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Electrode configurations for detection of intraventricular haemorrhage in the premature neonate

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Abstract

Intraventricular haemorrhage is a common cause of death in premature human infants. As preventative measures and treatments become available, a method for monitoring and detection is required. Electrical impedance tomography (EIT) is a viable monitoring method compared to modalities such as ultrasound, MRI or CT because of its low cost and contrast sensitivity to blood. However, its sensitivity to blood may be obscured by the low conductivity skull, high conductivity cerebrospinal fluid (CSF) and shape changes in the head and body. We estimated the sensitivity of three 16-electrode and impedance measurement configurations to bleeding using both idealized spherical and realistic geometry three-dimensional finite element models of the neonatal head. Sensitivity distribution responses to alterations in skull composition as well as introduction of conductivity anomalies were determined. Of the three patterns tested, a measurement scheme that employed electrodes at locations based on the 10–20 EEG layout, and impedance measurements involving current return over the anterior fontanelle produced superior distinguishabilities in regions near the lateral ventricles. This configuration also showed strongly improved sensitivities and selectivities when skull composition was varied to include the anterior fontanelle. A pattern using electrodes placed in a ring about the equator of the model had similar sensitivities but performed worse than the EEG layout in terms of selectivity. The third pattern performed worse than either the Ring or EEG-based patterns in terms of sensitivity. The overall performance of the EEG-based pattern on a spherical homogeneous model was maintained in a sensitivity matrix calculated using a homogeneous realistic geometry model.

Keywords

electrical impedance tomography; sensitivity; intraventricular haemorrhage; neonate

1. Introduction

While many aspects of birth outcomes have been vastly improved worldwide in the last 50 years, preterm birth is still the major cause of poor birth outcome (Goldenberg and Jobe 2001) and more than 50% of infants born at 24 weeks' or less gestation will die. Although survival rates of preterm infants have increased markedly, survivors of neonatal intensive care units suffer high rates of neurodevelopmental disability such as cerebral palsy and sensory impairment. A common cause of such disabilities is haemorrhage, hypoxia or ischaemia in the neonatal brain. Intraventricular haemorrhage (IVH) affects 35–50% of premature human infants who are born at 32 weeks' gestation and below (York and DeVoe 2002). Papile *et al* (1978) classified intra-ventricular haemorrhage into four grades (see table 1). Grades I and II are rarely associated with morbidity or permanent incapacity. Grades III and IV are associated with increased morbidity, conditions such as cerebral palsy (CP), periventricular leukomalacia (PVL) and post-hemorrhagic hydrocephalus (PHH).

1.1 Monitoring and treatment of IVH

While serious IVH may be detected from a bulging fontanelle, or encephalopathy of the patient, at present, the conventional diagnosis method for newborns at a risk of IVH is ultrasound (Hintz and O'Shea 2008). Ultrasound is 76–100% specific in at the detection of grade I lesions of greater than 5 mm width and almost 100% specific in grade III or IV haemorrhage. However, US detection of grade II haemorrhages (free blood in normal sized ventricles) is much less accurate (Babcock *et al* 1982, Mack *et al* 1981).

Low-dose indomethacin (Ment *et al* 1994), steroids or glutamine therapy are used experimentally as IVH preventatives or therapies (Anon 2004). As treatments become available, early warning of IVH becomes crucial.

1.2. Electrical impedance tomography

EIT monitoring involves the application of an electrode array (typically 16 to 32 electrodes) over the surface of an area of interest (Boone *et al* 1997, Cheney *et al* 1999). Sequences of current patterns are applied to a subset of electrodes, and voltages produced by the applied currents are measured. The currents employed are high frequency (about 50 kHz) and at a low enough magnitude that they are harmless. These measurements are used to create a cross-sectional maps of electrical impedance. Measurements can typically be made and images reconstructed rapidly. Although EIT images have a relatively poor spatial resolution compared to modalities such as MRI and CT they nevertheless contain valuable physical information, and can be collected with little cost. Unlike ultrasound, EIT can conveniently be used in monitoring mode. By comparing data collected at an earlier time to current data, images of impedance change, and therefore rates, of processes may be identified. Blood is generally easily detectable in the body because free blood has a significant impedance contrast with other tissues and fluids (Geddes and Baker 1967, Faes *et al* 1999). Determining absolute conductivity values is considerably more difficult and is hindered by other problems such as unknown boundary shape and the need to estimate electrode contact impedances. A major advantage of EIT in this clinical application is that it is non-invasive and not operator dependent. It may also be used in nurseries in remote areas or as a monitoring tool during transfer to larger care facilities.

EIT was briefly tested to the detection of neonatal intra-ventricular haemorrhage in the late 1980s (Murphy *et al* 1987, McArdle *et al* 1988), with a result showing detection of ventricular bleeding in one neonatal patient. These data were, however, significantly affected by artefacts. Murphy *et al* (1987) found that breathing movements and variations in cerebral blood flow caused the largest signal artefacts—up to 1% variation in their baseline signal at a frequency of approximately 0.5 Hz. The other significant artefact was synchronous with heart electrical activity, causing a variation of about 0.1% in signals. Large artefacts due to movement of the infant were also encountered. Nevertheless, these early studies of IVH using EIT demonstrated the potential of EIT for this application. More recently, researchers in EIT of the head have concentrated on the accurate detection of the position of epileptic foci, stroke and on functional studies, amongst other applications, and not on IVH (Tidswell *et al* 2001a, 2001b, Liston 2004). Their studies typically used 31 electrode patterns based on the 10–20 EEG system, with collection of data from all available configurations.

1.3. Sensitivity target

MRI and ultrasound have been used to measure volumes of lateral ventricles *in vivo* (Schierlitz *et al* 2001, Brann *et al* 1990, Gilmore *et al* 2001). Schierlitz *et al* (2001) measured lateral ventricle volume varied from 3.94 to 12.42 ml in a population of infants with gestational ages (GA) between 24 and 40 weeks. The correlation between GA and ventricular size found in the study was approximately linear (with correlation coefficient $r = 0.56$). Therefore, grade I bleeding constitutes ventricular bleeding of less than about 2 ml. The approximate diameter of

a preterm skull, determined from examining a set of neonatal and preterm MRI scans provided by the Department of Radiology at the University of Florida, is around 12 cm. Thus, an imaging technique that detects a grade I bleed must be capable of finding an amount of blood occupying less than 0.1% of skull volume. Further to this, the skull forms a barrier to electric signals, and the presence of this and cerebrospinal fluid surrounding the brain will combine to reduce the signal levels or sensitivity of external measurements to the presence of bleeding. Shape changes in the soft infant skull, breathing and cardiac activity will also tend to obscure these data. In order to optimize the threshold of detection for IVH, it is important to determine an EIT measurement configuration that is capable of producing a good signal from blood appearing in the centre of the head, and that is suitably configured to the anatomy of the neonate. Here we compare three 16-electrode layouts and measurement protocols on their ability to detect conductivity anomalies appearing in the centre of an idealized skull. To this end we compare sensitivity in the presence of blood anomalies in 3D models with and without bleeding, and in background conductivity configurations specific to the neonatal skull. We also compare the performance of one-electrode layout using sensitivity matrices calculated using both spherical and realistic neonatal head models.

2. Methods

Although blood has a significant conductivity contrast with other body tissues, the effect of the skull and cerebrospinal fluid (CSF) surrounding the brain, as well as electrical noise, will tend to obscure the effect of blood near the head's centre. *In vivo*, the effective sensitivity to the appearance of blood in the ventricles will be further reduced because of cardiac, breathing and motion artefacts.

Inspection of anatomical MRI scans of the neonatal head indicate that the ventricles are centrally located and if translated to a spherical geometry would lie in the upper hemisphere at an elevation of around 0.3–0.5 of the sphere radius into the upper hemisphere, as shown in figure 2(a). Therefore, we chose to study the sensitivities and contributions total measured resistances in both regions on the central axis and in regions having the same approximate size and location of the neonatal ventricles. The two central regions were spheres with relative radii of 0.25 compared to the background sphere radius and were centred at $z = 0.33$ and $z = 0.50$. We refer to the sphere at $z = 0.50$ as the *upper* sphere and the other as the *lower* sphere. The two lateral ventricles, approximated as ellipses, were located in relative radial terms at $(-0.2, 0, 0.3)$ and $(0.2, 0, 0.3)$ each having semiaxis lengths of 0.15, 0.44 and 0.15 in x , y and z axes, respectively. The four regions are illustrated in figure 2(a). The spherical regions (about twice as large as the ventricles) were chosen to compare with the lateral ventricle region properties and to determine the uniformity of each pattern.

2.1. Electrode configurations

Three 16-electrode configurations were compared on their sensitivity in a centrally located anomaly within the upper hemisphere of a homogeneous spherical model. A homogeneous spherical model was used as a simple means of determining the relative strengths and weaknesses of the methods and the properties displayed by each method may not be maintained in measurements made using realistic geometry models or *in vivo*. The three methods compared were a standard pattern involving a ring of equally spaced electrodes placed on the equator, a pattern based on the 10–20 EEG electrode layout and a third pattern with one electrode at the apex of the head (Cz) and the other 15 equally spaced around the equator. The three configurations are illustrated in figure 1. The *Ring* pattern shown in figure 1(a) has 16 electrodes placed equidistantly about the equator of the sphere, and is included here for comparison with other standard works in EIT and other work on EIT of the head (Gibson 2000, Murphy 1987). We used an adjacent current injection and voltage measurement protocol with this layout. The

adjacent current injection creates an approximate dipole field pattern. *EEG* electrodes were (in numbered order) T7, C3, P3, F3, O1, FP1, Oz, Pz, Cz, Fz, O2, FP2, P4, F4, C4, T8, with spherical coordinates as specified by EEG instrumentation manufacturer BioSemi B.V. (Amsterdam, Netherlands, www.biosemi.com). Locations of *EEG* electrodes in spherical coordinates are shown in table 2. We suspected that the location of an electrode over the anterior fontanelle (approximately at Cz) would produce an improved signal-to-noise ratio for centrally located anomalies. Therefore the measurement protocol used with this method involved currents sequentially applied between the electrode at Cz and all other electrodes, while voltage measurements are made between adjacently numbered electrodes. The electrode labelling shown in figure 1(b) was chosen to label from left to right and anterior to posterior, respectively, and was not chosen to be an optimal numbering scheme. The *CzRing* pattern shown in figure 1(c) has 15 electrodes placed around the equator perimeter and one at Cz. We applied all currents between the electrode at Cz and one of the perimeter electrodes, and measured voltages between available adjacent electrodes on the equator.

2.2. Sensitivity matrix calculations

The *sensitivity* of a particular measurement is defined as the dot product of lead fields formed where the source is applied through one of the two measurement pairs with the lead field generated using the second pair as a source, integrated over the anomaly volume (Geselowitz 1971, Lehr 1972). Thus we have

$$\frac{\delta Z_i}{\delta \sigma_j} = S_{ij} = - \int_{v_j} \frac{\nabla \phi(\sigma_j)}{I_\phi} \frac{\nabla \psi(\sigma_j)}{I_\psi} dv_j, \quad (1)$$

where i denotes a measurement and j is the region of interest. The gradient $\nabla \phi$ is that obtained when current is applied through the current injection electrode pair and $\nabla \psi$ is the gradient when current is applied through the measurement pair. δZ_i denotes the change in transfer impedance of the i th measurement.

The contribution of a particular region j to the i th impedance may be calculated by including the conductivity of the region:

$$Z_{ij} = - \int_{v_j} \sigma_j \frac{\nabla \phi(\sigma_j)}{I_\phi} \frac{\nabla \psi(\sigma_j)}{I_\psi} dv_j. \quad (2)$$

We created three spherical, uniform, second-order tetrahedral finite element models using Comsol (Comsol, Burlington MA) containing large numbers of elements (about 25 000) and defined electrode layouts and measurement schemes upon each one. Numbers of elements in each model differed because of the electrode locations but were in the range 20–30 000. We also calculated a sensitivity matrix using the *EEG* pattern and a uniform realistic geometry model. Sensitivity matrices were calculated for each scheme using the current application and voltage measurement schemes shown in table 3. Exact numbers of elements, condition numbers and ranks of the three matrices are also shown in table 3. Each matrix had a rank of 104, which is to be expected from the linearly independent number of measurements that can be made on a 16 electrode array when current application electrodes are excluded from measurements.

The condition number of the *Ring* measurement scheme was higher than for *CzRing* or *EEG* configurations, but similar to those found in two-dimensional matrices using this topology (Sadleir *et al* 2008).

2.3. Realistic geometry comparison

We compared sensitivity and specificity results obtained using the spherical EEG layout with those obtained from a sensitivity matrix calculated using a homogeneous realistic geometry model and this layout. This was done to determine the extent of differences that would be expected using this method *in vivo*, and also provides some insight into how differently the other candidate methods may perform on more accurate models. In making comparable regions to those in spherical models, we averaged maximal x , y and z deviations from the model anatomic centre and used these values to calculate approximate elliptical semi-axes for the model. These values were then used to create elliptical regions with the same relative locations and semi-axis values to those used in spherical background models. These regions are illustrated in figure 2(c). The total volume of the realistic geometry model was 790 ml, which compared well with values estimated from locally archived neonatal MRI images. Volumes of 11.1 and 11.0 ml were calculated for lower and upper central regions, respectively. Both left and right ventricle volumes were about 4.2 ml, similar to those observed *in vivo* (Gilmore *et al* 2001). Electrode locations were chosen to be similar as possible to those used on the spherical background model while adapting to the realistic head's orientation. The Cz electrode was shifted off the central axis to determine the effect of electrode location variation on the model's symmetry. The sensitivity matrix calculated using the EEG layout on a uniform realistic geometry model had 45 702 elements, and had a condition number of 6.3×10^5 . As for all other models, its rank was around 104.

2.4. Effects of layered head structure

Because the effects of the outer head composition may be significant, we also evaluated the sensitivity performance of the EEG-based pattern using a partial prior spherical model that incorporated layers of skin, skull (with and without the fontanelle) and CSF, the centre of the head being assigned to have the same conductivity as brain. Layer thicknesses were chosen according to data gathered from inspection of archived neonatal MRI models. The diameter of the model was chosen to be 11.9 cm (a volume of 884 ml), with a scalp thickness of 2.9 mm, skull thickness of 6.0 mm and CSF layer thickness of 3.0 mm. The archived MRI data set used as a reference had a total volume of 921 ml. A 41 626 element (62 118 node), tetrahedral finite element model was created using the layer scalings noted above. The model outline is shown in figure 2(b). Regions were assigned conductivities found in the literature: scalp (0.43 S m^{-1} , Burger *et al* 1943); CSF (1.3 S m^{-1} , Oostendorp *et al* 2000); brain tissue (0.172 S m^{-1} , Stoy *et al* 1982) and blood (0.67 S m^{-1} , Geddes and Baker, 1967). We chose the skull conductivity to be either that of adult skull (15 mS m^{-1} , Oostendorp 2000) or piglet skull (60 mS m^{-1} , Sadleir and Argibay 2006). We chose to use the conductivity for neonatal piglet skull in the absence of values for human neonatal skull in the literature, and because the neonatal piglet will be used as an animal model in future tests of this technique. The layered model was tested with the EEG layout only. In order to determine the interaction between the presence of the fontanelle and the EEG measurement layout, the part of the head designated as fontanelle in figure 2(b) was either chosen to have the conductivity of skull or skin, corresponding to either 'closed' or 'open' skull conditions respectively. Sensitivity matrices were calculated for each of these configurations. We found the condition numbers for closed skull cases were generally lower than for open skulls, with values for adult skull configurations being about 8.8×10^{15} and 1.6×10^{16} , respectively; and for piglet skulls 3.5×10^{15} and 4.2×10^{15} .

2.5. Data presentation

A useful current pattern should produce both a large selectivity and sensitivity in the region of interest, and produce a consistent response in the neighbourhood of the region of interest.

We compared the sensitivities, *selectivities* and maximum and half sensitivity volumes of the current patterns within both central and lateral ventricular regions. *Selectivity* denotes the

contribution of a region of interest to the total measured transimpedance. The reference transimpedance for a particular four-electrode configuration may be obtained from sensitivity matrix entries by summing all entries over all pixel locations. The contribution to the i th measurement from a particular region of interest R may be extracted by summing sensitivity matrix entries over this volume:

$$\text{Sel}_{i,R} = \frac{\int_{V_R} \sigma \frac{\nabla \phi(\sigma)}{I_\phi} \frac{\nabla \psi(\sigma)}{I_\psi} dv}{\int_{\Omega} \sigma \frac{\nabla \phi(\sigma)}{I_\phi} \frac{\nabla \psi(\sigma)}{I_\psi} dv} = \frac{\sum_{i \in R} Z_{ij}}{\sum_{j=1}^{TE} Z_{ij}}, \quad (3)$$

where Ω denotes the entire measurement volume and R is the region of interest. We investigated both sensitivity and selectivity characteristics in both axially located and approximate lateral ventricle structures. Analysis of both central and lateral structures allows us to determine the consistency of each methods' response as location of bleeding, the ventricles and electrodes may vary.

Sensitivity values were expressed as the mean sensitivity over the region of interest as a fraction of the maximum sensitivity over the entire domain, for each measurement configuration in the scheme. The *selectivity* (Kauppinen *et al* 2005, Yang and Patterson 2008) is the fraction of the total measured resistance contributed by the central region, and the half and negative sensitivity volumes measured over the sphere. The half sensitivity volume is defined as being the volume of the sphere that has more than half the maximum sensitivity for a particular pattern. The negative sensitivity volume is the volume of a particular measurement pattern occupied with negative sensitivity, i.e. the volume of elements that produce contributions to the total measured resistance in an opposite direction from that expected. In most four electrode measurement schemes there will be significant negative sensitivity regions (Kauppinen *et al* 2005, Yang and Patterson 2008). In principle, negative sensitivity is not bad. If an appropriate model is used, it should be possible to relate any observed change to the correct sign of conductivity change. However, negative and positive sensitivities will sum to form the total measured resistance, and it is possible that for some current patterns location of negative sensitivities may be a problem and produce contradictory information. This could occur if for example there is a mismatch between model and experimental conditions. The half sensitivity volume gives an indication of the breadth of the positive sensitivity distribution.

3. Results

3.1. Dynamic range and resistances of models

The largest resistances measured on uniform models at the scale of spherical head models were about 3.2 Ω for the *Ring* pattern, 4.94 Ω for the *CzRing* pattern and 13.9 Ω for the *EEG* pattern. Symmetries in the *CzRing* and *EEG* patterns produced several transimpedance values very near zero, 15 for the *CzRing* pattern (one for each current configuration) and 24 for the *EEG* pattern. Excluding these very small values, dynamic ranges observed in data were around 20, 50 and 10 for *Ring*, *CzRing* and *EEG* patterns respectively. We observed a total of 20 low measurements in the realistic geometry EEG sensitivity matrix. Eliminating these values, the dynamic range of measurements on this model was around 34. Analyses of selectivities presented below do not include normalization by these small impedance values, and we have set these values to be zero in subsequent selectivity plots.

3.2. Sensitivity in a single region of interest

Sensitivities to bleeding in the central part of the head were investigated by computing sensitivities in the regions of interest. This is similar to the approach used by Kauppinen *et*

al (2005) where sensitivities in the centre of a disk were compared using different measurement configurations. Sensitivity maps for the four regions of interest are shown in figure 3 for all three patterns on the uniform sphere. For a significant number of measurements sensitivity was negative. The *Ring* pattern had the most positive sensitivity values. The average sensitivity of the *EEG* pattern was greater than either *Ring* or *CzRing* patterns for the upper sphere, and comparable to the *Ring* pattern over the lower sphere. However, the *EEG* pattern had significant regions of negative sensitivity. On the basis of sensitivity alone, the *Ring* pattern was superior to *CzRing* and *EEG* patterns, as it demonstrated mostly positive values in the regions of interest. The *CzRing* pattern had reasonable amplitude sensitivities but demonstrated an average negative sensitivity for both regions of interest. Examination of figure 3(c) and (d) shows that in the ventricles *EEG* pattern sensitivities were similar to those in the lower sphere and that there were significant negative sensitivities. The *Ring* pattern sensitivity peaks were much larger than the *EEG* pattern peaks for both ventricle samples. The *CzRing* pattern plots for simulated ventricles shifted considerably compared to the spherical region results and there are many large negative sensitivity values.

3.3. Selectivity

Whereas the sensitivity of the models reflects the interaction between current and measurement pair geometries in the model, the selectivity shows the contribution from the target region to measured voltages and transimpedances. Absolute selectivities for the uniform spherical model within the regions of interest are shown in figure 4. On this comparison, the *EEG* pattern is clearly better than either *CzRing* or *Ring* layouts, with up to 15% of total resistance contributed by the upper sphere and 7–10% of total resistance typically contributed by the lower sphere. *EEG* pattern selectivities in the ventricles were slightly smaller than those observed in the lower sphere, with average values being around 4%, but these values were on average larger than those observed for the *Ring* pattern. Selectivities for the *CzRing* pattern in the ventricles were also relatively high, averaging about 3% with peaks up to 6% of total resistances.

3.4. Half and negative sensitivity volumes

Negative sensitivity volumes for all spherical background models were large, averaging around 50%. The negative sensitivity volumes of the three patterns were around 70%, 35% and 53% for *Ring*, *CzRing* and *EEG* patterns, respectively. By contrast, half-sensitivity volumes were usually very small, being about 0.6% for the *Ring*, 0.3% for *CzRing* and 0.2% for the *EEG* pattern. The results broadly agree with sensitivity analyses performed using four electrode configurations in 2D (Kauppinen *et al* 2005) and in 3D (Yang and Patterson 2008).

3.5. Realistic geometry comparison

Figure 5 shows (a) sensitivities and (b) selectivities for the *EEG* pattern in the ventricles and upper and lower spheres of the uniform realistic geometry model. These figures may be compared directly with *EEG* pattern results obtained for uniform spherical models shown in figure 3 and figure 4. Both sensitivities and selectivities were very similar to those observed in the spherical model. Volumes of central regions were around twice the ventricle volumes in the realistic geometry model and this is reflected in the absolute selectivity plots shown in figure 5(b). The relatively large sensitivity and selectivity in the left ventricle likely reflects the asymmetry of *Cz* electrode placement. Overall sensitivities were about one third of spherical geometry values in the realistic geometry model than in the spherical model for all four regions of interest. Maximum selectivities in the realistic geometry sensitivity matrix were similar in all four regions to those found for the spherical model matrix, but fewer large selectivities were observed.

3.6. Layered head composition

We compared sensitivity matrices calculated with or without included fontanelle in the layered model shown in figure 2(b). Comparison was made of the average value found in the upper spherical region and the maximum sensitivity found within the brain compartment of the model (within $r = 0.8$). Results for sensitivity and selectivity measures are shown in figure 6 and figure 7. We found that sensitivities were generally much larger for open skull than closed skull models. Expressing the sensitivity in the upper sphere as a fraction of maximum values observed within the whole model, the open skull sensitivity for an adult skull was around 17 times greater than for a closed skull, and around 1.9 times greater comparing open and closed piglet skulls. Considering sensitivities within the upper sphere compared with maxima observed within the brain compartment, open adult skulls had sensitivities in this region about 2.4 times that of closed skulls, and open piglet skulls had sensitivities about 25% larger than closed skulls.

Comparison of selectivities showed that measured voltages over the skull fontanelle had a much larger contribution from the brain compartment, as expected. In figure 7 we demonstrate selectivities by showing the resistances contributed by the both the entire domain and the brain compartment alone.

Figure 7 shows total measured resistances (left, top and bottom) and (right, top and bottom) contributions to the total resistance from the brain compartment ($r \leq 0.8$) for open and closed skull models with the skull comprised of (top) adult skull with conductivity of 15 mS m^{-1} or (bottom) piglet skull with conductivity of 60 mS m^{-1} . As expected, measured base resistances with closed skulls were generally larger than those with open skulls. For both skull conductivity conditions, the relative resistance contributed by the brain compartment was larger for open than closed skulls. Therefore, the selectivity of the *EEG* measurement strategy was enhanced using an open skull geometry.

Selectivity in the smaller upper spherical region was also enhanced. We found that for open adult skulls, average selectivities were improved about 20% within this region, and selectivities were improved about 5–10% for open piglet skulls.

4. Discussion

In this study we have assessed the sensitivity and selectivity parameters of three four-electrode measurement strategies. Detection of bleeding in the head will be compromised by the presence of electronic noise and the presence of *in vivo* noise sources such as cardiac, breathing and movement artefacts. This study does not take into account the presence of noise. However, the sensitivity and selectivity parameters of the different measurement strategies can be used to predict their ability to detect changes in the ventricular regions. A robust measurement method should have both high selectivity and high sensitivity in the region of interest. We have found that the *EEG* measurement pattern had similar magnitude sensitivities to the *Ring* pattern in the two spherical regions of interest, but higher selectivities. In addition, the *EEG* pattern (eliminating small values) had both higher maximum measured transimpedances and a lower dynamic range than the *Ring* pattern. We found that similar sensitivity and selectivity patterns were found in the realistic geometry model and spherical models using the *EEG* pattern, although sensitivities were somewhat smaller in the realistic geometry model.

Although the *Ring* pattern had a stable sensitivity distribution and good selectivity, it cannot be used to determine three-dimensional anomaly locations because of its planar symmetry. The presence or absence of the fontanelle, and skull conductivity greatly affect sensitivity distributions. The *CzRing* pattern had lower sensitivities than the *Ring* pattern in all cases, with the average sensitivity being negative for both regions of interest tested. Selectivities of the

CzRing pattern were comparable to those of the *EEG* pattern, but the *CzRing* pattern's overall negative and variable sensitivity in central regions indicates it may be unsuitable for practical measurements.

Because of the symmetry of the EEG layout, there were several very low values for resistance predicted (24 of the 182 used). In our analysis of selectivity we chose to ignore these data. In the realistic geometry model we found 20 low values (less than 5% of the maximum measurement), with many of these measurements the same as in the spherical background model. Accordingly, contributions of these measurements to reconstruction quality will still be marginal, even with sensitivity matrices calculated using realistic geometries. We expect that broadly similar patterns to those obtained for spherical model-based matrices would be found when testing the *Ring* or *CzRing* patterns on realistic geometries. For example, the negative and low sensitivity values found for the *CzRing* method would also be anticipated in a realistic geometry model as these are a consequence of measurement and current field patterns being orthogonal or opposed in central regions for many measurements. While the *EEG* pattern seems promising, it is possible that there are superior measurement strategies that can be devised using the same layout, but avoiding measurement of near-zero values. However, the number of independent measurements possible with this number of electrodes (104) will always limit the available information.

The increased ability of the method to detect changes within the brain and ventricle areas when currents were returned using an electrode at Cz provides a pointer to methods for efficiently delivering current through the skull to target areas. In this application there are definite advantages to applying current to the anterior fontanelle. It is possible that applying currents via other skull openings could have advantages in other applications, particularly where there are natural or surgically created openings in the adult skull.

5. Conclusion

Electrical finite element models of the head were used to compare utilities of three different electrode layouts and measurement strategies to detect conductivity changes occurring near the lateral ventricles. The effect of changes in head composition was also assessed. Both the *Ring* and *EEG* patterns were found to perform well in terms of sensitivity and selectivity, with the *EEG* pattern having generally superior selectivity performance. Further work is needed to determine an optimal configuration for measurements of central conductivity changes in the neonatal head.

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Figure 1.
Three-dimensional spherical models showing (a) *Ring*, (b) *EEG* and (c) *CzRing* electrode layouts. Anterior and posterior surfaces are marked in (b).

Figure 2.

(a) Three-dimensional EEG spherical models showing central and lateral ventricle regions. The spheres each have relative radii of 0.25 and are centred at $z = 0.33$ and $z = 0.5$. The elliptical lateral ventricle structures had relative radial locations of $x = (-0.2, 0, 0.3)$ and $(0.2, 0, 0.3)$ and semiaxes of 0.15, 0.44 and 0.15 in x , y and z axes, respectively. (b) Layered spherical model with EEG layout. Conductivities used in the layered model were CSF (1.3 S m^{-1}); scalp (0.43 S m^{-1}); skull (15 mS m^{-1}); brain (0.127 S m^{-1}). The fontanelle was either chosen as skull or skin. (c) Axial, sagittal and coronal views of realistic geometry model, showing ventricular and central structures.

Figure 3.

Average sensitivities, calculated as a fraction of maximum values observed over the domain, for EEG, Czring and Ring patterns in (a) upper and (b) lower spheres; and (c) left and (d) right ventricle structures shown in figure 2(a).

Figure 4. Absolute selectivities, calculated using (3) observed within a relative radius ≤ 0.8 , for *EEG*, *Czring* and *Ring* patterns in (a) upper and (b) lower spheres; and (c) left and (d) right ventricle structures shown in figure 2(a).

Figure 5. (a) Sensitivities and (b) selectivities in central and ventricular regions found using a uniform realistic geometry sensitivity matrix based on figure 2(c).

Figure 6.

Sensitivities in the upper spherical region for the *EEG* pattern in a prior model with either open or closed skull having the conductivity of (a) adult or (b) piglet skull, or sensitivities in left ventricle region with open or closed (c) adult or (d) piglet skull. Open skull models had the fontanelle assigned to the same conductivity as skin. Conductivities used in the layered model were CSF (1.3 S m^{-1}); scalp (0.43 S m^{-1}); skull (15 mS m^{-1} (adult), 60 mS m^{-1} (piglet)); brain (0.172 S m^{-1}).

Figure 7.

Resistances of two region for the *EEG* pattern in a prior model with either open or closed skull having the conductivity of (above) adult or (below) piglet skull. Open skull models had the fontanelle assigned to the same conductivity as skin. Plots in the left show total resistance over the entire model (measured trans-impedances), those in the right show resistance over the brain compartment only.

Table 1

Grades of intraventricular haemorrhage defined by Papile *et al* (1978), with associated risks, prevalence and ultrasound specificities.

Grade	Description	Prevalence	Consequences	Detection
I	Isolated germinal matrix haemorrhage with no extension to the ventricles	12–51% of infants at GA < 33 weeks	Unlikely to lead to morbidity or disability	75% of cases detectable using ultrasound
II	Intraventricular haemorrhage with normal ventricular size; blood occupies up to 50% of ventricular volume		Unlikely to lead to morbidity or disability alone	~50% of cases detectable using ultrasound
III	Intraventricular haemorrhage with ventricular dilatation; blood occupies more than 50% of ventricular volume	20–25% of infants at GA < 30 weeks	Increased risk of CP, PHH and PVL Morbidity possible	75–100% of cases detectable using US
IV	Intraventricular haemorrhage with parenchymal involvement; ventricles often dilated		Increased risk of CP, PHH and PVL High risk of morbidity	75–100% of cases detectable using US

Table 2
Spherical coordinates of EEG electrodes, in degrees. T7 is left temporal and T8 right temporal. Oz is Occipital central electrode.

Number	Electrode	Azimuth	Latitude	Number	Electrode	Azimuth	Latitude
1	T7	-92	0	9	Cz	0	0
2	C3	-46	0	10	Fz	46	90
3	P3	-60	51	11	O2	92	-72
4	F3	-60	-51	12	FP2	92	72
5	O1	-92	72	13	P4	60	51
6	FP1	-92	-72	14	F4	60	51
7	Oz	92	-90	15	C4	46	0
8	Pz	46	-90	16	T8	92	0

Table 3
Properties of uniform *CzRing*, *EEG* and *Ring* sensitivity matrices defined on spherical models.

Layout	No. of elements	Condition number	Measurement scheme	No. of measurements	Rank
Ring	29 804	1.3×10^{19}	Adjacent	208	104
EEG	21 896	2.8×10^{14}	Cz/adjacent	182	104
CzRing	31 280	9.7×10^{14}	Cz/adjacent	195	104