Chapter 5

Opto-electronic Plethysmography

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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 co-operation. 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.

All these problems have induced investigators to attempt to measure ventilation indirectly by external measurement of chest wall surface motion [3]. Chest wall is defined by all the anatomical structures surrounding the lung and moving with it: 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. A difficulty is that the chest wall not only changes volume during breathing, but also changes shape. Konno and Mead [4] showed that, as a first approximation, the changes in shape could be accounted for by treating the chest wall as a system with at least two degrees of freedom, the volume changes of the rib cage and the abdomen [5]. Therefore, measurements must be made at more than one site and a number of devices have been used in the past to measure rib cage and abdominal motion, including mercury in rub-
ber strain gauges [5], linear differential transducers [4, 6], magnetometers [7, 8], and respiratory inductive plethysmography (RIP) [9]. Among these systems, only the last two have been extensively used in research and clinical practice. Respiratory magnetometers, consisting of tuned pairs of electromagnetic transmitting-receiving coils, measure the anteroposterior or the lateral diameters, while RIP, consisting of two loops of wire that are coiled and sewed into elastic belts, measures changes in cross-sectional areas of the rib cage and abdomen. These techniques might be used to estimate lung volume variations and several methods of calibration have been proposed [10, 11]. However, the validity of the calibration coefficients obtained experimentally to convert one or two dimensions to volume is limited to the estimation of tidal volume under conditions matched to those during which the calibration was performed. Several investigators have reported the errors introduced by the application of these devices to other conditions [12]. In addition, particularly in a clinical setting, adequate calibration data may be difficult to obtain because of poor subject cooperation.

Moreover, the limits of the two degrees of freedom model of the chest wall are frequently exceeded in conditions other than breathing at rest. Both the rib cage and abdomen are complex structures that can be easily distorted. The forces acting on the upper part of the rib cage (apposed to the lung) are quite different from those acting on the lower part (apposed to the diaphragm). Also the abdomen is constituted by at least two regions, one immediately subcostal that is mechanically linked to the rib cage and one (lower abdomen) that has no interactions with the rib cage [13]. Thus, acquisition of volume change from changes in diameter or cross-sectional area of a single transverse section are problematic for both rib cage and the abdomen, and attempt to measure changes in lung volume is subject to errors.

Another possibility is to use imaging systems like three-dimensional X-ray computed tomography (CT) [14] and magnetic resonance imaging (MRI), which allow accurate measurement of chest wall compartments. However, these techniques have a low temporal resolution, constrain the subject under analysis to given postures and, also, CT uses ionizing radiation.

Optical techniques have been proposed [15-18], but they have not been introduced in clinical practice because of their low temporal resolution and time-consuming data processing procedures. Total body plethysmography is impractical for prolonged measurements, almost impossible to use in situations like exercise, mechanical ventilation, and different postures other than the seated position, and is unable to provide the contribution of the different compartments to the total volume variation. In a study published several years ago [19], a modified plethysmograph was applied to the rib cage or to the abdomen separately, but the interpretation of the data was very difficult due to the movement of rib cage and abdomen. This approach was never transferred into any clinical environment. In 1994, an opto-electronic motion three-dimensional motion analysis system (Elite system [20]) was originally applied to analyze respiratory kinematics in a study in which inspired volumes were computed from geometric models (tetrahedra) of the whole chest wall and compared with spirometric data.