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A Robust Current Pattern for the Detection of Intraventricular Hemorrhage in Neonates Using Electrical Impedance Tomography

T. Tang, Sungho Oh, and R. J. Sadleir

J Crayton Pruitt Family Department of Biomedical Engineering, University of Florida, Gainesville, FL, USA

Abstract

We compared two 16-electrode electrical impedance tomography (EIT) current patterns on their ability to reconstruct and quantify small amounts of bleeding inside a neonatal human head using both simulated and phantom data. The current patterns used were an adjacent injection RING pattern (with electrodes located equidistantly on the equator of a sphere) and an EEG current pattern based on the 10–20 EEG electrode layout. Structures mimicking electrically important structures in the infant skull were included in a spherical numerical forward model and their effects on reconstructions were determined. The EEG pattern was found to be a better topology to localize and quantify anomalies within lateral ventricular regions. The RING electrode pattern could not reconstruct anomaly location well, as it could not distinguish different axial positions. The quantification accuracy of the RING pattern was as good as the EEG pattern in noise-free environments. However, the EEG pattern showed better quantification ability than the RING pattern when noise was added. The performance of the EEG pattern improved further with respect to the RING pattern when a fontanel was included in forward models. Significantly better resolution and contrast of reconstructed anomalies was achieved when generated from a model containing such an opening and 50 dB added noise. The EEG method was further applied to reconstruct data from a realistic neonatal head model. Overall, acceptable reconstructions and quantification results were obtained using this model and the homogeneous spherical forward model.

Keywords

Neonatal; Intraventricular bleeding; 3D EIT reconstruction; Localization; Quantification

INTRODUCTION

Electrical impedance tomography (EIT) is a noninvasive imaging technique in which images of conductivity within a body can be inferred from surface electrode voltages.³⁹ In conventional EIT measurements, currents are injected into a volume through one pair of an electrode array, and the data consists of differential voltages measured between other array electrodes. Changes in the conductivity distribution can be reconstructed using a sensitivity-based method.^{17,18}

EIT has been applied to several different clinical areas, for example, the imaging of thorax, breast cancer, gastrointestinal tract, and abdomen.¹¹ Application of EIT to the head is a clinically relevant problem because of the conductivity changes that may occur as a result of bleeding, ischemia, and neural activity, but suffers the disadvantage of being hindered by the low conductivity of skull (around 15 mS/m²²). Even so, EIT has also been applied to the

Address correspondence to R. J. Sadleir, J Crayton Pruitt Family Department of Biomedical Engineering, University of Florida, Gainesville, FL, USA. sadleir@u.edu.

detection of brain activity, stroke, and epilepsy by a group based at University College London (UCL).^{33,34} Their studies have indicated the possibility of using EIT as an imaging tool in circumstances where blood accumulates in the brain. Phantom experiments performed by the UCL group demonstrated that the presence of a skull could reduce the impedance change to one-fifth that observed without the skull.³⁴ However, such small changes could still be imaged with EIT. In the same work, the group performed similar visual and sensory stimulation on humans, and collected the data with a circular ring of 16 scalp electrodes using an ‘opposite’ current pattern (where current electrodes were diametrically opposed on the ring). In another similar experiment,³³ data were collected with 31 electrodes applied in a modified 10–20 EEG system of electrode placement using a current pattern that included all the possible current positions and voltage measurements. Reproducible stimulus-related impedance changes could be seen in both studies. The reconstructed images from the 3D current pattern in the second study showed a better performance in localizing the stimulated area of cortex than the RING pattern. Subsequently the group tested improvements in this 3D method by incorporating realistic geometry (RG) and conductivity distributions into the forward model.^{3,14} In general, better results in terms of image quality and localization error were achieved with these modifications.

We are applying EIT to the problem of identifying and quantifying cerebral intraventricular hemorrhage (IVH) in neonates. IVH affects 35–50% of very premature human infants to some degree and is a significant cause of death and disability.⁴¹ Ultrasound (US), computer tomography (CT), and magnetic resonance imaging (MRI) are common modalities used to diagnosis IVH.¹² Comparative studies have shown that CT and MRI have a better interobserver agreement for IVH compared with US.⁴ However, CT and MRI have marked disadvantages because of the need to transport patients, which is undesirable for newborn infants connected to life support devices. US, being a bedside tool that does not interfere greatly with patient management, is usually preferred over CT and MRI.⁵ Nevertheless, a considerable drawback of US is that it requires an expert technician to operate the device, which makes it unsuitable for continuous monitoring. EIT may be a good alternative to US as a screening tool to monitor and alert clinicians to the presence and progress of bleeding, an increasingly important utility as therapies for IVH become available.⁸

Different current patterns have different sensitivity distributions in the head and within the ventricles, and will give different quantification and localization accuracies. We aim to determine a robust method to localize and quantify bleeding in the ventricles. One important consideration we used in assessing the validity of candidate methods for detecting conductivity changes inside the head is the presence of the skull. The skull forms a large barrier to detection of conductivity changes when using scalp electrodes because the relatively more conductive scalp tends to shunt applied current away from the skull. The neonatal skull is distinct from the adult skull because of the presence of openings in the skull at the anterior and posterior fontanel. Therefore, consideration of the presence of the fontanel in a neonatal skull might give benefit to certain current application patterns or electrode layouts.

Blood has a lower conductivity than the cerebrospinal fluid (CSF) that is normally present in the ventricles,¹⁰ and therefore a resistivity increase should be observed in the ventricles during bleeding. EIT is a promising tool to detect such impedance changes and has potential to quantify the amount of bleeding.²⁸ The degree of bleeding can be classified into four grades.²³ In this application, we have particular interest in detection of grade II bleeding, which US can only identify in about 50% of cases.²³ Grade II bleeding corresponds to filling of less than half the ventricle with blood.

The purpose of this study, therefore, is to evaluate the performance of EIT, and of different EIT current patterns, in terms of their ability to quantify and localize anomalies similar to grade

II IVH bleeding events in the neonatal human head. Another problem that has been frequently addressed in the EIT literature^{1,3,14–16} is the benefit of incorporating an accurate reconstruction model to improve the image quality. In this study, we also address the problem of whether the use of a simple homogeneous spherical might be sufficient for the purpose of quantification, and compare image and quantification results with and without realistic reconstruction modeling.

MATERIALS AND METHODS

EIT Image Reconstruction

We can divide the process of reconstruction into two parts: a forward problem solution and an inverse problem solution. In order to solve the inverse problem we first solve the forward problem on a prototype model so that measured data can be compared with predicted voltage measurements.¹¹ A numerical modeling method such as the finite-element method may be used to solve the forward problem for general cases. Initially, forward models and sensitivity matrices may be formulated using a simple model such as a sphere containing a homogeneous conductivity distribution. In principle, better results from the inverse solution should be obtained if the forward model (the prior distribution) exactly matches the imaged object in terms of internal conductivity distribution and external geometry.^{1,3,16} However, it is difficult to obtain an accurate model for an individual head because of the difficulty of determining the locations of electrically relevant structures such as scalp, skull, CSF, grey, and white matter from modalities such as MRI and CT. Even then, external neonatal head shapes may vary during measurement because of the flexibility of the infant skull.

To formulate the inverse problem solution, we use information generated from the forward model in combination with experimentally measured boundary voltage measurements to predict the internal conductivity distribution. In time difference imaging, which is the method used in this study, conductivity changes from an initial assumed distribution are calculated from boundary voltage measurement changes. The relationship between the boundary voltage measurement changes and internal conductivity changes can be expressed to first order with a sensitivity matrix.¹⁸ The EIT inverse problem may then be solved by inverting the sensitivity matrix. Because the inverse problem is ill-posed, inversion must be performed using regularization methods such as truncated singular value decomposition (TSVD) or Tikhonov methods.¹¹ Specific details of reconstruction schemes used in this study are given in the section “Reconstruction.”

Current Patterns

We investigated two 16-electrode layout strategies and current patterns in this article. The first one was a RING layout (Fig. 1a), where the 16 electrodes were placed equally spaced in a single plane about the equator of the sphere. An ‘adjacent’ current pattern was applied to this layout, wherein current was applied in turn to pairs of adjacent electrodes, and voltages were measured across other pairs of adjacent electrodes. The *RING pattern* therefore involved 16 current positions and a total of 208 voltage measurements (only 104 of which were unique). The second layout was a standard EEG pattern, following the 10–20 system of electrode placement (Fig. 1b). The current pattern applied to the EEG layout was denoted the *EEG pattern*. In this pattern, all currents applied were returned via an electrode on the apex of the head, and voltages between ‘adjacent’ electrodes (in terms of electrode numbering, Fig. 1b) were measured. The EEG pattern therefore involved 15 current positions and a total of 182 voltage measurements. The basis for formulating the EEG current pattern was the geometry of the infant human skull. As neonates have a fontanel located at the apex of the skull, we expected that when currents flow via an electrode situated over the fontanel, there should be

greater sensitivities to centrally located conductivity changes compared to cases where the skull forms a complete barrier.²⁸

To formulate sensitivity matrices, we constructed spherical head models with unit radius and homogeneous conductivity distributions for each 16-electrode layout using finite-element modeling software (Comsol, Burlington, MA, USA). We constructed 26,126 and 21,896 tetrahedral element models for the forward models using RING and EEG electrode configurations, respectively. The EEG pattern was also applied to a neonatal head model with a realistic external geometry (the RG model).³⁶ This model contained four different anatomical structures (skin, skull, CSF, and brain) which were manually segmented from MRI images of a normal term neonate (Fig. 2b). The total volume of the model was 791 mL. The RG model was meshed into 45,702 tetrahedral elements with 8703 nodal points (the RG mesh). A sensitivity matrix for this model was formulated based on the RG mesh while assuming a homogeneous conductivity distribution. After creating quadratic tetrahedral meshes for each case, we calculated sensitivity values corresponding to each mesh element and constructed overall sensitivity matrices. Details of sensitivity matrix calculations may be found in Sadleir and Tang.²⁸

Data Generation

We gathered simulated bleeding data using RG and spherical finite-element models and saline-filled phantoms.

Finite-Element Models of Bleeding—To obtain more realistic boundary measurements and evaluate the effects of other head structures on reconstructions, we constructed a spherical finite-element model that contained representations of electrically important anatomical structures in the infant skull (Fig. 2a). These included the brain, skull with anterior fontanel, scalp and a thin CSF layer between the brain and the skull. We placed two small elliptical spheroids in the upper hemisphere of the brain compartment to simulate the two lateral ventricles. The size and position of the ventricles were chosen from representative neonatal head MRI data.³⁰ From these data, we determined that the total lateral ventricle volume was about 1/70 of the head volume V_H ($V_H/140$ each) and that the ventricles should be placed at a relative radius 0.3 above the equator. The minor and major axes of each ellipse were chosen to be 0.3 and 0.88 of the sphere radius, respectively, giving a volume for each ventricle of $V_H/100$. Blood-like anomalies with relative radii of 0.1 (a relative volume of 1/1000) were moved to six different locations within the ventricle regions (Fig. 2a). The conductivity of each structure was assigned according to values from the literature. The values used were blood (0.67 S/m),⁷ CSF (1.3 S/m),²² brain (0.17 S/m),³² and scalp (0.43 S/m).⁶ The conductivity of the neonatal skull was assigned to be 0.06 S/m (four times adult skull value) according to preliminary data obtained using a neonatal piglet skull.²⁵

We also applied the EEG current pattern to the RG head model (Fig. 2b).³⁶ To simulate intraventricular bleeding, we added two ventricles within the brain compartment of the RG mesh. Similar to the ventricle structures used in the spherical model, they were elliptical with a total size of about 1/70 of the head volume. The ventricles were placed 18 mm above the anatomical center and were spaced laterally 18 mm apart about this point. We used six bleeding sites at similar relative locations within the ventricle regions, as for the spherical model (Fig. 2b, right). Tissue conductivities used were the same as for the spherical model. The 16 electrode positions were adapted to the RG model and placed on the scalp at the locations shown in Fig. 1b and detailed in Sadleir and Tang.²⁸ To get a more comprehensive understanding of quantification in this RG model, we also investigated multi-site bleeding. To do this we placed anomalies at combinations of the six locations, leading to a total of

$C_6^1 + C_6^2 + C_6^3 + C_6^4 + C_6^5 + C_6^6 = 63$ bleeding cases.

Phantom Models of Bleeding—Phantom experiments were conducted to allow testing of how each current pattern, electrode layout, and reconstruction performed in the presence of realistic noise. The phantoms and measurement system are shown in Fig. 3.

We used only two different conductivities in the phantom. A 0.6% saline solution (the conductivity of which was approximately 1.3 S/m, the conductivity of CSF²²) was used as the phantom background, and gel anomalies were made using TX151 powder.³⁸ TX151 is a solidifying medium that produces a stable gel-like material when mixed with water, agar, and saline. By carefully adjusting the percentage of each ingredient (deionized water 80.8%, sucrose 13.8%, agar 2.5%, TX151 2.5%, and sodium chloride 1.3%), we made the anomaly conductivities approximately 0.67 S/m (the conductivity of blood). Further details of anomaly construction may be found in Sadleir *et al.*²⁷ The spherical polycarbonate containers used as spherical phantoms were obtained from a manufacturer of lighting fixtures (Formed Plastics, NY, USA). Each spherical phantom was equipped with 16 stainless steel electrodes (diameter 14 mm) according to their electrode configuration (Fig. 3a). Both phantoms had inner diameters of 19.7 cm and anomalies had diameters of 2 cm (a relative radius of 0.1). The size ratio between the phantom and the anomaly was therefore approximately the same as those used in simulations. A single anomaly was placed accurately at the same six positions used in the spherical numerical model (detailed above), using a 3D Positioner (Velmex, Inc., NY, USA). The overall experimental layout is shown in Fig. 3b.

Data were collected using a MK1 16-electrode EIT system provided by Impedance Imaging Research Center at Kyung Hee University, Korea.²¹ This is a multi-frequency EIT system with a frequency ranging from 10 Hz to 500 kHz (Fig. 3c). We chose to use a single frequency of 10 kHz and a constant current output of 1 mA. This frequency was chosen because the impedance properties of both TX-151 and saline are flat over the interval 10 Hz–50 kHz²⁷ and 10 kHz was well calibrated and in the center of this range. The reference electrode for the system was a copper wire placed at the center of the phantom ‘neck.’ Ten data sets were collected and averaged as a reference before anomalies were introduced. In each of the six experiments, five data sets were collected and averaged as the perturbation data for that anomaly position.

Reconstruction, Quantification, and Localization of Reconstructed Anomalies

Reconstruction—EIT reconstruction is a non-linear problem. However, in difference imaging, if the conductivity changes are small enough, the relationship between conductivity perturbations $\Delta\sigma$ and changes in boundary voltage perturbations ΔV are approximately linear.¹⁸ This linear relationship can be represented by a sensitivity matrix S (1), which may be calculated from a spherical model with a uniform conductivity background.²⁸ After pseudo-inverting S to $S^\dagger$, the conductivity map can be calculated from the product of ΔV and $S^\dagger$ as in Eq. (2).

$$\Delta V = S \Delta\sigma \quad (1)$$

$$\Delta\sigma = S^\dagger \Delta V \quad (2)$$

In EIT, the sensitivity matrix is always severely ill-conditioned and will therefore have large condition numbers. Each of the three sensitivity matrices used in this study had a full rank of 104 but involved a measurement strategy that involved making a larger number of observations, as shown in Table 1. Typical condition numbers (ratio of maximum to minimum matrix

eigenvalues) were of the order of 10^{19} for the RING strategy and 10^{14} for the EEG strategy. Data errors will be magnified in the inversion process because of this ill-conditioning. Therefore, we regularized the inversion by employing TSVD.⁴⁰ The truncation point, k , will be less than or equal to the rank of the matrix, but in general will be less depending on the noise level in the voltage measurements and the rank of the sensitivity matrix. We chose k -values for our reconstructions by inspecting images generated using a range of truncation points and determining the maximum value that produced a coherent reconstructed image. The truncation values needed to be chosen carefully since inappropriate truncation values would produce inaccurate solutions. The optimal truncation numbers for the problems in this study are between 50 and 70, and were evaluated with an l -curve method.⁹ The bottom row of Table 1 lists the truncation points we chose for each case using this heuristic method. Truncation points of 60 and 50 were used in both EEG and RING cases for simulated and experimental data, respectively.

Quantification—Quantification was assessed with a characteristic parameter called the quantification index (QI), defined in Eq. (3), which has been found to correlate well with volumes of blood-like anomalies.²⁶ The QI is the sum of conductivity changes over the image volume. In Eq. (3), n_e is the total number of elements in the mesh, $\Delta\sigma_i$ is the conductivity change reconstructed in the i th element, and v is the volume of each element. QI values should be linear with anomaly volume over the range of volumes experienced. It should also have low spatial variability over the imaged object.²⁰

$$QI = \sum_{i=1}^{n_e} \Delta\sigma_i v_i \quad (3)$$

Localization—The reconstructed anomaly location is defined in Eq. (4). The expression for anomaly location (x, y, z) is effectively the ‘center of mass’ of the difference image, which is calculated by averaging the positions of all elements, weighted by their conductivity changes. The values x_i, y_i, z_i are the coordinates of the center of each mesh element.

$$x = \frac{\sum_{i=1}^{n_e} \Delta\sigma_i v_i x_i}{\sum_{i=1}^{n_e} \Delta\sigma_i v_i}, \quad y = \frac{\sum_{i=1}^{n_e} \Delta\sigma_i v_i y_i}{\sum_{i=1}^{n_e} \Delta\sigma_i v_i}, \quad z = \frac{\sum_{i=1}^{n_e} \Delta\sigma_i v_i z_i}{\sum_{i=1}^{n_e} \Delta\sigma_i v_i}, \quad (4)$$

Errors in reconstructed locations were calculated as the absolute location errors quoted as a fraction of the model or phantom radius (‘relative radial error’, denoted D_{xy}). We considered separately horizontal errors (errors within the xy plane) and axial errors (errors in z direction, denoted D_z), respectively, for each of the two current patterns. Therefore, we define these values as

$$D_{xy} = \sqrt{(\Delta x)^2 + (\Delta y)^2} / R \quad (5)$$

$$D_z = |\Delta z| / R \quad (6)$$

Image Resolution and Contrast—We compared image resolution using a method based on one suggested in Adler *et al.*,² wherein we estimated the relative volume contained in the set of voxels anomalies that had reconstructed values more than half the maximum within the image V_{HM}/V_0 , where V_0 is the entire model volume, and then took the cube root of this value, i.e., for each image we computed the value

$$R = \sqrt[3]{\frac{V_{HM}}{V_0}} \quad (7)$$

The half maximum threshold was also used to construct isosurfaces for display of reconstructed anomalies. Use of this measure allowed comparison of both the extent of artifacts within images (as large artifacts will tend to increase the apparent resolution) and the true resolution in ventricular regions. We examined the ability of our reconstruction to resolve two anomalies by examining the reconstructed profile through a line joining the two anomalies. We defined the two anomalies as resolvable if the value between the anomalies reached a value half the maximum conductivity found on the profile.

We compared image contrast c for different reconstructions by first finding the average image value over a region having the same size as the original anomaly that was placed in the lower hemisphere (c_B), far away from artifacts and the anomaly reconstruction. We then took the difference between this value and the average conductivity in the region of the reconstructed anomaly, centered on the maximum reconstructed voxel value (c_M). Thus we have $c = c_M - c_B$. Larger contrast values corresponded to better reconstruction qualities.

Comparison Between ‘Open Skull’ and ‘Closed Skull’ Models

The spherical model shown in Fig. 2a represents an ‘open skull’ model, since we assigned the scalp conductivity (0.43 S/m) to the fontanel. To evaluate the benefit of this fontanel to the EEG-based current patterns, we constructed another ‘closed skull’ model by setting the fontanel conductivity to be the same as the skull conductivity (0.06 S/m). Selectivity¹³ denotes the contribution of a region of interest to a total measured transimpedance. The greater the selectivity in a region, the greater is the signal-to-noise ratio (SNR) in data gathered from that region. Our earlier study²⁸ showed that the EEG pattern has better selectivities in the ventricle regions with an open fontanel included. In this study, we compared the performance of the EEG current pattern on both open and closed models by comparing the reconstructed images when different levels of white Gaussian noise were introduced to the measurements. The noise levels were determined by the SNR levels of practical EIT systems. The KHU Mark1 system was reported to have an average SNR of 80–90 dB.²¹ An SNR of 50–60 dB was observed with simple phantom experiments using the ITS P1000 system.¹⁹ Therefore, we made comparisons with 50, 65, and 75 dB introduced Gaussian noise. We added noise to model data by first measuring the signal power, and then adding noise at the specified SNR level using the MATLAB function AWGN. Noise was added first to the reference data and then to data from a case with an anomaly present. These two sets were subtracted from each other to form the voltage perturbation ΔV (2). For each noise level and skull composition, we created 100 data sets. We compared the resulting resolution and contrast values in reconstructed images at the same noise level for the two skull conditions *a priori* using *t*-tests.

RESULTS

Comparing Spherical Model and Phantom Data

Results for reconstructions of simulated and phantom data using both current patterns and sensitivity matrices are shown in Fig. 4. Figure 4a shows an ideal image of bleeding accumulation at position 1 (Fig. 2a), a relative radial position of $(-0.2, 0.25, 0.3)$, illustrated in axial and sagittal sections and as isosurfaces, with results for reconstructions of (b) simulated EEG data, (c) simulated RING data, (d) phantom EEG data, and (e) phantom RING data. Isosurfaces were calculated at a threshold set at half the maximum voxel value within the image. As expected, the RING pattern could not determine axial positions of anomalies because of its two-dimensional symmetry. Resolutions in reconstructions from all phantom data appeared similar to reconstructions from simulated data, but contained artifacts, as reflected in results shown in Table 2. In most cases, the characteristic size of reconstructions for EEG cases was smaller than those for the RING cases. Interestingly, although it is clear that in both comparisons of anomaly 1 reconstructions that those for the RING are more diffuse, the R measure gave a smaller value for RING resolutions than for EEG resolutions.

The EEG pattern could localize well in all three dimensions. Figure 5 shows that the RING pattern localized anomalies as well as the EEG pattern in the xy -plane, producing a maximum radial relative error of 0.035 in both cases. The EEG pattern localized the z -coordinate well in simulated data, with the maximum D_z less than 0.03 of the radius over all intraventricular locations. Localization errors were larger for reconstructions from both current patterns applied to the phantom case, with the worst value of D_{xy} being 0.17 for the EEG pattern and 0.1 for the RING pattern.

The QI values found were comparable to the ideal QI, which can be calculated as

$$QI_{\text{ideal}} = (\sigma_{\text{blood}} - \sigma_{\text{CSF}}) \cdot \frac{4}{3} \pi (r_{\text{blood}})^3 = -0.0026 [\text{Sm}^2]. \quad (8)$$

Quantification results for the RING pattern were similar to those for the EEG pattern for simulated data, with the largest variation not more than 5% for both cases. Quantification results from phantom data were also similar, although QI errors found using the RING pattern were very large in some cases. The worst QI error in reconstructions using the EEG pattern with phantom data was around 10%, whereas the worst QI error in corresponding RING pattern phantom reconstructions was almost 30% (Fig. 6).

Comparison Between ‘Open Skull’ and ‘Closed Skull’ Models

Comparisons of reconstructions of a blood-like anomaly at position 1 using the EEG current pattern with open and closed skull models are shown in Fig. 7. The advantage of using a fontanel-focused pattern was made visible when 50 or 65 dB Gaussian white noise was added to data. At 50 dB, the open skull model had resolution 0.1805 and contrast 0.5228, whereas the closed skull model had a resolution of 0.1959 and contrast 0.5158. At 65 dB, we could still locate a peak conductivity change at a location corresponding to the anomaly position in an open skull reconstruction (resolution $R = 0.187$, contrast $c = 0.523$) while the reconstruction on the closed skull model was almost completely dominated by artifact ($R = 0.205$, $c = 0.499$). Both images were still affected at 75 dB noise levels, with open skull resolution and contrast being 0.222 and 0.473, respectively, and closed skull resolution and contrast being 0.235 and 0.467, respectively. In the 50 and 65 dB cases, R and c values were significantly better ($p < 0.05$) for open skull cases than for closed skull cases. At higher SNR levels (75 dB and above) we found that resolutions and contrasts were not significantly different. We compared the ability

of the open skull EEG configuration to distinguish between closely separated blood-like anomalies within the ventricles and found that we could resolve anomaly pairs 1–6 and 1–4 (using the numbering scheme in Fig. 2a, left) by our criteria, but not anomaly pairs 2–5 or 3–6—even though clear dips were visible between reconstructed peaks in all cases.

Investigations of Model Mismatch

We investigated the effect of model mismatch by reconstructing EEG pattern images using a sensitivity matrix calculated for a spherical object (S_{sp}) and data generated from the RG model. These reconstructed images were then compared to the images reconstructed using a homogeneous RG sensitivity matrix. Figure 8 shows the reconstructions for an individual impedance change at position 1. Images in Fig. 8 reconstructed using the spherical matrix were similar to the second image in Fig. 4, which shows the reconstruction from EEG data calculated from a spherical object. Although resolutions were better for RG reconstructions than for spherical (0.138 and 0.168, respectively), the contrast for the spherical case (0.467) was better than for RG reconstructions (0.454). Quantification results for multi-site bleeding cases are compared in Fig. 9a. The correspondence between results obtained using the spherical reconstruction matrix was similar to those obtained from the RG matrix, both of them giving QI estimations having a maximum error not more than 0.0025 S cm^2 over all positions, a value that corresponds to a blood volume of 0.4 mL when considered relative to the total volume of the head. We also compared the performance of the two reconstruction matrices with respect to QI values when white Gaussian noise was added to data (Fig. 9b). In this figure, we compare the average and standard deviation of QI errors over all positions for both reconstruction matrices at 11 different SNR levels between 60 and 90 dB. We found that the variation in quantification errors became almost three times larger for both reconstruction methods as the SNR was decreased to 60 dB.

DISCUSSION

Comparison of Results from Simulated and Phantom Spherical Model Data

Both the RING and EEG patterns performed well in terms of xy -plane localization and quantification accuracy on reconstructions of simulated data generated from spherical models, with the localization results shown in Fig. 5 confirming our observations from the reconstructed images of Fig. 4. Localization accuracy was slightly worse for the phantom data, presumably because of the presence of random and systematic experimental noise. We believe that the large localization errors at positions 4, 5, and 6 in the EEG phantom experiment shown in Fig. 5 may have been caused by a degradation in anomaly quality after it had been immersed in the saline background for some time (each set of measurements at a single anomaly volume were performed by moving the anomaly around the tank over a period of approximately 10 min^{26, 27}), but they may also have been caused by the slight bilateral asymmetry of the sensitivity matrix combined with noise. As shown in Fig. 6, the QI variability in the noiseless simulated data was relatively flat in both cases, with the largest variations being not more than 5%. As with localization performance, phantom results tended to agree with the simulation results, but we observed larger QI variations for phantom cases. Figure 6 also shows that the EEG pattern performed better than the RING pattern in phantom experiments, confirming results of our earlier investigation.²⁸ In this study, sensitivity analysis of the two patterns showed that the RING current pattern had a much lower selectivity than the EEG pattern in ventricular regions. Therefore for the same conductivity perturbation, smaller voltage changes were produced, as a consequence of which the noise played a more significant role in the RING phantom measurements. On closer inspection of data vectors, we found that the normed sum of the voltage perturbation caused by a single blood-like anomaly within a ventricle was about 1/5000 that of the normed sum of all reference measurements for the RING pattern, and about 1/1000 for the EEG pattern. The SNR level of the KHU Mark1 system was reported to be 80–90 dB.

²¹ This implies that noise accounted for 17–50% of RING phantom data, while accounting for only 3–10% of EEG phantom data. As a result, the QI values were more severely contaminated by the noise-generated artifacts in reconstructions based on the RING current pattern.

Anomaly Quantification

Absolute QI values do not only depend on the size of the anomaly, but also depend on the current pattern's sensitivity matrix and the truncation point used. For example, the RING and EEG patterns produced different sensitivities in the ventricle regions, and therefore had different QI accuracy for the same anomaly presented in this particular region. A truncation number of 50 was used for all the phantom experiments in this study. A different truncation point would shift all the absolute QI values to another level. In practice, we can determine the relationship between QI and anomaly volume with an EIT system calibrated for a particular configuration and truncation number using a phantom and blood-like anomalies.^{26,29}

Variations of up to 30% were observed in QI values for anomalies moved within ventricular regions of the RING spherical phantom, and up to 10% using the EEG pattern and phantom, an observation that was correlated with poor localization in the *xy*-plane. This may have been related to the overall smaller selectivity found for RING pattern in ventricular regions,²⁸ combined with measurement noise. Even though the EEG pattern showed generally good results, some method of QI regularization that reduces its shift variance may be appropriate. We have results of implementing such a method in 2D 16-electrode reconstructions.^{20,29} This algorithm gave very promising results in 2D, resulting in a reduction in QI spatial variability to about half. The method may be straightforwardly extended to 3D. We believe that the use of a similar approach with the EEG pattern would likewise make QI accuracy greater in the present context.

Comparison Between 'Open Skull' and 'Closed Skull' Models

We chose to compare the open skull and closed skull models using three different levels of Gaussian noise (50, 65, and 75 dB), which were chosen from the range of SNR values reported for practical EIT systems.^{19,21} Our sensitivity analysis²⁸ showed that the EEG pattern has on average a 7–10% larger selectivity near the ventricle region on the open skull model than on the closed skull model. This, in conjunction with the overall larger sensitivities observed in open skull cases, indicates that perturbations inside the lateral ventricle region should contribute larger relative voltage changes with an open skull than with a closed skull. Therefore, the EEG pattern should have a greater potential to detect and localize a perturbation within the ventricle regions when the fontanel-like opening is included. The comparison of the reconstructed images in Fig. 7 demonstrated that the EEG pattern produced overall better images on the open skull model than on the closed skull model, especially at low-SNR levels. The implication of this is that the EEG pattern should produce superior images on neonatal skulls than (closed) adult skulls. Images produced above 75 dB SNR were similar because the noise only accounted for about 6% of the voltage changes. However, this indicates that the EEG pattern may have the potential to detect smaller amounts of bleeding than the RING pattern above 75 dB.

Comparing Localization Results with Earlier Studies

The UCL group performed phantom experiments with a hemispherical tank and a modified 10–20 EEG electrode configuration that involved 31 electrodes.³⁵ Their results showed that the localization errors were as low as 3.5% of the image diameter when the anomaly was placed at the anterior or posterior of the midline of the hemisphere ($z = 0.5$), which is overall better than our localization accuracy. Another study on EEG dipole localization⁴² showed that using 32 electrodes instead of 19 improved the localization by 2.7 mm on average in the upper hemisphere of the adult head. Localization errors can be further improved by weighting the

sensitivity matrix. Liston *et al.*¹⁵ normalized the sensitivity matrix by dividing each column by the sum of this column before inverting the matrix, and localized a Perspex rod within 20% of the tank diameter. The localization error was further reduced to 5% of the tank diameter when an additional post-weighting and correct row normalization method was implemented.¹⁴ Therefore, increasing the number of electrodes and normalizing the sensitivity matrix may possibly improve localization errors in our context. However, in application of our method to infants, we believe that using a 16-electrode configuration is preferable than using 32 due to the small size of neonatal heads. In addition, the purpose of this context is to make a clear comparison of the two different layouts with the same approach. Therefore, we chose to use a simple, unfiltered sensitivity matrix. Although we have not used any weighting method, we still have a valid comparison between the two cases, and even with this simple approach, achievement of a 17% relative radial error is quite acceptable when compared to the minor diameter of a ventricle, which is about 30% of the head radius.

Investigations of Model Mismatch

Earlier studies^{1,3,16} showed that incorporating a more accurate forward model that matches the conductivity structures of the imaged object can improve image quality in general. For example, inclusion in forward models of a bone model,¹⁶ a four-shell head model,³ or including anisotropy¹ all demonstrated improvements in image quality, image resolution, and localization errors. However, realistic models need to be used carefully, since with inaccurate prior information, they may yield images worse than those reconstructed with a simple model.¹⁴ In Bagshaw *et al.*,³ it was noted that using a full-shelled head model produced worse images than using the homogeneous head model when reconstructing the head phantom data, because the resistivity ratio between the skull and brain region for the tank was too dissimilar to the value assumed in the model. Meeson *et al.*¹⁶ also concluded that using a uniform matrix seems less likely to generate artifacts when the underlying conductivity distribution is unknown. Bagshaw *et al.*³ also studied the effect of using RG in the forward model. They found that both localization accuracy and image quality were improved, but only slightly. In practice, the head geometry varies from patient to patient, which makes it difficult to find a forward model that is really accurate unless each model is constructed individually. In addition, the neonatal head can vary in shape depending on posture. Exactly how accurate a model should be is a matter of discussion.³⁷ Therefore, it is of great interest to investigate the quality of reconstructions and quantity index accuracy using a homogeneous spherical model as a forward model. Figure 9 shows that the QIs calculated using the spherical matrix were similar to those calculated using the accurate RG matrix with a maximum error of 0.0025 S cm² (0.4 mL) in both cases. The response curve was not exactly linear, and appeared to have a zero intercept at a volume larger than 0 mL. This suggests that sensitivity is low at very small volumes. Both spherical and RG matrix results tended to shift away from the expected QI when the total blood volume became larger than 1.5 mL. This tendency agreed with the observations from Sadleir and Fox²⁶ when using anomalies less conductive than the background. Both these inaccuracies may be caused by the non-linear characteristics of the EIT problem itself.³¹ As Fig. 9b shows, use of the RG reconstruction matrix did not produce an advantage over the spherical reconstruction matrix at all SNR levels. The largest QI variability found using the spherical matrix increased to 0.006 S cm² (corresponding to a blood volume of 0.95 mL) at 60 dB. However, this level of variability is not likely to lead to an incorrect diagnosis of type II IVH, in which the blood volume may be up to 3 mL.

Contrast values were similar in both reconstructions (S_{RG}) and for those reconstructed with the spherical matrix (S_{Sp}), whereas resolutions were slightly better for S_{RG} reconstructions than S_{Sp} reconstructions. The spherical matrix reconstructions could of course not localize anomalies as well as the RG matrix cases. However, images reconstructed with the spherical matrix showed the anomalies coherently, and allowed us to verify their location—something

that could be of practical benefit in verifying diagnoses made from QI information. In considering whether it is appropriate to use RG or spherical models for diagnosis in this application, it is important to bear in mind the hazards inherent in obtaining RG data in unstable neonates (involving transport to MRI or CT machine and the inherent risks during these steps, as well as the time needed to produce an RG model) and weighing them against the clinical need to confirm bleeding. It is probably true that use of a reconstruction based on an appropriate geometry will produce better images; however, we have shown here that postprocessing of images using both methods obtained similarly accurate QI values. If QI is used as a primary indication of bleeding status, images could be used as a secondary measure to verify the presence of a coherent anomaly and to discount the possibility of QI variations being caused by electrode contact problems or other equipment errors. While images from spherical reconstructions are not quite as clear as in those based on accurate models of the head, sufficient information should be present in these reconstructions to allow verification of diagnoses.

Other Possible Applications

Based on our observations in this study, the EEG current pattern may also have potential for use in diagnosis of other bleeding conditions such as intraparenchymal hemorrhage (IPH). In intraparenchymal bleeding, an increase in blood volume should result in a decrease in impedance, as blood (0.67 S/m) has a higher conductivity than brain tissue (0.172 S/m).⁷ Another possible application is the detection of acute stroke in neonates. Romsauerova *et al.*²⁴ studied the feasibility of using multi-frequency EIT for the detection of stroke with numerical models and phantom experiments. Both applications would be expected to be easier than the application to detection of blood in the ventricles, since they involve less central locations and blood becoming present in an area where it is effectively masked by surrounding CSF. Application of EIT methods in adult humans is in general made very difficult because of the barrier formed by the adult skull. However, exploitation of the fontanel space may make it quite practical to diagnose several conditions in brains of neonates with EIT.

CONCLUSION

Two conductivity reconstruction methods were compared on their ability to reconstruct, localize, and quantify anomalies with simulations and phantom experiments. The EEG current pattern was found to be the more robust than the RING current pattern, having an overall better quality in terms of reconstruction, localization, and quantification for objects in regions corresponding to the brain ventricles. Although the RING pattern could not distinguish the height (*z*-value) of an anomaly, it may be of some use in terms of volume estimation or *xy*-localization and in some applications it may be a more practical pattern to use because of the relative simplicity of this layout. Because of the small errors observed in volume estimates and because of uncertainties inherent in the use of RG models, we consider it is satisfactory to use a homogeneous spherical model as a general forward model for practical reconstructions and quantification. All the findings in this study indicated that the fontanel- based EEG current pattern has a great potential to detect and quantify IVH and may possibly be applied to pathologies such as IPH and hemorrhagic stroke on neonates.

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FIGURE 1.

Three electrode configurations viewing from the top and demonstrations of the current pattern applied in each case (a) RING layout and RING current pattern (b) EEG layout and EEG current pattern.

FIGURE 2.

(a) A spherical model with scalp, brain, ventricles, and skull shells, including a structure similar to the fontanel (highlighted). The figure at left is an axial view, showing the six positions occupied by a blood-like anomaly with a relative radius of 0.1 within the ventricles. The figure at right is a sagittal view, showing simulated ventricles placed at a relative radius 0.3 above the equator. Minor and major dimensions of the simulated ventricles were 0.3 and 0.88, respectively. (b) A neonatal head model with a realistic external geometry and internal structures. The figure at left shows the scalp, skull, CSF, and brain structures. The figure on the right shows interior ventricles and blood-anomaly structures. The relative sizes of the ventricles and blood-like anomalies were approximately the same as for the spherical model.

FIGURE 3.

(a) Spherical phantoms with EEG layout (left) and RING layouts (right). (b) The complete experimental system, showing IIRC-EIT system and phantom.

FIGURE 4.

Axial (relative radial position $z = 0.3$) and sagittal ($x = -0.2$) conductivity cross sections (first and second columns) and isosurfaces (third column) of (a) an blood anomaly at position 1 in the spherical model, and (b) reconstructions of EEG simulated data, (c) RING simulated data, (d) EEG phantom data, and (e) RING phantom data, all using reconstruction matrices based on homogeneous spherical forward models. Isosurfaces were calculated at half maximum voxel values within each image.

FIGURE 5.

Radial relative localization errors D_{xy} and D_z of the six anomaly positions for simulated and phantom data for RING and EEG patterns.

FIGURE 6. Quantification Indices of the six anomaly positions with simulated and phantom data for the RING and EEG patterns.

FIGURE 7.

Axial (relative radial position of $z = 0.3$) and sagittal ($x = -0.2$) conductivity cross sections (first and second columns) and isosurfaces (third column) showing (a) ideal image of bleeding at position 1 within a spherical model, and the reconstructions using the EEG current pattern on 'open skull' or 'closed skull' models at three different noise levels. The noise levels used were 50 dB (b) open skull and (c) closed skull, 65 dB (d) open skull and (e) closed skull and 75 dB (f) open skull and (g) closed skull. Isosurfaces were calculated at half maximum voxel values within each image.

FIGURE 8.

Axial (relative radial position of $z = 0.3$) and sagittal ($x = -0.2$) conductivity cross sections (first and second columns) and isosurfaces (third column) for (a) ideal bleeding at position 1 of the RG model and reconstructions of the data collected from the RG model using the (b) S_{SP} and (c) S_{RG} . Isosurfaces were calculated at half maximum voxel values within each image.

FIGURE 9.

(a) QIs (with the corresponding volume estimation labeled on the right) for the multi-site bleeding cases using the homogeneous spherical sensitivity matrix and the homogeneous RG matrix. The bleeding could accumulate at any single or combinations of the six bleeding sites.

There were $C_6^1 + C_6^2 + C_6^3 + C_6^4 + C_6^5 + C_6^6 = 63$ cases in total (8). (b) Comparison of average QI errors (with the corresponding errors in volume estimation labeled on the right) for the 63 cases using the two sensitivity matrices under 11 different SNR levels ranging from 60 to 90 dB. Error bars show the range of QI values obtained at each noise level.

TABLE 1

The condition number, rank, and size for the sensitivity matrix used for each current pattern, showing truncation points chosen for each reconstruction.

Current pattern	Ring (spherical)	EEG (spherical)	EEG (RG)
Condition number	1.43×10^{19}	2.84×10^{14}	6.3×10^5
Rank	104	104	104
Number of voxels	26126	21896	45702
Number of measurements	208	182	182
Truncation point (simulated/phantom data)	60/50	60/50	60/NA

TABLE 2
Comparison of resolutions R , in relative radial units, calculated for each of the cases shown in Fig. 4.

Anomaly position	1	2	3	4	5	6
EEG simulation	0.285	0.243	0.253	0.212	0.159	0.208
RING simulation	0.231	0.313	0.243	0.286	0.324	0.309
EEG phantom	0.276	0.268	0.281	0.230	0.137	0.260
RING phantom	0.259	0.326	0.211	0.242	0.177	0.274