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Electrical impedance tomography: a new strategy to estimate V/Q relationship

Abstract

The perfusion ventilation imbalance isolated (V/Q mismatch) or in association with the intrapulmonary shunt (V/Q ratio) is the fundamental mechanism that determines the gasometric alterations present in the pathologies of the pulmonary parenchyma, the airways, and the pulmonary circulation. As the ventilation imbalance can be corrected with oxygen administration, the intrapulmonary shunt is refractory to oxygen therapy; therefore it is considered a remarkable tool to evaluate the magnitude of pulmonary compromise, especially when the patients are receiving mechanical ventilation with positive airway pressure.

Keywords: electrical impedance tomography, chronic obstructive pulmonary disease, pulmonary parenchyma, alveolar hemorrhage, atelectasis, V/Q ratios

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Abbreviations: EIT, electrical impedance tomography; COPD, chronic obstructive pulmonary disease; MRI, magnetic resonance imaging; CT, computerized tomography; AU, arbitrary units; VT, tidal volume; TIV, Tidal impedance variation; SPECT, single-photon emission computed tomography; PET, positron emission tomography; PCA, principal component analysis; LH, left heart; RH, right heart

Introduction

In pathologies that affect the airways (asthma, COPD) or the pulmonary parenchyma (pulmonary edema, ARDS (Acute Respiratory Distress Syndrome), alveolar hemorrhage, atelectasis), ventilation gradients are created by altering their distribution in different areas of the lung.[1] Some alveolar units are poorly ventilated but well perfused, with reduced V/Q ratios, which produces an increase in the venous admixture that impoverishes the O2 content of arterial blood.

Hypoxic pulmonary vasoconstriction attempts to cushion this V/Q imbalance so that poorly ventilated areas are also the worst perfused and the venous admixture is reduced. In pathologies with pulmonary vascular involvement (severe emphysema, pulmonary hypertension, and pulmonary embolism) there are well-ventilated but poorly perfused areas of the lung, with a high V/Q ratio, which increases the physiological dead space but have less gasometric repercussion as there is no effect in venous admixture.

In recent years there has been substantial progress in the imaging assessment of patients with lung disease who require mechanical ventilation. This has been demonstrated by the inclusion of pulmonary ecotomography, positron emission tomography, electrical impedance tomography (EIT) and magnetic nuclear resonance.[3]

The EIT uses electrical current to evaluate the distribution of the conductivity of the alternating current within the thoracic cavity. The advantage of the latter is that it is not invasive, does not emit radiation and can remain as long as necessary by acquiring real-time images facilitating the interpretation of pulmonary ventilation. During an apnea, the injection of saline hypertonic fluid could measure pulmonary perfusion.

The impedance is a physical variable that describes the resistance to the passage of the current offered by an inanimate body (minerals) or in biological tissue (bioimpedance) according to its characteristics. The unit of impedance is the ohm (Ω). This method is based on the repeated measurement of surface voltages, resulting from a high frequency (50-80 kHz) and low intensity (5mA) of alternating current rotary injection that circulates between the electrodes around the thorax (4th or 5th intercostal space) through a multichannel system. With a computer and 16 or 32 electrodes (with 1 or 2 neutral or reference) applied to the patient's chest, the bioimpedance signal can be processed, which allows describing the characteristics of the tissues (conductivity) in the selected body circumference (thoracic mapping). These electrodes collect the impedance information from a cross-section of approximately 5 to 10cm.

During an EIT test, cyclic injections of electric currents are performed sequentially, usually between all pairs of adjacent electrodes. This means that, in a 16-electrode system, the current is injected 16 times during each measurement cycle. During each application, the resulting potential differences are between the 13 pairs of passive electrodes. This pattern of current injections and potential difference measurements results in a set of 208 voltages acquired during each cycle. The voltage data calculated on the surface of the body is used to calculate the distribution of the electrical impedance within the thorax. This process is called image reconstruction since the calculated regional impedance values can be used to generate cross-sectional images, that is, scans of their distribution in the thorax. The scanning frequencies of the EIT are high, and currently available sampling rates of 25-50 cycles per second are feasible. These scanning rates are much higher than those used in conventional radiology, for example, in MRI and CT.

To continue, we must first clarify some basic concepts related to EIT. A) The concept of spatial resolution refers to the ability of a sensor to distinguish the smallest object, that is, the area represented by a pixel. B) The temporary resolution corresponds to the observation frequency of the sensor (according to the cycles per second: 25-50c/sec). C) The signal-to-noise ratio is the one that exists between the

power of the signal being transmitted and the power of the noise that corrupts it (decibels). The resolution of the images is better when the more electrodes are used (spatial resolution); however, the greater the number of electrodes the proximity of these increases the signal to noise ratio, which can interfere with the quality of the signal. We must also consider the contact impedance that is produced, as the name implies, at the junction between the electrode and the skin.

The impedance mathematically is a complex number formed by a real part (the resistance) and an imaginary part (reactance). If we apply this variable to biological tissue, we have different resistance to the passage of current. Measurements are generally based on the application of an alternating current by two contiguous electrodes (emitters) while the rest of the electrodes (receivers) detect the electrical signal. This signal travels through the thorax following routes that vary according to the shape of the chest wall and the distribution of the obstacles offered by the intrathoracic structures. Immediately after, the next pair of electrodes injects the current, and the rest of the electrodes receive the resulting voltage and so on; this information is registered by a computer.

Tissue densities determine the electrical potentials that will be measured on the surface of the thorax. These potentials are used to obtain the distribution of the electrical impedance using some reconstruction algorithm that solves the problem of the non-linearity of the signal, which we will explain later. Each complete lap of measuring the impedance of the analyzed section is called a cycle. Current tomography, as stated above, generally complete 25-50 cycles per second (temporal resolution), that is, 25 to 50 images per second (frames or raw images) and with 16 electrodes, an image of the plane defined by a matrix of 1,024 pixels (spatial resolution) allowing the evaluation of lung function under dynamic conditions. For the reconstruction of the information, the change of the relative impedance in each pixel is used. This (dimensionless) value derives from the difference in tissue impedance between two instants in time or between inspiration and expiration (expressed in arbitrary units [AU]).

The method used for the reconstruction of information in images is of crucial importance. The most used in clinical practice is the Sheffield back-projection algorithm and its subsequent modifications. The reconstruction algorithms refer to the calculations made by the computer that transform the voltages acquired on the surface of the thorax into an impedance image of a cross-section of the thorax. The back-projection filter uses a linear approximation to solve the problem of the characteristics of the impedance distribution.

Sinton et al. suggested an approach based on dynamic impedance changes, that is, relative changes in impedance at times or during breathing, assuming that the shape of the chest did not change between measurements. The bioimpedance measurements can be classified into two types: The first type implies a reduction in the impedance observed, as in tissues that have an increase in extracellular water content, high concentration of electrolytes (hypertonic NaCl) and a high number of cell junctions; and the second implies an increase in impedance as occurs with fat, bone, and air, which act as resistor elements. Thus, reconstruction is used based on the changes in impedance that is compared with a reference (relative impedance). The reconstruction algorithms discard measurements from poorly positioned electrodes and ignore redundant data.

The change in thoracic bioimpedance is fundamentally influenced by two cyclic mechanisms: ventilation and perfusion. The increase in the amount of air during inspiration, together with the increase in lung volume and the change in the volume of the rib cage, leads to an increase in impedance that is proportional to the volume of inspired gas. On the other hand, pulmonary perfusion causes small changes, of the order of 3%, in the thoracic impedance between systole and diastole, a percentage that is optimized with the injection of hypertonic serum by central venous line (Figure 1).

Ventilation/Perfusion monitoring

Ventilation monitoring

Global ventilation has been shown to correlate with global impedance changes (ΔZglobal) which result from sequential EIT measurements. Tidal impedance variation (TIV) represents ΔZ which is generated during a tidal breath and equals the difference between the maximum and minimum impedance at end-inspiration and end-expiration. Global TIV can be scaled and converted to volume (mL) with sufficient accuracy using the measurements of tidal volume (VT).

Regional ventilation which has been assessed by regional impedance changes (ΔZregional) using EIT correlates with data derived by electron beam computed tomography (R2=0.81–0.93). In ARDS patients, an excellent correlation (R2=0.96) was observed between ΔZregional and regional air content changes determined by CT, dynamic X-ray CT, single-photon emission computed tomography (SPECT), and positron emission tomography (PET).[4–6]

Perfusion monitoring

The quantification of pulmonary perfusion by EIT can be performed by a brief respiratory pause, followed by rapid infusion of hypertonic saline, which will dramatically reduce chest impedance, thus acting as an intravascular contrast. Pulmonary perfusion by EIT proved to be capable of producing significant and well-agreed V/Q maps as compared to perfusion maps produced using single-photon emission tomography (SPECT) (Figure 1).

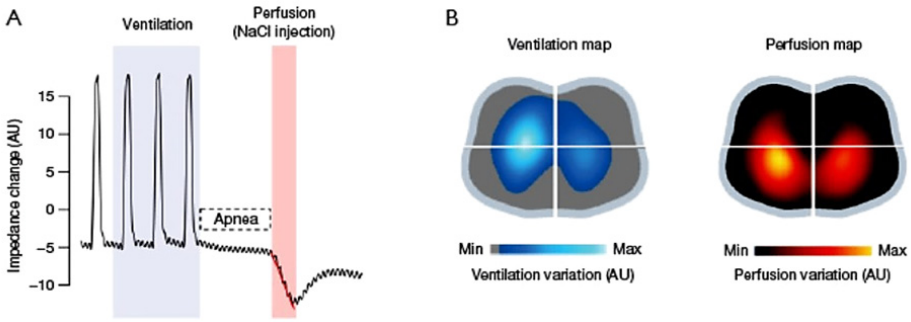

Figure 1 In A. The máximum slope of impedance reduction upon injecting of 10mL 7.5% NaCl indicates the passage of the indicator through the lung territory (rose colour). In B. the map of ventilation and perfusion obtained with EIT, showing limited perfusion and ventilation of the left Lung.[2]

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Another possibility is based on the separation of the cardiac signal from the ventilation signal through electrocardiogram gating or through the principal component analysis (PCA). Since the conductivity of the bolus differs widely from blood, it leaves a trace of impedance change while it passes across the vascular compartment. Regional lung perfusion can then be extracted from the indicator based-signal (IBS) by gamma-variate model fitting. The gamma-variate function has been often used to describe the dispersion of a bolus as it passes through a series of compartments. For this reason, it is frequently chosen to fit first-pass data in studies quantifying cardiac output and left to right cardiac shunts. The IBS can be separated into disjoint image streams that correspond to the activity in the right heart (RH), the lung (L) and the left heart (LH) (Figure 2).[7]

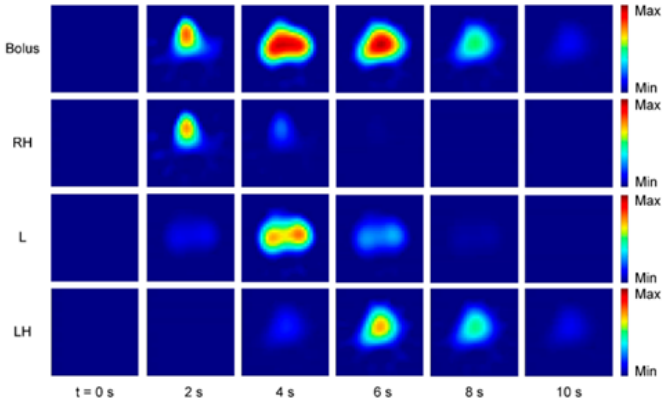

Figure 2 Bolus injection and transport of contrast agent (NaCl 7.5%) through the cardio-pulmonary system. The bolus is injected at time =0 seconds and transported from the right heart (RH) to the lung (L) into the left heart (H). Separation of the bolus is achieved using gamma-variate model fitting.

EIT has been shown useful for monitoring of regional lung perfusion as well as stroke volume estimation. Continuous and noninvasive perfusion monitoring can be performed badly on the EIT waveforms resulting from cardiac activity in both the heart and lung region, whereas intermittent and invasive monitoring is possible using contrast-agents such as hypertonic saline injected through a central venous line. Monitoring of regional lung perfusion and V/Q imaging using EIT might become available especially in patients who are receiving mechanical ventilation.

Conclusion

EIT-based perfusion monitoring may allow an image of local perfusion and mapping of regional ventilation-perfusion ratios. Some of the approaches described to separate cardiac from respiratory impedance changes are making an expiratory pause, by use of modeling techniques that may separate both signals while breathing (heartbeat-related EIT signal variations) or by intravenous injection of hypertonic saline (which serves as contrast agent due to its much higher conductivity). Nevertheless, clinical validation in this field is still pending.

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Conflicts of interest

The authors declare there are no conflicts of interest related to the article.

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