Monitoring changes in lung air and liquid volumes with electrical impedance tomoography

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Adler, A., R. Amyot, R. Guardo, J. H. T. Bates, and Y. Berthiaume. Monitoring changes in lung air and liquid volumes with electrical impedance tomoography. J. Appl. Physiol. 83(5): 1762–1767, 1997.—Electrical impedance tomoography (EIT) uses electrical measurements at electrodes placed around the thorax to image changes in the conductivity distribution within the thorax. This technique is well suited to studying pulmonary function because the movement of air, blood, and extravascular fluid induces significant conductivity changes within the thorax. We conducted three experimental protocols in a total of 19 dogs to assess the accuracy with which EIT can quantify changes in the volumes of both gas and fluid in the lungs. In the first protocol, lung volume increments from 50 to 1,000 ml were applied with a large syringe. EIT measured these volume changes with an average error of 27 ± 6 ml. In the second protocol, EIT measurements were made at end expiration and end inspiration during regular ventilation with tidal volume ranging from 100 to 1,000 ml. The average error in the EIT estimates of tidal volume was 90 ± 43 ml. In the third protocol, lung liquid volume was measured by instilling 5% albumin solution into a lung lobe in increments ranging from 10 to 100 ml. EIT measured these volume changes with an average error of 10 ± 10 ml and was also able to detect into which lobe the fluid had been instilled. These results indicate that EIT can noninvasively measure changes in the volumes of both gas and fluid in the lungs with clinically useful accuracy.

EIT

We obtained EIT images with a custom-designed and -built system. Technical details of the design and construction have been previously published (10), and a block diagram of the steps involved in obtaining EIT images with the system is shown in Fig. 1. Briefly, 16 electrocardiograph (ECG)-style electrodes are attached circumferentially around the subject’s thorax and connected to our EIT data-acquisition system, which sends the voltage data from each electrode to a computer for image calculation. EIT data were acquired by successively applying low-amplitude alternating current (300 μA at 10 kHz) across a pair of adjacent electrodes and measuring the voltage differences produced between each of the remaining pair combinations formed by the remaining 14 electrodes. The complete set of voltage measurements obtained from all electrodes for current injection at each pair of adjacent electrodes comprises one EIT data set. The electrical signals produced by physiological processes, such as the ECG, are eliminated by the EIT data-acquisition system, which discards all electrical signals not at the frequency of current injection. Three additional electrodes are used to record the

Invasive bedside cardiopulmonary monitoring is an integral and almost routine procedure in the intensive care unit (ICU) and affects decisions concerning treatment (15, 27). Presently, the status of the respiratory system in an ICU patient is assessed by using several techniques. Pulmonary mechanics parameters are obtained from measurements of flow and pressure at the mouth, and sometimes pressure in the esophagus (14). Fluid movement in the lungs (edema formation) is monitored by both chest X-ray and wedge pressure obtained with a Swan-Ganz catheter (12, 27). However, although these techniques provide essential information, they all have serious drawbacks related either to their invasiveness or to lack of specificity.

Ideally, cardiopulmonary monitoring in the ICU would be best served by a technique that combines the anatomical specificity of the chest X-ray, the continuous physiological-monitoring characteristics of the Swan-Ganz catheter, and the noninvasiveness of mouth pressure and flow measurements. A recently developed imaging technique called electrical impedance tomoography (EIT) shows promise in this regard. EIT produces a cross-sectional image of the conductivity distribution within a body by using electrical measurements made at a series of electrodes spaced around the body surface (4, 19, 28). Consequently, EIT should be well suited to imaging the distribution of air, plasma, and blood within the thorax, given the widely differing conductivities of these three materials with respect to the average conductivity of the lungs. EIT is also completely noninvasive and is relatively inexpensive and noncumbersome compared with other tomographic techniques (4, 19, 28). We therefore felt that EIT would be ideal for following changes in gas and fluid volumes within the lungs. Although some studies have reported the feasibility of using EIT to monitor lung edema (5, 18) and total lung volume (22), its sensitivity to changes in these quantities has not been evaluated. Therefore, in this study we investigated the accuracy with which EIT could quantify the changes we induced in lung gas and fluid volumes in anesthetized dogs.

METHODS

EIT

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ECG, which allows synchronization of EIT data acquisition with the cardiac cycle.

A cross-sectional image is calculated of the change in conductivity distribution in the thorax between the taking of any two data sets (1). This imaging approach is called “dynamic EIT,” to distinguish it from static EIT, which aims at reconstructing the absolute conductivity distribution. Dynamic imaging is significantly better at compensating for measurement errors and variations in electrode position than static EIT and is also more sensitive to small conductivity changes (2). The dynamic image is generated from any two data sets by using the algorithm described previously (1). The EIT image represents a cross section of the change in conductivity distribution in the plane of the electrodes between the time of acquisition of the data sets. Because electric currents can flow out of the electrode plane, the width of the cross section is about one-half the thorax diameter (2), which in this application is ~10 cm.

Estimation of Lung Volume Changes

Figure 2 shows the steps involved in estimating the change in lung gas or fluid volume with respect to some initial image. A volume of fluid or gas is introduced into the lung, and an EIT data set is collected. An image of the change in conductivity distribution within the thorax is then calculated by using this data set and a reference set taken before the lung volume change. A region of interest (ROI) is chosen surrounding the area in the image undergoing the change in conductivity, and the sum \( y \) of all the pixel values in the ROI is calculated. The ROI comprises 75% of the image, excluding areas close to the thorax surface, which tend to be most affected by electrode movement artifacts. Finally, an estimate of the volume change is calculated from \( y \) on the basis of a mathematical model.

Numerical simulations indicate that the relationship between the image pixel sum and the thoracic volume change should be essentially linear for physiological realizable conductivity changes (26) thus

\[
y = kx
\]

The constant of proportionality \( k \) depends on the conductivity of the material causing the lung volume change, its position relative to the plane of the EIT electrodes, and the particular ROI chosen. Thus a separate value of \( k \) must be determined experimentally for each material (e.g., air, plasma, saline) in each subject by regressing a range of values of \( y \) against independent estimates of the corresponding values of \( x \).

Given a measured \( y \), an estimate \( \hat{x} \) of the lung volume change is calculated as

\[
\hat{x} = \frac{y}{k}
\]

The error (e) in this estimate is

\[
e = \hat{x} - x
\]

The average value of \( e \) obtained in an experimental animal by using all volume increments of either air or saline was taken as the EIT measurement error for that material (denoted \( E_{\text{air}} \) and \( E_{\text{sal}} \) for air and saline, respectively) in that animal. \( E_{\text{air}} \) and \( E_{\text{sal}} \) were calculated for all animals studied.

Experimental Protocol

All the experimental protocols were reviewed and approved by the animal ethics committee of the Centre de Recherche Hôpital-Dieu de Montréal and of McGill University. The dogs (20–25 kg) used in protocol 1 were deeply anesthetized with a bolus of pentobarbital sodium (25–30 mg/kg iv) and maintained by an additional bolus injection of 130 mg/kg. The dogs (20–40 kg) used in protocols 2 and 3 were deeply anesthetized with a bolus of pentobarbital, and the anesthesia was maintained with halothane (2–2.5%). All dogs were paralyzed with Pavulon, and the level of anesthesia was evaluated by monitoring the changes in heart rate and/or blood pressure. After tracheostomy, the animals were mechanically ventilated for 1 h before experimental protocols were begun. Hemodynamic stability was assessed by monitoring heart rate, blood-gas contents, and pulmonary arterial and wedge pressures.

Sixteen ECG-style electrodes for obtaining EIT data were spaced evenly around the shaved thorax 10 cm above the base of the rib cage. An additional three electrodes were attached for recording the ECG. To avoid any contribution from cardiac activity, all EIT data acquisitions were triggered 100 ms after the QRS peak of the ECG. All EIT measurements were repeated three times.

Protocol 1: Estimation of incremental changes in gas volume. Nine dogs were studied. After a period of mechanical ventilation, the lungs were inflated twice to total lung capacity (transpulmonary pressure 3 kPa) and then disconnected from the ventilator. After animals passively expired to functional residual capacity (FRC), EIT data were obtained. A 2-liter syringe was then connected to a tracheal cannula, and three different gas volumes were introduced. Dynamic EIT images were reconstructed at each level with respect to the EIT data obtained at FRC. This protocol was repeated for volumes of 50, 100, 150, 200, 300, 400, 500, 600, 700, 800, 900, and 1,000 ml.
Protocol 2: Estimation of lung volume changes during ventilation. In another 10 dogs similarly prepared, we applied regular mechanical ventilation (10–18 breaths/min) by using tidal volumes of 200, 500, 700, and 1,000 ml. In three of these animals, we used tidal volumes in increments of 100 ml from 100 to 1,000 ml. At each tidal volume level, a dynamic image was calculated between EIT data sets taken at end expiration and end inspiration. Between each of the tidal volume settings we applied regular ventilation with a tidal volume of either 500 or 700 ml to allow the animal to stabilize.

Protocol 3: Estimation of changes in lung fluid volume. Normal saline solution containing 5% bovine albumin and Evans blue dye was instilled into a lobe of the right lung through a catheter positioned by using a bronchoscope. Volumes of 10, 25, 50, 75, and 100 ml of the fluid were given. At each volume level, EIT data sets were acquired at end expiration during regular mechanical ventilation (tidal volume 500 ml). An EIT data set was also collected before any fluid instillation. Just before the animal was killed, 10 ml of the saline were injected subcutaneously under the electrode positioned just to the left of center on the animal’s back. EIT images obtained before and after the subcutaneous injection allowed all the images to be oriented with respect to the animal. At the end of the experiment, the lungs were extracted and examined for the location of the Evans blue dye to confirm the site of fluid instillation.

In the animals we also investigated how ventilation to the two lungs would be affected by the unilateral fluid loading. Images were obtained from EIT data obtained at end expiration and end inspiration (protocol 2), and separate ROIs were defined to encompass the right and left lungs. Separate y values (Eq. 1) were then calculated for each ROI, and their ratio was taken as an index of relative ventilation to the two lungs. This quantity was calculated both before fluid loading and after the animals had received their full 100 ml of fluid, which allowed us to estimate the percent decrease in ventilation to the instilled lung after it received the fluid.

RESULTS

Representative EIT images obtained in one animal are shown in Fig. 3. Figure 3A shows the change in conductivity distribution because of a tidal volume of 500 ml (protocol 2) and shows the two lungs with approximately equal ventilation. Figure 3B shows the image obtained relative to control after instillation of 100 ml of fluid into the right lung (protocol 3). The increase in conductivity in the right lung can be clearly seen, whereas no change is apparent in the left lung. Figure 3C shows the differential image obtained before and after subcutaneous saline injection at the end of the experiments. The light area in the image identifies the left back of the animal. These images are similar to those obtained in all animals studied and show that movement of air into the lungs causes a localized decrease in conductivity, whereas a movement of fluid into the lungs produces a localized increase. In some cases this was not always possible to see two distinct lung regions after inflation with air. However, after fluid instillation it was always possible to identify the affected lung.

Figure 4 shows \( \hat{x} \) (Eq. 2) vs. incremental gas volume obtained from protocol 1, whereas Fig. 5 shows a similar plot obtained from protocol 2. In both cases, the value of \( k \) (Eq. 1) was chosen so that the regression line through these points was the line of identity. The average correlation coefficients \( (r^2) \) of the regression lines were 0.997 and 0.987 for protocols 1 and 2, respectively. The means of the vertical displacements of the data points from these lines give estimates of \( E_{air} \), which had values of 27 ± 6 ml (range 19–36 ml) from protocol 1 and 90 ± 43 ml (range 40–168 ml) from protocol 2.

Figure 6 shows \( \hat{x} \) vs. incremental instilled fluid volume from protocol 3. Again, \( k \) was adjusted to have the regression line equal the line of identity (average correlation coefficient 0.961). \( E_{sal} \) was 10 ± 10 ml (range 0–20 ml) from protocol 3.

![Fig. 3. Each image shown on left is a wire-frame plot where magnitude of conductivity change corresponds to height and on right is a gray-scale image where magnitude of conductivity change is indicated by increasing contrast. A: images of conductivity change in thorax because of a tidal volume of 500 ml. B: image of conductivity changes because of instillation of 100 ml into lower right lobe. C: image of conductivity changes because of subcutaneous injection of 10 ml of saline under dorsal electrode. All wire-frame images are shown at same scale, whereas each gray scale image was individually normalized to use full range of gray-scale contrast. Units on wire-frame images are arbitrary, upward deflexion indicates increasing conductivity. Neutral gray regions of image indicate areas that did not undergo conductivity change, whereas dark areas indicate a reduction in conductivity and light areas indicate an increase.](http://jap.physiology.org/Downloadedfrom)
3–38 ml). After the instillation of 100 ml of fluid into the lung, we observed that the ventilation shifted away from the instilled lung in 8 of 10 dogs, with the ventilation in the instilled lung decreasing by 9.3 ± 11.1 (SD)%.

**DISCUSSION**

Although cardiopulmonary monitoring in the ICU is routine, the methods used all have major drawbacks. For example, pressure and flow measurements at the airway opening are noninvasive and may be employed continuously, but they give only a generalized indication of mechanical abnormalities in the respiratory system without any anatomic localization of these abnormalities. Similarly, the Swan-Ganz catheter provides important ongoing physiological information but is invasive and could lead to complications (27). Furthermore, recently it has been shown that patients who had a Swan-Ganz catheter inserted have an increased mortality (6). In contrast, chest X-rays provide precise information concerning the location of an abnormality (25) but are limited by the radiation exposure they entail and so cannot be used for continuous monitoring. Thus there is a clear need for new ICU monitoring techniques that combine the various advantages of present techniques but avoid their limitations. EIT shows promise in this regard. Indeed, Morice et al. (17) have shown that EIT can identify pneumothorax as an unchanged conductivity during inspiration and aspiration of pleural effusion. Newell et al. (18) used EIT to monitor the development of pulmonary edema in dogs after oleic acid injection and demonstrated a correlation between their measurements and the standard wet-to-dry weight ratio of the lung. EIT is thus suited to monitoring conditions that involve changes in air and water volumes within the lungs. However, no study has yet addressed the issue of how accurately these volumes can be measured by EIT. Consequently, we set out to determine the accuracy with which our own EIT system can make such measurements.

The first purpose of our study was to assess the accuracy of EIT measurements of lung gas volume. It has been suggested that lung injury induced by mechanical ventilation is related more to excessive lung volume than to high pressure (8), so being able to accurately assess lung volume relative to FRC is clearly important in managing patients at risk for barotrauma and volutrauma. Monitoring the state of hyperinflation of the lungs is also crucial for optimal weaning of patients with chronic obstructive pulmonary disease (12, 24). Present methods for measuring lung volume in ventilated patients are typically related to measurement of airway pressure, which may be difficult to relate accurately to volume when compliance is changing as in acute respiratory disease syndrome patients. Other lung volume-measuring techniques such as helium dilution are difficult to perform in ventilated patients and cannot be applied continuously.

Our study has shown that EIT is able to measure changes in gas volumes with an accuracy of 30 ml when the lung is inflated with a calibrated syringe. This is well within the 5% or 100-ml error considered acceptable for spirometers by the American Thoracic Society (7). We also measured tidal volume between the beginning and end of a normal mechanically ventilated breath because we felt this was much more representative of the clinical situation and found larger errors of ~90 ml. The difference between this result and the smaller error obtained with the syringe was most likely due to errors in the ventilator estimate of tidal volume and to timing variability of the operator damping the airflow tube at the beginning of inspiration and expiration. Nevertheless, the EIT error obtained during mechanical ventilation still meets American Thoracic Society standards.

The second purpose of our study was to determine how accurately EIT can assess changes in lung fluid volume. Recently, Mitchell et al. (16) have shown in a prospective randomized trial that fluid management based on measurement of extravascular lung water leads to a reduction in the number of days that a patient remains on the ventilator in the ICU. This trial measured lung water by using the double-indicator dilution technique (25), which is relatively invasive, does not permit continuous monitoring, and does not show where in the lung the fluid has collected. EIT, by contrast, does not suffer from any of these limitations and is able to provide estimates of lung water with an error of 10 ml. This compares favorably with that of the gravimetric method, which has been shown to have a coefficient of variation of 11.1% (3). The idea of using changes in thoracic conductivity to monitor air and water movement in the lung is not new. Transthoracic impedance, obtained by measuring the voltage-current relationship between electrodes placed
at the neck and abdomen, has been proposed as a tool for diagnosing pulmonary edema (21, 23, 29). However, the impedance values obtained are sensitive to many factors and have no clear physical relationship to the quantities of interest. Thus it is difficult to separate the effects due to changes in such variables as posture and skin moisture from those due to changes in air or fluid in the lungs. Furthermore, enthusiasm for thoracic impedance measurements decreased after the results of a clinical study by Fein et al. (9), which showed large variations in normal impedance values, suggesting that the technique might not be reliable as a diagnostic tool. However, dynamic EIT extends the transthoracic impedance concept and achieves greater specificity by imaging the anatomic regions in which changes in conductivity take place. Although the outer edges of the image tend to be affected by changes in posture and electrode-skin resistance, these effects can be excluded by restricting the analysis to a central ROI. Furthermore, by using dynamic EIT we can compensate for errors because of electrode movement and variations in thorax shape that influence static impedance measurements (2).

Our EIT system, like most others, provides a single cross-sectional image in the plane of the electrodes. However, this image is affected by thoracic conductivity several centimeters above and below the image plane (2). We have shown (2) that the thickness of the cross section is about one-half not confined to the image plane (2). We have shown (2) that the thickness of the cross section is about one-half the diameter of the thorax, which corresponds to ~10 cm in the dogs we studied. Thus, with only one ring of electrodes, we effectively obtain an average image over ~20 cm of the length of the thorax. This represents a significant portion of the thorax in dogs and may partly explain why our estimates of lung volume change were so accurate even though they came from only a single cross-sectional image. Nevertheless, the sensitivity and specificity of the electrical signal decrease significantly as we move out of the image plane. One could thus imagine obtaining a significantly improved lung volume estimate by using multiple layers of electrodes to generate a three-dimensional thoracic image (13).

EIT is still an underdeveloped technology with much room for improvement. For example, an increase in impedance could be interpreted either as an increase in liquid content in the thoracic cavity (pleural effusion, pulmonary edema) or as a decrease in air volume (atelectasis). However, it has recently been suggested that interstitial edema may be distinguishable from alveolar edema on the basis of different changes in conductivity obtained by injecting currents of different frequencies (20), so future development of the technology may improve some of its present limitations. Nevertheless, by using the present technology we were able to demonstrate a decrease in ventilation in the instilled lung by modifying our analysis technique. Our present EIT system is thus able to provide estimates of changes in lung gas and liquid volumes that rival standard clinical methods.

In summary, we have estimated the errors associated with EIT measurements of changes in gas and liquid volumes in the lung in dogs. We found that gas volumes were estimated to within 30–90 ml, whereas liquid volumes were accurate to within 10 ml, all of which are clinically acceptable. These measurements were made completely noninvasively with a relatively inexpensive and noncubersome system and demonstrate that EIT has the potential to become a useful monitoring modality for ICU patients.

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