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Infant lung function testing in the intensive care unit

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Abbreviations ARDS adult respira-
tory distress syndrome ·
BPD bronchopulmonary dysplasia ·
Crs total respiratory system compli-
ance · ETT endotracheal tube ·
FD forced deflation · FV flow-vol-
ume · FRC functional residual
capacity · MEF maximal expiratory
flow · PEEP positive endexpiratory

pressure · Rrs total respiratory
system resistance · RSV respiratory
syncytial virus · RV residual
volume · TAA thoracoabdominal
asynchrony · TLC total lung
capacity · UAO upper airway
obstruction · VC vital capacity ·
 V_T tidal volume

Introduction

Until recently, there has been limited ability to assess objectively the deviations from normal in lung function in infants and children, either within or without the Intensive Care Unit (ICU). However, the application of the rapid data acquisition and processing abilities of the personal computer and miniaturization of equipment has led to major changes in this field, particularly for the newborn-4-year-old child age range. The modifications of old and development of new pulmonary function tests have allowed those involved in the care of such patients novel measurements and new perspectives in both the assessment and management of respiratory failure.

This review will focus on techniques which are used to measure thoracoabdominal asynchrony, tidal breathing flow-volume loops, small airway function (forced expiratory maneuvers), respiratory mechanics and lung volumes in critically ill infants and children. The major goals of these techniques in an Intensive Care setting are to:

- i) understand the underlying pathophysiology,
- ii) aid with diagnosis,
- iii) provide assessment of therapeutic response, and
- iv) provide a guide to changes in a patient's condition which will allow timely interventions to support the patient. Ideally, such devices should also provide a measure

of the disease and a prediction of outcome, in addition to being inexpensive and noninvasive. However, at this point in the evolution of Intensive Care, our most useful monitors, diagnosticians and medical prognosticators are almost certainly highly trained and "disease-smart" physicians, nurses and respiratory therapists at the bedside.

Some applications of pulmonary function testing generally require that the infants are heavily sedated, and under neuromuscular blockade when mechanically assisted with ventilation. The latter situation also requires (usually) that there be no leak around the ETT in order to obtain good quality studies. This can be achieved in most cases with either a cuffed ETT or with an uncuffed tube with pharyngeal packing. Cuffed ETT are not recommended for use in children under the age of 8 years [1], but in a recent prospective study involving 250 infants and children, using cuffed ETT one-half size smaller than the calculated uncuffed ETT for age, we had no greater incidence of complications either short-term (post-extubation stridor) or long-term (tracheal stenosis) [2]. We now use cuffed ETT routinely in our Pediatric ICU in infants and young children with pulmonary disease.

It is not the intent of this report to justify as "useful" or "essential" any of the techniques which are subsequently mentioned. At one end of the spectrum, demonstration of a significant reduction of mortality and morbidity associated with a change in practice provides a dra-

matic justification for the use of a diagnostic or monitoring tool. Unfortunately, unlike therapeutic interventions, it is rarely possible to provide evidence of such changes with routine monitoring devices in Intensive Care. In addition, little has been published on the efficacy or cost effectiveness of the various procedures. At the other end of the spectrum, one can argue that any intervention which provides more understanding of disease processes can be justified, providing there is little or no deleterious effect on the patient.

Non-invasive respiratory monitoring

Intermittent arterial blood gas analysis is regarded as the standard diagnostic tool for respiratory failure and is fundamental for accurate assessment of pulmonary gas exchange and ventilator management [3]. However, technologic advances have made non-invasive devices available for continuous monitoring of oxygen (O_2) and carbon dioxide (CO_2). These include end-tidal CO_2 , transcutaneous CO_2 and O_2 , and pulse oximetry. If the caregiver is aware of their limitations, they allow quick feedback on rapidly changing conditions and are helpful in the continuous supervision of respiratory therapy. Non-invasive blood gas monitoring has the potential to reduce significantly the frequency of ABG sampling. Detailed descriptions of their operation and limitations are beyond the scope of this paper, but are available in recent reviews [4, 5].

Thoracoabdominal asynchrony (TAA) and paradoxical breathing are often observed in infants and children with various forms of respiratory diseases including upper airway obstruction (UAO), parenchymal processes (such as hyaline membrane disease, pneumonia and pulmonary edema), obstructive lower airways disease (asthma, bronchiolitis, BPD) and neuromuscular diseases. This phenomenon has generally been descriptive and was referred to as chest wall retractions in clinical scoring systems. However, phase angle analysis of the Lissajous figure allows us to easily detect, quantify and monitor TAA in a non-invasive manner [6]. In this technique, rib cage (RC) and abdominal (ABD) movements are recorded by use of an uncalibrated respiratory inductance plethysmograph, the bands of which are placed at the levels of the nipples and upper abdomen. The analog output of the RC and ABD movements is acquired by a computerized data acquisition system that is programmed to calculate continuously phase angles utilizing the method of Agostoni and Mognoni [7]. Phase angle (θ) is thus calculated according to the equation:

$$\sin \theta = \frac{m}{s}$$

where m is the length of the midpoint of the RC excursion and s is the length depicting the ABD excursion (Fig. 1). In addition, RC and ABD movements can be continuously displayed as an X-Y plot giving optical information about changes in loop shape and loop direction.

Except during REM sleep, the RC and ABD expand and decrease in synchrony in normal full-term infants and children, producing a closed or very narrow loop with a positive slope on the X-Y plot (mean $\theta = 8^\circ$, range = 0 to 25°). However, during TAA the loop opens and becomes progressively wider as TAA increases. Paradoxical breathing also creates a closed or very narrow loop, but with a negative slope. Important information can further be obtained from the loop direction. This indicates which compartment (RC or ABD) precedes the other. Counterclockwise loops indicate that the ABD compartment (diaphragm) leads the RC as usually observed in normal quiet breathing and most forms of respiratory distress in children. Clockwise loops signify the opposite which is typically associated with diaphragmatic paralysis [8].

Continuous phase angle measurement is a promising non-invasive technique for the objective assessment of TAA in a variety of respiratory diseases. Phase angles are elevated in UAO and decrease after α -agonist therapy (Fig. 2), but demand further research and a better understanding of their predictive value for respiratory failure [6]. We have recently validated phase angle measurements by showing that they correlate with the level of the imposed respiratory load, but do not detect respiratory muscle fatigue, in UAO [9]. However, it has been demonstrated that phase angle measurements correlate with transcutaneous CO_2 in infants with severe laryngotracheobronchitis and provide a useful, additional way to

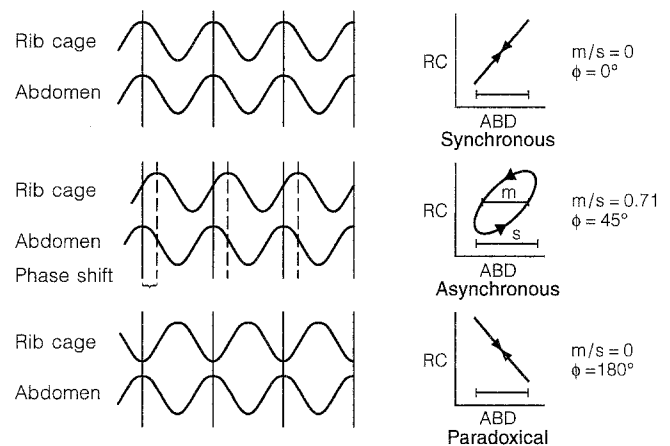


Fig. 1 On the *left hand side* of the diagram, the RC and ABD movements are drawn as sine waves to show temporal relationships as they move out of phase (*X-axis* represents time). On the *right hand side*, the movements are plotted as Lissajous figures to demonstrate the calculation of corresponding phase angles and also to show the direction of the loop (e.g. anti-clockwise in second panel, where abdominal movement precedes thoracic – as in group)

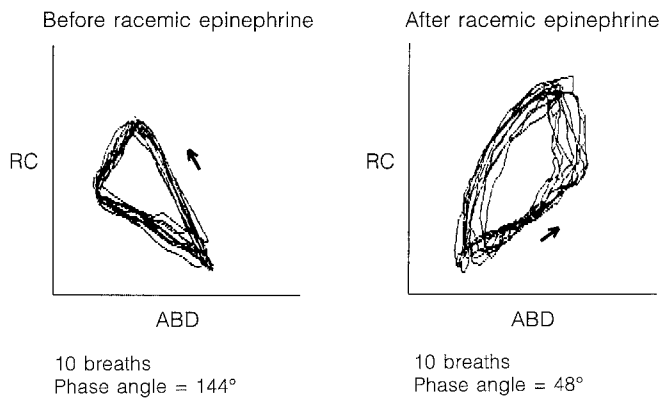


Fig. 2 Phase angle analysis and Lissajous figures of an infant with croup before and after the application of racemic epinephrine. Lissajous figures are anti-clockwise (as indicated by the *arrow*) and confirm thoracic and abdominal motion are not sinusoidal. The change (improvement) in phase angle ($144^{\circ} \rightarrow 48^{\circ}$) was associated with a decrease in clinical signs of UAO and an increase in V_T from 3 to 5.4 ml/kg

monitor such patients [10]. Phase angle measurements also correlated with improvement after bronchodilator therapy in children with obstructive airway disease such as asthma and BPD [11]. The use of phase angle analysis to monitor TAA in infants and children with neuromuscular disease (e.g. infant botulism, Guillain-Barré syndrome, myopathies, neuropathies, spinal cord injuries) needs to be further clarified, but offers a promising tool to monitor sprinting or weaning processes from mechanical ventilation. Diaphragmatic paralysis can be easily detected at the bedside by the characteristic generation of clockwise loops, even when this may not be obvious clinically as in the case of unilateral paralysis (e.g. post cardiothoracic surgery) [8].

A problem often encountered with this technique is that at various times phase angle loops are not based on clear sinusoidal RC and ABD movements and produce numerous types of non-sinusoidal patterns [12]. We have recently suggested that a sine-wave independent mathematical approach in loop analysis improves the accuracy of phase shift calculations under such circumstances [13, 14].

Other methods used in the assessment of respiration involve techniques measuring the “work of breathing”. The classic method for “work of breathing” has been thoroughly studied in adults by Collett and co-workers [15]. However, modified approaches have been used in infants. The “work of breathing” can be measured relatively non-invasively by use of an oesophageal balloon to record pleural pressure changes and subsequent calculation of the pressure-time index. This index is an estimate of the energy cost of the “work of breathing” because O_2 consumption by muscle is proportional to the integral of muscle tension (or pressure) with respect to time. Klein

and Reynolds demonstrated that when the unintegrated pleural pressure signal was used in the index, they were able to show a response to therapy with continuous inflating pressure in sleep-related UAO [16]. Since breathing slows and inspiratory pressure is greater with UAO, the “raw” pressure-time index underestimates the true integrated pressure-time index. Nonetheless, this seems a simple and effective objective measurement for upper airway obstruction. A further variation was suggested by Wolfson et al. [17] in their study on helium breathing in infants with BPD. These techniques await further validation.

Flow-volume loops

Tidal breathing flow-volume loops

Tidal breathing flow-volume or pressure-volume loops are increasingly used in continuous or intermittent monitoring of mechanical ventilation in both Neonatal and Pediatric ICUs. They can be measured on special “stand-alone” equipment, or are increasingly incorporated into modern ventilators. These allow the measurement of tidal volume (V_T), tidal flows and pressures generated during mechanical and spontaneous breaths. Thus, impacts of alteration in ventilator settings or lung physiology on these parameters can be readily detected [18, 19]. Spontaneous tidal volume breaths and ventilator breaths can be compared (especially along the expiratory flow limb where flow-limitation may be readily seen) which enables an estimate of the ventilatory reserve provided by mechanical ventilation (Fig. 3). The generation of inadver-

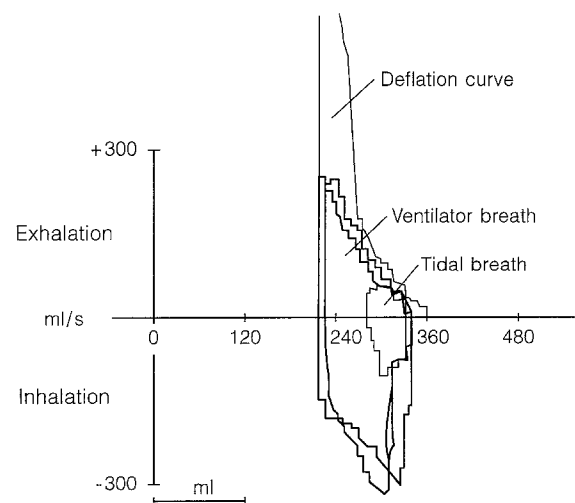


Fig. 3 Superimposed tidal breath, ventilator breath and forced deflation FV-loops of a 9.5 month old infant with BPD illustrate the lack of ventilatory reserve. In this patient airway obstruction is so severe that flow limitation (proven by forced deflation) occurs even during tidal breathing

tent PEEP with high respiratory rates or short expiratory times can be graphically displayed and assessed. Loop patterns are useful to detect various types of airway pathology or partially obstructed endotracheal tubes [20].

Forced expiratory flow-volume loops

The measurement of maximal expiratory flows (MEF) is a sensitive test of abnormalities in the tracheobronchial tree and contributes greatly to the diagnosis and treatment of lung disease. The forced deflation (FD) technique allows examination of forced expiratory flow-volume loops (FV-loops) and the measurement of maximal expiratory flows (MEF) and vital capacity (VC) in intubated infants and children who are unable to generate a voluntary maximal expiratory maneuver [21]. However, it is important that techniques measuring MEF rates reach flow limitation for precise interpretation of the acquired data. Unless it is certain that flow is limited (effort independent) at a particular lung volume, changes in flow rates after a therapeutic maneuver (e.g. bronchodilator) may still be attributed to factors other than a simple response to therapy. We have recently shown that the FD technique is capable of producing forced expiratory flows at flow limitation in intubated animals and infants with normal and obstructed airways [22, 23].

For the test procedure, the lungs are inflated by squeezing a breathing bag filled from a continuous compressed O₂ supply to +40 cmH₂O inflation pressure, defined as total lung capacity (TLC). Inflation pressures are held static for at least 3 s, after which a sliding valve is activated to expose the airways to a 100-L capacity, constant negative pressure source of -40 cmH₂O deflation pressure. The lungs are deflated until expiratory flow ceases at residual volume (RV) or for at least 3 s. VC and MEF at various subdivisions are measured by an interposed pneumotachograph. Throughout the procedure the individual is usually under neuromuscular blockade and/or heavy sedation.

Normal values for VC and MEF at the various subdivisions still need to be defined, but in our laboratory lie in the range of 50–70 ml·kg⁻¹ for VC and 24–38 ml·kg⁻¹·s⁻¹ for MEF₂₅ and 6–15 ml·kg⁻¹·s⁻¹ for MEF₁₀. Since it has become standard of care to use inhaled bronchodilators on intubated and ventilated patients in a variety of diseases, their effectiveness with respect to bronchodilatation can easily be documented by the FD technique [24, 25]. Obstructive airway and restrictive lung diseases produce very characteristic patterns (Fig. 4) which are helpful in assessing the underlying pathophysiology (e.g. RSV infection causing bronchiolitis, ARDS or pneumonia). Serial VC and MEF assessments are helpful in a variety of lung diseases like BPD and ARDS requiring long term mechanical ventilation and document the resolution or progression of the disease process [26, 27].

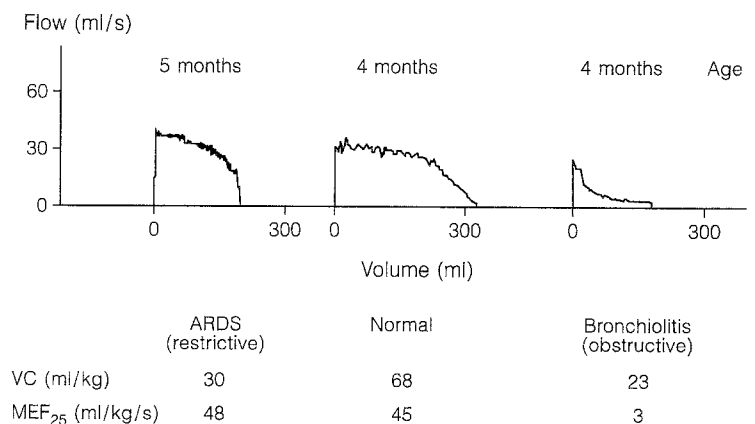
Respiratory mechanics

Compliance and resistance reflect the mechanical properties of the lungs and require the measurement of flow, volume and pressure. Compliance (C) is defined as the change in volume per unit change in pressure:

$$C = \frac{\Delta \text{ volume}}{\Delta \text{ pressure}} = \frac{\text{ml}}{\text{cmH}_2\text{O}}$$

It must be emphasized that compliance is a function not only of the elastic properties of the respiratory system, but also of its volume. In other words, the value obtained is different at various lung volumes, dependent on the shape of the pressure-volume curve, which in turn depends on the amount of lung disease and therapeutic maneuvers such as PEEP or surfactant administration. Sudden changes in compliance often reflect the opening and closing of individual lung units rather than changes in lung tissue and surface tension characteristics [28]. Thus, ideally compliance should be corrected for total lung capacity (TLC) and body weight [29]. Compliance refer-

Fig. 4 Typical forced deflation FV-curves from TLC (+40 cmH₂O) to RV (-40 cmH₂O). Each curve generated from a child with cuffed 3.5 mm I.D. ETT under neuromuscular blockade



enced to functional residual capacity (FRC) is termed specific compliance.

Resistance (R) is calculated from the equation:

$$R = \frac{\text{pressure}}{\text{flow}} = \frac{\text{cmH}_2\text{O}}{\text{ml/s}}$$

and represents the resistive properties of the airways, lung tissue and chest wall.

Several methods have been designed to measure compliance and resistance in ventilated infants which has led to a confusing nomenclature for the practitioner. Compliance is referred to as either dynamic compliance (C_{dyn}) when it is measured when ventilation is in motion, or as static (passive) compliance (C_{rs}) when respiratory muscles are inactive during the test procedure. The same applies to the resistance of the respiratory system, which is referred to as either dynamic (R_{e}) or total respiratory system resistance (R_{rs}). C_{dyn} can be simply calculated by dividing V_{T} by the total change in pressure necessary to deliver that volume. These numbers can be easily extracted from mechanical ventilation. However, it is understood that C_{dyn} is related to both elastic and flow resistive characteristics according to the equation of motion of the single compartment model of the respiratory system:

$$P = \frac{V_{\text{T}}}{C} + RF$$

where P = transpulmonary pressure, V_{T} = tidal volume and F = tidal flow. Thus, C_{dyn} changes with alteration of mechanical ventilation settings including respiratory frequency, inspiratory and end-expiratory pressure [30].

The classic technique of determining C_{dyn} is based on the measurement of oesophageal pressure as a quantification of pleural pressure [31]. This allows differentiation of C_{dyn} into its components of lung compliance (C_{L}) and chest wall compliance (C_{CW}). However, C_{CW} is usually very high in infants and its contribution to total respiratory compliance (C_{TOT}) can often be neglected [32, 33], since C_{L} and C_{CW} are related as follows:

$$\frac{1}{C_{\text{TOT}}} = \frac{1}{C_{\text{L}}} + \frac{1}{C_{\text{CW}}}$$

This technique is invasive by virtue of the need of an oesophageal catheter and the accuracy of such measurements in intubated infants and children is controversial [34, 35].

Newer methods measure static compliance (C_{rs}) and resistance (R_{rs}) and are based on relaxation of both inspiratory and expiratory muscles during brief airway occlusions during exhalation. The most widely used methods are the passive deflation and the multiple occlusion techniques [36–38]. Muscle relaxation is achieved either by invoking the Hering-Breuer inflation reflex or by use of neuromuscular blockade. We favor the use of short-term neuromuscular blockade together with sedation for

a mechanically ventilated patient in the controlled setting of an ICU because it guarantees complete muscle relaxation during the whole expiratory phase.

In the following we will concentrate on the discussion of the passive deflation technique (single breath occlusion) and refer the reader to recent literature for the other methods [39]. The passive deflation technique involves measuring pressure during occlusion of the airway at end-inspiration and fitting a straight line to the FV-curve obtained during the subsequent passive exhalation [40, 41]. If there is no muscle activity during exhalation, the expiratory time constant (Tr_{s}) or emptying time of the respiratory system will be entirely dependent on the mechanical properties of the lungs and can be described as follows:

$$\text{Tr}_{\text{s}} = C_{\text{rs}} \times R_{\text{rs}}$$

Thus, both C_{rs} and R_{rs} can be obtained from a single breath. The determination of Tr_{s} gives some idea of how rapidly the lung empties following a mechanical breath. A single time constant is defined as the time required to exhale 63% of the tidal volume. Three time constants are needed to exhale 95% of the delivered tidal volume. This permits the determination of respiratory rates allowing complete exhalation or the detection of rate settings which lead to inadvertent PEEP.

The passive deflation technique relies on the assumption that the respiratory system can be regarded as a single compartment model. This is valid in most healthy infants especially over the tidal volume range. However, in the presence of lung disease, the respiratory system will not always behave like a single compartment model and a single time constant will not adequately describe all the respiratory mechanics [42, 43]. We have noted multiple time constants in infants with restrictive lung diseases such as acute ARDS or pulmonary edema, or with severe obstructive airway disease [44]. In all these circumstances C_{rs} and R_{rs} are best measured over the longest linear fit of the passive expiratory FV-curve. However, calculation of time constants at different intercepts may give additional information and better describe the respiratory mechanics over the whole expiration phase [45]. Pattern recognition adds valuable information to the interpretation of results obtained by measuring respiratory mechanics. While obstructive lung disease is characterized by a concave slope of the passive expiratory FV-loop, restrictive lung disease often results in convex loop patterns. It is important to note that in the case of intubated patients, C_{rs} and R_{rs} measurements include the physical properties of the ETT. Unfortunately, there is still a great lack of normal values for C_{rs} and R_{rs} in intubated infants and children with normal lungs. According to our studies, such normal data lie in the range of $0.8\text{--}1.2 \text{ ml} \cdot \text{cmH}_2\text{O}^{-1} \cdot \text{kg}^{-1}$ for C_{rs} and $0.4\text{--}0.8 \text{ ml} \cdot \text{cmH}_2\text{O}^{-1} \cdot \text{s}$ (up to 1.0 with $\text{ETT} < 3.5 \text{ mm I.D.}$) for R_{rs} [25].

Pfenninger and Aebi [46] have recently used the passive deflation technique for R_{rs} and C_{rs} to compare

the response to inhaled and intravenous salbutamol in ventilator-dependent infants with chronic lung disease, and concluded there was no difference. Such careful objective physiological measurements should influence how weaning such infants from the ventilator is approached.

Functional residual capacity

Although the most fundamental interest in lung volume measurements in infancy and childhood relates to the assessment of normal and abnormal lung growth [47, 48], the determination of lung volumes is an important part of the respiratory management of infants and children [49]. Lung volume measurements can help in diagnosing respiratory disorders, in evaluating responses to therapy, and in finding suitable ventilator settings with respect to rate and ventilating pressures [50–52]. Lung volume is also an important variable when lung mechanics are measured [53] because specific compliance and specific resistance are normalized by lung volume, i.e. the functional residual capacity (FRC).

Currently, FRC and V_T are the only lung volumes that can be accurately, repeatedly, and reliably measured in infants and small children. Hence, they are the only lung volumes that can be routinely determined for clinical reasons either in the ICU or in the Out-Patient Clinic. Other lung volumes such as TLC, VC and RV can also be measured, but the techniques are employed mainly for research, and require an endotracheal tube.

FRC can be measured by three techniques: plethysmographic (infant body box), helium (He) dilution (a closed-circuit method), and nitrogen (N_2) washout (in its modern form, an open-circuit method). Except in the smallest of infants, sedation is required for each technique. Sulphahexafluoride has recently been used in a promising washin-washout technique [54, 55], and has now been validated [66].

Infant plethysmography

The body plethysmograph technique is labor-intensive, as is the calibration of the infant body box. Edberg and colleagues [56–58] have recently applied modern computing techniques to the infant body box and obtained useful data in the neonatal ICU setting. However, their lung volume data was measured using the nitrogen (N_2) washout method. For most workers, the use of plethysmographic methods is impractical in the ICU environment, particularly for mechanically ventilated patients. The recent advent of a commercially available computerized infant box (Jäger) may modify this opinion.

Helium (He) dilution

The closed-circuit He dilution method has been adapted to measure FRC on ventilated patients by Heldt et al. [59]. The patient is connected via the ETT and a sliding valve to both a bag (which is situated inside a transparent plexiglas box) and to the ventilator. In normal pre-test position the patient is ventilated directly by the ventilator through the valve. The bag, which contains a known amount of gas with known He concentration (and thus a known amount of Helium) is sealed and is not connected to the patient. At end-exhalation the valve is switched so that the patient is directly connected only to the He-containing bag while the ventilator ventilates the box surrounding the bag and compresses the bag accordingly. The patient is thus ventilated by the bag which is externally compressed by the ventilator cycle. After several breaths, equilibration of He concentration between the lungs and the bag is achieved and FRC can be calculated in the same way as in non-ventilated subjects. Helium dilution can be used in patients on very high inspired oxygen concentrations ($FIO_2 = 0.97$). However, most thermal conductivity based He analyzers are inaccurate when O_2 concentrations are high. Moreover, calibration depends on the O_2/He mixture ratio and should be repeated each time. Leak-free connections in intubated infants are more difficult to achieve in this age group, where cuffed tubes are rarely used. If the leak is minimal, it may be eliminated by gentle tracheal pressure during the recording period. Although a method for correcting leaks during FRC_{He} measurements has been described by Fox et al. [60] and is currently incorporated into some automated systems. This has not been fully evaluated nor validated and may result in significant errors.

Nitrogen (N_2) washout

The technique is based on washing out the N_2 from the lungs by giving the subject 100% O_2 to breathe. If the amount of N_2 washed out is measured and the initial alveolar N_2 concentration is known, then the lung volume from which point the washout started can be derived. In this open circuit method, the patient is switched to breathing 100% O_2 and from this point the volume of N_2 exhaled is determined by integration with respect to time of the instantaneous N_2 concentration flowing in the exhalation circuit multiplied by the instantaneous flow. In 1985, Gerhardt and co-workers [61] devised a new open washout system to which a constant background O_2 flow was delivered. The patient inhaled from and exhaled to that circuit with background flow. Although the instantaneous flow rate of the washout circuit changes continuously as the subject breathes, the average flow leaving the system over time remains unchanged because the volume of gas subtracted during inspiration is added back to the

system during exhalation (this is true as long as the temperature and humidity of the inhaled and exhaled gas are equal – a condition which is easy to meet by using a humidifier). Because the method ignored the instantaneous change in flow and used only the average constant flow for calculation, it was essential that sampling of N_2 for concentration measurements would “see” a continuous decrease of N_2 concentration as the washout proceeds, without the effect of the respiratory phase. This was achieved by incorporating a mixing chamber in the exhalation circuit before the sampling port from which mixed expired gas was sampled for N_2 analysis. The technique developed by Gerhardt et al. for spontaneously breathing infants [62] is not immediately applicable to ventilated children mainly because the gas flow during calibration does not equal the flow during the test. In order to overcome this difficulty, Sivan and co-workers [63] used the respiratory mass spectrometer already “in-line” for measuring the instantaneous N_2 concentration, to record the minute ventilation by the argon dilution technique [64]. At FRC the patient is switched to a second ventilator delivering 100% O_2 (washout ventilator) and washout starts.

This N_2 washout technique (which unlike the He dilution method is limited to patients at $FIO_2 < 0.65$) allows accurate determination of FRC during mechanical ventilation and correlates well with those values produced using the Douglas bag technique [52]. In patients with restrictive lung disease, including a group with ARDS [44], FRC measured at clinically chosen levels of PEEP (4–10 cmH_2O) was 45% below predicted FRC for nonintubated normal children and 60% below that of ventilated children with normal lungs at physiological levels of PEEP (2–4 cmH_2O). The use of progressively greater levels of PEEP produced increases in FRC towards predicted normal values [63]. However, this suggests that in ARDS at least, normalization of FRC would require sufficient PEEP to contribute to barotrauma or to compromise cardiac output and systemic oxygen transport. In spontaneously breathing infants and children, FRC is the same whether determined by He dilution or N_2 washout methods [65], and is in the range of 16–22 $ml \cdot kg^{-1}$ (mean = 20.4 $ml \cdot kg^{-1}$). The only published data on ventilated infants and children with normal lungs demonstrated FRCs up to 50% more than the nor-

mal values for spontaneously breathing (i.e. not ventilated) children on PEEPs of 2–4 $cm H_2O$ [63].

Summary

As a result of the previous shortage of tools to assess objectively the overall physiological status of the respiratory system in infants and young children, it has been difficult to measure the degree of physiological disorder or the response to therapy in respiratory diseases such as BPD, the pediatric version of ARDS, bronchiolitis, pneumonia, asthma and croup in this patient population.

The newborn – four-year old child is particularly difficult to study because of their lack of cooperation and size. The recent progress in computer technology made pulmonary function testing available for this age range and opened up new possibilities for monitoring changes in disease processes affecting the respiratory system. This may improve medical management of infants and children with lung and heart diseases in particular.

In 1989, Shannon [49] proposed in this Journal that the minimum physiological information needed for the intelligent use of mechanical ventilation (particularly if lower airway and/or pulmonary parenchymal disease was apparent) required the measurement of at least 4 variables: i) arterial partial pressure of carbon dioxide; ii) arterial oxygen saturation; iii) the mechanical time constant of the lung and iv) FRC. In many circumstances, arterial CO_2 is approximated by alveolar (end-tidal) CO_2 and the arterial oxygen saturation is obtained from pulse oximetry accurately if perfusion is adequate. The mechanical time constant and FRC are easily measured by the techniques described above and together provide important information concerning appropriate ventilator settings for a given disease.

The described techniques bring new insights and awareness, but also new responsibilities in the management of infants and children with respiratory compromise. Not all of these techniques need to be applied to all infants in the ICU. Not all the assumptions upon which some of the techniques we have described are based will prove true. Any such methods which do not withstand solid scientific testing must be quickly discarded and replaced with better and (hopefully) easier methods.

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