The primary function of whole-body plethysmographs is the measurement of intrathoracic gas volume (TGV) and volume change. Different models of whole-body plethysmograph (or “body box”) are used to measure changes in lung volume, from mL to L.

The first reports of plethysmography described measurement of thoracic gas volume (TGV) [1] and airway resistance (R_{aw}) [2]. Volume changes of mL were measured in terms of associated changes in box and alveolar pressures (P_{A}), using the constant-volume variable-pressure box. Changes in lung volume during compression and decompression of thoracic gas were measured while the subject breathed entirely within the box.

An alternative volume-displacement body box measured volume changes of the thorax directly, including the mass of gas flowing into and out of the lung and simultaneous compression and decompression of thoracic gas (“Mead box”), with the subject breathing in and out across the wall of the box [3]. The Mead box...
In the combination box, displacement or integrated-flow box the above approaches, using a subsequent technology combined inspiratory effort at RV.

Trapped gas during initial expiration and decompression of this compression of thoracic gas during volume (RV), which includes volume between TLC and residual flow is less than the difference in box that the VC estimated from mouth obtained by integration of airflow.

Figure 1 illustrates schematically measured changes in TGV during vital capacity (VC) efforts. This results in a “different” VC to that obtained by integration of airflow. Figure 1 illustrates schematically that the VC estimated from mouth flow is less than the difference in box volume between TLC and residual volume (RV), which includes compression of thoracic gas during expiration and decompression of this trapped gas during initial inspiratory effort at RV.

Subsequent technology combined the above approaches, using a pressure-compensated volume-displacement or integrated-flow box [4–9]. In the combination box, subjects breathe either across the wall of the box to the outside (total thoracic displacements) or within the box to measure compression volumes only. Pressure change in the box is added to the volume displaced through the box wall to provide a measure of thoracic volume displacements. The pressure-compensated integrated flow is commonly called a “transmural” box and offers the good frequency response of the pressure box to be analyzed by the manufacturer’s technical experts. Different manufacturers state their requirements in rather different ways. Verification with manufacturer’s representatives is required.

The pressure-compensated integrated-flow (“transmural box”) model is constructed separately (as an extra-cost option) from the usual pressure-compensated integrated-flow (“transmural box”) model.

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Unique feature
Regression $R_{aw}$ Number of tests $a$
Pantring $R_{aw}$ Unique sequencing $b$
“instantaneous” $R_{aw}$ (within breath) $c$
Therm/Pres comp
Variable orifice flow $d$

Wheelchair access $e$
Yes
Yes
Yes
No

Pressure-compensated integrated-flow model $f$
No
Yes
No
Yes

Warm-up required $g$
No
Yes
Yes
No

Compensation chamber
Optional
Optional
Yes
No

Optional PFTs
Add unit
Yes
Yes
Yes

Information from manufacturers (Cosmed, Medisoft/Pulmolink, Viasys/Jaeger, and ZAN/nSpirehealth) was provided in May, 2007. Specifications are subject to change from descriptions provided by manufacturers at that time. Specifications for Biomedin and MedGraphics plethysmographs were not available in response to direct enquirers, but may be available from their websites. All manufacturers provide all alternative measures of $R_{aw}$ and $R_{aw}$ as described below. All provide box calibration with a motorised syringe. All correct measured TGV for equipment deadspace. All provide user-choice of measuring sequence. All provide a wide variety of optional additional pulmonary function testing, requiring extra costs for built-in or added-on PFT equipment. Manufacturers’ specifications included below were verified by personal correspondence with representatives of each company.

1: Unique features of each manufacturer are best explained and verified by direct correspondence with company representatives.
2: Cosmed provides all the usual measures of $R_{aw}$ listed in the text in addition “Total Regression and Balanced Regression” measures. Any number of $R_{aw}$ tests can be included in one measure. Medisoft (mainland Europe) and Pulmolink (UK) provide for panting $R_{aw}$ as primary choice, but also offer tidal breathing methods. They allow VC to be done before or after shutter closure, and to restart procedure if desired. VC measures can be imported and aligned at measured TLC. Software in US models differs from European models.
3: The “instantaneous $R_{aw}$ described in the text below is provided with the same technical procedures and parameter listings. An additional graphic and numerical display of resistance as a function of absolute thoracic gas volume (R/V display) within the tidal breathing excursions is provided. Pre- and post-bronchodilator measures can be superimposed on the same graphic representation.
4: Rigid security glass walls with good thermal conductivity provide for patent pending electronic compensation of both thermal and pressure signals, allowing rapid testing to begin without the usual 2-min warm-up. Unique variable-orifice flow meter provides very low flow resistance and independence from humidity effects. Electronics in US models differs from European models.
5: Extra-cost option where available.
6: Rigid security glass walls with good thermal conductivity provide for patent pending electronic compensation of both thermal and pressure signals, allowing rapid testing to begin without the usual 2-min warm-up. Unique variable-orifice flow meter provides very low flow resistance and independence from humidity effects. Electronics in US models differs from European models.
7: Cosmed provides all the usual measures of $R_{aw}$ listed in the text and in addition “Total Regression and Balanced Regression” measures. Any number of $R_{aw}$ tests can be included in one measure. Medisoft (mainland Europe) and Pulmolink (UK) provide for panting $R_{aw}$ as primary choice, but also offer tidal breathing methods. They allow VC to be done before or after shutter closure, and to restart procedure if desired. VC measures can be imported and aligned at measured TLC. Software in US models differs from European models.
8: The “instantaneous $R_{aw}$ described in the text below is provided with the same technical procedures and parameter listings. An additional graphic and numerical display of resistance as a function of absolute thoracic gas volume (R/V display) within the tidal breathing excursions is provided. Pre- and post-bronchodilator measures can be superimposed on the same graphic representation.
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Principles of whole-body plethysmography

The body box consists of a rigid chamber, comparable in size and shape to an enclosed telephone booth, in which the subject sits. Pressure transducers measure the pressure across a pneumotachograph (flow), pressure across the wall of the box, and pressure at the airway opening.
In the variable-pressure box, changes in $P_A$ are inferred from changes in box pressure. A shutter close to the mouth can be closed to occlude the airway transiently, during which voluntary respiratory efforts performed against the closed shutter are recorded (Mueller and Val Salva manoeuvres), allowing changes in $P_A$ to be estimated by the change in mouth pressure ($P_m$). $P_m$ is plotted against simultaneous box pressure changes during respiratory efforts against a closed shutter to measure absolute TGV. The relationship between alveolar and box pressure measured during respiratory efforts against a closed shutter is extended to dynamic events during breathing to measure $R_{aw}$, defined as the relationship between airflow and $P_A$.

Three different types of whole-body plethysmograph have been used to measure changes in thoracic volume: large volume changes during the VC, or only those that result from compression/decompression of gas in the lungs. Different transducer sensitivities and mechanical arrangements are used to achieve this. The constant-volume (variable-pressure) body box measures small volume changes due to compression/decompression of gas in the lungs. The constant-pressure (volume-displacement) body box (Mead box) measures only large changes in lung volume due to gas flow into and out of the lungs. The pressure-corrected variable-volume body box, (transmural), combines the advantages of the other types. The sensitivity and rapid frequency response of transmural boxes provide measures of large slow volume changes in the lungs during breathing as well as TGV and $R_{aw}$, but not all manufacturers currently provide this technology (table 1). The use of the pure volume-displacement body box (Mead box) does not provide the *sine qua non* of clinical plethysmography, namely measurement of TGV and $R_{aw}$. It will not be further discussed.

Box pressure transducers must register very small changes in pressure. If the total chamber volume of the box is ~800 L, and TGV is ~4 L, and respiratory efforts against the closed shutter produce changes in alveolar pressure of 2 kPa, then the change in TGV due to compression/decompression is ~80 mL. This 80 mL compression volume “signal” will cause a change in box pressure of 0.01 kPa. Thus mouth pressure changes are 200 times box pressure changes. This ratio is directly related to the ratio of TGV to box chamber volume.

With such sensitive transducers to measure changes in box pressure, it is logical that small changes in ambient room air pressure may cause an artefact in box pressure during patient testing. If the walls of the box are very rigid and the box is properly sealed, this can be compensated for by computer algorithms. Alternatively, a physical compensation can be made for changes in ambient room air pressure, using a “compensation chamber” (similar to the original DuBois box). Table 1 lists methods used by different manufacturers. An additional source of possible error in box pressure is slowly occurring pressure changes unrelated to respiratory manoeuvres, such as heating of the interior caused by the presence of a person inside the chamber. These changes may be compensated for by software algorithms or by periodically venting the box to the room, or both.

In practical use, the box pressure transducer is calibrated in terms of changes in TGV by rapid introduction and withdrawal of 30–50 mL air into the box chamber, commonly using a motor-driven syringe. After calibration, changes in box pressure reflect changes in TGV due to compression/decompression of thoracic gas. Changes in calibrated box pressure are usually recorded in terms of a volume change known as “shift.
volume”. Shift volume is the change in TGV due to compression/decompression during occluded respiratory efforts, and also during breathing within the body box. Calibration of the body box is normally done without a subject in the chamber, and must be corrected for the subject’s body volume, using the subject’s body mass to calculate the final adjusted calibration coefficient.

Figures 2a and b are schematic representations of a variable-pressure constant-volume plethysmograph and a pressure-corrected integrated-flow plethysmograph. Frequency response of box volume changes is achieved by adding a signal proportional to box pressure to the integrated pneumotachograph signal in the wall of the box. Such a pressure-corrected plethysmograph permits accurate measurement of changes in TGV during forced expiration manoeuvres. Occlusion of the pneumotachograph in the wall of the plethysmograph and routing the mouthflow pneumotachygraph back into the box chamber converts the flow plethysmograph back into a variable-pressure plethysmograph, allowing more sensitive measurements of TGV and $R_{aw}$.

The box pressure change required to drive air through a flow meter in the wall of the box reflects compression/decompression of box air that does not reach equilibrium until box pressure has returned to atmospheric, as noted by Mead [5]. Thus, as the subject breathes room air through a tube across the wall of the box, changes in TGV expand or compress box air, and simultaneously displace some air in or out of the box across the flow meter in its wall. The volume change due to compression or decompression of box air is accounted for by adding an electrical signal proportional to box pressure to the integrated flow across the wall of the box (thus, “pressure-corrected”) [4–7].

Figure 2. Schematic representation of: a) a variable-pressure constant-volume plethysmograph; and b) a pressure-corrected integrated-flow plethysmograph. (a) illustrates the controlled mechanical leak to room air and optional reference chamber. The subject breathes through a pneumotachograph entirely within the box chamber. Recording of volume displacements of the thorax is limited to compression and decompression of thoracic gas. Calibration of plethysmographic pressure is done via a motorised syringe inserting and withdrawing 30–50 mL of air into the box chamber at a frequency of 1–3 Hz. (b) Arrows illustrate differences in the pressure-corrected integrated-flow plethysmograph showing the subject’s breathing through the wall of the box to room air with an additional pneumotachograph recording changes in box volume. Recording of volume displacements of the thorax includes mass of gas airflow into and out from the lung, and compression/decompression of thoracic gas $V'$: flow; $P_A$: alveolar pressure; $P_m$: mouth pressure; $V_L$: lung volume; $R_{aw}$: airway resistance; TGV: thoracic gas volume; $\Delta V$: change in volume; $\Delta P$: change in pressure.
Pressure-corrected integrated-flow body (transmural) boxes provide sensitive recordings of pressure and volume events over a wide range of volume displacements, including maximal expiratory flow volume curves and measurement of TGV, specific airway resistance (sR\textsubscript{aw}) and R\textsubscript{aw} with the same instrument. They permit evaluation of differences between thoracic gas compression and airway closure, the so-called “trapped gas” [10].

Clinical use of plethysmography

After the subject has entered the plethysmograph, the door is closed with an airtight seal. Some body boxes require ~2 min for chamber pressure to equilibrate during warming and humidification of chamber air by the subject. During this initial period, the plethysmograph cabin is vented periodically to room air via a solenoid-operated valve. After equilibration of box pressure, the subject is asked to close his/her lips tightly around the mouthpiece and breathe normally through the pneumotachygraph. The patient sits erect with head and neck in a neutral posture. A nose-clip is applied to close the nares. Patients may require reassurance in adapting to breathing within an airtight box, prior to making clinical measurements.

Clinical body box measurements include three measuring sequences whose order may be defined by the user. sR\textsubscript{aw} is usually measured first, followed by measurement of TGV and finally, measurement of slow and forced vital capacities.

Measurement of TGV

As described initially by DuBois et al. [1], measurement of TGV is done in the body box using Boyle Mariotte’s law relating pressure and volume changes under isothermal conditions. During compression of thoracic gas, P\textsubscript{A} rises, and the product of pressure and volume remains constant. In the body box, respiratory efforts against the closed shutter produce changes in alveolar pressure, which are closely similar to P\textsubscript{m} changes, and are associated with reciprocal changes in TGV. TGV is decompressed and compressed, causing corresponding changes in box pressure, which are recorded in terms of the change in TGV, denoted as “shift volume”.

In normal subjects, change in P\textsubscript{m} closely approximates that in P\textsubscript{A} during panting efforts [1]. However, in patients with airflow obstruction, significant differences between changes in oesophageal pressure and P\textsubscript{m} occur during panting efforts against a closed shutter [11–16]. If panting efforts are done slowly (1 Hz), it is possible to measure changes in P\textsubscript{A} from P\textsubscript{m}. P\textsubscript{m} (P\textsubscript{A}) is plotted against simultaneous box pressure changes (measured as shift volume) during respiratory efforts against a closed shutter to measure absolute TGV.

Boyle Mariotte’s law states that the product of pressure and volume is constant under isothermal conditions. In the body box, respiratory efforts against the closed shutter change P\textsubscript{A} and TGV by small amounts. These changes are shown in equation (1):

$$P\textsubscript{A} \times TGV = (P\textsubscript{A} – \Delta P\textsubscript{A}) \times (TGV + \Delta TGV) \quad (1)$$

After expanding and rearranging equation (1):

$$TGV = \frac{\Delta V \times (P\textsubscript{A} – \Delta P\textsubscript{A})}{\Delta P\textsubscript{A}} \quad (2)$$

The terms in equation (2) are defined as follows:

TGV is the absolute intrathoracic gas volume to be measured (calculated).

P\textsubscript{A} is alveolar pressure at rest (before any occluded respiratory efforts). It is important to note
that at resting end-expiratory lung volume (EELV) with the airway open to atmosphere, \( P_A \) is not “zero” but rather is equal to atmospheric pressure, \( \sim 100 \) kPa.

\( \Delta P_A \) is the change in alveolar pressure during the occluded respiratory effort, measured as change in \( P_m \).

\( \Delta TGV \) (or \( \Delta V \)) is the change in thoracic gas volume during the occluded respiratory effort. This change in volume, as noted above, will be 40–80 mL. It is equal and opposite to the change in box volume occurring simultaneously in the airtight body box. This “volume” change is measured by the box pressure transducer, as the shift volume.

Because \( \Delta P_A \) is very small compared to \( P_A \) (\( \sim 2\% \)) it is usually omitted in the differential term.

The measurement procedure for determining TGV requires substantial subject cooperation, making respiratory efforts while ventilation is interrupted by the closed shutter. Therefore, TGV may vary from test to test.

In contrast, residual volume (RV) and total lung capacity (TLC) require less subject coordination. An inspiratory capacity (IC) effort may be made immediately after shutter opening following the TGV measurement, to measure TLC. RV may then be obtained by subtracting a subsequent slow VC (SVC) from TLC, or alternatively by subtracting an independently measured SVC from TLC, to avoid integrator drift during exhalation to RV in the box.

Control of shutter closure in most body boxes includes a program to reopen after a defined occlusion duration or number of zero-pressure crossings, to minimise patient discomfort and avoid the subject experiencing fear when (s)he is trying to breathe but the shutter is closed! Subjects should always be informed to remove the mouthpiece from their mouth in the event the shutter does not open or if the subject senses substantial difficulty breathing. The shift volume and the corresponding \( P_m \) changes during shutter closure are displayed on an X–Y graph in figure 3, demonstrating excellent quality and patient cooperation.

As in all pulmonary function evaluations, it is recommended that three replicates of the measurement of TGV are recorded and saved. Measurement quality is reflected in large part by the variability of replicate trials. QUANJER et al. [17] suggest a maximal deviation of 5% between the individual trials.

**Measurement of sRaw**

Thermal and humidity effects arise during inspiration of plethysmographic air and expiration of warm humid alveolar air. Electronic compensation for thermal and humidity effects permits tidal breathing during measurement of \( s_{Raw} \) [18]; and current whole-body plethysmographs commonly incorporate algorithms to compensate for these effects.

Mouth flow during spontaneous breathing is continuously recorded from the pneumotachygraph and displayed on a graphic X–Y.
display versus the shift volume produced by thoracic compression and decompression as shown in figure 4. As noted above, shift volume excludes lung volume change due to gas flow in and out of the lung. The sRaw loop is influenced by Raw and TGV and its inclination rotates clockwise if either Raw or TGV or both are increased.

During assessment of sRaw, the relationship between airflow and shift volume described by Dubois et al. [2] defines not Raw, but sRaw. Currently, sRaw is most often defined during tidal breathing, without rapid shallow respirations. sRaw is given by the slope of the mouth pressure versus shift volume loops shown in figure 4 (corrected by the body box computer to refer to “dry” gas). Different numerical “slopes” may be calculated from the sRaw loop shown for a patient with chronic airflow obstruction (fig. 4c), depending on which portion of the loop is chosen.

Subjects should sit upright and avoid neck flexion or rotation. Five to ten sRaw loops should be recorded as one trial. Normally, three replicate trials are recorded and saved. Optimal quality of the recording is achieved when sRaw loops are regular and reproducible with the loop nearly closed, although patients with significant airflow obstruction manifest open loops during expiration.

While loops shown in figures 4a and b appear as nearly linear flat loops, the sRaw loop is complex in the presence of peripheral airway disease, as described by Dubois et al. [2]. Since the sRaw loop includes varying flows throughout the tidal breathing respiratory cycle, different investigators have used different portions of the loop to approximate a “representative” value for the entire cycle.

The total specific resistance (sRtot) [19] and effective specific resistance (sReff) [20] are both utilised in clinical laboratories in Europe. In North America, the linear portion of the sRaw loop between inspiratory and expiratory flow rates of 0.5 L·s⁻¹ [21, 22] provides a linear approximation of sRaw.

This last linear approximation (sR0.5) is shown in figure 5. It is measured between near end-inspiratory volume and the first portion of expiration. sR0.5 standardises the flow at which resistance is measured; and offers less interindividual variability.

In contrast, the sRtot, as described by Islam and Ulmer [19], is determined by a straight line between maximal inspiratory and maximal expiratory shift volume, and is more sensitive to peripheral airways disease, but manifests greater variability from test to test.

The sReff was introduced by Matthys and Orth [20]. This numerical parameter is used to integrate effects of variable flows and nonlinearities of mouth
flow-shift volume loops during tidal breathing, using the quotient of the integrated shift volume–volume loop (flow resistive work of breathing) and the integrated flow–volume loop. This ratio reflects larger central airways somewhat more prominently than sRaw.

The reciprocal of sRaw is denoted specific conductance (sGaw). The conversion of sRaw to sGaw is based on the original observations of Brusasco and DuBois [23] that the major determinant of Raw in normal subjects is lung volume and, accordingly, that the relationship between lung volume and conductance is linear within and between individuals. Thus, sGaw is a “volume-normalised” expression for airway conductance.

Since both resistance and resting EELV may change during bronchial or therapeutic challenge, sRaw and sGaw provide useful practical assessments of airway responsivity, even in the absence of a determination of absolute TGV. Such measures of airway response during tidal breathing are often considered preferable to spirometric assessments [24].

Calculation of airway resistance, RRaw, is simply by dividing sRaw by TGV. Thus as noted above, calculation of RRaw requires choosing a numerical approximation to the sRaw loop, and knowledge of TGV, determined by respiratory efforts against a closed shutter.

In practice, measurements of TGV are conveniently performed immediately after the sRaw breathing loops; and three replicates are recommended. Quality of the measurement is reflected in part by the variability of replicate trials and, in part, by how closely the Pm–plethysmograph pressure tracing approximates a straight line.

By definition, inaccuracy in the determination of TGV will cause a proportional error in RRaw calculations. For this reason, and because it is technically more demanding for patients with airflow obstruction to make respiratory efforts against a closed shutter than for tidal breathing, some clinicians restrict their attention to sRaw and sGaw [19, 20, 25, 26].

In many patients with COPD, RRaw appears to be nearly within normal limits, due to compensatory lung hyperinflation, especially when measured between 0.5 L·s⁻¹ inspiratory and expiratory flow. In these cases, sRaw and sGaw still show abnormality, because of the increased TGV maintained during tidal breathing.

Spirometric measurements may be made in the body box after TGV has been determined. An IC effort immediately after shutter reopening defines absolute TLC. This may be followed immediately by a maximal forced expiratory effort to define forced expiratory volume in one second and forced VC (FVC). These spirometric data are recorded from the flow meter in the pressure box.

Using the transmural box it is possible to view the maximal expiratory flow–volume curve with respect to volume displacements of the thorax, including compression effects, during forced expiration [27]. This is more reliable for detecting expiratory flow limitation (EFL) during resting breathing than spirometry using integrated mouth flow as the volume axis.

The VC measured from thoracic wall displacements in a transmural box is greater than that measured for integrated flow. This is not an artefact, and reflects compression of trapped gas, as shown schematically in figure 1.

Further discussion is provided in the European Respiratory Monograph [28].

Clinical interpretation and emerging concepts of body plethysmography

Clinical utility of whole-body plethysmography is discussed by BRUSASCO and PELLEGRINO [29] and physiological considerations are presented in detail by PRIDE and MACKLEM [30].

The raison d’être of whole-body plethysmography is the measurement of lung volumes. Accordingly, the first acknowledged clinical benefit of body plethysmography is the definition of restrictive lung disease [31]. Normative data for TGV and pulmonary subdivisions allow definition of restrictive lung disease as distinct from obstructive, in the presence of a reduced VC.

Definition of abnormally increased lung volumes in obstructive lung disease is a further appropriate clinical use of whole-body plethysmography. Because lung volumes measured by gas dilution techniques measure only the volume of ventilated airspaces, when plethysmography is combined with dilution measures of lung volumes, the volume of trapped gas is estimated by the difference between plethysmographic and dilutional TGV. The difference between FVC measured by transmural box volume change and that measured by integrated mouth flow also provides an estimate of trapped gas volume.

The voluntary rapid shallow obstructed respiratory efforts described by DuBois et al. [1] overestimate TGV because the change in Pm underestimates the change in PA [12, 13], in the presence of intrathoracic airway obstruction. In patients with airflow obstruction, changes in Pm significantly underestimate those in the oesophagus, taken to be equal to PA changes during respiratory efforts against a closed shutter. Increased airflow obstruction, ▶
increased compliance of the upper extrathoracic airways and increased rate of panting all combine to cause the underestimation of Pa change by $P_m$, and consequent overestimation of TGV. These studies have resulted in a recommendation of panting at 1 Hz to optimise the measurement of TGV.

The simplest form of Boyle Mariotte’s law used in manual calculations of TGV [1] has been evaluated by Coates et al. [32] who included calculation of TGV using the complete Boyle Mariotte’s law equation and demonstrated errors in the order of 3% during panting and 2-9% during a single inspiratory effort against a closed shutter as recommended for children [33]. Such discrepancies are not likely to influence clinical decisions, but are easily avoided using modern computational methods in automated whole-body plethysmographs [34].

The foregoing review of the measurement of TGV emphasised the cooperation required of the patient, including panting efforts against a closed shutter at a controlled low frequency and maintenance of an open glottis during obstructed respiratory efforts. Emerging concepts may avoid these constraints in the future by measuring TGV during tidal breathing without obstructed respiratory efforts [35]. Since $sR_{aw}$ is expressed numerically by the product of TGV and $R_{aw}$, addition of a known resistance in the respiratory path would permit determination of TGV by subtraction. Because $R_{aw} = sR_{aw} / TGV$, $sR_{aw}$ is the product of TGV and $R_{aw}$. By adding a known resistance it is possible to compute:

$$sR_{aw1} = R_{aw} \times TGV \quad (3)$$

and

$$sR_{aw2} = (R_{aw} + \text{Radded}) \times TGV \quad (4)$$

Thus:

$$sR_{aw2} - sR_{aw1} = \text{Radded} \times TGV \quad (5)$$

Or:

$$\text{TGV} = (sR_{aw2} - sR_{aw1}) / \text{Radded} \quad (6)$$

The advantage of estimating lung volume in this manner is that tidal breathing only is required. However, TGV must be constant between tidal breathing without and with the added resistance. It remains to be determined whether modern computer-assisted body boxes will provide comparable TGV results during respiratory efforts against a closed shutter and during tidal breathing without and with added resistance. This approach appears worthy of further investigation as it presents a convenient approach to the measurement of TGV that is likely to be more easily applicable to a wide variety of patients.

Calculation of $R_{aw}$ in a body box demands the constraints and linear approximations described above, and a single number defining “resistance” is not entirely satisfactory in patients with substantial peripheral airflow obstruction. Nonhomogeneous lung mechanical properties, EFL and airway closure all contribute to the complex shapes of $sR_{aw}$ loops. The complex shape of the $sR_{aw}$ loop itself provides more information than approximations of the $sR_{aw}$ slope in determining patients’ pathophysiology.

Plethysmographic $sR_{aw}$ can be measured both during rapid shallow breathing (panting) and during tidal breathing. The initial description of $sR_{aw}$ [2] utilised rapid shallow breathing to minimise thermal effects. This had the advantage of resulting in full opening of the vocal cords [36]. However, controlling panting frequency at a rate of 1 Hz [17–22], increases the likelihood of variable glottic opening [36].

The clinical utility of plethysmographic measurements of $R_{aw}$ and $sR_{aw}$ is attested to by the fact that they have been considered the “gold standard” for decades for assessing airway function. In patients with significant airflow obstruction, $sG_{aw}$ is commonly assessed. This permits lung hyperinflation to be taken into account. Normative values are available for $R_{aw}$, $sR_{aw}$, and their reciprocals, $G_{aw}$ and $sG_{aw}$ [37–40].

The choice of which measure of resistance is clinically most useful varies among different investigators and between countries. Some investigators emphasise $R_{tot}$ because it includes effects of multiple mechanical abnormalities associated with advanced peripheral airway obstruction. Against this is the disadvantage of test-to-test variability, due to its derivation from only two points (maximal inspiratory and expiratory shift volumes) of the $sR_{aw}$ loop.

Other investigators prefer $R_{eff}$, because it integrates the entire ranges of flow, shift volume and lung volume of the complete tidal breath, and may thus be expected to offer less within-individual variability.

Others argue against both these approaches because of their sensitivity to nonflow-resistive mechanical effects due to compression of nonventilating air spaces, dynamic expiratory intrathoracic airway compression, and EFL during tidal breathing.

These mechanical abnormalities, are largely excluded from the calculation of $R_{0.5}$. North American clinicians utilise $R_{0.5}$, from a standardised flow range between late inspiration, +0.5 L·s⁻¹, and early expiration, 0.5 L·s⁻¹, on the $sR_{aw}$ loop (fig. 5). This calculation results in a lower $R_{aw}$ than either $R_{eff}$ or $R_{tot}$ because it is minimally affected by dynamic airway compression or compression of nonventilating airspace. It may manifest less test-to-test variability within an individual.
The effects of dynamic airway compression and compression of nonventilating airspaces lead to a dependence of $R_{tot}$ and $R_{eff}$ on breathing pattern itself, namely the degree to which patients with chronic airflow obstruction “force” their expiratory effort.

During resting tidal breathing in normal individuals, expiratory airflow is largely produced by stored elastic energy in the chest wall. In patients with chronic airflow obstruction, active expiratory muscle recruitment is much more likely. Such patients commonly utilise active expiratory muscle effort to aid expiratory airflow and manifest EFL, even during resting tidal breathing [41]. The degree of expiratory muscle effort may change with bronchial or therapeutic challenge, and will directly influence calculated $R_{eff}$ and $R_{tot}$ because greater efforts cause greater shift volumes without corresponding increases in expiratory airflow in the presence of expiratory flow limitation.

Thus, there are marked differences between “instantaneous” airflow resistance during inspiration and expiration in patients with chronic airflow obstruction. Numerical representation of the mechanical abnormalities that occur separately during expiration is not possible using $R_{0.5}$, $R_{tot}$ or $R_{eff}$ due to the definition of these quantities based on the $sR_{aw}$ loops. Instead, graphic display of the $sR_{aw}$ loop is required to appreciate the prominence of such abnormalities during the expiratory phase [2, 42-44].

Current computer-assisted plethysmography makes it possible to calculate “instantaneous” values of airflow resistance, provided TGV is known. As noted above, during breathing within the body box, airflow resistance in the lung requires compression of thoracic gas during expiration and expansion of thoracic gas during inspiration, resulting in the “shift volumes” measured by the pressure change in the plethysmograph. Calculation of $R_{aw}$ requires measures of $P_{A}$ and airflow. During free breathing, shift volume can be used to record an index of changes in $P_{A}$, because shift volume is the product of TGV and the change in alveolar pressure, divided by initial $P_{A}$. In other words, the fractional change in $P_{A}$ integrated over TGV causes a change in TGV equal to shift volume, which, in turn, results in box pressure change. In this way, shift volume provides an index of changes in $P_{A}$ provided TGV is known. It is again emphasised, however, that box pressure change during breathing is not equal to change in $P_{A}$. It is much smaller in magnitude, and reflects the fractional change in $P_{A}$ modified by the ratio of TGV to box volume.

Instantaneous $P_{A}$ during free breathing may be estimated by computer, continuously in time, from measured signals of shift volume, volume and airflow after respiratory efforts against a closed shutter have been utilised to calculate TGV. Instantaneous $R_{aw}$ ($iR_{aw}$) is then defined by the ratio of instantaneous $P_{A}$ to instantaneous airflow. This computer calculation has only recently been implemented, and displays $R_{aw}$ throughout the tidal breath, except at end-expiration and end-inspiration, where $iR_{aw}$ is undefined because airflow is zero.

$R_{aw}$ calculated in this manner includes nonlinearities in flow resistance and effects of EFL and a contribution of compression of trapped gas to flow resistance. EFL contributes variably to apparent $R_{aw}$ as a function of respiratory effort: the greater the expiratory muscle effort, the larger the calculated expiratory $R_{aw}$ at a fixed flow rate. Compression of trapped gas during expiration and decompression during inspiration also contribute to the total dynamic $P_{A}$ burden during breathing. The degree of trapped gas in patients with airflow obstruction is likely to reflect small airway obstruction more importantly than obstruction of larger more central airways.

In summary, most commercially available body boxes at present provide useful measures of absolute lung volumes at RV, FRC, and TLC, and measures of resistance to airflow. Absolute lung volumes (at TLC) are the gold standard definition of restrictive lung disease. Airflow resistance is a simple name, but its evaluation during breathing in a body box requires careful attention to the complex effects of dynamic compression of intrathoracic airways during expiration, expiratory flow limitation during resting breathing, EFL, and compression of nonventilated airspaces. It is up to the individual physician to choose calculated “airway resistance” measures that incorporate his or her perceptions of the important physiological issues encountered in patients with obstructive lung disease.
REFERENCES


REFERENCES Continued


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**Elite™ Series Plethysmograph**

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**MasterScope Body**

MasterScope Body, the Gold-Standard for airway resistance measurements with optimised computer assisted loop compensation and wireless bluetooth connection to the box.

- Comprehensive PFT differential diagnostics including
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  - Spirometry, Flow/volume, MVV
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