The Cardiff Mk2b MIT head array: optimising the coil configuration

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Abstract A hemispherical MIT helmet coil array for imaging cerebral haemorrhage has been designed using a realistic 12-tissue finite-difference model of the head including a large peripheral haemorrhage (volume 49 ml). The coil array was first optimised by reaching a compromise between the quality of the reconstructed images and the financial cost of the digital detection system. The practical implementation of the helmet is partially complete.

1. Introduction
Recently, several groups have proposed the use of a hemispherical MIT array of coils for imaging cerebral stroke. Such an array will conform more closely to the shape of the head than a conventional cylindrical array and should increase the inductive coupling to the brain and hence the sensitivity to conductivity changes within it. Zolgharni et al [1] modelled a hemispherical coil array consisting of 56 coils using a finite-difference method combined with an anatomically-realistic head model comprising 12 tissue types. Frequency-difference images (1 – 10 MHz) were reconstructed from the modelled data with different levels of the added phase noise and two types of a-priori boundary errors that are likely to arise in a practical imaging system. It was concluded that a noise level of 3 m° (standard deviation) was adequate for obtaining good visualisation of a large peripheral haemorrhage (volume ≈ 50 ml). The simulations further showed that quite small discrepancies between the true skin surface and that used for computing the sensitivity matrix could give rise to significant artefacts on the images. Eichardt et al [2] have recently confirmed the better sensitivity distribution of the hemispherical arrangement than the cylindrical, using either 16 or 31 pairs of exciter and sensor coils. Zheng et al [3] have reported the construction of a hemispherical helmet MIT device using 15 axial gradiometers. Measurements (40-120 kHz) were made of a bowl filled with saline solution representing the brain and an immersed agar block of higher conductivity representing a haemorrhage. The inverse problem was not solved but 2D pseudo-images of the conductivity distribution at different frequencies were generated by interpolation.

In this paper we report the first stage of the practical realisation of the array of Zolgharni et al [1].

2. Concept of the helmet array
Following our earlier simulations and the work of others, the helmet was planned as an array of coils (separate excitors and sensors) positioned on a hemisphere, inside an outer, concentric, hemispherical, electromagnetic screen: this type of screen is known to be effective in rejecting extraneous interference and minimising inter-coil capacitive coupling [4]. An inner, plastic, hemispherical shell was to be fixed inside the coil array so as to completely enclose and protect the coils.
3. Determining the diameter of the plastic shell

The inner plastic shell needs to be small enough to allow the coils to be close to the scalp, in order to maximise the MIT signals, and large enough to accommodate most adult heads. The 1988 U.S. Army Anthropometry Survey [5] provides head measurements from 1774 males and 2208 females. The most relevant dimension is the anterior–posterior distance from between the eyebrows to the back of the cranium. The measurement fell within 209.5 mm for 96.5% of the males and 99.95% of the females. A further margin of 10 mm was allowed for the later addition of cushioning, leading to an inside diameter of 230 mm. The coils were positioned further out by another 10 mm, on a hemisphere 250 mm in diameter.

4. Determining the number of coils

The simulations of Zolgharni et al [1] configured the 56 coils as 28 exciters and 28 sensors. The new MIT electronics developed by our group can achieve a sufficiently low level of phase noise (1 m°) but is not compatible with the use of 56 coils. The number of exciters must be a multiple of 16 as the power multiplexers have 16 channels. The number of sensors must be a multiple of 7 because each PXI measurement board has 8 inputs, one of which is used for a phase reference [6]. The number of coils in the array determines the maximum coil diameter that will fit onto the hemisphere. Three arrays were considered: 30 coils (16 exciters, 14 sensors, coil diameter 45 mm); 46 coils (32 exciters, 14 sensors, coil diameter 35 mm); 60 coils (32 exciters, 28 sensors, coil diameter 30 mm). The realistic 12-tissue head model was discretized into a finite-difference mesh of 2.5 mm voxels for computing the simulated MIT signals. For speed of computation, the electromagnetic screen was initially not taken into account. The sensitivity matrix was computed by setting the conductivity to a uniform value of 1 S m⁻¹ throughout the head.

For the 30-coil array, the number of possible ways of arranging the sensors and exciters is \(30C_{14} = 30!/16!14! = 145,422,675\). Admitting only those arrangements symmetrical about the sagittal and coronal plans passing through the centre of the helmet reduces the number to 35. These were tested for quality according to the method of Gursoy and Scharfetter [7]. The quality factor, \(Q_r\), (equation 12 in their paper) was computed for each coil arrangement: \(Q_r\) is a measure of the orthogonality of the rows of the sensitivity matrix and reflects the independence of the different sensor/exciter combinations. For the helmet, the values of \(Q_r\) spanned only a small range, 0.775-0.799, indicating that there was no clear best arrangement. A distributed arrangement was therefore chosen for this and for the 46- and 60-coil arrays (fig 1).

MIT measurements were then simulated at frequencies of 1 and 10 MHz and 1 m° (SD) Gaussian phase noise was added. Frequency-difference images were reconstructed by a single-step Tikhonov-regularized method. The value of the regularisation parameter, \(\lambda\), was chosen so as to minimise the error

\[
E = \frac{\|\Delta\sigma_{\text{true}} - \Delta\sigma_{\text{rec}}\|}{\|\Delta\sigma_{\text{true}}\|}
\]

where \(\Delta\sigma_{\text{true}}\) is the true volume distribution of conductivity changes between 1 and 10 MHz and \(\Delta\sigma_{\text{rec}}\) is the reconstructed image. Full details of the forward and inverse solutions and the tissue conductivities are provided elsewhere [8]. From a qualitative assessment of the images, the visualisation of the simulated haemorrhage improves when the number of coils is increased from 30 to 46, but there is less improvement by increasing the number of coils further to 60 (fig 1). The 46-coil array was chosen for practical implementation. Adding a further 14 sensors to form a 60-coil array would have increased the cost of the PXI digital detection system substantially.

5. Setting the diameter of the electromagnetic screen

Our previous work has shown that a standoff distance of 60-80% of the coil diameter gave reasonable suppression of inter-coil capacitive coupling without excessively damping the inductive signals [4]. A standoff distance of 32 mm was chosen (91% of the diameter) as this allowed the use of a
commercially available size of spun aluminium hemisphere (outside diameter 320 mm, thickness 3 mm - fig 2a). Fig 2b shows the first stage of construction of the helmet containing the 46 coil formers fixed inside the aluminium screen and the vacuum-formed inner plastic shell.

6. Simulating the electromagnetic screen

The electromagnetic screen cannot be simulated explicitly by the finite difference model which assumes a skin depth large compared with the dimensions of the conductive region. Whilst this is true for the head, it is not for the screen (skin depth \( \approx 85\) \(\mu\)m at 1 MHz). The effect of the screen can however be included by making use of the expression derived by Kistis et al [9] for the voltage change, \(\Delta V\), in a sensor coil, \(a\), when an exciter coil, \(b\), is activated with a current, \(I\), at a frequency, \(\omega\).

\[
\Delta V = \frac{j\omega}{I} \int_v A_v \cdot J_b dv
\]

where \(J_b\) is the current density induced in the target (the head) and the integral is performed over the volume of the target, \(v\). \(A_v\) is the vector potential induced within the empty coil array, with no conductive target present, when now coil \(a\) is used as the exciter. The vector potential field within the empty coil array was pre-computed by Comsol Multiphysics for each of the 46 coils activated in turn as the exciter. \(A_v\) is then known directly and \(J_b\) is computed with the finite difference model using the vector potential, \(A_b\), pre-computed for coil \(b\) as the exciter. Consistent with the assumption of a large skin depth in the tissue is that in computing the eddy-current distribution, the finite difference model makes the approximation that the vector potential is always equal to the applied value that exists with no tissue present: the model neglects the contribution to the vector potential produced by the eddy currents in the tissue themselves. The reason for using the finite-difference model at all, rather than the
potentially-more-accurate Comsol Multiphysics, was its much shorter computation time, by a factor of approximately 60.

Qualitatively, the frequency-difference image reconstructed from the data simulated including the e.m. screen allows at least as good a visualisation of the stroke as before (fig 1, column 5).

7. Conclusions

Simulations have shown that for a realistic level of random phase noise, it should be possible to visualise a large peripheral cerebral haemorrhage by frequency difference MIT imaging (1-10 MHz) with the 46-coil helmet. The construction of the helmet continues. The simulations assumed that the boundary of the conducting region (skin surface) was known precisely. Development of a boundary-measuring system meeting the requirements set by Zolgharni et al [8] is an essential next step towards clinical application of the MIT helmet.

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