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A compensated radiolucent electrode array for combined EIT and mammography

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Abstract

Electrical impedance tomography (EIT), a non-invasive technique used to image the electrical conductivity and permittivity within a body from measurements taken on the body's surface, could be used as an indicator for breast cancer. Because of the low spatial resolution of EIT, combining it with other modalities may enhance its utility. X-ray mammography, the standard screening technique for breast cancer, is the first choice for that other modality. Here, we describe a radiolucent electrode array that can be attached to the compression plates of a mammography unit enabling EIT and mammography data to be taken simultaneously and in register. The radiolucent electrode array is made by depositing thin layers of metal on a plastic substrate. The structure of the array is presented along with data showing its x-ray absorbance and electrical properties. The data show that the electrode array has satisfactory radiolucency and sufficiently low resistance.

Keywords

electrical impedance tomography; x-ray mammography; radio-lucent electrode; tomosynthesis

1. Introduction

At present, x-ray mammography is the gold standard imaging modality for breast cancer screening in clinical practice. However, the sensitivity of mammography has been reported to be 74% with specificity 60%, which means a 26% false-negative rate and 40% false-positive rate (Kopans 1998). About 70–80% of women who undergo biopsy have benign lesions. Mammograms also require cumulative x-ray exposure and they are difficult to interpret, especially with the dense breast tissue prominent in younger women.

Increasing evidence is found showing significant differences in impedance between malignant breast tumors and normal tissues (Surowiec *et al* 1988, Jossinet 1996). Electrical impedance tomography (EIT), a non-invasive technique used to image the electrical conductivity and permittivity within a body from measurements taken on the body's surface, could thus be used as an indicator for breast cancer. Surowiec *et al* (1988) studied the dielectric properties of breast tissue, Jossinet (1996) measured the impedance of breast tissues and Da Silva *et al* (2000) used the impedance measurements to distinguish carcinomas from normal and benign tumors. These studies have shown that EIT is a promising modality to image breasts for malignancies. Recently, many reports have shown that the EIT technique can improve the sensitivity and specificity when used as an adjunct to mammography (Kerner *et al* 2002, Assenheimer *et al* 2001, Cherepenin *et al* 2001). These findings on the impedance properties of breast tumors encourage the investigation of the application of EIT in breast cancer detection.

Because of the low spatial resolution of EIT, combining it with other modalities may enhance its utility (Zou and Guo 2003). X-ray mammography, the standard screening technique for breast cancer detection, is the first choice for that other modality. To combine these two techniques, one could take mammograms and EIT images separately and use a mapping-based method to co-register. However, changes in the breast shape between the x-ray and EIT exams would make the registration mapping complex and difficult. An alternative is to use the same geometry in EIT and mammogram. Not only would it be easier to analyze the EIT images, but also the EIT images could augment the two-dimensional x-ray mammograms by providing three-dimensional images of the tissue being imaged. This could provide additional information on the location of the tumor. For this mammography geometry, EIT distinguishability has been studied experimentally by Kao *et al* (2003). A simplified reconstruction algorithm has been built and successfully tested in a saline-filled test tank by Choi *et al* (2004).

In order to take the x-ray and EIT measurements at the same time, the major task is to build electrodes which can allow the x-rays to pass through. Here, we describe a radiolucent electrode array that can be attached to the compression plates of a mammography unit, enabling EIT and mammography data to be taken simultaneously and co-register the images of these two modalities. The idea of using radiolucent electrodes has been patented by Steinberg *et al* (2004), but we know of no reports detailing their manufacture or electrical properties.

2. Methods

2.1. Mammography geometry

In x-ray mammography, a breast is compressed between two plates and x-ray images of the breasts are made as shown in figure 1. In order to image the electrical impedance distribution of the breast, electrodes need to be placed so that they are in contact with the upper and lower flattened surfaces of the breast. The electrode arrays must allow x-rays to pass through without significant attenuation, avoiding the need for an increase in radiological dose, or an artifact in the x-ray image. If a metal layer were to attenuate the x-ray to such an extent as to require an increase in the source power this would not of necessity translate into an increased x-ray exposure to the patient, since it would be absorbed or attenuated in the metal layer, but image artifacts would be likely to remain.

2.2. Fabrication of radiolucent electrodes

In selecting the materials for the electrode arrays, we investigated the radiolucency and the resistivity of several metals. Figure 2 shows the resistivity versus the x-ray mass attenuation coefficient with a photon energy of 250 keV for several candidate metals (NIST 2004). The plot shows that aluminum has both a low radiodensity, i.e. has the lowest attenuation coefficient, and a low resistance. However, aluminum oxide, Al_2O_3 , which will form on aluminum electrodes, has very poor conductivity. Consequently, to protect the surface from oxidation, we cover the aluminum with a very thin layer of gold. The metal is deposited on a 2 mil (50 μm) Kapton polyimide film which is an electrical insulator with a high x-ray radiolucency. To get the desired radiolucency for the electrode array, we have to have a very thin metal layer. This thin layer is produced using E-beam evaporation (figure 3) in the Microfabrication Clean Room (MCR) at RPI. The sample (Kapton) is placed in the deposition chamber and once the vacuum in the chamber reaches approximately 5×10^{-7} Torr (53 μPa), the metallic target is heated to high temperature by the electron beam. This process leads to the evaporation of the metal and its deposition onto the sample. The evaporation material is contained in a water-cooled copper hearth eliminating problems related to crucible contamination. Isolated crystal rate sensors ensure process accuracy and reliability. A mask is placed over the sample at the top of the deposition chamber to produce the desired metal

patterns to form the electrode array. Several crucibles are available to enable multiple layers of different metals to be deposited sequentially on the sample.

To produce an array of electrodes, we implement a stacked structure much like a multilayer printed circuit board. Each layer in this stack consists of metal deposited on Kapton in a desired pattern. An example of a single layer, consisting of 300 nm thick aluminum strips which are extended with thicker copper strips, is shown on the left in figure 4. A thick deposition of copper is used to make the narrow electrical leads from the electrode which have a low resistance. To enhance the adhesion of the metals to the Kapton, we deposit 20 nm titanium before the aluminum layer (Russell *et al* 1995). We add 50 nm of gold on top of the aluminum in the region that will form the electrode surface to reduce the contact impedance between skin and electrode. The right portion of figure 4 shows the side view of the stack. A ground layer (green) was added between the electrode layers (black) to reduce the noise and cross-talk between the electrodes.

The complete electrode array in the mammography geometry consists of two electrode arrays, top and bottom. Figure 5(b) shows one electrode array with its two associated ribbon cables. Each ribbon cable is connected to a bundle of coaxial cables with driven shields connected to channels of our EIT instrument (ACT 4) via a 75-circuit coaxial connector and an SMB connector.

2.3. Results

The radiolucent electrode arrays were tested in the Department of Radiology at the Massachusetts General Hospital (MGH) with a tomosynthesis digital mammography machine (Dobbins and Godfrey 2003). Figure 5(a) shows our EIT instrument (ACT 4) connected to the mammography machine and figure 5(b) shows the electrode arrays attached to the compression plates. Figure 5(c) shows a tomosynthesis image from one patient. The reconstructed layer shown is the location of the bottom electrode array. The copper part of the array can be easily seen at the top and bottom of the right-hand side of figure 5(c), but the electrode is not visible even though the gold has a relatively high radiodensity. The exposure time to obtain the mammogram was not increased. Figure 5(d) shows the image of figure 5(c), with the position of the electrode array indicated by a superimposed grid. It is at this location in figure 5(c) that no electrodes are seen.

For a test of durability, we stuck scotch tape on the electrodes and peeled it off slowly. The electrode array did not peel off from the Kapton film. In patient studies, we have found that the electrode arrays can be used for approximately 5–10 studies before the electrode surface starts to wear away. The measured resistance of each electrode is less than 2Ω and the capacitance to ground from each electrode is between 300 pF and 550 pF as shown in figure 6.

3. Cable and array compensation

The results above show that the radiolucent electrode arrays present an electrical impedance characterized by a significant series resistance and a large shunt capacitance. What may be even more critical is that this resistance and capacitance varies from electrode to electrode within the array, meaning that it must be accounted for if the impedance seen by the electrode is to be measured accurately. We have taken two approaches to modeling and compensating for these electrical characteristics of the arrays. In each case, each cable and electrode is characterized as a collection of one or more impedance values and these values are used to compute the voltage and current at the electrode from those measured at the instrument end of the cable.

3.1. Simple model

In the simple model, the complete cable and electrode for each channel is characterized as the single complex impedance that is measured by the EIT instrument when nothing is attached to the electrode. This measured impedance is then assumed to be in parallel with the load impedance applied to the electrode. This model is shown in figure 7(a), where the excitation source for the electrode drives the cable and electrode impedance, denoted as $Z_{\text{cable\&array}}$, in parallel with the unknown load resistance, Z_{load} . The values of $Z_{\text{cable\&array}}$ measured for the radiolucent electrode arrays are primarily capacitive.

3.2. Complex model

A more accurate approach is to model both the cable and electrode as a series and shunt impedance as shown in figure 7(b). The individual impedances in this model are measured through a series of open- and short-circuit measurements. The impedances for the cable, Z_{parC} and Z_{serC} , are determined by disconnecting the electrode array from the end of the cables and measuring the impedance seen by the source with the electrode array end open-circuited (Z_{parC}) and short-circuited (Z_{parC} in parallel with Z_{serC}). Using the values for the cable impedances along with open- and short-circuit impedances with the electrode array attached, the electrode array impedances, Z_{parC} and Z_{serC} , can be determined. We have built devices that make it easy to short all electrodes to ground to facilitate easy measurement of the short-circuit impedances.

We use the four complex impedance values directly in computing the voltage and current values at the load from those measured at the source. It is interesting to note, however, that Z_{parC} for our 2 m, shield-driven, coaxial cables is well modeled as a capacitor of approximately 7 pF while Z_{serC} is well modeled as a small resistance ($<1 \Omega$) in series with an inductance of approximately 1 μH . At high frequencies, this series inductance has a significant impact on the measured impedance. The series and parallel impedances for the electrode array correspond closely to the resistances and capacitances shown in figure 6.

3.3. Measurement results

Figure 8 shows the measured impedances when a 750 Ω resistor was placed between three different pairs of electrodes on the radiolucent electrode array. The resistor was attached to a pair of platinized platinum–iridium-surfaced titanium probes which were manually held in contact with the desired pair of electrodes for the measurements. The ACT 4 system, without the lead cables, has an automated calibration system which calibrates all channels of the instrument to a common standard, meaning that without the variations introduced by the cables and electrodes, the same impedance should be measured when the resistor is placed between any pair of instrument channels. The plot on the left of figure 8 shows the magnitude of the measured resistance as a function of frequency for each of the three channel pairs. Results are provided without any compensation for the cables/array, for compensation by the simple model and for compensation by the more complex model. Without any compensation, the apparent magnitude of the impedance drops at high frequencies due to the large shunt capacitance within the electrode array. Furthermore, the variation between channel pairs is high due to the variation in this shunt capacitance from electrode to electrode. Using the simple model substantially improves the performance by reducing both the variation with frequency and the variation between channel pairs. However, there is still some loss in performance at high frequencies. The plot on the right of figure 8 shows the measured angles for the 750 Ω resistor. Without any compensation, the angle deviates greatly from the actual value of 0, while both compensation techniques bring the angles near zero. Once again, the more complex approach is better at high frequencies.

4. Conclusions

The radiolucent electrode array has been built and tested successfully. The electrical properties of the array are acceptable and can be compensated for after taking measurements with open- and short-circuited arrays. We have designed, built, tested and applied an EIT system to perform regional impedance spectroscopy on breasts simultaneously and co-register with mammograms. The system has been used to study breast cancer patients at Massachusetts General Hospital. The study, involving a small number of patients undergoing biopsies, is designed to establish the ability of EIT to detect cancer by directly comparing EIT results with biopsy results.

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Figure 1.
Dual mode of EIT and mammography.

Figure 2.
Resistivity versus x-ray mass attenuation for various candidate electrode materials.

Figure 3.
A picture and sketch of the deposition machine.

Figure 4.

Electrode strips and a sketch of the electrode layer in cross-sectional view (not to scale). Green: ground layer (200 nm aluminum); black: electrode strips (300 nm aluminum); yellow: electrode array (50 nm gold); white: Kapton layer (2 μm); not shown: adhesive layer (10–20 nm titanium under aluminum and gold).

Figure 5. (a) Radiolucent electrode array and the wire connection; (b) tomosynthesis mammography machine with radiolucent electrode array; (c) a tomosynthesis image with a electrode array from a studied patient; (d) the location of the electrode array registered in mammogram.

Figure 6.
The measured resistance and capacitance from each electrode to ground.

Figure 7.

(a) Compensation model: simple model; (b) compensation model: complex model.

Figure 8.
Magnitude and angle of measured 750 Ω load.