Assessment of respiratory mechanics can play a central role in the management of critically ill patients undergoing artificial ventilation because of acute respiratory failure (ARF) [1]. This is a condition defined by a rapid deterioration in pulmonary gas exchange that may be due either to alterations in the mechanical properties of the respiratory system leading to ventilation–perfusion mismatching or shunt, or to neuromuscular insufficiency causing alveolar hypoventilation.

Assessment of respiratory function and mechanics is of crucial importance:
- to understand the pathophysiology of the disease underlying ARF;
- to assess the status and progress of the disease;
- to provide guidelines for therapeutic measures (positive end-expiratory pressure, bronchodilators, fluids);
- to improve patient–ventilator interaction;
- to prevent ventilator-related complications;
- to plan the discontinuation of mechanical ventilation.

Despite the great importance of monitoring lung mechanics in ventilator-dependent patients, these measurements are not regularly performed. This is probably due to the general prejudice that these measurements are difficult to obtain in the intensive care unit (ICU).

This chapter will be focused on respiratory mechanics rather than on general respiratory physiology reflecting the attitude and expertise of the authors.

The purpose of this chapter is to review briefly the most common methods and techniques for measuring and monitoring respiratory mechanics at the bedside of the patient in the ICU.

We will go through a three point analysis: 1) measurements during controlled mechanical ventilation, i.e. in the relaxed, passive patients; 2) evaluation of respiratory mechanics during assisted mechanical ventilation; 3) the issue of the patient’s evaluation in the weaning process from mechanical ventilation.

Before discussing the monitoring techniques, it is necessary to provide a brief overview of the mechanical properties of the respiratory system.

The act of breathing is performed against several impediments: elastic, resistive, viscoelastic, plasticelastic, inertial and gravitational forces, compressibility of intrathoracic gas, and distortion of the chest wall from its relaxed configuration. Despite its apparent complexity, the dynamics of breathing have been satisfactorily represented, at least for clinical purposes, by a single-compartment model consisting of a rigid tube and a compliant balloon [2, 3].
The equation of motion for inspiration during spontaneous breathing is the following:

\[ P_{\text{mus}} = P_{\text{el,rs}} + P_{\text{res}} + (P_{\text{in}}) + \text{PEEPi} \] (1)

\[ P_{\text{el,rs}} = E_{rs} \times \Delta V \] (2)

\[ P_{\text{res}} = R_{\text{tot}} \times V' \] (3)

\( P_{\text{mus}} \) represents the pressure developed by the inspiratory muscles, \( P_{\text{el,rs}} \) and \( P_{\text{res}} \) are the pressure dissipated respectively to overcome the opposing elastic and resistive forces, \( P_{\text{in}} \) is the inertial force, PEEPi is the intrinsic positive end-expiratory pressure, \( i.e. \) the elastic pressure that is present if inspiration does not begin from the relaxed functional residual capacity, \( E_{rs} \) is the elastance of the respiratory system, \( R_{\text{tot}} \) is the total resistance of the respiratory system (plus the endotracheal tube, if present); \( \Delta V \) is the inspired volume and \( V' \) the inspiratory flow.

Since \( P_{\text{in}} \) is normally negligible it can be removed from the equation:

\[ P_{\text{mus}} = (E_{rs} \times \Delta V) + (R_{\text{tot}} \times V') + \text{PEEPi} \] (4)

During assisted ventilation the equation changes to:

\[ P_{\text{mus}} - P_{\text{vent}} = (E_{rs} \times \Delta V) + (R_{\text{tot}} \times V') + \text{PEEPi} \] (5)

Whereas during controlled mechanical ventilation (CMV), when the patient is relaxed, the \( P_{\text{mus}} \) is 0 and all the pressure is developed by the ventilator:

\[ P_{\text{vent}} = (E_{rs} \times \Delta V) + (R_{\text{tot}} \times V') + \text{PEEP}_{\text{tot}} \] (6)

Where PEEP_{tot} is the sum of PEEPi plus the external PEEP applied by the ventilator.

### Controlled mechanical ventilation

Measurements of respiratory function during CMV can be performed using different techniques. In the following it will be assumed that the patient is a relaxed, passive patient, \( i.e. \) a patient without a significant spontaneous respiratory activity. Monitoring during CMV can be performed off-line or on-line (see table 1).

### Monitoring off-line:

**Interrupter technique.** The most widely used technique for assessing respiratory mechanics in patients during CMV is the rapid airway occlusion technique. This procedure, albeit introduced at the beginning of the last century, gained popularity in the
last 10–15 yrs after a series of studies that have elucidated the theoretical aspects of the technique as well as its physiological basis [4, 5]. This technique requires either ventilators equipped with special software options or a specific "button" to control the inspiratory and expiratory valves or additional equipment (i.e. pneumotachograph, pressure transducer, occlusion valve) inserted in line to the ventilator circuit. When applied at the end of expiration (end expiratory occlusion, EEO, fig. 1) it provides a measure of the static intrinsic positive end expiratory pressure (PEEPi,st), also known as auto-PEEP.

In patients with PEEPi flow stops abruptly before the next mechanical inflation, producing a characteristic "truncated" appearance on the expiratory flow curve. In this condition, during CMV, dynamic intrinsic PEEP (PEEPi,dyn) can be calculated by measuring the amount of pressure (airway pressure) that need to be developed by the ventilator to reverse the flow from expiration to inspiration. This can be easily assessed by recording simultaneously flow and airway pressure and averaging many breaths by superimposing flow–pressure loops and looking at the point where pressure tracings cross the zero flow (fig. 2) [6]. PEEPi,dyn represents the lowest regional PEEPi which has to be counterbalanced by the positive pressure of the ventilator to start inspiratory flow [6].

If the rapid airway occlusion is applied just before the end of the inspiration (end inspiratory occlusion, fig. 3) it enables measurement of most respiratory mechanics parameters. As it can be observed in figure 3 the rapid end-inspiratory occlusion is

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**Fig. 1.** – Representative record with measurement of intrinsic positive end-expiratory pressure (PEEPi) by end-expiratory airway occlusion (EEO) in a mechanically ventilated patient during controlled ventilation with constant inspiratory flow. Records of a) flow and b) pressure at the airway opening. Inspiration is upward. At the end of the tidal expiration, the expiratory circuit of the ventilator is occluded and airway pressure becomes positive, reflecting the end-expiratory elastic recoil of the respiratory system due to incomplete expiration. The value of PEEPi is provided by the difference between the EEO airway pressure plateau and atmospheric pressure. Visual detection of the plateau on airway pressure provides direct evidence of absence of leaks in the circuits, respiratory muscle relaxation, and equilibration between alveolar and tracheal pressure. Modified with permission from [6].
characterised by an immediate drop in airway pressure from a peak ($P_{\text{peak}}$) to a lower value ($P_1$), followed by a slow decay to an apparent plateau achieved after 3–5 s [7].

From these manoeuvres it is possible to compute static elastance ($E_{\text{st}}$) and interrupter...
resistance ($R_{\text{int}}$), as well as $R_{\text{tot}}$, according to the following equations:

$$E_s = \frac{P_{\text{plat}} - P_{\text{PEEP},s_t}}{V_T}$$  \hspace{1cm} (7)

$$R_{\text{int}} = \frac{P_{\text{peak}} - P_1}{V'}$$  \hspace{1cm} (8)

$$R_{\text{tot}} = \frac{P_{\text{peak}} - P_{\text{plat}}}{V'}$$  \hspace{1cm} (9)

Where $V_T$ is tidal volume and $V'$ the flow immediately preceding the occlusion. Calculations should be based on at least three manoeuvres. From the values of $R_{\text{tot}}$ and $R_{\text{int}}$ the resistance of the endotracheal tube need to be subtracted to obtain the $R_{\text{tot},\text{rs}}$ and $R_{\text{int},\text{rs}}$. The resistances of the endotracheal tubes are markedly curvilinear and can be computed from the following equation:

$$R = K_1 V' + K_2 V'^2$$  \hspace{1cm} (10)

Where $K_1$ (laminar flow) and $K_2$ (turbulent flow) are two constants and can be directly measured "in vivo" or obtained from published data computed "in vitro" and $V'$ is the flow.

**Relaxed expiration.** Measurement of respiratory mechanics during relaxed expiration may be useful, being potentially less influenced than inspiration by settings on the ventilator. When the patient is sedated or completely relaxed the expiration is mainly governed by the mechanical properties of the respiratory system. If expiration continues below the tidal end-expiratory volume, dynamic pulmonary hyperinflation exists and can be computed from the difference in volume between the end of relaxed expiration and the end-expiratory volume of the preceding breath.

Normal subjects and patients with increased elastance show a smooth decrease in expiratory flow throughout expiration, whereas patients with airflow limitation exhibit a curvilinear pattern, convex to the volume axis [8], often associated with a sudden drop of flow, when there is dynamic hyperinflation.

Expiratory flow limitation can be confirmed by the lack of change in flow with modification in pressure at the airway opening or in the added flow resistance or with application of a negative expiratory pressure (NEP). First, external resistance may be increased [9] or decreased [10], for example, by adding an expiratory resistance or removing the expiratory circuit of the ventilator. Secondly, pressure at the airway opening may be increased [11] by setting PEEP by the ventilator or decreased with the negative expiratory pressure technique [12].

**Monitoring on-line**

Despite the great importance of monitoring lung mechanics in ventilator-dependent patients, the measurements previously illustrated are not continuous [13]. The rapid airway occlusion interferes with the ventilator settings, requires a valve or a specific "button" on the ventilator, and thus is not suitable for continuous monitoring. Continuous monitoring enables the early detection of changes in patient’s status, thus allowing a rapid therapeutic response, as well as the evaluation of its effectiveness. This requires noninvasive, breath-by-breath monitoring and the microprocessor-based
ventilator have shown the potential for continuous monitoring of respiratory mechanics in ventilator-treated patients [14, 15].

The monitoring systems currently used rely on estimation techniques that are not up to date: isovolume method for calculating resistance [16], measurements of dynamic compliance at points of zero flow [11], and measurement of dynamic PEEPi [17]. All these techniques are based on the assumption of the first-order model of respiratory mechanics, i.e. on a linear model, while resistance and elastance are known to be volume and airflow dependent.

Tracking respiratory parameters in time is possible using these mathematical models of breathing mechanics and recursive estimation techniques [15, 16]. A parsimonious model is needed because the performance of recursive methods for real-time identification sharply deteriorates with increasing model complexity [18]. This leads to the selection of the first-order viscoelastic model for on-line monitoring of breathing mechanics. However, the parameters are allowed to change during the breath to provide a better description of the data, thus accounting for the non-linear behaviour of respiratory mechanics during artificial ventilation. The algorithm used in the literature is the recursive least square (RLS), which has been recently modified to account for the non-linear behaviour of respiratory mechanics during artificial ventilation [19, 20]. Briefly, the current authors have adopted an RLS algorithm, combined with the classical first-order model of respiratory mechanics and the continuous measurement of airflow and airway pressure, to quantify respiratory mechanics in real time [6]. The method constructs, from the recursive parameter estimates during inspiration, a weighted mean and standard deviation of dynamic resistance, elastance and PEEPi. The mean values are updated on a cycle-by-cycle basis to allow real-time monitoring of these clinical indexes in ventilator-dependent patients with acute respiratory failure of different origins, including chronic obstructive pulmonary disease (COPD).

Recently, Volta et al. [21] applied a different method using the least square fitting with the first order model keeping resistance and compliance constant over the whole breathing cycle. They applied the method in patients with and without expiratory flow limitation. They concluded that data weighted on inspiration were acceptable in both patient populations.

We can conclude that the proposed technique, although it has been limited so far to patients without any respiratory activity, during controlled mechanical ventilation, is a simple and robust tool for clinical use in the routine of the ICU.

**Assisted mechanical ventilation**

*Work of breathing*

To achieve normal ventilation, work needs to be performed to overcome the elastic and frictional resistances of the lungs and chest wall. Determinants of work of breathing (WOB) in ventilator-dependent patients are shown in table 2.

During controlled mechanical ventilation the external WOB is performed by the ventilator and can be easily computed from the area subtended by the inflation volume and applied pressure (either airway or tracheal). Changes in WOB during CMV reflect changes in the mechanical properties of the respiratory system (provided that the ventilator setting is unchanged).

During assisted mechanical ventilation due to the activity of the inspiratory muscles it is necessary to use the oesophageal balloon to compute the WOB.

During assist-control ventilation, as proposed by Marini and coworkers [22, 23] and
WARD et al. [24], it is possible to measure noninvasively the WOB done by the patient by comparing the pressure–volume (or pressure–time) relationship with that during CMV. This technique of measuring WOB assumes the unproved supposition that the mechanical properties of the respiratory system remain essentially identical during "passive" and "active" conditions. Difference in WOB may be used to compare the effects of the ventilator settings, response to therapy, such as bronchodilators [25, 26].

Measurement of WOB may help in deciding the appropriate level or type of ventilator assistance and avoid both excessive or insufficient support. However, the superiority of the measurement of WOB, over more simple measurements (such as peak or plateau airway pressure, PEEPi, etc.), has not been proved. Although used in research, this procedure does not enjoy wide popularity in clinical practice because of its "invasiveness", apparent complexity and the difficulty of its interpretation.

### Pressure time product

A significant limitation of measurements of respiratory work is that it may underestimate the energy expenditure of the respiratory muscles during isometric contraction. To overcome this problem many authors have suggested the use of the pressure time product for the respiratory muscles. This requires the placement of an oesophageal balloon and is calculated as the time integral of the difference between oesophageal pressure ($P_{oes}$) measured during assisted ventilation and the recoil pressure of the chest wall measured during passive mechanical ventilation with $V_T$ and flow setting identical to the assisted breaths. This can be easily performed during assist/control ventilation, but not during other ventilatory modalities, such as pressure support ventilation, when there is a great variability in $V_T$ and flow. This can be overcome using the method proposed by JUBRAN et al. [27]. The recoil pressure of the chest wall ($P_{el,cw}$) can be computed by multiplying the chest wall elastance, measured during passive ventilation, by the volume signal. Then the pressure time product is calculated as the time integral of the difference between $P_{oes}$ and $P_{el,cw}$. Another obstacle is the possibility that the rapid drop in $P_{oes}$ with the beginning of inspiration is not due to the activity of the inspiratory muscles but instead to the cessation of the activity of the expiratory muscles. This can lead to an overestimation of the pressure time product. This can be avoided by also placing a gastric balloon and measuring the transdiaphragmatic pressure ($P_{di}$), i.e. the differential pressure between gastric and oesophageal pressure. The time integral of $P_{di}$ measures only the respiratory effort of one inspiratory muscle, the diaphragm (i.e. the major inspiratory muscle), but does not need any correction for expiratory muscles activity or the chest wall elastic recoil [28, 29].

<table>
<thead>
<tr>
<th>Table 2. – Determinants of work of breathing in ventilator-dependent patients</th>
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<tr>
<td>Patient’s abnormal mechanics</td>
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<tr>
<td>Low compliance</td>
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<td>High flow resistance</td>
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<td>PEEPi</td>
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| Diameter of the endotracheal tube  |
| Ventilator circuit, valves and devices  |
| Ventilatory pattern  |
| $V_T$ and $V_E$  |

**Inspiratory flow rate and waveform.**

PEEPi: intrinsic positive end expiratory pressure; $V_T$: tidal volume; $V_E$: minute ventilation.
Weaning from mechanical ventilation

Airway occlusion pressure

The decrease in airway pressure 0.1 s after initiating an inspiratory effort against an occluded airway \( (P_{0.1}) \) has commonly been used as an index of neuromuscular ventilatory drive [30–32]. This measure is not a great task to be performed in the ICU and requires only a pressure transducer, a two-way valve and a data acquisition system. This measurement is also automatically performed by some modern ventilators without any direct physician’s intervention. Furthermore, during assisted ventilation, in some ventilators (with a trigger delay around 100 msec) \( P_{0.1} \) can be measured breath by breath by reading the pressure developed by the patient during the inspiratory effort needed to trigger the mechanical breath [33, 34].

The value of \( P_{0.1} \) depends not only on the inspiratory centre output and the intact neural pathway to the inspiratory muscles, but also to the electro-mechanical coupling and the efficiency of the muscles. If an elevated \( P_{0.1} \) is present it undoubtedly indicates an increase in the activity of the respiratory centre, but a low value may be related not only to a reduced centre activity, but also to an impairment in one of the factors mentioned above, particularly in the pressure generating capacity of the inspiratory muscles due to weakness or fatigue. Furthermore, if PEEPi is present, i.e. there is dynamic hyperinflation, the \( P_{0.1} \) measured at the airway opening did not take in consideration the extra-pressure required to overcome PEEPi. \( P_{0.1} \) is usually increased in patients with acute respiratory failure.

Monitoring of \( P_{0.1} \) in mechanically ventilated patients may help to identify patients with high probability of successful weaning (\( P_{0.1} \leq 4 \text{ cmH}_2\text{O} \)) among those with high risk of weaning failure (\( P_{0.1} \geq 6 \text{ cmH}_2\text{O} \)) [35, 36], or may help to adequately set the level of ventilatory support [37, 38].

Maximal inspiratory pressure

The inability of the respiratory muscles to sustain spontaneous ventilation is the primary indication for the main therapeutic modalities in the ICU, namely mechanical ventilation [39]. The balance between the maximal inspiratory muscle strength and the mechanical load is the major determinant of the ability to sustain indefinite alveolar ventilation. Respiratory muscle performance is one of the major issues in deciding the timing and pace with which mechanical ventilation can be discontinued [40, 41].

Measurement of inspiratory muscle strength can be assessed by measuring maximum inspiratory airway pressure (\( P_{I,\text{max}} \)) while the patients makes a maximum inspiratory effort against an occluded airway [42]. This manoeuvre requires the patient’s cooperation and coordination. This is not an easy task in ventilator-dependent, acutely ill patients. To obtain more reproducible measurements in ventilator-dependent patients, MARINI et al. [43] suggested the use of a unidirectional valve to permit exhalation while inhalation is blocked. This permits exhalation proximal to residual volume, where \( P_{I,\text{max}} \) is expected to be higher. The highest values of \( P_{I,\text{max}} \) are usually reached after 15–20 s of occlusion [43]. Despite this manoeuvre values of \( P_{I,\text{max}} \) measured in critically ill patients are usually underestimated and show a poor reproducibility [44].

Breathing pattern

Abnormalities in respiratory frequency (\( f \)) and \( V_T \) are common in acutely ill patients admitted to the ICU. \( V_T \), \( f \) and minute ventilation are easy to measure in intubated
patients, whereas they are more difficult to obtain in spontaneously breathing subjects
due to the low tolerance of a face mask or mouthpiece, and moreover during noninvasive
ventilation due to the leaks always present. All modern ventilators have a "monitoring
area" in which breathing pattern parameters are continuously displayed. However, to
ensure the accuracy and reliability of volume measurements it is preferable to have a
simple handheld spirometer.

Rapid shallow breathing is a frequent finding in critically ill patients and it has been
related to respiratory muscles fatigue by some [45, 46] but not all [47, 48] investigators.
An elevated frequency often is the earliest sign of impending respiratory distress, and the
degree of elevation is proportional to the severity of the underlying lung disease [49].
Breathing pattern is commonly monitored during the titration of ventilatory support. In
particular \( f \) and \( V_T \) are used to identify the appropriate level of pressure support [50–54].

**Conclusions**

Respiratory function testing, either off- or on-line, due to its simplicity, its
noninvasiveness and clinical usefulness should be a common practice in the critically
ill patients. Its use may help to adjust ventilator settings, medical treatment and assist the
clinician in the weaning process.

**Summary**

Assessment of respiratory mechanics can play a central role in the management of
critically ill patients undergoing artificial ventilation because of acute respiratory
failure (ARF). This assessment is of crucial importance to understand the
pathophysiology of the disease underlying ARF and to improve the patient–ventilator
interaction and the medical treatment of the disease.

Despite the great importance of monitoring lung mechanics in ventilator-dependent
patients, these measurements are not regularly performed.

The purpose of this chapter is to review briefly the most common methods and
techniques for measuring and monitoring respiratory mechanics on-line and off-line at
the bedside of the patient in the intensive care unit (ICU) and to be persuasive about
the usefulness and the feasibility of monitoring respiratory mechanics in the congested
rooms of the ICU. The chapter includes a three point analysis: 1) measurements
during controlled mechanical ventilation, *i.e.* in the relaxed, passive patients; 2)
evaluation of respiratory mechanics during assisted mechanical ventilation; 3) the
issue of the patient’s evaluation in the weaning process from mechanical ventilation.

**Keywords:** Acute respiratory failure, intensive care unit, mechanical ventilation,
monitoring, respiratory mechanics.

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