
Optoelectronic Plethysmography: Principles of Measurements and Recent Use in Respiratory Medicine

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11.1 Introduction

Although the measurement of pulmonary ventilation by a spirometer or a pneumotachograph may appear to be a simple procedure, it is much more complicated than most realize. Temperature, humidity, pressure, viscosity, and density of gas influence the recording of its volume. Mouthpieces, face masks, and noseclips may introduce leaks and therefore cause losses, are impractical for prolonged measurement, limit the subject's mobility, introduce additional dead space, and thereby increase tidal volume. They also make the subject aware that his breathing is being measured and therefore interfere with the natural pattern of breathing and its neural control [1, 2]. Breathing through a mouthpiece and flowmeter or from a spirometer is extremely difficult in children or uncooperative adults; it cannot be used during sleep, to analyze phonation, and during weaning from mechanical ventilation may require excessive patient cooperation. During exercise, rebreathing from a spirometer or a bag-in-box system can only be done for short time periods, while integration of flow at the mouth suffers from integration drift, so that changes in absolute lung volume are not accurately recorded. A possible approach to solve this problem is to collect the expired gas, breath by breath, in a large spirometer (e.g., a Tissot spirometer) or in a large, gas-tight bag (e.g., a Douglas bag), which are then emptied through a precision gasometer. But even emptying the spirometer or the bag causes problems due to the gasometer, which may require intermittent calibration over time.

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11.1.1 Measurements of Chest Wall Motion

All these problems have induced investigators to attempt to measure ventilation indirectly by external measurement of chest wall surface motion [3]. The chest wall is defined by all the anatomical structures surrounding the lung and moving with it: the rib cage, diaphragm, abdominal content, and abdominal wall. Displacements of the lung are transmitted to the chest wall and vice versa, and, therefore, measurements of thoracoabdominal surface movement can be used to estimate lung volume variations. In the last decades, a number of devices and methods have been developed in order to allow measurements of the rib cage and abdominal motion, and, in parallel, several attempts have been made to define calibration methods able to estimate volume changes of the single compartments of the entire chest wall and of the lung from measurements of diameters, circumferences, or cross-sectional areas, such as the isovolume method [4], changing posture [5], natural breathing [6]. The validity of the calibration coefficients obtained experimentally to convert one or two dimensions to volume is generally limited to the estimation of tidal volume under conditions matched to those during which the calibration was performed. Numerous devices based on sensing belts positioned on the rib cage and abdomen or wearable garments embedding different kinds of sensors have been proposed. The sensor technology used in sensing belts can be quite different and includes mechanical transducers, such as capacitive elastic strain gauges [7] and piezoelectric films [8], ultrasound waves in rubber tubes [9], and optical fibers [10]. Respiratory inductive plethysmography (RIP) allows to measure changes of rib cage and abdominal cross-sectional areas, by two coils of insulated wire sewn inside elastic bands which are usually placed below the axillary line and above the umbilicus [11]. Variations in the self-inductance of the coil are proportional to the cross-sectional area enclosed by the coil, and, therefore, it varies as the rib cage and the abdomen expand and contract during respiration.

A variety of optical techniques using multiple video cameras combined with either light projected on the chest surface or reflective markers positioned on it have been proposed to track the changing shape of the thoracoabdominal surface during breathing and from this to calculate the enclosed volume. Optical methods based on structured light to analyze chest wall movement during breathing have been pioneered by Peacock et al. [12] and Saumarez [13], who proposed a technique for mapping the size and shape of the thoracoabdominal wall by projecting a grid on sheets of light creating contour lines on the visible surface of the torso, recording them by still or video cameras and reconstructing the shape from digital information. These systems, however, remained confined in few research applications. More recent advances, including color structured light systems [14], are nowadays opening new perspectives for the development of more automatic procedures to process the data and to obtain chest wall surface movement and volume variations during breathing. These systems are still in their development phase, however.

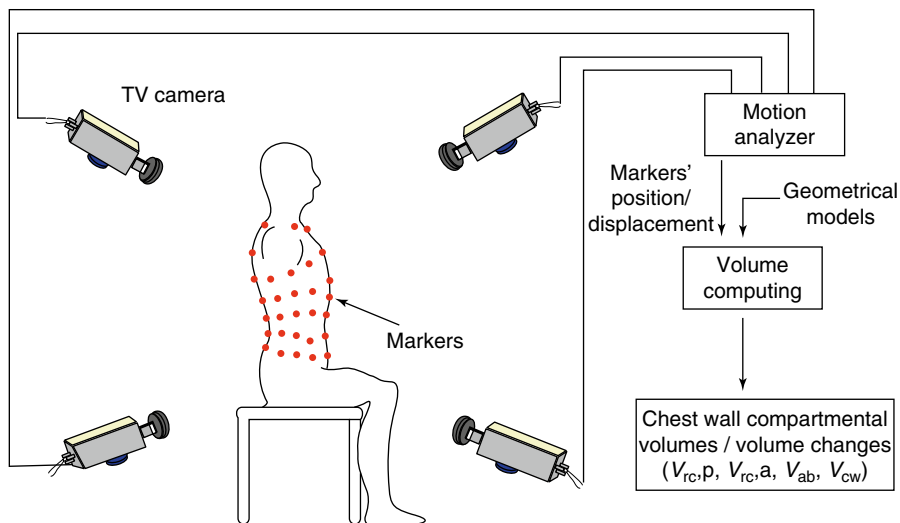


Fig. 11.1 Optoelectronic plethysmography (OEP): principle of measurement

11.1.2 Optoelectronic Plethysmography

Optoelectronic plethysmography (Fig. 11.1) is a now-established technique, based on typical methods for optical motion analysis, that allows to measure the variations of the volume of the chest wall and its compartments during breathing [15, 16]. A number of reflective markers are positioned by a hypoallergenic tape on the trunk of the subject in selected anatomical reference sites of the rib cage and the abdomen.

A set of cameras is placed nearby the subject under analysis. Each camera is equipped with an illuminator (infrared light-emitting diodes) that determines a high contrast between the reflective marker and the rest of the scene on the recorded image, thus allowing the fully automatic recognition of the markers. When a single marker is seen by two or more cameras, its position (defined by the three-dimensional coordinates in the reference system of the laboratory) can be calculated by stereophotogrammetry, being known the position, orientation, and the internal parameters of each camera. Once the 3D coordinates (X , Y , Z) of the points belonging to the chest wall surface are acquired with reference to an arbitrary coordinate system (Fig. 11.2), a closed surface is defined by connecting the points to form triangles (mesh of triangles) (Fig. 11.3). For each triangle, the area (A_i) and the direction of the normal of the plane defined by that triangle are determined. Successively, the internal volume of the shape is computed using Gauss' theorem (or divergence theorem, or Green's theorem in space).

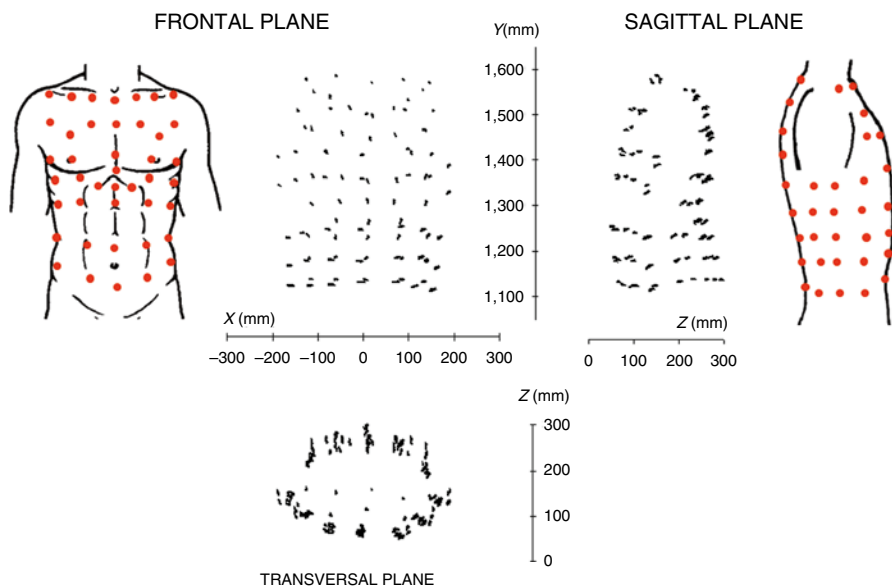


Fig. 11.2 Trajectories in the frontal (X - Y axes), sagittal (Y - Z axes), and transversal (X - Z axes) planes of the markers placed on the chest wall surface (during quiet spontaneous breathing). From markers' displacement, the variations of the enclosed volume is computed (see Fig. 11.3)

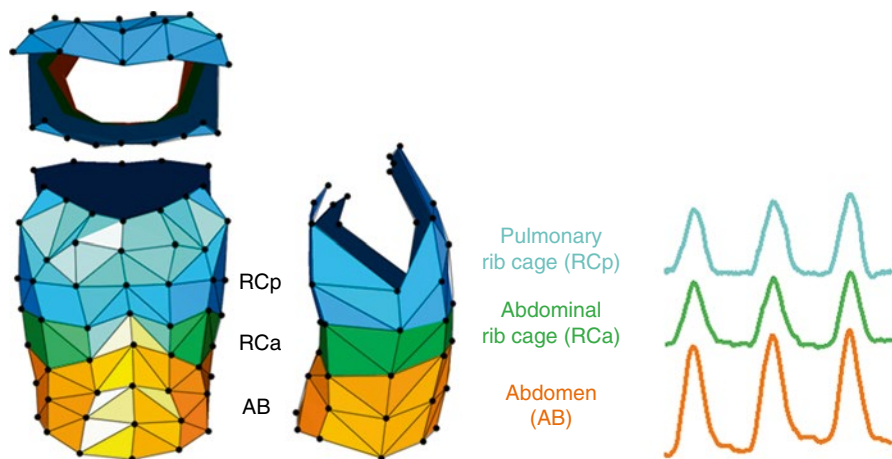


Fig. 11.3 Geometrical models of the three chest wall compartments: pulmonary rib cage (RCp), abdominal rib cage (RCa), and abdomen (AB) (left, three views) and their volume changes during quiet spontaneous breathing, respectively, $V_{rc,p}$; $V_{rc,a}$; and V_{ab} . Chest wall volume (V_{cw}) is equal to $V_{rc,p} + V_{rc,a} + V_{ab}$

Let S_{cw} be the closed chest wall surface S_{cw} enclosing the volume V_{cw} , and let \vec{F} be a vector field defined at every point of V_{cw} then

$$\int_{S_{cw}} \vec{F} \cdot \vec{n} dS_{cw} = \int_{V_{cw}} \nabla \vec{F} dV_{cw} \quad (11.1)$$

where:

S_{cw} is the chest wall surface

V_{cw} is the volume enclosed by S_{cw}

\vec{F} is an arbitrary vector

\vec{n} is the outward-pointing unit normal vector at the different points of S_{cw}

∇ is the divergence operator

If we choose an arbitrary vector with a unit divergence, Eq. 11.1 becomes

$$\int_{S_{cw}} \vec{F} \cdot \vec{n} dS_{cw} = \int_{V_{cw}} dV_{cw} = V_{cw} \quad (11.2)$$

and the volume integral is computed by means of an easier surface integral.

Passing from continuous to discrete form, Eq. 11.2 becomes

$$\sum_{i=1}^K \vec{F} \cdot \vec{n}_i A_i = V_{cw} \quad (11.3)$$

where:

K is the total number of the triangles

A_i is the area of the i th triangle

\vec{n} is the normal unit vector of the i th triangle

This procedure allows the direct computation of the volume enclosed by the thoracoabdominal surface approximated by a closed mesh of triangles.

11.2 Measurement of Chest Wall Compartmental Volumes

The markers are positioned on approximately horizontal rows at the levels of the clavicular line, the manubriosternal joint, the nipples, the xiphoid process, the lower costal margin, the umbilicus, and the anterior superior iliac crest [15]. Surface landmarks for the vertical lines are the midlines, both anterior and posterior axillary lines, the midpoint of the interval between the midline and the anterior axillary line, the midpoint of the interval between the midline and the posterior axillary line, and the midaxillary lines. Extra markers are added bilaterally at the midpoint between the xiphoid and the most lateral portion of the 10th rib and in corresponding posterior positions.

Markers' positioning is designed to allow an adequate sampling of the complex shape of the thoracoabdominal surface and an adequate subdivision of total chest

Fig. 11.4 Three-compartment chest wall model schematic diagram of the respiratory system, where different pressures of interest are shown: P_{ao} pressure at the airway opening, P_{alv} alveolar pressure, P_{pl} pleural pressure, P_{ab} abdominal pressure, P_{bs} pressure at the body surface. The different functional respiratory muscle groups displace the different chest wall compartments

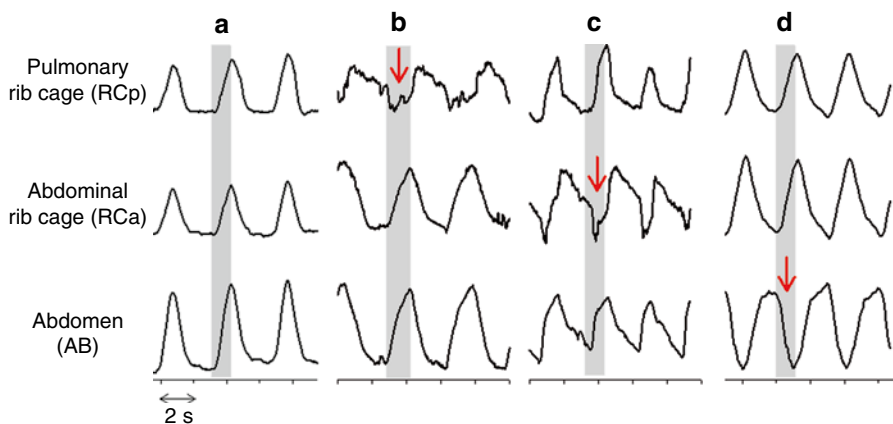
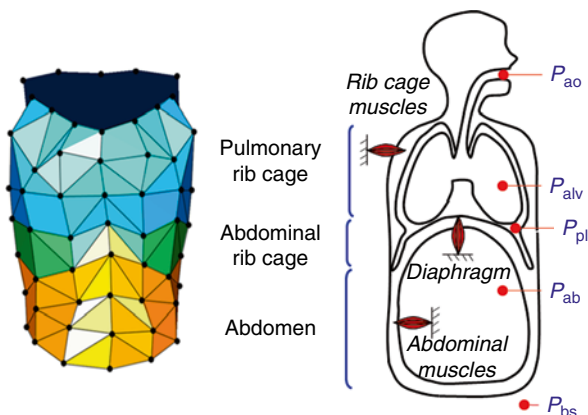


Fig. 11.5 Representative examples of volume variations of the pulmonary rib cage (*RCp*), abdominal rib cage (*RCa*), and abdomen (*AB*) during quiet breathing. **(a)** Normal healthy subject. **(b)** Patient with type III osteogenesis imperfecta, presenting inspiratory paradoxical inward motion of the *RCp*. **(c)** Patient with chronic obstructive pulmonary disease (COPD), presenting inspiratory paradoxical inward motion of the *RCa*. **(d)** Patient with late-onset type II glycogenosis, presenting inspiratory paradoxical inward motion of the *AB* due to diaphragmatic severe weakness. *Gray area*: inspiration. Paradoxical motion is shown by the *red arrows*

wall volume into different compartments. The geometrical models of the compartments that have been developed for OEP volume measurements follow the three-compartment model, composed by the pulmonary rib cage (*RCp*), the abdominal rib cage (*RCa*), and the abdomen (*AB*) [15–18]. The rib cage is separated from the abdomen by the line of markers placed on the lower costal margin. The subdivision of the rib cage into *RCp* and *RCa* is defined by the transverse section at the level of the xiphoid [19]. Precisely, the surface that encloses *RCp* extends from the clavicles to the line of markers extending transversely at the level of the xiphisternum, while *RCa* extends from this line to the lower costal margin. *AB* extends caudally from the lower costal margin to the level of the anterior superior iliac crest.

The three-compartment model of the chest wall takes into consideration that the lung- and diaphragm-apposed parts of the rib cage (RCp and RCa, respectively) are exposed to substantially different pressures on their inner surface during inspiration (pleural pressure and abdominal pressure, respectively), that the diaphragm acts directly only on RCa and not on RCp, and that non-diaphragmatic inspiratory muscles (i.e., external intercostal, parasternals, scalene, neck muscles) act largely on RCp and not on RCa. Abdominal volume change is defined as the volume swept by the abdominal wall, as described by Konno and Mead [20], and it is the result of the action of the diaphragm and expiratory abdominal muscles (internal and external obliques, rectus, and transversus) (Fig. 11.4). As these three compartments are exposed to different pressures and to the action of different respiratory muscle groups, their movement is in principle independent. This is shown in Fig. 11.5. While in healthy subjects the three compartments generally move synchronously, in different diseases, different asynchronous movement between the three chest wall compartments can be found, typically as paradoxical inward motion in either the pulmonary rib cage (e.g., in osteogenesis imperfecta, due to rib cage deformities), the abdominal rib cage (e.g., in chronic obstructive pulmonary disease, COPD, due to lung hyperinflation and the consequent flattened diaphragm), or the abdomen (e.g., in late-onset type II glycogenosis, due to diaphragm paralysis). OEP allows also to split each compartment into its right and left parts, where asynchronies can also be found, e.g., in the presence of diaphragm hemiparalysis or hemiplegia.

11.3 Accuracy of Measurement of Lung Volume Variations

In the last years, different protocols of OEP have been developed for different experimental and clinical situations. The validation of these measurement protocols was always performed by comparing the chest wall volume variation, measured by OEP, with lung volume variations measured by a spirometer or integrating a flow measurement at the airway opening. In the first studies, volume changes were compared in healthy subjects while sitting or standing and wearing 89 markers, during quiet breathing, slow vital capacity maneuvers [15], and incremental exercise on a cycle ergometer [18]. In these conditions, the coefficient of variation of the two signals was always lower than 4 %. Successively, OEP was validated in constrained postures, like the supine and prone position. In these situations, the analysis is performed placing the markers only in the visible part of the trunk surface, while the inferior part is considered fixed with the support (e.g., the bed). The volumes measured using OEP were also compared with measurements taken using spirometry and pneumotachography both in healthy subjects during quiet and deep breathing on rigid and soft supports in supine and prone position [16] and in sedated and paralyzed patients with acute lung injury and acute respiratory distress syndrome while receiving continuous positive pressure ventilation or pressure support ventilation [21]. Tidal volume measurements of OEP, spirometry, and pneumotachography were always highly correlated with discrepancies lower than 5 %.

More recently, simultaneous measurements of tidal volume by OEP and pneumotachography demonstrated that OEP is able to provide accurate measurements of tidal volume values in newborns at rest [22] and in healthy men and women during submaximal (-2.0 ± 7.2 %) and maximal (2.4 ± 3.9 %) cycling exercise [23]. OEP has also recently shown to accurately evaluate vital capacity in the supine position in patients with respiratory muscle dysfunction of variable severity, including those with paradoxical abdominal movements [24]. Intra-rater and inter-rater reliability of OEP was evaluated on at rest and during cycle-ergometer submaximal exercise [25], and results showed intraclass correlation coefficient values higher than 0.75 and coefficient of variation of method error values less than 10 % for most variables in both conditions.

11.4 Double Plethysmography

Although the results obtained in all the above-cited validation studies have shown a very good agreement between chest wall and lung volume variations, it is important to remind that the two measurements are not necessarily equal. Gas compression and dilation and possible blood shifts into and out of the thorax might produce relevant differences between variation of gas and chest volumes, particularly during maneuvers in which intrathoracic pressures change significantly. When, for example, the respiratory system is subjected to large positive or negative pressures (e.g., during mechanical ventilation, during active expiration with occluded airways or in the presence of expiratory flow limitation, during inspiratory efforts with occluded airways), changes in V_{cw} equal changes in lung gas volume (ΔV_L), plus the volume of any blood shifts from the thorax to extremities or vice versa (V_B): $\Delta V_{cw} = \Delta V_L + V_B + \Delta V_L$ is the sum of the volume of gas expired (or inspired) at the mouth (V_M) plus the volume of gas compressed (or dilated) in the lung (V_C) and therefore $\Delta V_{cw} = V_M + V_B + V_C$. Recently, OEP was combined to whole-body plethysmography (WBP) which measures changes in body volume which are equal to $V_M + V_C$. WBP is insensitive to blood shifts, whereas OEP measures the same variables as WBP plus any blood shifts between the trunk and the extremities (Fig. 11.6).

In a series of experiments, simultaneous measurements of OEP and WBP allowed to measure VB continuously with the so-called double plethysmography [26, 27]. It was shown that outflow from the splanchnic blood reservoir is controlled by abdominal pressure, and that during quiet breathing with diaphragm descent, the diaphragm serves two functions, i.e., to ventilate the lung and to shift blood from the splanchnic vascular bed to the extremities. With simultaneous contraction of abdominal muscles, such as occurs during exercise [17], the circulatory function of the diaphragm can be considerably enhanced. Under appropriate circumstances, the diaphragm's circulatory function combined with abdominal muscle contraction can act as an abdominal circulatory pump, capable of acting as an auxiliary heart.

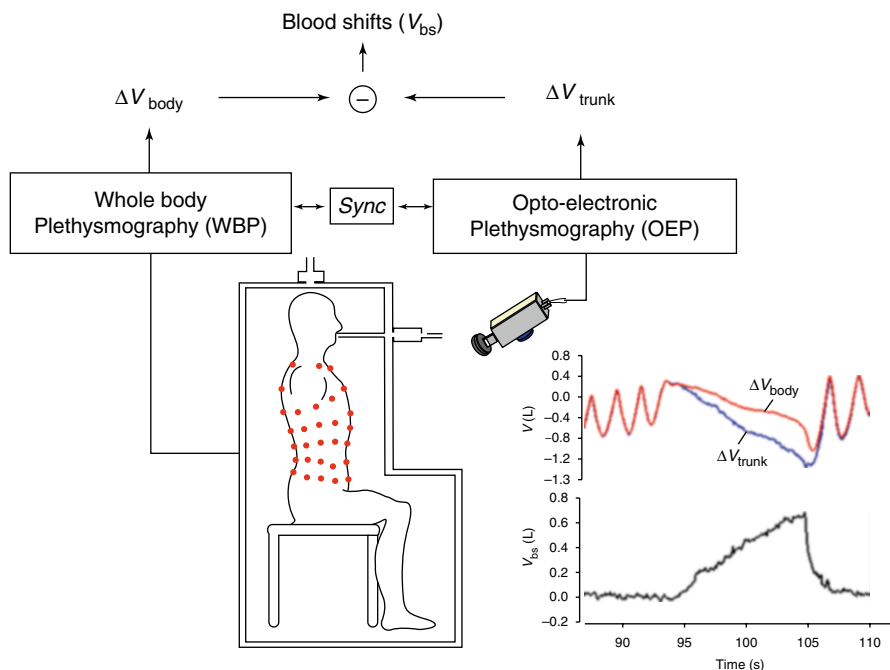


Fig. 11.6 Double plethysmography: principle of measurement (see text). The graphs on the bottom right show simultaneous recordings of total body volume variations (*red line*) and trunk volume variations (*blue line*) (top tracings) during a ramp increase in P_{ab} , while P_{pi} remained unchanged. The volume of blood shifted from the trunk to the extremities (V_{bs}) is shown in the bottom tracing (Modified from Aliverti et al. [26])

11.5 Measurement of End-Expiratory Lung Volume Variations

The measurement of end-expiratory lung volume (EELV) is of scientific and clinical importance in understanding respiratory mechanics. Unfortunately, continuous monitoring of EELV in normals and patients presents technical difficulties. Gas dilution techniques cannot be used for continuous monitoring because of the long wash-in/washout time, while the accuracy of RIP in monitoring EELV has also been questioned. Flowmeters cannot be used for long periods, mainly because integration of flow at the mouth suffers from integrator drift, so that changes in absolute lung volume are not accurately recorded. In fact, when flow is integrated to provide volume, an upward or downward drift in the volume baseline is invariably seen due to both physiological reasons and methodological errors: not unitary respiratory exchange ratio, differences in temperature and gas composition between inspiration and expiration, leaks between the airway opening and pneumotachometer, zero offset in flow calibration, and imperfections

in the pneumotachometer response. In principle, it might be possible to avoid drift in volume by preconditioning the inspired gas to BTPS conditions, continuously monitoring gas partial pressures in both the alveoli and the pulmonary arterial and venous blood to correct for respiratory exchange ratios not equal to unity, and eliminating all the factors mentioned above. However, this is extremely difficult, if not impossible, in practice. Consequently, it is never known how much of the baseline drift in volume is due to drift and how much represents a true change in absolute lung volume. Drift correction algorithms first assess the upward or downward trend in functional residual capacity (FRC) over a period containing many breaths in which the subject is assumed to be in the physiological steady state and the assumption generally made is that FRC remains more or less constant; successively, the algorithm removes the trend. Although OEP cannot provide the absolute lung volume unless the subdivisions of lung volumes are known, it appears to be a suitable method for estimating its variations and measuring breath-by-breath EELV changes, as well as its distribution in the different chest wall compartments. End-expiratory volume variations of the chest wall ($\Delta\text{EEV}_{\text{cw}}$) measured breath by breath by OEP before, during, and after an increase/decrease in positive end-expiratory pressure were compared with the corresponding variations of EEVL measured by the helium (He) dilution technique [28]. The regression line between EEVL changes measured by He and EEV_{cw} changes measured by OEP was very close to identity. OEP measurements of EEV_{cw} accurately reflect the changes of EEVL. Furthermore, OEP allows a continuous compartmental analysis, even during unsteady conditions, and this feature to track end-expiratory and end-inspiratory chest wall volume is extremely useful during incremental exercise as an alternative to serial measurements of inspiratory capacity (IC). This procedure, usually adopted for determining EELV changes during exercise, has inherent problems, namely, (a) the requirement of a high level of patient's collaboration, (b) the assumption that total lung capacity is reached at every maneuver performed by the subject, (c) that the maneuver is started from a volume that is representative of EELV, (d) the assumption that TLC does not change, and (e) the impossibility of track EELV on a breath-by-breath basis. Vogiatzis et al. [29] compared simultaneous measurements of inspiratory capacity (IC) at rest and during incremental cycle exercise in a group of male and female patients with stable COPD. Changes in IC from quiet breathing measured by the spirometer were in good relationship with end-expiratory chest wall volume variations obtained by OEP during exercise and recovery from exercise, with a mean difference between these two measurements throughout all stages equal to 7.0 (5.8)% or 35 (24) ml.

11.5.1 Studies Based on Optoelectronic Plethysmography in Health

In the last decade, a high number of studies employing OEP to study chest wall kinematics in various conditions have been performed, both in health and disease.

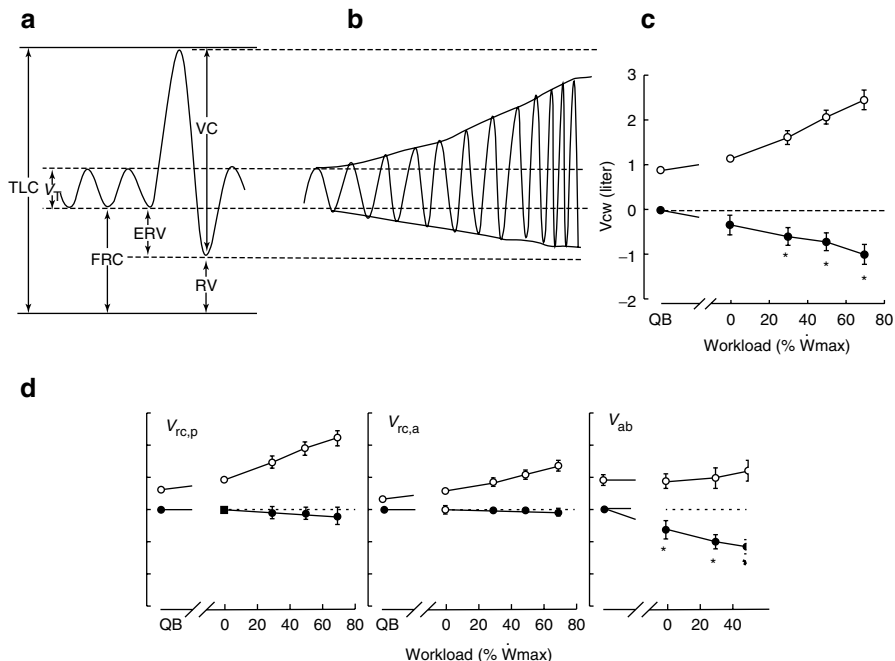


Fig. 11.7 (a) Normal spirogram (TLC total lung capacity, FRC functional residual capacity, ERV expiratory reserve volume, RV residual volume, VC vital capacity, V_T tidal volume). (b) Schematic diagram of the variations of V_T during incremental exercise in healthy subjects. (c) Chest wall volume (V_{cw}) during exercise. The difference between end-inspiratory (open circles) and end-expiratory V_{cw} (closed circles) is V_T . Dashed line, end-expiratory V_{cw} during quiet breathing (QB). (d) End-inspiratory (open symbols) and end-expiratory (closed symbols) mean volumes of the pulmonary rib cage ($V_{rc,p}$), abdominal rib cage ($V_{rc,a}$), and abdomen (V_{ab}) during exercise (Modified from Aliverti et al. [17])

Here below is not provided a comprehensive literature analysis, but only a summary of the main findings achieved so far.

Healthy subjects were analyzed during exercise, both on a cycle ergometer [17, 18, 30, 31] and during walking on a treadmill [32]. In both modalities, end-expiratory lung volume is reduced because of the recruitment of the expiratory muscles, and this reduction increases with the intensity of exercise. During heavy exercise, about one third of the tidal volume is accomplished below FRC , and about 40 % of the increase in tidal volume is attributable to the recruitment of expiratory reserve volume. The reduction in end-expiratory total chest wall volume is almost entirely due to a decrease in end-expiratory volume of the abdomen (Fig. 11.7).

End-expiratory volumes of both rib cage compartments do not change significantly. As the end-inspiratory displacement of the abdomen is nearly constant during exercise, the increase in end-inspiratory lung volume is almost entirely due to rib cage expansion. In other words, during exercise, the increase in rib cage tidal volume results from recruiting only its inspiratory reserve volume, while the

increase in abdominal tidal volume largely resulted from recruiting only its expiratory reserve volume. Despite different recruitment patterns, the relative contributions to tidal volume from the different compartments remain nearly constant, although the tidal volume more than triples during exercise.

In other conditions than exercise, a detailed analysis of total and compartmental chest wall volumes allows to better understand mechanisms underlying different functions provided by the respiratory muscles. In a recent study investigating cough mechanics in health, it was shown how operating total and compartmental chest wall volumes are the most important determinant of the peak flow achieved and volume expelled during coughing [33]. Binazzi et al. [34] studied chest wall kinematics during reading, singing, and whispering. They found that the activity of the control of expiration during phonation is more complex than during exercise and all the three chest wall compartments contribute to the decrease of the total end-expiratory chest wall volume. Similar complex mechanisms of interaction between non-diaphragmatic inspiratory muscles and expiratory muscles are adopted during professional flute playing and allow the control of sound production that results in “breath support” which in turn is associated with high-quality playing [35].

11.5.2 Optoelectronic Plethysmography Studies in Disease

11.5.2.1 Chronic Obstructive Pulmonary Disease (COPD)

OEP has been extensively used to evaluate total and compartmental volume variations during incremental exercise in COPD patients. It was found that the patients with more severe COPD showed dynamic hyperinflation during incremental exercise, but other patients, specifically those with a greater expiratory flow reserve at rest, adopted the more “normal” approach of reducing EEV_{cw} when they exercised [36]. The presence in COPD patients of two distinct groups, those that strongly recruited abdominal muscles and those that did not, was confirmed in a subsequent study [37]. Among the patients who developed DH at maximal exercise, at least two significantly different patterns of change in EEV_{cw} were observed during the test [29, 38]. Some patients had a progressive significant increase in end-expiratory volume of the chest wall (“early hyperinflators”) (Fig. 11.8), while in other patients, this remained unchanged up to about 66 % of maximum workload and increased significantly only in the last third period of exercise (“late hyperinflators”).

The essential difference between euvolumics and hyperinflators is the degree of expiratory muscle recruitment. The kinematic difference, in fact, is in the behavior of the abdomen. During exercise, in some patients with COPD, excessive expiratory muscle recruitment occurs and the high expiratory pressures have adverse circulatory effects, namely, decreased cardiac output and blood shifts from the trunk to extremities. In addition, the oxygen cost of breathing is so high in COPD that it can become a very large percentage of total body oxygen uptake. This can establish competition between respiratory and locomotor muscles for the available oxygen supply at low exercise workloads and this can be a potent factor limiting exercise

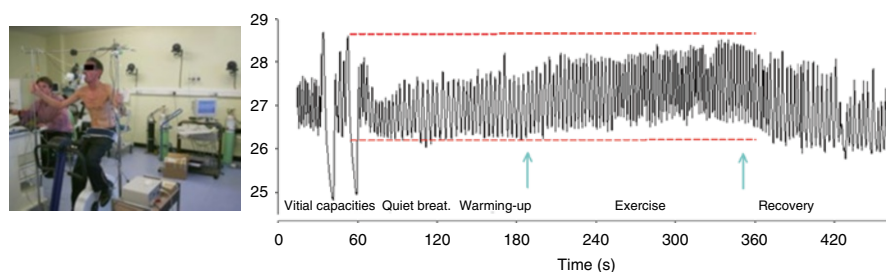


Fig. 11.8 *Left:* a COPD patient analyzed by OEP during exercise on a cycle ergometer at Aintree University Hospital, Liverpool, UK. *Right:* representative tracing showing chest wall volume variations during two vital capacity maneuvers, spontaneous quiet breathing at rest, warming up, incremental exercise, and recovery from exercise. Note dynamic hyperinflation of the chest wall during the period of exercise. *Red dashed* tracings indicate chest wall volume at FRC and TLC

performance in COPD [39]. In a series of complex experiments employing OEP, cardiac output, and peripheral muscle oxygen measurements, it has been recently shown that the heliox breathing during exercise improves peripheral muscle oxygen availability with different mechanisms in hyperinflator and non-hyperinflator COPD patients. In hyperinflators, heliox increases arterial oxygen content and quadriceps blood flow at similar cardiac output, whereas in non-hyperinflators, heliox improves central hemodynamics and increases systemic vascular conductance and quadriceps blood flow at similar arterial oxygen content [40, 41].

It is not yet clear why COPD patients adopt different patterns of end-expiratory rib cage and abdominal volume variations during exercise. Nevertheless, this seems at least partially dependent of the presence of lower ribcage inspiratory inward paradoxical motion that is present at rest in several COPD patients [42, 43]. In a recent study, it was shown that total end-expiratory chest wall volume increased immediately when exercise began (“early hyperinflation”) in patients presenting lower rib cage paradox at rest, but later (“late hyperinflation”) in patients with synchronous rib cage motion [42].

11.5.2.2 Thoracic Surgery

Three different, distinct patterns of breathing and chest wall volume regulations were found in severe patients with COPD, interstitial pulmonary fibrosis (IPF), and cystic fibrosis (CF) adopted by the ventilatory pump to cope with chronic respiratory failure [44]. The same authors demonstrate that after lung transplantation (LTx), the chronic adaptations of the ventilatory pattern to advanced lung diseases are reversible, and that the main contributing factor is the lung itself rather than systemic effects of the disease. Another study performed in patients affected by cystic fibrosis treated with bilateral LTx demonstrated a rearrangement of the volumes of the different compartments, with a significant abdominal volume reduction, and lower rib cage increase, suggesting diaphragm repositioning [45].

Other studies employed OEP to assess the effects of laparoscopic surgery on chest wall kinematics and inspiratory muscle activity [46] and to study the effects of

Nuss surgical technique for pectus excavatum on chest wall function at rest and during exercise [47–49].

11.5.2.3 Neuromuscular Disorders

In Duchenne muscular dystrophy (DMD) patients, abdominal motion during spontaneous breathing in awake conditions and in supine position has been proven to be not only an important indicator of the degree of respiratory muscle impairment and disease progression but also an early indicator of nocturnal hypoxemia [50]. Inefficient cough in DMD, moreover, is associated with reduced operating lung and chest wall volume secondary to weakened inspiratory muscles. Abdominal contribution to tidal volume during spontaneous breathing has been shown to be a non-volitional and noninvasive index able to discriminate efficient and inefficient cough [51]. In adolescent and adult DMD patients who present either no sign or only mild nocturnal oxygen desaturation, a reduced abdominal contribution to inspiratory capacity is a specific marker of the onset of diaphragm weakness and should be considered to identify the correct timing for the institution of nocturnal NIV [52].

OEP, also, revealed mild initial modifications in the respiratory muscles in other muscular dystrophies, namely, limb-girdle muscular dystrophy, Becker muscular dystrophy, and facioscapulohumeral dystrophy, which could be helpful for functional and new therapeutic strategy evaluation [53].

11.5.2.4 Other Diseases

In a study performed on a group of patients with late-onset type II glycogenosis (Pompe disease), it has been shown that the abdominal percentage contribution to tidal volume in supine position well correlates with the postural change of forced vital capacity and therefore represents a noninvasive non-volitional index to detect diaphragmatic weakness in these patients [54]. Lanini et al. studied a group of hemiplegic patients due to a cerebrovascular accident [55]. The expansion of the paretic and healthy sides was similar during quiet breathing, but paretic displacement was higher during hypercapnic stimulation in most patients, suggesting that hemiparetic stroke produces asymmetric ventilation with an increase in carbon dioxide sensitivity and a decrease in voluntary ventilation on the paretic side. More recently, Lima et al. [56] have shown that incentive spirometry is able to promote an increased expansion in all compartments of the chest wall and to reduce the asymmetric expansion between right and left pulmonary rib cage, and, therefore, it should be considered as a tool for rehabilitation. Possible alterations in chest wall kinematics due to obesity have been recently investigated by Barcelar et al. [57]. Compared to controls, obese women are characterized by strongly altered increased abdominal volume variations. No differences were found between central and peripheral obese women suggesting that the amount of fat in the abdominal compartment, and not the peripheral, alters the respiratory system and determines restriction.

In osteogenesis imperfecta, an inherited connective tissue disorder characterized by bone fragility, multiple fractures, and significant chest wall deformities, the

restrictive respiratory pattern is closely related to the severity of the disease and to the sternal deformities. Patients classified as type III patients show structural rib cage deformities (pectus carinatum) that alter respiratory muscles coordination, leading to chest wall and rib cage distortions (i.e., paradoxical inspiratory inward motion of the pulmonary rib cage, see Fig. 11.3) and an inefficient ventilator pattern [58].

11.5.3 OEP Studies in Intensive Care Patients, During Mechanical Ventilation and Anesthesia

As reported above, OEP has been introduced in the ICU and the method has been validated in patients during both pressure support ventilation (PSV) and continuous positive pressure ventilation (CPPV) [21]. Using OEP in the supine position, significant differences in the distribution of tidal volume in the different chest wall compartments between normal subjects and mechanically ventilated patients and between patients receiving CPPV and patients receiving PSV were observed. For spontaneously breathing subjects, distribution of chest wall volume changes is dictated by the mechanical characteristics of the respiratory system and the relative activity of the diaphragm and inspiratory rib cage muscles. For paralyzed CPPV patients, only the mechanical characteristics of the system are involved. Conversely, during PSV, a part of the breathing pattern is controlled by the patient, and the situation is more complex. The synchronization of respiratory muscle action and the resulting chest wall kinematics are valid indicator of the patient's adaptation to the ventilator.

In a study performed on nine patients with acute lung injury/acute respiratory distress syndrome [59], it was shown that at pressure support levels lower than 15 cmH₂O, all the following parameters increased: the pressure developed by the inspiratory muscles, the contribution of the rib cage compartment to the total tidal volume, the phase shift between the rib cage and abdominal compartments, the post-inspiratory action of the inspiratory rib cage muscles, and the expiratory muscle activity. PSV, therefore, should not be considered a "unique form" of ventilation, because its effects may be quite different depending on the pressure support level and pressure support levels greater than 10 cmH₂O are necessary to allow homogeneous recruitment of respiratory muscles, with resulting synchronous thoracoabdominal expansion.

OEP was also recently used to study the effects of anesthesia on chest wall motion during spontaneous breathing and positive pressure ventilation [60] in a group of subjects undergoing elective surgery requiring general anesthesia (Fig. 11.9). Chest wall volumes were continuously monitored by optoelectronic plethysmography during quiet breathing (QB) in the conscious state, induction of propofol anesthesia, spontaneous breathing during anesthesia (SB), pressure support ventilation (PSV), and pressure control ventilation (PCV) after muscle paralysis (Fig. 11.7). The total chest wall volume significantly decreased immediately after induction by equal reductions in the rib cage and abdominal volumes. During QB, rib cage volume displacement corresponded to 34.2 ± 5.3 % of the tidal volume.



Fig. 11.9 OEP measurements at Uppsala University Hospital, Sweden, during the analysis of chest wall volume variations during induction of anesthesia

During SB, PSV, and PCV, this increased to 42.2 ± 4.9 , 48.2 ± 3.6 and 46.3 ± 3.2 %, respectively, with a corresponding decrease in the abdominal contribution. Breathing was initiated by the rib cage muscles during SB. Propofol anesthesia decreases end-expiratory chest wall volume, with a more pronounced effect on the diaphragm than on the rib cage muscles, which initiate breathing after apnea.

More recently, OEP was used to compare different modes of jet ventilation (JV), including high-frequency jet ventilation (HFJV) that is today routinely used for airway surgery (e.g., laryngo-microscopic or bronchoscopic procedures) and that can be beneficial for thoracic surgery and during procedures requiring minimal respiratory excursion (e.g., radiofrequency ablation of small tumors in the lung or liver) [61]. In this situation, OEP is the only system that can estimate operating volume and ventilation by monitoring chest wall volume variations. In fact, when evaluating different HFJV techniques, tidal volume and minute ventilation are difficult to assess because HFJV is applied in open systems, and standard pneumotachography does not provide accurate values.

Conclusions

Optoelectronic plethysmography has the great advantage that it can measure breathing patterns in any condition (e.g., rest, exercise, phonation, sleep) where the chest wall can be visualized totally noninvasively. It is highly accurate in the measurement of total chest wall volume variations, allowing partitioning of the complex shape of the chest wall into different compartments. In addition to being noninvasive, it requires no connection to the patient. Furthermore, it can be used without a subject-specific calibration, because it provides a direct measurement of the volumes in a three-dimensional reference frame and the calibration is not based on particular respiratory maneuvers requiring subject cooperation.

All these features make the OEP not only a reliable system for basic physiological and pathophysiological studies but also an attractive tool for evaluating breathing under a wide variety of circumstances both in health and disease.

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