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Practical assessment of respiratory mechanics

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More than 30 years after its first description, mortality associated with the acute respiratory distress syndrome (ARDS) is still great, with reported rates between 30 and 60%.[80] The range in mortality is partly because of the loose definition[6] of both ARDS and acute lung injury (ALI), based on bilateral lung infiltrates on X-ray, a left ventricular filling pressure <18 mm Hg and a Pa_{O_2}/FI_{O_2} of <300 mm Hg (ALI) or <200 mm Hg (ARDS). No factors relating to lung mechanics are used in these definitions. Use of PEEP can easily move the patient from ARDS to ALI or even out of the definition completely. The vagueness of the definition is also indicated when mortality is the same in ALI and ARDS, not related to ventilator settings.[49 50] However, Amato and colleagues[2] showed that a 'protective' form of ventilation, in contrast to 'normal' ventilator treatment, reduced mortality from 71 to 38% in ARDS, and the ARDSNetwork study found that mortality decreased from 40 to 31% when tidal volumes were decreased from 12 to 6 ml kg^{-1}.[1] It may be argued that a tidal value of 12 ml kg^{-1} is not normal, and these studies may only show that the use of large tidal volumes is harmful, as seen in the study of Amato and colleagues,[2] in which the control group mortality was very high (71%), rather than small tidal volumes being protective.[18] A link between lung mechanics and morbidity and mortality was suspected in 1998,[72] and in 1999 Ranieri and colleagues reported that inflammatory markers in lung lavage fluid and blood were less with a protective ventilatory strategy[65] and that the risk of developing organ failure was reduced.[65 66] The concept of ventilator-induced lung injury is well established. Lung injury can cause a systemic inflammatory reaction leading to multiple organ failure and death in patients with ARDS.[1] Despite the clear link between lung mechanics and outcome in ALI and ARDS, ventilator settings are normally based on blood gases rather than measurements of lung mechanics.[37]

Rarely is the gap between research and clinical practice as wide as in the assessment of lung mechanics. In research, static measurements are considered ideal and the ventilator settings are supposed to be made on this basis, i.e. end-expiratory pressure should be above the lower inflection point and tidal volume below the upper inflection point on the static pressure–volume curve to avoid cyclic collapse, reopening of alveoli and overstretching. In clinical practice, however, lung mechanics are assessed during ventilation so that pressure measurements are affected by the resistance of the endotracheal tube. At best, the ventilation is set with an end-inspiratory pause giving semistatic conditions, so that an adequate plateau pressure indicates the maximal alveolar pressure. Intermittent application of a prolonged hold in expiration gives information about intrinsic PEEP. The discrepancy between the static methods used for research and clinical reality and the dynamic measurements of lung mechanics raises the question of whether static measurements are truly superior to dynamic measurements, if the latter could be obtained in such a way that the effects of endotracheal tube and airway resistance were minimized.

Airway pressure

In clinical practice, airway pressure, measured in the ventilator or at the patient connection, and tidal volume measurements are the most common values used for assessment of lung mechanics. However, pressure measured proximal to the endotracheal tube is affected to a great extent by the tube resistance. Peak inspiratory pressure (PIP) is much greater than peak tracheal or peak alveolar pressure.[12 81] The magnitude of the difference depends on the tube resistance and the inspiratory flow at end of inspiration. This means that the smaller the inspiration:expiration (I:E) ratio, the greater the end-inspiratory flow and the more the PIP will overestimate the maximum alveolar pressure. If an end-inspiratory pause is used, the pressure during the pause (plateau pressure) represents the maximum pressure in the airway below the tube.[73] Recording PIP only is inadequate, but recording both PIP and plateau pressure gives useful information; for example, increased resistance caused by narrowing of the tube (secretions, kinking) will increase PIP and not affect plateau pressure.[75] If the increased resistance is caused by the lungs of the patient below the tube, then both the PIP and the plateau pressure may increase. In modern ventilators there is no end-inspiratory pause when pressure control mode is used and the proximal inspiratory pressure will always be greater than the maximum tracheal pressure. With normal I:E ratios (<1:2) there is little risk of incomplete expiration and the proximal pressure will give correct information about the end-expiratory pressure of the patient. However, if the I:E ratio is greater, or the tidal volume is large, or the frequency is high, or respiratory compliance is great, or resistance is great, there may be incomplete expiration and intrinsic PEEP can develop.[68] In such cases, the end-expiratory pressure can only be measured accurately if the end-expiratory pause is prolonged. This creates static conditions in the airway, so that airway pressure is the same from the proximal measurement point down to the alveoli. It is a disadvantage that this intrinsic PEEP level cannot be monitored continuously, as it affects oxygenation and outcome,[15 17] recruitment of lung and the maintenance of recruitment. The efficacy of triggering the ventilator is enhanced in patients with intrinsic PEEP if the extrinsic PEEP is set just below the intrinsic level. Mean airway pressure calculated from the proximal pressure usually gives correct values as overestimates of pressure during inspiration are balanced by underestimates of pressure during expiration.

Calculated tracheal pressure

By using measurements of flow, proximal pressure measurements and an expression for the resistance of an endotracheal tube developed from laboratory testing of clean tubes and connectors,[27] the tracheal pressure can be calculated continuously (Fig. 1). This is a great step forward in the monitoring of ventilator patients. It has been used in commercially available ventilators for automatic tube compensation. Particularly during inspiration, ventilator flow is increased to overcome the endotracheal tube resistance.[26 30 43 57] However, there are some drawbacks with this method. The calculation of tube resistance is based on measurements with clean tubes and a specific set of connectors. Adding a humidifier increases resistance by 15% and different connectors can have different resistances, which affects the calculated tracheal pressure. Loring and colleagues[47] found that changes in position or angulation of the endotracheal tube caused large changes in resistance of the tube, so that calculated tracheal pressure may differ considerably from directly measured tracheal pressure. Clean tubes are rare in the intensive care unit environment.[25] Secretions in the tube will cause overestimation of the inspiratory pressure and underestimation of the expiratory pressure below the tube. In children, where small endotracheal tubes do not allow the use of catheters to measure pressure, Guttmann and colleagues[29] calculated tracheal pressure from proximal pressure and flow by using

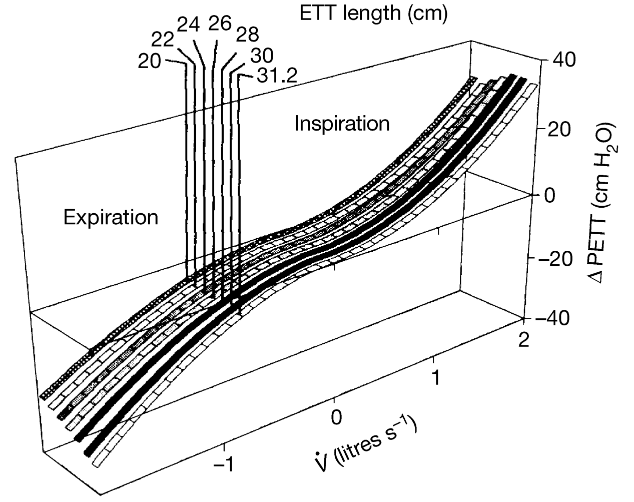

Fig 1 Decrease in pressure (ΔPETT) at different inspiratory and expiratory flows ($\dot{V}$) over endotracheal tubes of different lengths with an inner diameter of 7.5 mm. The pressure difference increases with increasing length, as expected. The pattern is most prominent during expiration. (Reproduced with permission from *Anesthesiology* 1993; **79**: 513.)

constants calculated from bench tests of endotracheal tubes of specified length, curvature and diameter. They included an expression for inertia,[27] that was not used in their adult method.

Direct tracheal pressure

Direct tracheal pressure measurements can be made by passing a catheter through the endotracheal tube and connecting it to a pressure transducer.[40] It is not clear whether the catheter should have an end-hole or a side-hole, or what should be the exact position of the tip of the pressure catheter. Holst and colleagues[34] showed a small difference in pressure readings between side- and end-hole catheters. Kárason and colleagues[40] found that the difference depended on the position of the catheter tip, and Navalesi and colleagues[60] found large differences in pressure according to whether the catheter was inside the endotracheal tube or below the tip of the tube. Direct measurements are made to obtain values for the tracheal pressure in the trachea where flow changes caused by the transition from tracheal pipe diameter to endotracheal tube diameter are minimized. This is between 2 cm above the carina down to the carina, which is a reasonable place for a reference measurement. With a side-hole catheter, the pressure readings will indicate this reference value only if the side-hole is placed below the tip of the endotracheal tube. When the hole is inside the tube, the static (lateral) pressure will be much higher than the pressure in the trachea because the flow velocity is ~10 times greater inside an endotracheal tube with an inner diameter of 7 mm than in the trachea with a diameter of 22 mm. Thus, the correct position of a side-

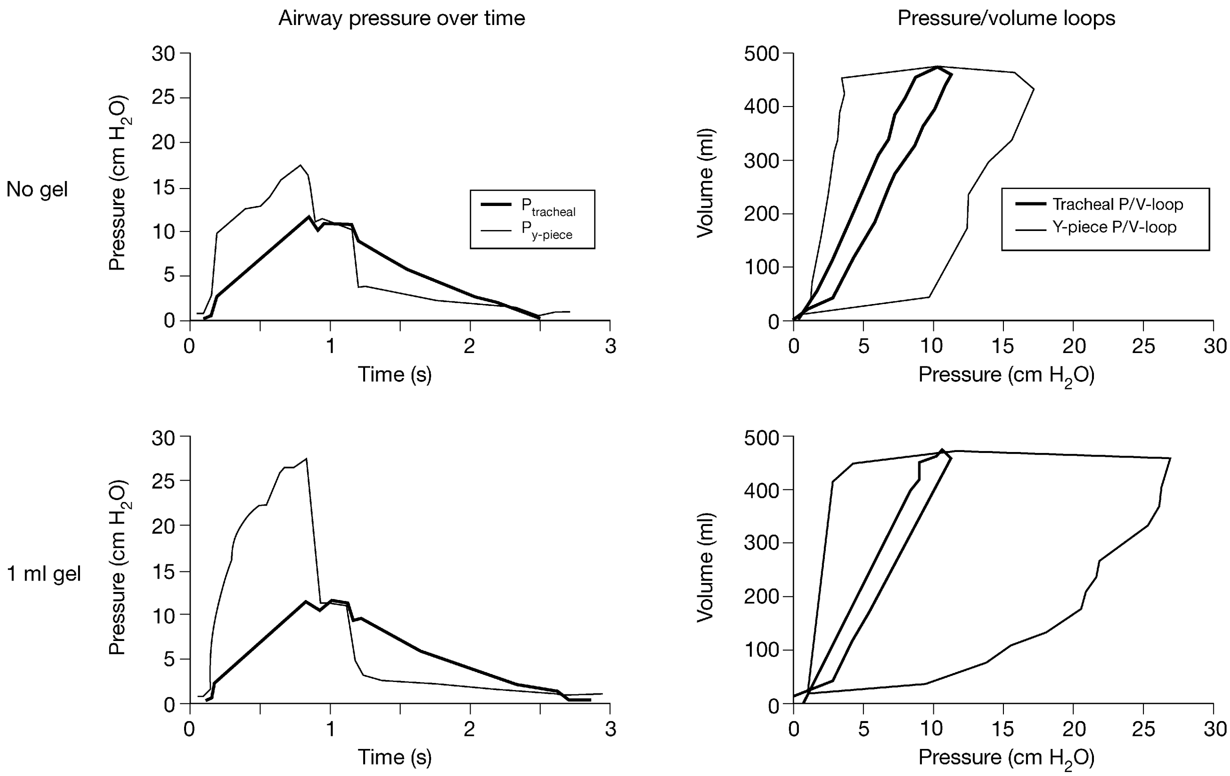

Fig 2 Proximal and distal airway pressure. The left column shows the directly measured tracheal pressure (thick black line) and Y-piece pressure (thin black line) in an endotracheal tube with inner diameter 7 mm without and with 1 ml of gel (mimicking secretions) deposited in the tube. The corresponding P–V loops are shown in the column to the right. Note that the peak tracheal pressure is not affected by the gel, but the peak inspiratory pressure measured at the Y-piece is markedly increased. The plateau pressure from the Y-piece measurement corresponds with the peak tracheal pressure because it is measured during static conditions. The decrease in tracheal pressure during expiration is much slower than the corresponding pressure at the Y-piece, which decreases almost immediately to the extrinsic PEEP level. In the P–V-loop graphs, a large increase in Y-piece loop area is noted after gel administration, but the tracheal P–V loop is almost unchanged. The main difference in the Y-piece loop is seen during inspiration, when the decrease in pressure across the tube is markedly increased by the gel in the tube. (Modified and reproduced with permission from *Acta Anaesthesiologica Scandinavica* 2001; **45**: 173.)

hole catheter is crucial and difficult to verify clinically. An end-hole catheter, on the other hand, measures static minus kinetic pressure during inspiration and static plus kinetic pressure during expiration. This would underestimate pressure during inspiration and overestimate pressure during expiration if the kinetic pressure was large. However, the kinetic energy of gas is so small that the effect on the pressure readings in comparison with the reference pressure is negligible.[40]

The pressure catheter can be gas-filled or fluid-filled with a slow drip passing through to keep it clear of secretions. The advantage of a gas-filled catheter is that it gives correct absolute pressure readings irrespective of the transducer position, whereas the fluid-filled catheter requires that the transducer is at the same level as the catheter tip to measure absolute pressure accurately. The fluid-filled catheter is less sensitive to secretions and occlusion. The gas-filled catheter can be occluded or the signal can be dampened by secretions, and has to be flushed with air from a syringe. Söndergaard and colleagues[74] found it easy to detect occlusion from the pressure signal and waveform for long-term tracheal pressure monitoring, and flushing every 2 h was necessary. The increase in airway resistance imposed by the pressure cannula in the trachea is small but intrinsic PEEP can occur if the resistance is increased too much. However, direct tracheal pressure measurements provide correct end-inspiratory and expiratory pressures, including intrinsic PEEP caused by ETT resistance, without stopping ventilation and irrespective of ventilatory mode. This is particularly important when an inverse I:E ratio or high respiratory frequencies are used. The difference between end-inspiratory and end-expiratory pressure is needed to calculate respiratory system compliance correctly, and correct values can be obtained only when intrinsic PEEP is considered in addition to the semistatic end-inspiratory pressure. In addition, the pressure waveform will show clearly that the changes in pressure during inspiration and expiration below the tube are much slower than above the tube during mechanical ventilation (Fig. 2). The difference is most marked during inspiration with pressure control ventilation, where proximal pressure increases promptly to the pressure value set in the ventilator and the tracheal pressure increases more slowly because of the endotracheal tube resistance.

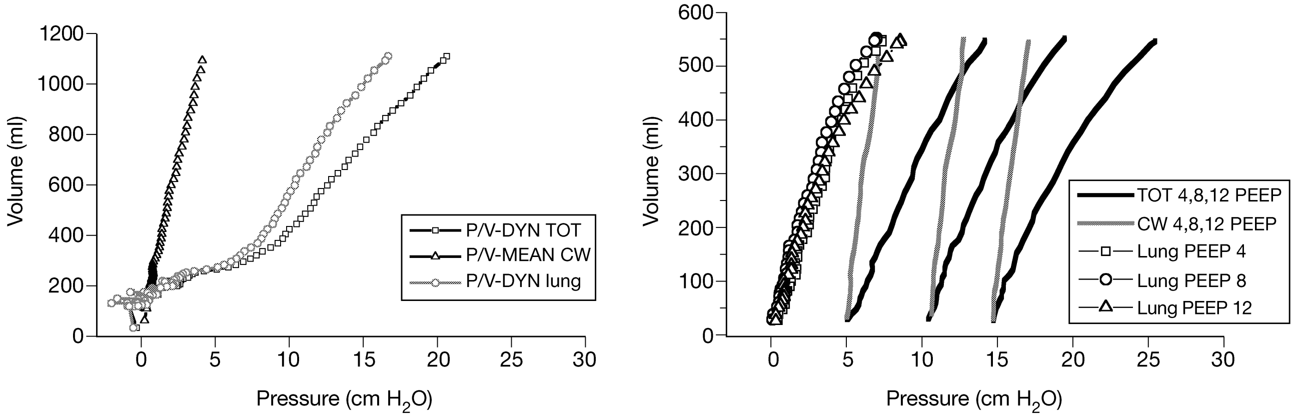

Fig 3 The left graph shows development of respiratory mechanics in a patient with ALI during low-flow inflation. Lower inflection points are noted at different positions in the total respiratory system (P/V-DYN TOT) and the lung dynostatic P–V curve (P/V-DYN Lung). In the P–V curve of the chest wall (P/V-MEAN CW) no LIP is seen. In the column to the right the respiratory mechanics are shown during normal tidal breathing in the same patient. No LIP can be seen in the P–V curves of the total respiratory system (TOT), chest wall (CW) or lung. The tidal breathing graph shows a successive decrease in volume-dependent compliance (Fig. 10) in the total respiratory system and lung within each breath. (Reproduced with permission from *Intensive Care Medicine* 2001; **27**: 1328.)

Söndergaard and colleagues[73] described direct tracheal pressure monitoring using a fibre-optic pressure transducer with a fibre dimension of 0.25 mm, which has almost no effect on resistance when it is placed in the endotracheal tube, even with one as small as 3 mm in inner diameter. This method is still under development but it could be used in combination with proximal pressures and flow to measure inspiratory and expiratory tube resistance continuously. This could provide a warning for tube obstruction, which is a constant concern in paediatric intensive care.

Pressure–volume curves

Rahn and colleagues[63] and Fenn[20] established the modern analysis of respiratory mechanics by describing the pressure–volume (P–V) curve of the respiratory system. They described a sigmoid static P–V curve where compliance was small below the functional residual capacity (FRC). At FRC, compliance was greatest and then decreased progressively with increasing lung volume. The transition from small compliance at low volumes to greater compliance is the lower inflection point (LIP), and the transition from greater compliance to smaller at the end of inflation is the upper inflection point (UIP). In subjects with normal lungs there is no LIP when starting inspiration from FRC. In 1984, Matamis and colleagues[53] related the pattern of the chest X-ray, the stage and the total respiratory system P–V curve in patients with ARDS. Very early in the disease, compliance was normal, with little hysteresis and no LIP. Normal compliance, increased hysteresis and an LIP were found in the early period, decreased compliance, marked hysteresis and an LIP occurred in the later stages of the disease, and in the end-stage a marked lowering of compliance, no increase in hysteresis and no LIP were noted. The lower inflection point was believed to reflect the pressure level at which collapse of alveoli occurred during expiration and then re-opened during the next inspiration. This was considered to cause damage, referred to as atelectrauma. At the top of the P–V curve, compliance decreased and this upper inflection point was considered to occur when the lung was overinflated. It was considered that tidal ventilation and PEEP should be set so that end-expiratory pressure was above the lower inflection point to prevent cyclic collapse of alveoli and the end-inspiratory pressure kept less than the UIP to avoid over-stretching.[3,13,15,59] Ranieri and colleagues[65,66] showed that avoiding collapse and overstretching reduced ventilator induced lung injury and multi-organ failure. However, there was no consensus on how to set PEEP.[9,24,44,58,76] The lower inflection point is still a much debated concept, and has been considered in full by Hickling.[31–33] It is puzzling that the first portion of gas, often not more than 100–150 ml, is sufficient to create enough pressure to open a great number of alveoli or terminal bronchioli. The compliance of the P–V curve below the LIP is often not more than 10 ml cm H_2O^{-1}. If all terminal bronchioli were closed, the event could perhaps be explained, but if some airways are not closed and still in continuity with open alveoli, where compliance would definitely be >10 ml cm H_2O^{-1}, then the first part of the gas would be delivered to this part of the lung rather than the least compliant parts. Also, why are lung mechanics quite different during dynamic conditions, i.e. normal tidal breathing, compared with static measurements? Neumann and colleagues[61] showed that the time to collapse alveoli is very short when pressure is reduced in the airways. When dynamic and semistatic P–V curves are compared (Fig. 3), a lower inflection point that is quite visible during a low flow inflation P–V curve disappears when tidal breathing with a frequency of 20 is used.[39] It seems that the time available for collapse is too short under these circumstances (Fig. 4). There are differences in dynamic and static P–V curves obtained with the SLICE

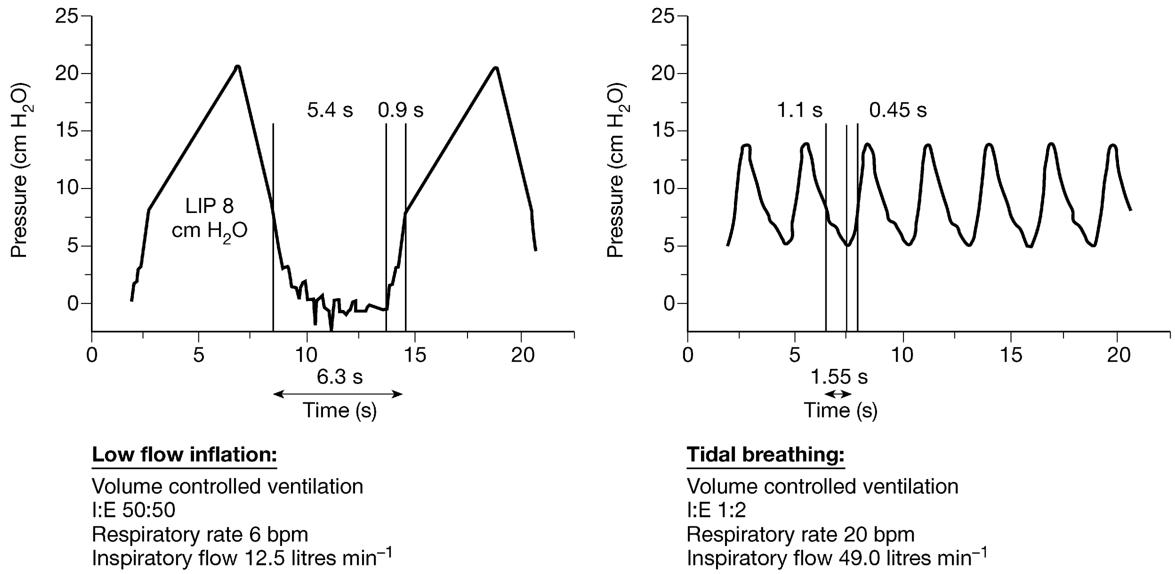

Fig 4 Changes in alveolar pressure over time in an ALI patient (same patient as in Fig. 3) during low-flow inflation and normal tidal breathing. During low-flow inflation, the time when the alveolar pressure is below the lower inflection point (LIP) (~8 cm H_2O) is 6.3 s, but it is only 1.55 s during normal tidal ventilation. In CT studies of end-expiratory breath-holding at zero end-expiratory pressure, airway collapse begins after 0.6 s and is almost over after 4 s.[22] During low-flow inflation, this allows time for the airways to collapse, but this might not be the case during normal tidal breathing. Thus, the respiratory system is not only volume- and pressure-dependent but also time-dependent.

method[28 55] and the dynostatic algorithm,[41] which are two breath-by-breath methods for alveolar P–V curves (see below). These show the greatest compliance at the start of inspiration and a progressive decrease in compliance during the whole inspiration.[39 42 46] This contrasts with the static P–V curve, where compliance is extremely low below the inflection and then increases after the LIP. The initial compliance of a dynamic P–V curve is often five to eight times greater than in the static P–V curve. Assessment of respiratory mechanics seems to need dynamic, continuous monitoring rather than static, intermittent measurements.

Static/semistatic methods for P–V curves and loops

The super-syringe

The static reference method is the super-syringe method, applying increments of volume with 50–100 ml gas up to a total volume of 1000–2000 ml. After each increment, the static airway pressure is measured during a pause of a few seconds when there is no flow, and the pressure is the same in the entire system from the super-syringe to the alveoli. The lung is then deflated in the same way and an inspiratory and expiratory P–V curve is plotted.[8] The whole procedure takes 45–120 s, during which oxygen uptake will continue from the alveoli. Carbon dioxide elimination will be close to zero during the inflation. Inspiration resembles a long apnoea, so that alveolar and arterial carbon dioxide partial pressures increase. This reduces the evolution of carbon dioxide from blood to alveoli as the content of carbon dioxide increases with increasing partial pressure. Thus, during stepwise slow deflation the carbon dioxide output from the alveoli will be less than before the start of the measurement, while oxygen uptake will continue unchanged. This causes expired volume to be less than inspired volume and the P–V curve has a marked hysteresis. In a patient with ARDS, with a small functional residual capacity, and sepsis with an increased oxygen uptake of ~300 ml min^{-1} and carbon dioxide production of 250 ml min^{-1}, if the inflation–deflation procedure lasts for 60 s the deflation volume would be 200–300 ml smaller than the inflation volume. In addition to this apparent hysteresis, which is an artefact from the mechanical point of view, the slow, stepwise inflation to a high pressure will also cause hysteresis because very slow compartments of the lung will be inflated and collapsed alveoli will be recruited. Both these phenomena mean that the deflation volume will be substantially smaller than the inflation volume. Compression of inflated gas and changes in humidity and temperature may cause further artefacts. Mathematical correction of these factors is advised but may be insufficient.[22 78] Thus, this method creates results which would not be seen during normal ventilator treatment.

The multiple occlusion technique

To mimic conditions during normal ventilation and ventilate the patient during the measurement, the multiple occlusion technique has been proposed.[36 45 77] It uses a sequence of different-sized volume-controlled inflations with an end-inspiratory pause to allow semistatic pressure measurements. Pressure and volume are plotted for each end-inspiratory pause to form a static P–V curve. If expiratory interruptions are also done, a static expiratory P–V curve is

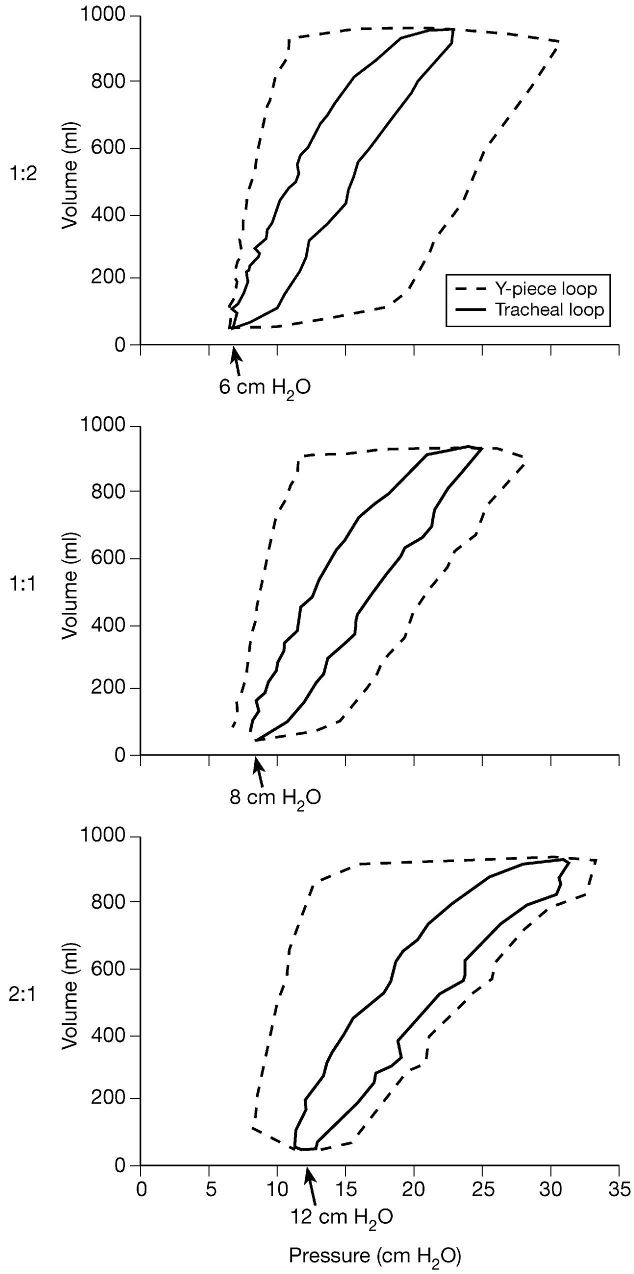

Fig 5 A patient with volume-controlled ventilation with extrinsic PEEP 8 cm H_2O and increasing I:E ratio. Development of intrinsic PEEP (indicated by arrows) is detected in the tracheal P–V loop but not in the Y-piece loop. Evidence of successive overinflation and decreasing compliance after increasing the I:E ratio, despite unchanged tidal volume, is explained by the development of intrinsic PEEP. (Reproduced with permission from *Acta Anaesthesiologica Scandinavica* 2000; **44**: 571.)

obtained. In spite of the fact that these inspiratory and expiratory P–V curves are truly static, no hysteresis is found even in ARDS patients. This is explained by how the measurement points on the curves are obtained: they are obtained by a series of normal, but varying breaths and the artefacts caused by gas exchange do not occur. The largest tidal volumes used with this technique may cause recruitment, but this is limited because of the short time the pressure is very high during tidal breathing. The entire sequence of measurements may take 5–10 min and all the tidal volumes must be in volume control mode. The multiple occlusion technique is the superior semistatic method and the inherent measurement artefacts are minimal.[69 70] This method can be regarded as static if the end-inspiratory pause is long enough also for viscoelastic pressure equilibrium. Even with a short end-inspiratory pause the pressure at the end of such a pause will be within 1 cm H_2O of a static measurement.

The low-flow inflation technique

The low-flow inflation technique uses a very small constant inspiratory flow to generate a large total volume. The low flow causes a minimal but recognizable pressure decrease over the endotracheal tube, which means that the dynamic inspiratory pressure volume curve will be shifted to the right[48 70] depending on the endotracheal tube resistance and the flow selected compared with a true static P–V curve. The slope (compliance) of the curve is parallel with a static P–V curve only if airway resistance is constant throughout the inspiration. This is doubtful, as reliable data show that resistance decreases as the lung is inflated and the airways widen.[4] As the flow is low, the duration of the inspiration will be long and the same artefacts described with the super-syringe technique are present to some extent. If the total inflated volume is great, with a large end-inspiratory pressure, this procedure will recruit collapsed lung. The low-flow inflation method does not analyse expiration and no information on hysteresis, real or artefactual, is obtained. Another drawback of the method is that the procedure requires resetting the ventilator and only volume-control ventilation can be used. Thus, lung mechanics estimated with a low-flow inflation do not reflect the lung mechanics of normal tidal breathing.

A common feature of these static and semistatic methods is that they require stopping therapeutic ventilation for a longer (super-syringe) or shorter period (low-flow inflation) or using a sequence of breaths that is different from the selected therapeutic settings. Such intermittent and static or semistatic methods may not be relevant in predicting the mechanical behaviour of the lung under dynamic conditions, where resistance and compliance depend on volume, flow and respiratory frequency.[4 14 19 71]

Dynamic methods for P–V loops

The obvious advantage of dynamic methods is that they are continuous and can be used as ventilator treatment is applied. Dynamic methods are more methods of monitoring than of measurement, and their lack of precision is counterbalanced to some degree by their capacity to follow trends.

Dynamic, proximal (to the endotracheal tube) P–V loops

Continuous display of pressure from the patient connection or ventilator can be plotted against volume to give breath-

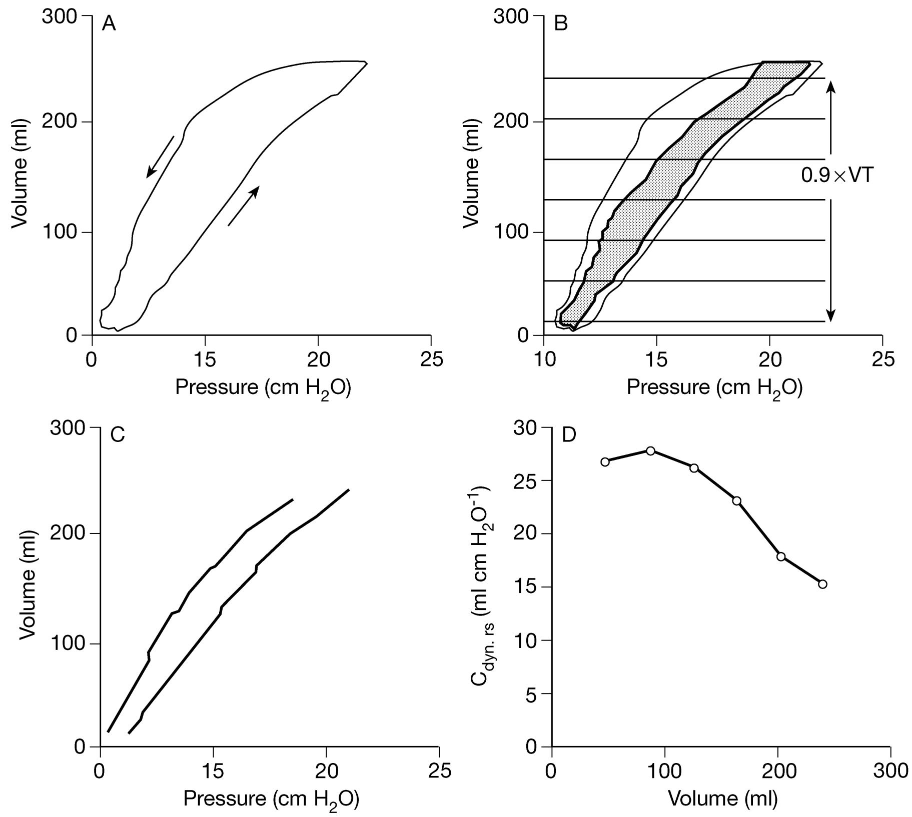

Fig 6 Proximal and tracheal loops and the SLICE method-based compliance–volume curve in a surfactant-deficient animal. (A) Proximal loop with up-arrow indicating inspiratory limb and down-arrow indicating expiratory limb of loop. (B) Calculated[27] tracheal pressure *vs* volume displayed inside the proximal loop. The 5% lowest and highest volume slices are excluded from analysis because of interference from the ventilator valve and large volume acceleration. The remaining 90% of the tidal volume is divided into six slices and compliance and resistance are calculated by multiple linear regression for each slice. (C) Calculated and measured tracheal pressures. They coincide almost totally, indicating good agreement. (Reproduced with permission from *British Journal of Anaesthesia* 2000; **85**: 577.) (D) Compliance–volume plot obtained using the SLICE method.

by-breath P–V loops. The effect of endotracheal tube resistance makes the interpretation of these loops very difficult. In pressure control mode, the inspiratory part of the loop shows an immediate increase in pressure to the set inspiratory pressure level and then an increase in volume with constant pressure level. However, the pressure in the lungs increases much more slowly and the square waveform of the loop recorded from ventilator pressure has little in common with the P–V events in the alveoli during inflation. In volume-control mode the inspiratory part of the proximal loop has an almost parallel but right-shifted course compared with the alveolar P–V curve. The magnitude of the shift depends mainly on the endotracheal tube resistance and the flow through it. As the inspiratory part of the proximally recorded loop is parallel to the inspiratory part of a loop from pressures measured below the tube, changes in the waveform show events occurring in the lungs. Thus, a curving of the plot during late inspiration reflects decreasing compliance or even overstretching of the alveoli. In neither pressure-controlled nor volume-controlled ventilation does the upper end of the loop reflect true peak alveolar pressure, because the pressure decrease over the tube is not distinguishable.

The expiratory part of the proximally recorded loop has the same form in pressure-control and volume-control mode and mainly reflects the low resistance of the expiratory pathway of the ventilator. Pressure measured in the ventilator tube will decrease almost immediately to atmospheric pressure or the end-expiratory pressure level set in the ventilator. The pressure in the trachea decreases more slowly. Intrinsic PEEP cannot be detected in the proximal loop. Because neither peak airway pressure at end-inspiration nor end-expiratory pressure is measured accurately when the proximal pressure is taken, compliance will be underestimated.

The stress index method

The stress index method[67] analyses the pattern of proximal pressure change during volume-controlled inspiration. A decrease or deflection of the slope during inspiration indicates increasing compliance and possible recruitment of lung, and an increase or inflection in the slope suggests overinflation and a decrease in compliance. The technique assumes that airway resistance in the endotracheal tube and patient airway is constant during inspiration. This may not be correct in the patient airway, where airway resistance decreases as the airway widens with lung inflation, but endotracheal tube resistance will be almost constant during constant-flow inspiration. As the endotracheal tube contributes ~70% of the resistance between the Y-piece and the alveoli, changes in the resistance of the patient airway during inspiration probably have a negligible effect on the accuracy of this method, which so far has been used only in isolated rat lung preparations.

Dynamic, distal (to the endotracheal tube) P–V loops

Plotting tracheal pressure against volume provides dynamic distal (tracheal) P–V loops (Fig. 5)[41] where the influence of endotracheal tube resistance during inspiration is removed, and during expiration endotracheal tube resistance is accounted for. Such a loop has an area that is only 30% of a loop made using proximal pressures, and the difference is caused by endotracheal tube resistance (Fig. 6). With the distal loop, the inspiratory part increases more rapidly during pressure-control mode as the flow is great at first, compared with volume-control mode. The expiratory limb shows a slow decrease in pressure and volume as the endotracheal tube resistance delays the pressure decrease in the airways. Overdistension can be detected even in pressure-control ventilation with decelerating flow, not as an upper inflection point but rather as a progressively decreasing compliance with increasing inspiration—an overdistension zone. The endpoints of the tracheal loop give the pressure values used for calculation of conventional endpoint compliance. It is clear, especially where marked overdistension is present and the loop is convex upwards, that a line through the endpoints will lie outside the tracheal loop. Thus, compliance must vary with volume, and compliance will be greater at the beginning of the breath and then decrease with increasing volume.

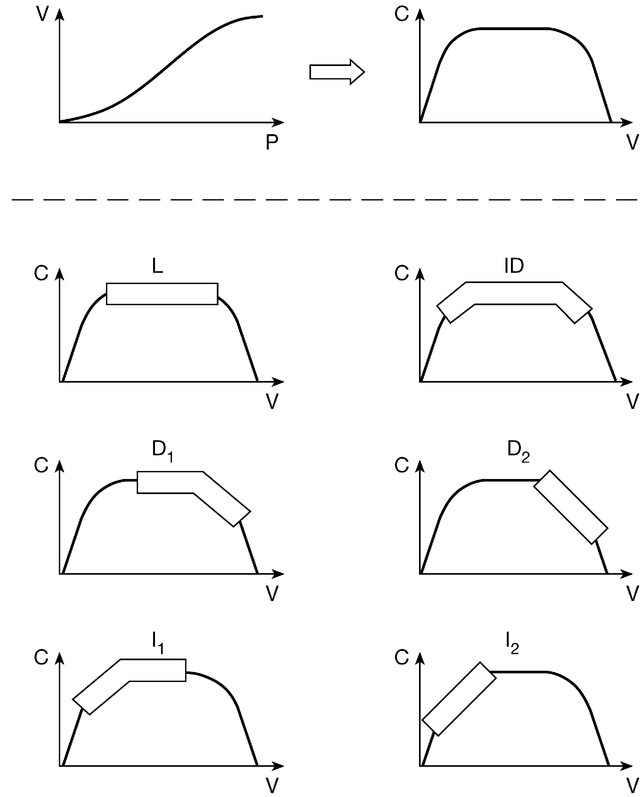

Fig 7 Top panels show an S-shaped P–V curve, with a corresponding trapeziform compliance–volume curve. Below are six pre-defined shapes of parts of the compliance–volume curve. L is the linear part, with constant compliance, of the SLICE method. ID indicates the increasing–decreasing compliance pattern in the tidal volume. D_1 and D_2 respectively show moderately and severely decreasing compliance with increasing tidal volume. I_1 and I_2 respectively show moderately and severely increasing compliance with increasing tidal volume. The SLICE method gives information on volume-dependent compliance on a breath-by-breath basis. (Reproduced with permission from *Intensive Care Medicine* 1999; **25**: 1084.)

Dynamic alveolar P–V curves

Multiple linear regression

The equation of motion is:

$$\Delta P = V/C + \dot{V} \times R + \text{PEEP}$$

where V/C is the pressure necessary for expanding the lung, $\dot{V} \times R$ is the pressure necessary to overcome the resistance of the airways and PEEP is the pre-existing pressure in the lungs. Values for C and R can be found by a statistical method, multiple linear regression, also known as the least-squares fit method (LSF or LSM).[35 79] This method assumes constant compliance and resistance throughout the whole breath or for the inspiratory and expiratory parts separately. Because this assumption may not be valid, the method has been developed by dividing the breath into six slices and applying the method for each slice (Fig. 7).[55] The assumption is that resistance and compliance remain constant within each slice (Fig. 8). The method estimates changes in compliance in six parts of the breath.[56] This forces the inflection points to be positioned at the intersection between the slices. The number of slices can be increased, but this reduces the data available for the multiple linear regression computation and the precision of the estimates of C and R.

The dynostatic algorithm

Kárason and colleagues proposed a method for breath-by-breath analysis of alveolar P–V curves.[42] The method is based on the tracheal P–V loop and analysis of pressure and flow at the same lung volume during inspiration and expiration. Inspiratory and expiratory resistance at each volume are assumed equal. Alveolar pressure is calculated using the equation of motion, which is the basis of the dynostatic algorithm:

$$P_{\text{insp}} = P_{\text{elastic}} + \dot{V}_{\text{insp}} \times R_{\text{insp}} \rightarrow R_{\text{insp}} = (P_{\text{insp}} - P_{\text{elastic}})/\dot{V}_{\text{insp}}$$

$$P_{\text{exp}} = P_{\text{elastic}} + \dot{V}_{\text{exp}} \times R_{\text{exp}} \rightarrow R_{\text{exp}} = (P_{\text{exp}} - P_{\text{elastic}})/\dot{V}_{\text{exp}}$$

As $R_1 \approx R_E$,

$$P_{\text{dynostatic}} = P_{\text{elastic}} = (P_{\text{exp}} \times \dot{V}_{\text{insp}} - P_{\text{insp}} \times \dot{V}_{\text{exp}})/(\dot{V}_{\text{insp}} - \dot{V}_{\text{exp}})$$

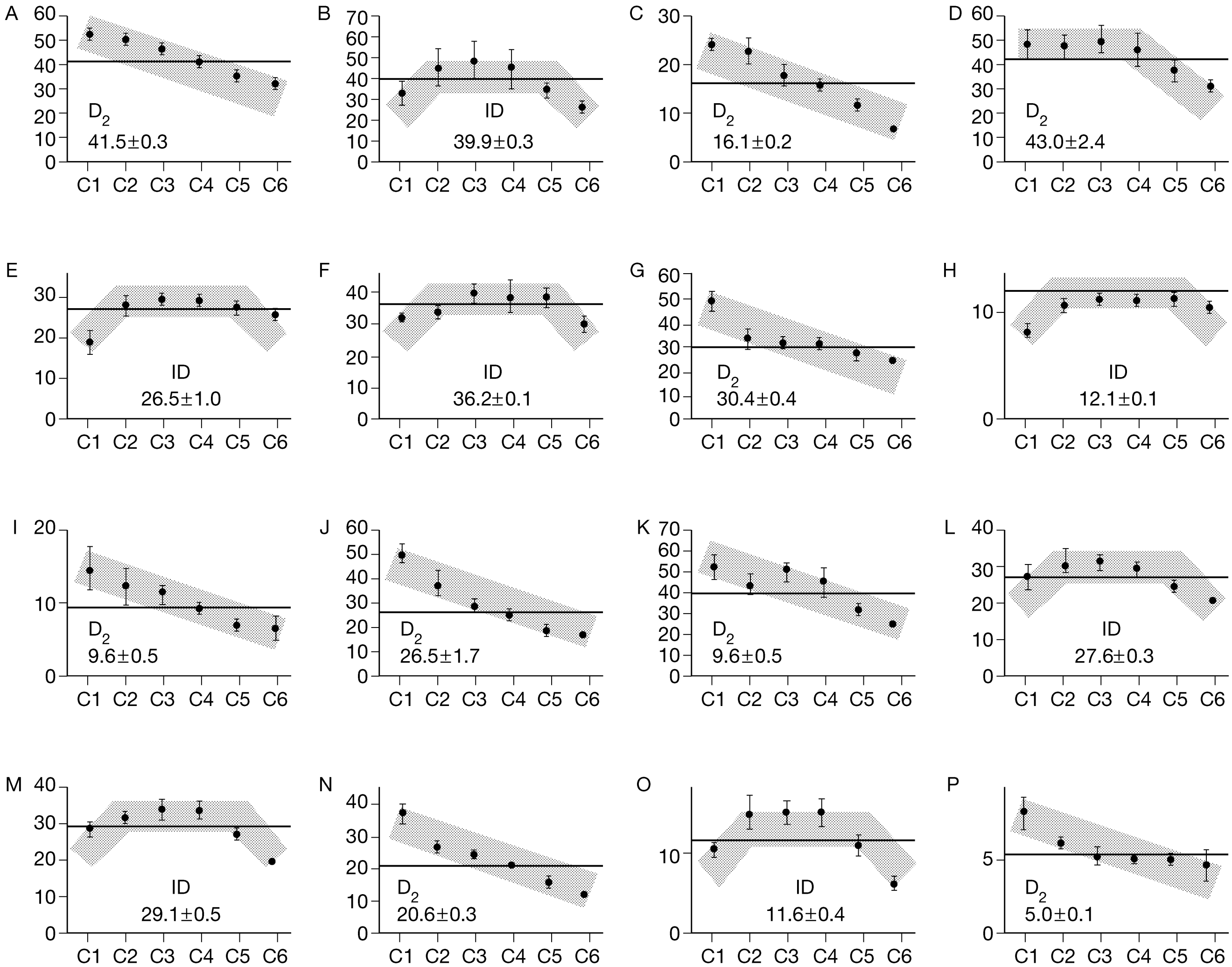

Fig 8 Volume-dependent compliance and single-value compliance in 16 patients (A–P). Compliance is on the *y* axis and tidal volume slice on the *x* axis, lowest to the left and greatest to the right. Volume-dependent compliance is calculated by the SLICE method and shows compliance (ml cm H_2O^{-1}, mean and standard deviation) for six different volume parts of the tidal volume. Single-value compliance, calculated by regression, is shown as a straight line parallel to the *x* axis. Note that in patients B, E, F, H, L, M and O there are signs of both a lower and an upper inflection point. In the remaining nine patients, compliance decreases with increasing volume, indicating possible continuous stretching or overstretching. In these patients, with the possible exception of patient D, no upper inflexion point is seen. (Reproduced with permission from *Intensive Care Medicine* 1999; **25**: 1084.)

The calculation is repeated for a number of volume values and alveolar pressure is estimated. A plot of alveolar pressure against volume can be made during dynamic ventilation (Fig. 9). In contrast to other methods of dynamic calculation of the alveolar P–V curve, such as the multiple linear regression method, the only assumption is that inspiratory and expiratory resistance are equal at each isovolume, and do not have to be constant throughout inspiration or expiration. The dynostatic algorithm does not require constant flow and gives correct dynamic alveolar P–V curves irrespective of ventilatory mode, without needing an end-inspiratory pause to calculate alveolar pressures.

To test the effect when inspiratory and expiratory resistance values are unequal, which contradicts a basic assumption of the dynostatic algorithm, the authors used a lung model in which the resistances could be varied independently. Within a wide range of R_{insp}:R_{exp} ratios from 2.3:1 to 1:2.3, the dynostatic algorithm P–V curve related very well to the reference alveolar P–V curve in the lung model. In ARDS patients, R_{insp} and R_{exp} are nearly equal.[7] The reason for the good agreement is that the tracheal P–V loop used for the calculations and the inspiratory–expiratory tracheal pressure difference is small (5–7 cm H_2O at mid-tidal volume) where the difference is usually greatest. Thus, even if the expiratory resistance is twice the inspiratory resistance, the alveolar pressure has to be between the inspiratory and expiratory limbs of the tracheal loop and the possibilities for deviation are small.

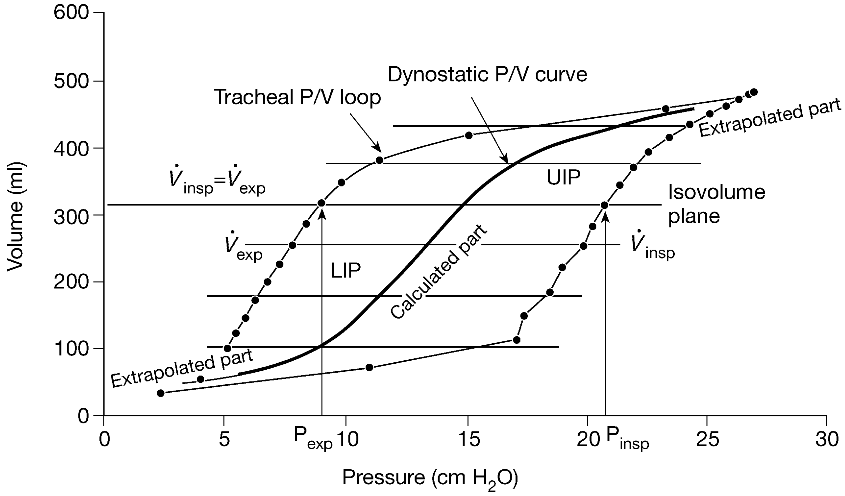

Fig 9 Schematic graph of the mathematical background for the calculation of the dynostatic P–V curve. Alveolar pressure is calculated according to the equation

$P_{\text{dynostatic}}=(P_{\text{exp}}\times\dot{V}_{\text{insp}}-P_{\text{insp}}\times\dot{V}_{\text{exp}})/(\dot{V}_{\text{insp}}-\dot{V}_{\text{exp}})$

Every point on the dynostatic curve (Px_{dyn}) is calculated using pressure and flow values at isovolume levels (indicated by broken lines) during inspiration and expiration in the tracheal P–V loop. (Reproduced with permission from *Acta Anaesthesiologica Scandinavica* 2000; **44**: 578.)

A typical feature of the dynostatic algorithm P–V curve is that compliance is great at low volume and decreases as the volume increases. The decrease in compliance as lung volume increases is much less when compliance is calculated in the conventional way, whereas the final compliance of the dynostatic algorithm P–V curve shows a marked decrease (Fig. 10). Kárason and colleagues[42] found no lower inflection points in patients with ALI or ARDS, but this may have been because they did not decrease PEEP to zero during measurements.

The dynostatic algorithm P–V curve can be regarded as an inspiratory P–V curve superimposed on an expiratory twin P–V curve. Because the time when alveolar pressure value is known, at each lung volume during inspiration and expiration, a plot of alveolar pressure can be made (Fig. 11).

Separation of respiratory mechanics into lung and chest wall elements

During the late 1990s, interest grew in the separation of lung and chest wall mechanics. Gattinoni and colleagues[23] suggested classifying ARDS into primary and secondary ARDS: primary ARDS was a primary disease of the lung and the secondary type was caused by extrapulmonary disease such as peritonitis. The response to PEEP was different in the two forms of the disease[23] and it seemed that the PEEP required could be determined by measurements of abdominal pressure. The lower inflection point present in static P–V curves of the total respiratory system in 13 patients was found only in the chest wall P–V curve in seven of these patients.[54] Kárason and colleagues[39] showed that lower inflection points were not found in two patients when alveolar P–V curves were obtained during dynamic conditions, and in one of the patients the LIP was only present in the chest wall P–V curve. These findings suggest that respiratory mechanics should be partitioned into lung, chest wall and/or abdominal mechanics. As a surrogate for pleural pressure, we measure pressure in the oesophagus and bladder pressure to estimate abdominal pressure.

Oesophageal pressure measurements

Oesophageal pressure has been used to estimate pleural pressure. Oesophageal pressure can be measured using a thin catheter with a 10 cm long balloon at the tip. The balloon is filled with 0.5 ml of air and usually positioned in the middle third of the oesophagus. It is crucial to confirm correct positioning by the occlusion test.[5] The method has not reached widespread acceptance as it is difficult to introduce the relatively soft balloon catheter into the oesophagus of an unconscious patient, particularly if the patient has a gastric tube in place already. Kárason and colleagues[37] suggested the use of a double-lumen gastric tube, the narrow lumen being connected to a standard pressure transducer and filled with fluid. Correct positioning was checked by compression of the rib cage during an expiratory hold, an occlusion–compression test, which can be done in paralysed patients. When this is done, the changes in oesophageal and airway pressure should be equal.[16] The advantage of this method is the easy introduction of the device, but it is difficult to position the transducer at the correct level for correct absolute oesophageal pressure measurements. However, the absolute value is less important than the difference between the end-inspiratory and end-expiratory oesophageal pressures, which is used to estimate the compliance of the chest wall.

Gastric pressure

Similar methods can be used for gastric pressure measurements, which will reflect intra-abdominal pressure. When a

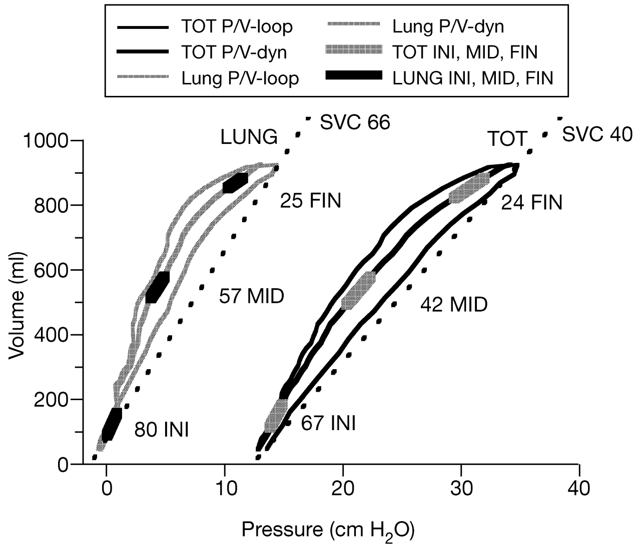

Fig 10 P–V loops in a patient with ALI from the lung and total respiratory system, based on tracheal pressure measurements. Conventionally, compliance is calculated as tidal volume divided by the difference between peak and end-expiratory pressure. The initial (INI), middle (MID) and final (FIN) volume-dependent compliances within the breath for the total respiratory system (TOT) and the lung (LUNG) are shown. Note that initial compliance is markedly greater and final compliance at the end of inspiration is markedly less than conventional single-value compliance (SVC). Calculation of volume-dependent compliance for different parts of the tidal volume could be a sensitive marker for collapse and overinflation of lung. (Reproduced with permission from *Intensive Care Medicine* 2001; **27**: 1328.)

fluid-filled measurement system is used, the mid-axillary line can be used as zero reference for the transducer.[11] In most cases of secondary ARDS the abdominal elastance is greater than that of the chest wall, indicated by intra-abdominal and oesophageal pressure measurements respectively. For gastric pressure monitoring, positioning of the pressure line is easy and the measured pressure is probably more reliable than from the oesophagus.

Bladder pressure

Abdominal pressure can be measured via a urinary catheter. The bladder is first drained of its content, 50–100 ml of saline is instilled and the catheter is clamped distal to the pressure measurement position.[52] Collee and colleagues[11] compared measurements of intra-abdominal pressure using a gastric tube and using bladder pressure, and found that the two pressures were within ±2.5 cm H_2O. Variations in bladder and central venous pressure correlate well with oesophageal and gastric pressures respectively.[10]

Imaging methods

CT scanning

Great advances in understanding lung mechanics, ALI and ARDS have been made with CT scanning methods.[23 24 62] It is mainly a research tool. The anatomical resolution is very good and the introduction of rapid spiral CT scanning has made it possible to measure the whole of both lungs rather than just selected slices.[61] Differences in aeration with gravity in a vertical axis can be shown and the effect of lung recruitment manoeuvres on different lung regions can be seen.[51 62] The drawbacks of the method are obvious: transport of a critically ill patient, and intermittent information. However, comparison between CT scanning and bedside assessment of lung mechanics can help in interpreting information from the non-imaging methods.

Electrical impedance tomography

Functional electrical impedance tomography (f-EIT) is a bedside imaging technique in which an electric current is passed through electrode pairs across the chest. The electrode pairs chosen are rotated around the chest using 16 electrodes placed around the circumference of the rib cage.[21] The resulting voltages are used to construct an impedance image. The resolution of the image is not as good as the CT scan, but gas distribution differences between spontaneous and controlled ventilation can be detected, as can changes in distribution caused by application of PEEP. The temporal resolution is rapid. The f-EIT technique is being developed with promising results.

Implications of respiratory monitoring for clinical management

Measurements of respiratory mechanics in basic research, to generate the ideas for clinical outcome studies, differ substantially from measurements used in clinical studies. Clinical study measurement methods are often crude and inadequate for the ventilatory strategy chosen in the study.[17 18] In standard clinical practice the monitoring of respiratory mechanics is even more superficial, often choosing what is easy to monitor rather than what is adequate. For example, peak inspiratory pressure proximal to the tube is often recorded rather than end-inspiratory pause pressure. In the near future the most important advance will be the transfer of methods used in measuring lung mechanics for research studies to clinical practice, to monitor the critically ill patient.

The first step should be improved tracheal pressure measurement, preferably by direct measurements to give breath-by-breath information on intrinsic PEEP and peak tracheal pressure, without interrupting ventilation and irrespective of the mode of ventilation. By measuring proximal and tracheal pressure, together with flow, endotracheal tube resistance can be monitored.

Rather than static and semistatic methods for P–V loops, dynamic P–V loops using the SLICE method or the dynostatic algorithm will show immediately if a lower inflection point is present, and indicate clearly any tendency to overstretch the lungs. This on-line information will help

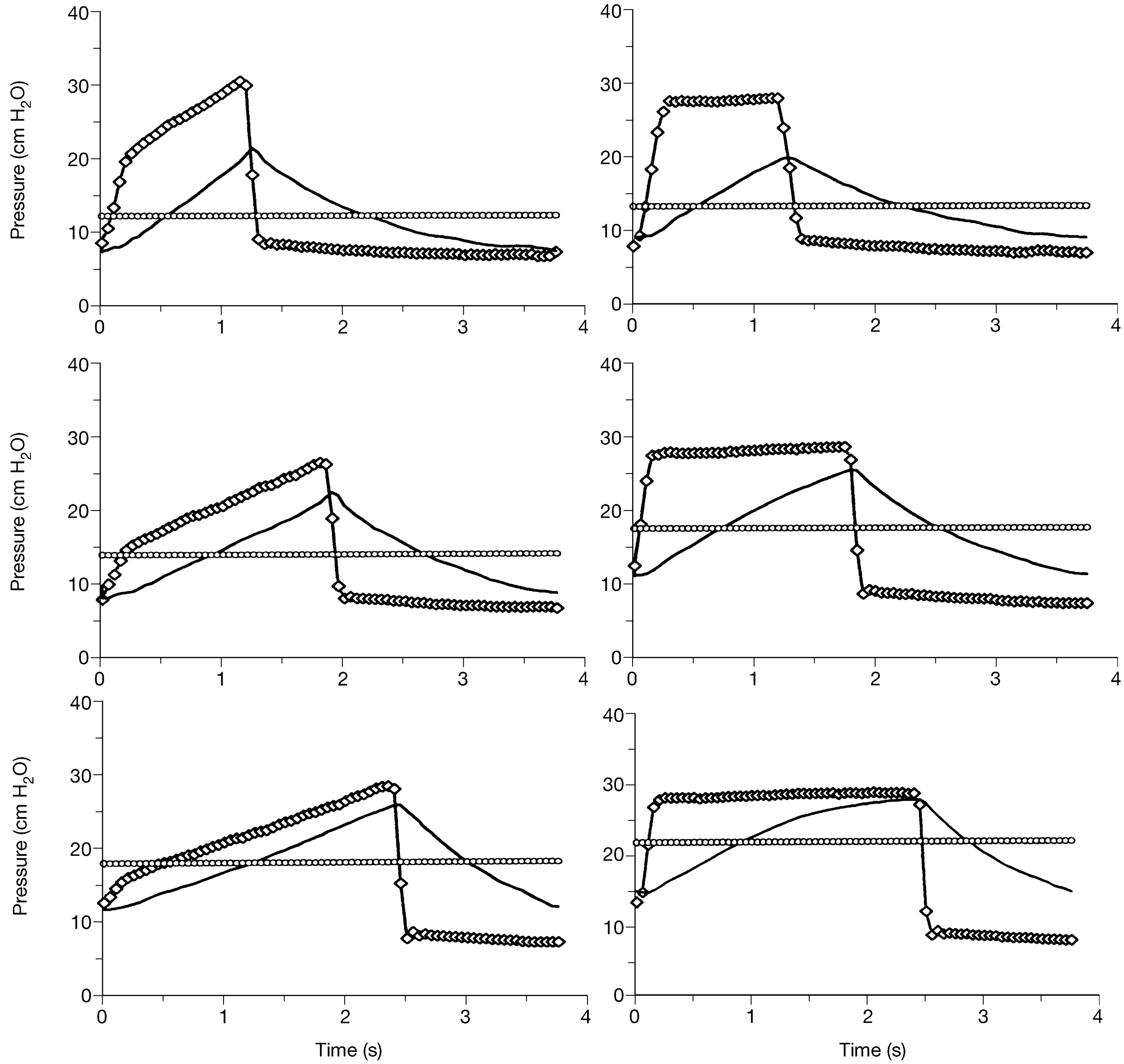

Fig 11 Proximal Y-piece pressure (diamonds), dynostatic alveolar pressure (continuous line) and mean alveolar pressure (circles). Alveolar pressures are calculated using the dynostatic algorithm. The I:E ratio is 1:2, 1:1 and 2:1 in the upper, middle and lower panels respectively. Left column shows volume-controlled ventilation and right panel shows pressure-controlled ventilation. Note the marked difference in alveolar pressure and the pressure normally measured above the tube at the Y-piece or in the ventilator. During inspiration, the alveolar pressure is markedly less than the proximal pressure. The difference between the two pressures is constant in volume-controlled ventilation, but in pressure-controlled ventilation the difference decreases throughout inspiration because the flow is decreasing. Mean alveolar pressure at the same tidal volume and I:E ratio is greater in pressure-controlled than in volume-controlled ventilation because alveolar pressure increases more rapidly.

in choosing the best ventilator setting and allow adjustment with changes in the respiratory mechanics of the patient.

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