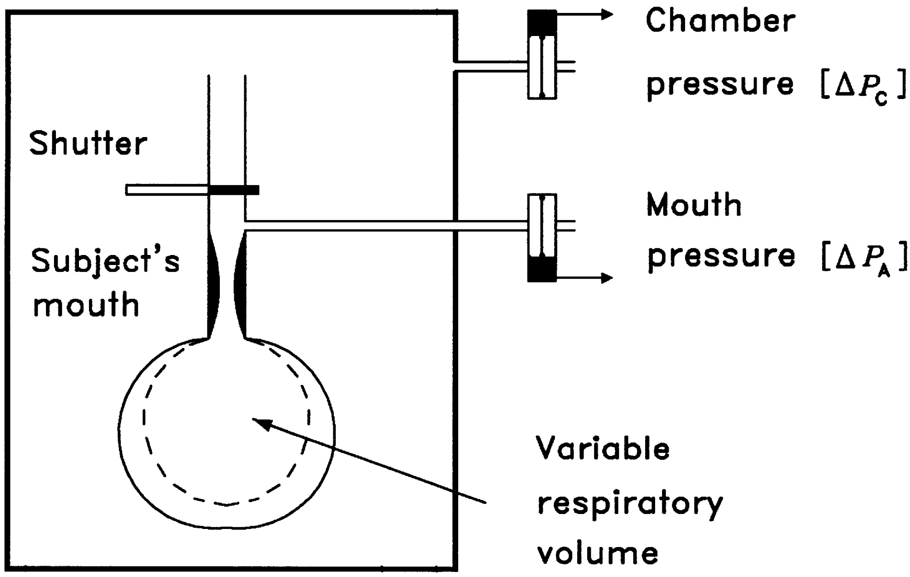

FIGURE 77.3 The respiratory system of a patient sitting inside an airtight chamber is schematized as an expandable balloon. The quantities of air inside and outside the balloon are separately constant: only their volumes and pressures change as the subject attempts to breathe against the closed shutter.

$$V_A = -P_A \frac{\Delta V_A}{\Delta P_A}$$

while for the adiabatic air in the chamber,

$$\Delta V_C = -V_C \frac{\Delta P_C}{\gamma P_C}$$

where γ is the usual ratio between molar heat capacities: $\gamma = c_p/c_V$.

Since the system is globally closed ($\Delta V_A = -\Delta V_C$), the lung volume V_A is related to the alveolar pressure P_A (which is taken to be the barometric pressure minus the water vapor pressure at 37°C) and to the differential pressures ΔP_C and ΔP_A (respectively equal to the chamber pressure fluctuation and to the alveolar pressure fluctuation) by the formula:

$$V_A = kP_A \frac{\Delta P_C}{\Delta P_A}$$

The coefficient k can be obtained through a calibration procedure that creates sinusoidal volume variations by means of a reciprocating pump. The pressure changes appearing on the right-hand side of the last equation may be measured by sending the outputs of the two sensors of Figure 77.3 to a data acquisition module installed in a host computer, typically a personal computer.

Correcting to Standard Conditions

Measurements of gas volumes make reference to lung BTPS (body temperature and pressure, saturated with water vapor) conditions. Volume measurements (V_A) made at ATP (ambient temperature and pressure) conditions should be corrected to BTPS conditions (V_B).

When bell or bellows spirometers are employed, the following formula [1] is often used:

$$V_B = \frac{310.2\left(P_B - P_W\right)}{\left(273.2+t\right)\left(P_B - 6.3\right)} V_A \tag{77.1}$$

where t = inside gas temperature (°C), P_B = barometric pressure (kPa); P_W = saturated water vapor pressure (kPa) of the gas. The temperature t of the air inside the spirometer should be measured accurately during each breathing maneuver (Table 3 in Reference 7).

Measurements of Flow

Linear Resistance Pneumotachometer

The standard LRPTM contains the following elements (Figures 77.4a and b):

1. A fixed resistive load;
2. A differential pressure sensor;
3. Electronic instrumentation for processing and displaying the output of the pressure sensors.

The relationship between flow rate and pressure difference should be linear within the range of useful flow rates; the maximum flow value, determined by the onset of turbulence, and the linearity range are given by the manufacturer's specifications.

It is good practice to calibrate the PTM periodically, by connecting it in series with a bell spirometer, and then use it without changing the geometry of the immediately adjacent tubing. In fact, accurate flow rate measurements are possible only if information about the flow–pressure characteristics over the useful flow range is available; furthermore, this information should have been obtained in the same conditions in which the flowmeter will be used [8].

Figure 77.4(a) depicts the Fleisch PTM, whose resistive element consists of a bundle of capillary tubes. The function of the heater is to avoid the condensation of water vapor inside the tubes; the two chambers of the differential pressure sensor are connected at the two ends of the resistive element, whose resistance is lower than 0.1 kPa s/L (1 cm H_2O s/L). Since typical spirometric flows, for quiet breathing and slow maneuvers, are of the order of 1 L/s, they generate pressure differentials of about 1 cm H_2O. Considerably stronger flows, exceeding 10 L/s, may be produced by adults in some maneuvers; in these cases, Fleisch PTMs of larger dimensions should be employed in order to keep the spirometric flow within the linearity regime.

Figure 77.4(b) illustrates the Lilly PTM, whose resistive element consists of a wire mesh and is of the order of 0.05 kPa s/L (0.5 cm H_2O s/L) to which pressure differentials of the same order as the Fleisch are associated. The Lilly PTM is much less exposed than the Fleisch to the risk of nonlinearity at higher flow rates.

The linear resistance PTMs, in conjunction with an appropriate differential pressure transducer, an amplifier, and an analog or, more often, digital integrator, form the most widely used instrument for the combined measurement of flow and volume. In fact, recent recommendations on the standardization of lung function tests are limited to Fleisch and Lilly tachometers only [1].

Hot Wire Pneumotachometer

The HWPTM works on the principle that the cooling rate of a heated wire depends on the speed at which the surrounding fluid is flowing [9-11]. The rate at which thermal energy is lost by the wire may be written as

$$Q = hS\left(T_w - T\right) \tag{77.2}$$

where Q is the heat transfer rate; S is the surface area of the wire; T_w is the wire mean temperature; T is the temperature of the fluid surrounding the wire; h is the transfer coefficient between sensor and fluid and is defined by the equation above.

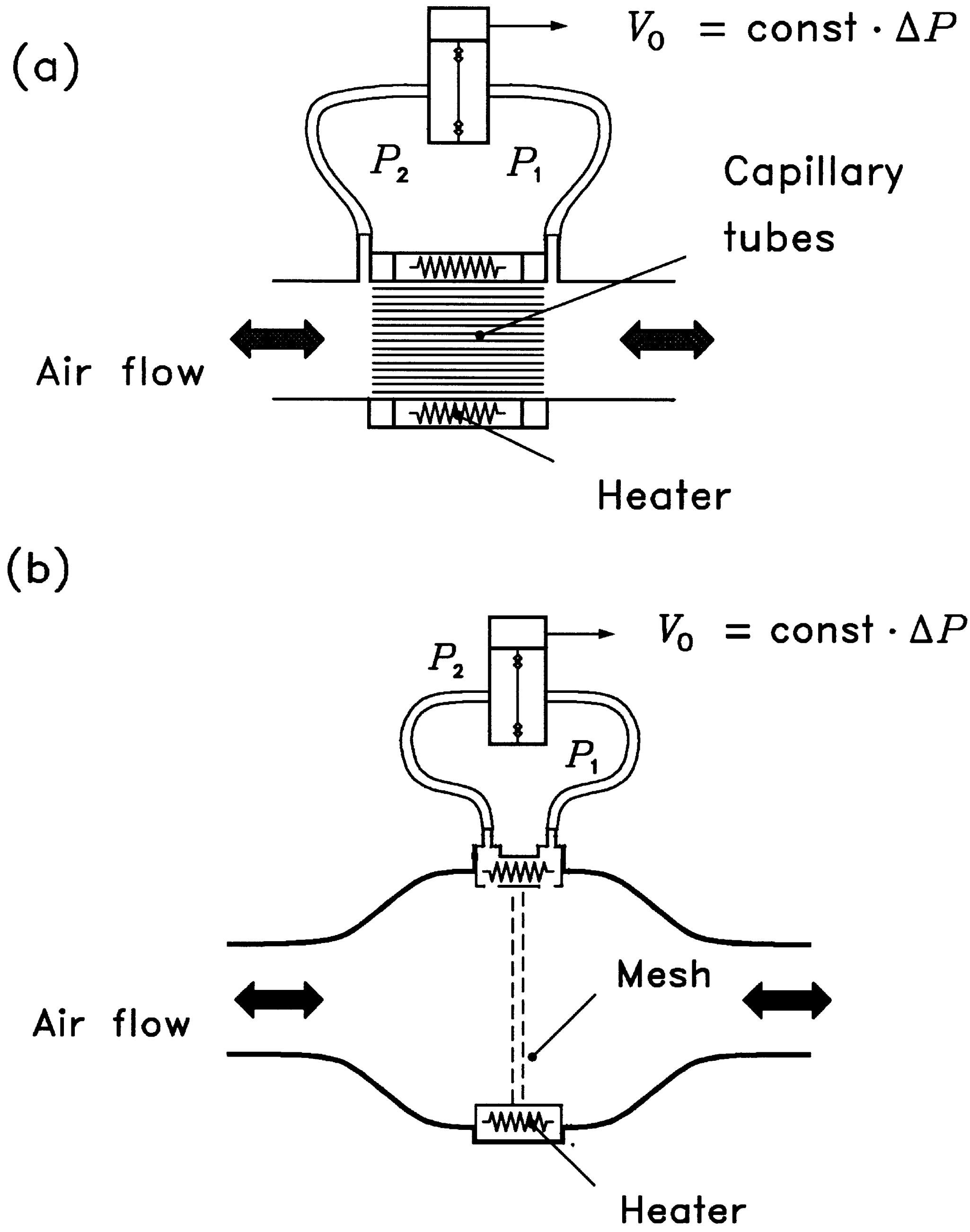

FIGURE 77.4 (a) The Fleisch (capillary) pneumotachograph. (b) The Lilly (wire-mesh screen) pneumotachograph. In both sensors, the task of the heater is to reduce water vapor condensation.

The speed v of the fluid is linked to h by an important empirical relation whose analytical form has withstood the test of time; however, owing to the complexity of the formula and to the number of physical quantities involved in it, it is convenient to make direct reference to Reference 11 where all necessary details can be found.

It is wise to remember, in fact, that from the operational point of view the last word on these instruments is always entrusted to a sound calibration procedure.

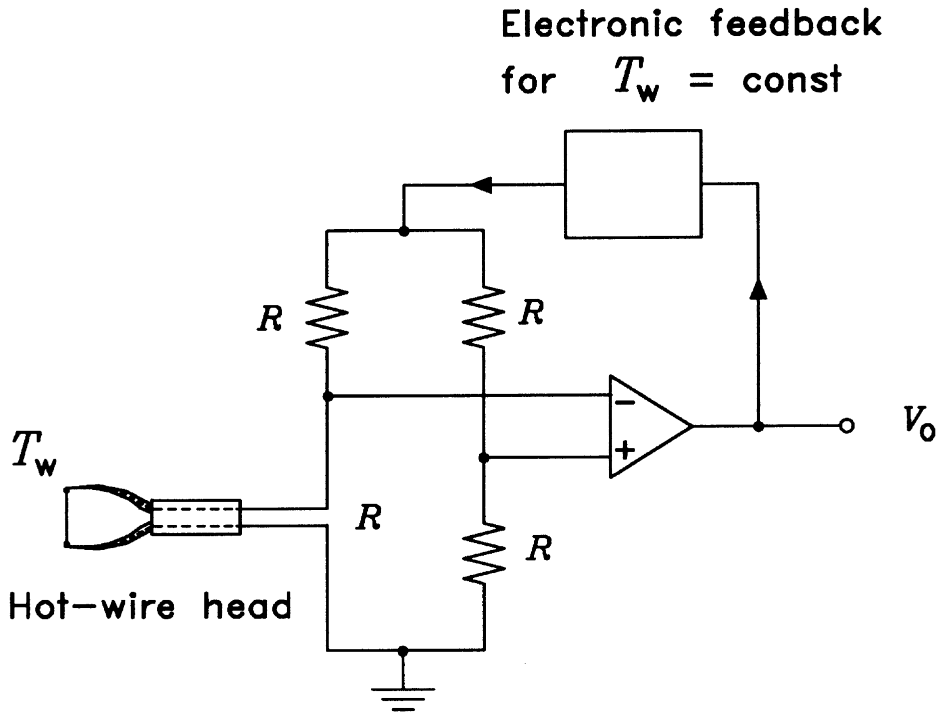

FIGURE 77.5 Block diagram of the constant-temperature hot-wire flowmeter. The electronic feedback system allows one to maintain the sensing wire temperature constant.

For spirometric purposes, it is recommended to use the HWPTM in the *constant temperature* mode of operation, according to the diagram of Figure 77.5.

Using a suitable feedback system the current is adjusted so as to maintain the wire temperature (hence, its resistance) at a constant value through the entire range of velocities allowed by the flowmeter. Once this condition is satisfied, the output voltage value is a measure of the heat transfer rate. The constant-temperature design has several advantages, e.g., the wire is protected against burnout and it is easy to compensate for environment temperature change either by using a temperature-sensitive element in the bridge or by measuring the actual temperature and correcting the results.

Accurate measurements require periodic calibration procedures (for example, using the syringe mentioned in connection with the spirometer) because dirt accumulating on the wire may significantly alter the heat transfer coefficient; it is easy, however, to remove dirt thoroughly from the sensor by manually superheating the wire for a few seconds or by immersing the sensor head into a suitable cleaning liquid. The calibration procedure should be repeated at intervals comparable to those of the LRPTM.

The setup in Figure 77.5 measures only unidirectional flows.

Hot-Film Pneumotachometer

More recently, hot-film PTMs have been developed which can be used in place of the hot wire, the rest of the PTM remaining unchanged. The sensing element is deposited onto a nonconducting support (quartz) by vacuum sputtering, which ensures uniform thickness (about 0.1 μm) of the sensing element (platinum or nickel). In these sensors, heat can be conducted through the substrate and lost by convection to the ambient gas. The length-to-diameter ratio of the sensor is smaller; consequently, the temperature distribution along the sensor is less uniform. These drawbacks, which are more theoretical than real since calibration is always necessary, are largely compensated for by a lower fragility and sensitivity to particulate contamination.

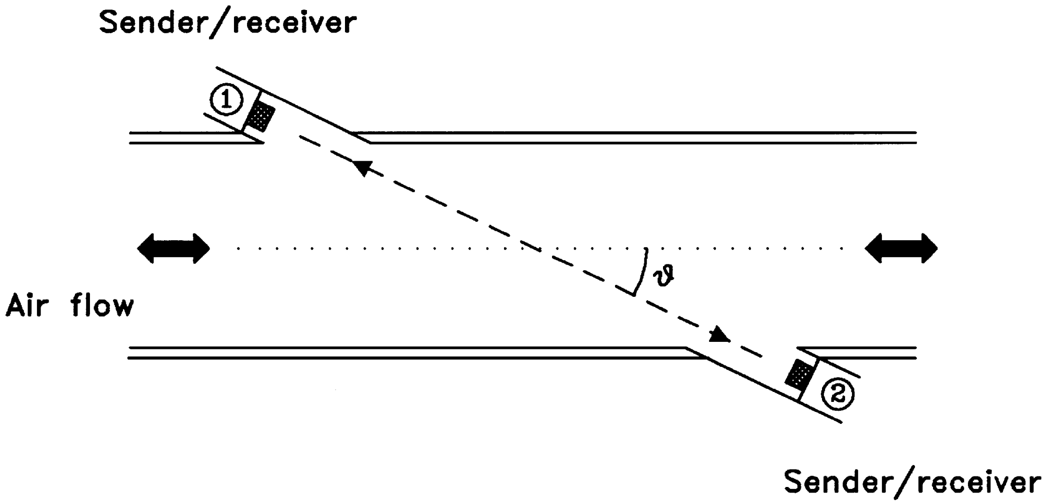

FIGURE 77.6 Transit time ultrasonic pneumotachograph. Owing to the large Q factor of the resonators, the central resonant frequencies of the two piezoelectric crystals should closely match.

Ultrasonic Pneumotachometer

The basic working principles of UPTMs [10] are (1) sound is sped up or slowed down as it propagates through a moving medium, (2) the back and forth transit time of a sound signal is related to flow velocity and turns out to be largely independent of the acoustic velocity of the medium. UPTMs can be classified according to whether they use *pulsed* or *continuous* ultrasound signals.

In the first case, the transmitter is driven by a short pulse of sine waves; the round-trip transit time of individual pulses or sequences of pulses is measured.

In the second case, a continuous ultrasonic signal is transmitted along a closed path and either the phase shift or the frequency shift is measured. As a specific example, a pulsed transit time UPMT (Figure 77.6) is described.

Two piezoelectric crystal transducers are recessed into the wall of a conduit at an angle ϑ to the flow axis. Provisions are made in order to reduce the level of acoustical and electromagnetic external interferences. Moreover, moderate heating (at about 40°C) prevents water from condensing on their surfaces. The two devices are made to function alternately as transmitter and receiver of a short burst of 50 to 200 kHz sound wave.

With no gas flowing through the conduit, the time required for sound transmission in either direction is the same. When gas is flowing, the two times are, respectively,

$$t_{12} = L \big/ \left(c + v \cdot \cos \vartheta\right) \qquad t_{21} = L \big/ \left(c - v \cdot \cos \vartheta\right)$$

where L is the transmitter–receiver distance and c is the speed of sound.

The gas velocity v turns out to be independent of the actual value of c (which in turn depends on the gas type and on the working conditions and can range from around 200 m s^{-1} to nearly 1000 m s^{-1}); an expression for v is easily obtained from the last equations by assuming $c^2 \gg v^2$

$$v \approx \frac{L}{2 \cos \vartheta} \cdot \frac{\Delta t}{t_a^2} \tag{77.3}$$

where $\Delta t = t_{21} - t_{12}$ and $t_a = (t_{21} + t_{12})/2$.

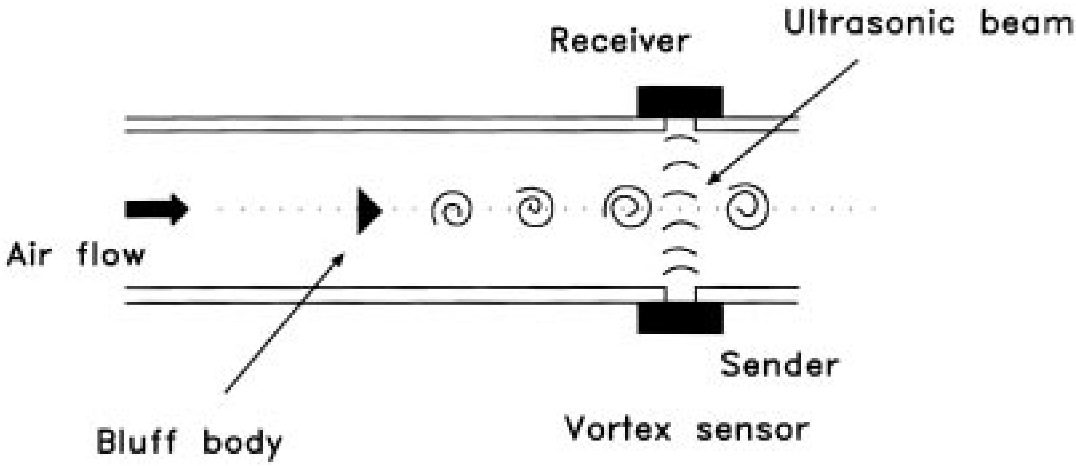

FIGURE 77.7 A vortex shedding flowmeter equipped with an ultrasound vortex sensor. Vortices generated by the bluff body are sensed by the ultrasonic beam located downstream. In the optical model, the functions of vortex generation and counting are both taken on by an optical fiber.

The flowmeter measures the average flow velocity of the gas along the path of the ultrasound beam.

In using the UPTM it should be remembered that the working frequency is very close to the resonant frequencies of the two crystals which should be closely matched: small differences would cause a strong reduction of the instrument sensitivity owing to the sharpness of the resonance curves.

Vortex-Shedding Pneumotachometer

When the Reynolds number is sufficiently high, vortices of regular periodicity are born downstream of a bluff body located in the fluid flow (see Figure 77.7).

The frequency at which these eddies are generated is a linear function of flow velocity and depends weakly on the dimensionless Strouhal number [12,13]. Different types of VSPTMs use different methods for measuring the vortex frequency; to this end, ultrasonic beams and optical fibers are widely employed.

Ultrasonic: An acoustic beam (Figure 77.7) is sent across the flow by an ultrasound transmitter located on one side of the tube (downstream of the bluff body); an ultrasound receiver is located on the opposite side of the tube. In this case, eddies generated by the bluff body act as modulators of the sound wave intensity. By doubling the setup [13] the ultrasonic VSPTM can also be used for bidirectional flows.

Optical fiber: In this model, an optical fiber is stretched across the tube and acts both as the bluff body and the vortex frequency sensor [14]. The light from an LED enters the fiber at one end, the other end being coupled to a photodiode. The intensity of the light propagating through the fiber is modulated by the mechanical vibrations of the fiber produced by the eddies moving with the fluid and fluctuates in step with the vortex generation frequency.

Both in the ultrasonic and the optical fiber VSPTM, the fluctuating intensity of the beam (respectively, sound and light) is processed to output a flow indication.

The main limitation of all types of VSPTMs is a "blind zone" at low fluid flows corresponding to the absence of vortices at low Reynolds numbers. In compensation, these devices are quite insensitive to the thermodynamic conditions of the gas and to the presence of particulate matter. Similarly to the case of HWPTMs, the last word on VSPTMs is entrusted to a sound calibration procedure.

Measurement of Volume Using a PTM Plus an Integrator

Another kind of plethysmograph can be built by combining a PTM and a digital integrator that calculates volume as follows:

$$V(\tau) = \int_0^\tau \dot{v}(t)\,dt \tag{77.4}$$

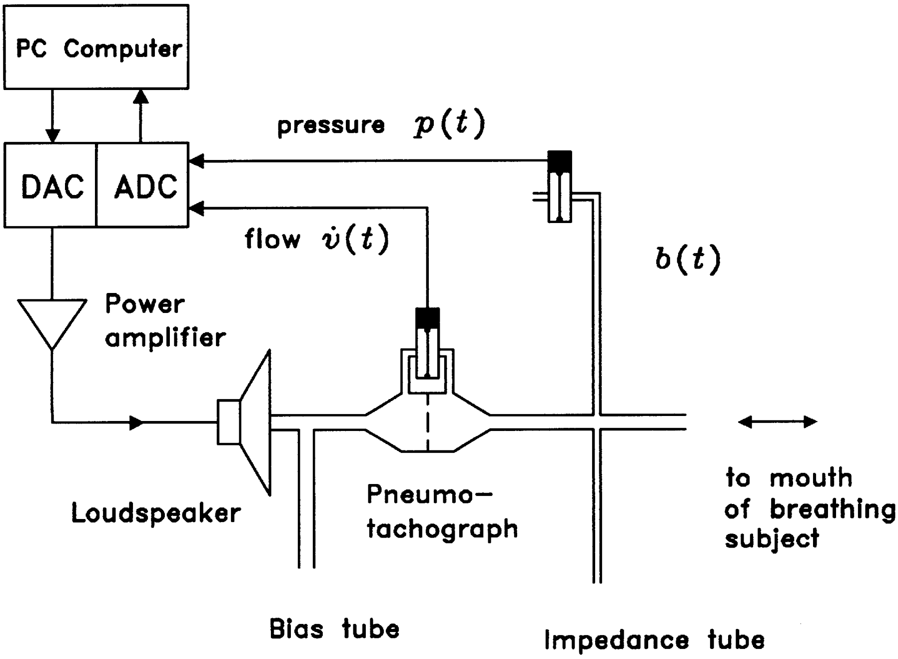

FIGURE 77.8 Experimental setup for the computer-based measurement of respiratory input impedance.

where $\dot{v}(t)$ is the instantaneous flow. The problems exhibited by this kind of instrument are the same as those of the PTM around which it is built, since the process of integration is invariably digital; moreover, correction to BTPS requires considerably more care [1]. Finally, when a heated PTM is employed, correction to BTPS is often done by assuming instantaneous thermalization of the gas within the tachometer (this means that the gas passing through the instrument assumes completely and without delay the same temperature of the tachometer [1]); this assumption may be quite unrealistic, especially in the case of the wire mesh screen type. The correction should thus be different for different types of PTMs.

The Forced Oscillation Technique

The forced oscillation technique (FOT) is a modern tool [15] used to gain information on the structural and mechanical properties of the respiratory system by measuring how the latter responds to an externally imposed excitation. When linearity is assumed, well-known correlation techniques (borrowed from conventional signal analysis) can be exploited to measure the system complex impedance $Z_R(f)$, which is defined as the ratio of complex input pressure difference to complex output flow (i.e., the ratio of amplitudes and the difference of phases) and is equal to the inverse of the system transfer function $H_R(f)$.

Depending on the part of the respiratory system actually investigated, i.e., on the positions where the excitation pressure difference is applied and the flow is measured, different values for $Z_R(f)$ can be obtained:

- *Input impedance:* Both the excitation pressure (measured relatively to the pressure on the body) and flow are measured at the mouth;
- *Transfer impedance:* The excitation pressure is measured at the mouth, whereas flow is measured at the thorax (or vice versa) by means of a body plethysmograph.

An experimental setup for the measurement of respiratory input impedance is shown in Figure 77.8. A suitable software provides

- Generation of a pseudorandom sequence (typically obtained by summing the first 25 harmonics of 2 Hz with random phases) which, after digital-to-analog conversion and proper power amplification, drives a large (20 to 30 cm diameter) loudspeaker;

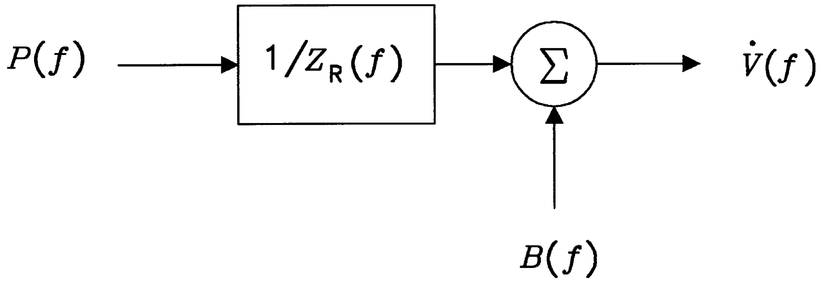

FIGURE 77.9 The respiratory system modeled in the frequency domain.

- Acquisition of two blocks of pressure and flow samples: 2×4096 data are collected in a 16-s run at a sampling rate of 256 sample/s;
- Processing of data in order to obtain the desired $Z_R(f)$ function in the typical interval 2 to 48 Hz.

The bias tube allows a flow of fresh air to be drawn through the system to minimize rebreathing of expired air. The impedance tube (2 m long, 2 cm diameter) allows spontaneous breathing; moreover, because its electrical equivalent is an inductor, it acts like a shunt for the low-frequency breathing components, thus reducing their level of disturbance on the pressure and flow measurements (noise), without affecting in a significant way the imposed pressure input (signal).

The respiratory apparatus is modeled as a system whose input is a random noise pressure $p(t)$ and whose output is a flow signal $\dot{v}(t)$, both corrupted by a breathing disturbance $b(t)$ uncorrelated to $p(t)$. Any text on random signal analysis [16] may provide useful information on the mathematical quantities employed in this section.

Using the following Fourier transform (FT) quantities:

- $P(f)$ = FT of the random noise pressure input $p(t)$
- $B(f)$ = FT of the subject's breathing noise $b(t)$
- $\dot{V}(f)$ = FT of the linear system output $\dot{v}(t)$

we have (Figure 77.9)

$$\dot{V}(f) = P(f) \cdot 1/Z_R(f) + B(f) \tag{77.5}$$

A valid estimate of $Z_R(f)$ cannot be obtained from this relation because of the presence of the large breathing disturbance. According to the theory of signal processing, a nonbiased estimate of $Z_R(f)$ can be obtained by using autospectra and cross spectra of input $[p(t)]$ and output $[\dot{v}(t)]$ signals; if $G_{PP}(f)$ and $G_{\dot{V}\dot{V}}(f)$ the autopower spectrum of pressure and flow, and $G_{\dot{V}P}(f)$ their cross power spectrum, the estimated impedance $Z_M(f)$becomes

$$Z_M(f) = \frac{G_{PP}(f)}{G_{\dot{V}P}(f)} = \frac{G_{PP}(f)}{\dfrac{1}{Z_R(f)} \cdot G_{PP}(f) + G_{BP}(f)} \tag{77.6}$$

which becomes almost coincident with the true value $Z_R(f)$ when a long time average allows sensible reduction of the contribution of the $G_{BP}(f)$ term.

Moreover, since the systematic bias depends mainly on the presence of the breathing noise on both pressure and flow signals, all efforts should be made to reduce this disturbance as early as possible, i.e., at the sensor level. This is achieved by using the impedance tube and by keeping the value of the pneumotachograph impedance as low as possible. Further reduction is usually obtained by high-pass analog filtering (2 Hz corner frequency) before the ADC and/or by digital filtering before FFT calculations. It is worth noting that the choice of a pseudorandom excitation makes possible the use of a digital comb filter (a multiple narrow bandpass, i.e., a bandpass for each 2 Hz harmonic) [17].

When the FOT is used, it is necessary to ascertain that the coherence [16] between input and output, expressed by the γ^2 function defined below, is sufficiently large:

$$\gamma^2(f) = \frac{\left|G_{\dot{V}P}(f)\right|^2}{G_{\dot{V}\dot{V}}(f) \cdot G_{PP}(f)} \tag{77.7}$$

As a rule of the thumb, the calculated $Z_R(f_0)$ (the impedance for a particular value of frequency) is accepted only when the corresponding $\gamma(f_0)$ is larger than 0.9 to 0.95.

Actually, the value of $\gamma(f)$ is the only form of control used; at any rate, some care should be taken to interpret the coherence function as an index for the reliability of respiratory impedance data [18, 19]. Obviously, higher values of the coherence function can be achieved by increasing the pressure input amplitude, but this may lead to an increase of nonlinearity effects [20].

In conventional setups, flow is evaluated by measuring the pressure drop ΔP that develops across the pneumotachograph, whose impedance is known and usually small with respect to the subject's input impedance Z_R. Since these two impedances are in series, the differential pressure ΔP measured by the pressure sensor is small compared with the pressure P applied to both sides of the transducer. Accurate measurements then require the use of highly symmetric sensors [21], where symmetry is expressed by the value of the common-mode rejection ratio, CMRR = 20 log $P/\Delta P$. In addition to this, however, other sources of error (such as the finite impedance of pressure sensors [22]) are usually present which render the correction procedure ineffective.

All these problems can be globally overcome by the following dynamic calibration procedure [23] that requires two simple preliminary measurements and the availability of a reference impedance Z_{REF}.

If Z_M^{∞} denotes the impedance measured when the measuring system is occluded ($Z_R = \infty$), Z_M^{REF} denotes the impedance measured when a known reference impedance is used ($Z_R = Z_{REF}$) and Z_M denotes the impedance measured when the subject is connected to the measuring device, then the subject's corrected impedance is given by

$$Z_R = Z_{REF} \frac{1/Z_M^{REF} - 1/Z_M^{\infty}}{1/Z_M - 1/Z_M^{\infty}} \tag{77.8}$$

Z_M^{∞} and Z_M^{REF} need not be evaluated each time a new subject is connected to the device, but only when some physical change (e.g., length variations of connecting tubes, replacements of sensors, etc.) has occurred since the last time the instrument was used.

Reference Impedance

The availability of a reference impedance is recommended [24]:

1. To correct the measured impedance using the procedure described above;
2. To compare measurements obtained by different devices, different techniques, and/or different groups.

For these purposes, a compact calibrator [25] has been proposed which displays the following features:

- It is simply reproducible;
- Its impedance value is not too far from typical human values;
- It is tractable from the mathematical point of view;
- Its lowest resonances fall outside the usual working frequency range.

TABLE 77.2 Complex Values of Calibration Impedance

	Impedance (10^5 Pa s m^{-3})		
Frequency (Hz)	Real	Imaginary	Modulus
0	3.09	0.00	3.09
5	3.11	1.11	3.30
10	3.14	2.17	3.82
15	3.20	3.22	4.54
20	3.28	4.26	5.37
25	3.37	5.28	6.26
30	3.47	6.29	7.18
35	3.58	7.28	8.12
40	3.70	8.26	9.05
45	3.82	9.23	9.98
50	3.94	10.17	10.91

The calibrator consists of a bundle of 30 tubelets (each 20.0 cm long and of 2.00 mm inside diameter) and its complex impedance is given in Table 77.2.

77.3 Future Perspectives

Measurements in ventilation strongly depend on the availability of reliable sensors and detectors (pressure, temperature, optical and ultrasonic beams, etc.). The use of these components has been rising steadily in the last decade and this trend is expected to continue in the near future. The foreseeable consequences of this trend on the development of dedicated instrumentation may be

1. The increasing use of transducers of various kinds in all spirometric measurements.
2. The universal use of personal computers (and specialized software) to process and display data.
3. The growing importance of calibration procedures, because the accurate response of a particular transducer is usually reproducible but not always theoretically predictable; on the other hand, this is fundamental to all measurement systems.

Defining Terms

Plethysmography: Method employed to evaluate volume variations, with particular reference to human organs.

Pneumotachography: Methodology for measuring gas flows.

Respiratory impedance: Complex ratio between input pressure difference and induced flow at a particular frequency, when the respiratory apparatus is viewed as a complex network of linear components (resistive, inductive, capacitive).

Spirometry: The measurement of volume changes of the ventilatory system (lungs and chest wall) usually inferred from the movement of gas to and from the system.

Ventilation: Physical interaction of a body of air with the respiratory system, either through spontaneous or mechanically assisted breathing.

References

1. P.H. Quanjer, G.J. Tammeling, J.E. Cotes, O.F. Pedersen, R. Peslin, and J.C. Yernault, Lung volumes and forced ventilatory flows, *Eur. Resp. J.*, 6: 5–40, 1993.
2. R.A. Wise, J. Connett, K. Kurnow, J. Grill, L. Johnson, R. Kanner, P. Enright, and the Lung Health Study Group, Selection of spirometric measurements in a clinical trial, the Lung Health Study, *Am. J. Respir. Crit. Care Med.*, 151: 675–681, 1995.

3. T.F. McAinsh, Ed., *Physics in Medicine & Biology Encyclopedia*, Oxford, U.K.: Pergamon Press, 1986.
4. D.I. Plaut and J.G. Webster, Ultrasonic measurement of respiratory flow, *IEEE Trans. Biomed. Eng.*, BME-27: 549–558, 1980.
5. W.S. Linn, J.C. Solomon, H. Gong, Jr., E.L. Avol, W.C. Navidi, and J.M. Peters, Standardization of multiple spirometers at widely separated times and places, *Am. J. Respir. Crit. Care Med.*, 153: 1309–1313, 1996.
6. American Thoracic Society, Standardization of Spirometry 1994 Update, *Am. J. Respir. Crit. Care Med.*, 152: 1107–1136, 1995.
7. L.R. Johnson, P.L. Enright, H.T. Voelker, and D.P. Tashkin, Volume spirometers need automated internal temperature sensors, *Am. J. Respir. Crit. Care Med.*, 150: 1575–1580, 1994.
8. K.E. Finucane, B.A. Egan, and S.V. Dawson, Linearity and frequency response of pneumotachometers, *J. Appl. Physiol.*, 32: 121–126, 1972.
9. N.H. Cook and E. Rabinowicz, *Physical Measurement and Analysis*, Reading, MA: Addison-Wesley, 1963.
10. L. Marton and C. Marton, Eds., *Methods of Experimental Physics. Fluid Dynamics*. Vol. 18, New York: Academic Press, 1981.
11. L.M. Fingerson, Thermal anemometry, current state, and future directions, *Rev. Sci. Instrum.*, 65: 285–300, 1994.
12. D.F. White, A.E. Rodely, and C.L. McMurtrie, The vortex shedding flowmeter, in R.B. Dowdell, Ed., *Flow — its Measurement and Control in Science and Industry*, Pittsburgh, PA: Instrument Society of America, 967–974, 1974.
13. Ch. Buess and W. Guggenbuhl, Ultrasonic airflow meters in medical application, *Proc. Annu. Conf. IEEE Eng. Med. Biol. Soc.*, 13–16, 1987.
14. S. Webster, R. McBride, J.S. Barton, and J.D.C. Jones, Air flow measurement by vortex shedding from multimode and monomode optical fibers, *Meas. Sci. Technol.*, 3: 210–216, 1992.
15. E.D. Michaelson, E.D. Grassman, and W.R. Peters, Pulmonary mechanics by spectral analysis of forced random noise, *J. Clin. Invest.*, 56: 1210–30, 1975.
16. J.S. Bendat and A.G. Piersol, *Random Data: Analysis and Measurement Procedures*, New York: Wiley & Sons, 1986.
17. R. Farré and M. Rotger, Filtering the noises due to breathing in respiratory impedance measurements, *Eur. Respir. Rev.*, 1: 196–201, 1991.
18. E. Oostveen and A. Zwart, Reliability of the coherence function for rejecting respiratory impedance data, *Eur. Respir. Rev.*, 1: 218–221, 1991.
19. H. Franken, J. Clément, and K.P. van de Woestijne, Systematic and random errors in the determination of respiratory impedance by means of the forced oscillation technique: a theoretical study, *IEEE Trans. Biomed. Eng.*, BME-30: 642–651, 1983.
20. M. Rotger, R. Peslin, R. Farré, and C. Duvivier, Influence of amplitude, phases and frequency content of pseudorandom pressure input on impedance data and their variability, *Eur. Respir. Rev.*, 1: 178–182, 1991.
21. R. Peslin, P. Jardin, C. Duvivier, and P. Begin, In-phase rejection requirements for measuring respiratory input impedance, *J. Appl. Physiol: Respir. Environ. Exercise Physiol.*, 56: 804–809, 1984.
22. R. Farré, R. Peslin, D. Navajas, C. Gallina, and B. Suki, Analysis of the dynamic characteristics of pressure transducers for studying respiratory mechanics at high frequencies, *Med. Biol. Eng. Comput.*, 27: 531–537, 1989.
23. R. Farré, D. Navajas, R. Peslin, M. Rotger, and C. Duvivier, A correction procedure for the asymmetry of differential pressure transducers in respiratory impedance measurements, *IEEE Trans. Biomed. Eng.*, BME-36: 1137–1140, 1989.
24. M. Cauberghs and K.P. van de Woestijne, Calibration procedure of the forced oscillation technique, *Eur. Resp. Rev.*, 1: 158–162, 1991.
25. L. Basano and P. Ottonello, A calibrator for measurements of respiratory impedance, *Meas. Sci. Technol.*, 6: 982–987, 1995.