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Influence of body parameters on gastric bioelectric and biomagnetic fields in a realistic volume conductor

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Abstract

Electrogastrograms (EGG) and magnetogastrograms (MGG) provide two complementary methods for non-invasively recording electric or magnetic fields resulting from gastric electrical slow wave activity. It is known that EGG signals are relatively weak and difficult to reliably record while magnetic fields are, in theory, less attenuated by the low-conductivity fat layers present in the body. In this paper we quantified the effects of fat thickness and conductivity values on resultant magnetic and electric fields using anatomically realistic torso models and trains of dipole sources reflecting recent experimental results. The results showed that when the fat conductivity was increased there was minimal change in both potential and magnetic fields. However, when the fat conductivity was reduced the magnetic fields were largely unchanged, but electric potentials had a significant change in patterns and amplitudes. When the thickness of the fat layer was increased by 30 mm the amplitude of the magnetic fields decreased 10 % more than potentials but magnetic field patterns were changed about 4 times less than potentials. The ability to localize the underlying sources from the magnetic fields using a surface current density measure was altered by less than 2 mm when the fat layer was increased by 30 mm. In summary, results confirm that MGG provides a favourable method over EGG when there are uncertain levels of fat thickness or conductivity.

1. Introduction

Propagating electrical waves in the stomach, known as slow waves, coordinate the mixing and propulsion of foods. The slow wave activity is known to originate in a pacemaker region located in the upper/mid-corpus on the greater curvature of the stomach. In humans, the slow waves normally initiate at approximately 3 cycles per minute (cpm), form organized rings, and propagate towards the pylorus (Hinder and Kelly 1977, O'Grady *et al* 2010). The disorganized slow wave activity has been associated with motility disorders such as gastroparesis and functional dyspepsia (Farrugia *et al* 2008, Koch *et al* 1989, Lin *et al* 1999). The ability to non-invasively measure and characterize the electrical activity in the GI system has significant clinical implications.

Electrogastrograms (EGG) and magnetogastrograms (MGG) provide two complementary methods for non-invasively recording the electric or magnetic field resulting from gastric electrical slow wave activity (Alvarez 1922, Bradshaw *et al* 2009). The slow wave electrical activity in the stomach musculature produces both extracellular electric potential and

magnetic fields. The EGG uses multiple electrodes placed on the abdomen to record cutaneous electric potentials, while an MGG uses a highly sensitive Superconducting QUantum Interference Device (SQUID) to record the corresponding magnetic field. Magnetic fields from gastric slow waves are incident on an array of pickup sensors housed in a liquid-helium-filled dewar positioned just above the abdomen of a supine subject. These sensors are coupled to SQUIDs which convert the incident magnetic flux to a voltage that can be digitized and recorded. MGGs have been used to examine uncoupling and the effects of elevated glucose in the gastric slow wave (Bradshaw *et al* 2003, 2007, 2009). Recent studies from our group have used surface current density (SCD) methods to localize the underlying sources and attempt to calculate propagation direction and velocities from magnetic fields (Bradshaw *et al* 2009b, Kim *et al* 2010).

Unlike the cardiac field where electrocardiography (ECG) is commonly used to assess cardiac function, to date, EGGs have received limited clinical acceptance and MGGs are largely limited to research centers. This is in part due to a lack of detailed knowledge about the underlying slow wave activity and both technical and physiological limitations related to recording EGGs and MGGs (Du *et al* 2010, Verhagen *et al* 1999). One significant issue is that EGG signals are relatively weak and difficult to reliably record. Magnetic fields, however, are, in theory, less attenuated by the low-conductivity fat layers present in the body, and thus may provide a significant advantage over EGGs (Bradshaw *et al* 1999, 2001).

A recent study showed that BMI and waist circumference modestly influenced the sensitivity of the MGG (89.8 ± 3.0 % in normal subjects vs. 82.5 ± 5.4 % in obese) but increasing BMI or waist circumference resulted in a substantial decrease in sensitivity in the EGG (61.5 ± 6.4 % in normal vs. 19.0 ± 7.7 % in obese subjects) (Obioha *et al* 2011). Ongoing experimental investigations are looking at the effects of body habitus on the relationship of the amplitude of EGG and MGG to internal slow wave amplitude.

We introduce anatomically realistic models that include different degrees of fat thickness and conductivity values to examine the resultant effect on bioelectro-magnetic fields calculated at the 110 sensors/electrodes located on or near the anterior surface of the torso due to gastric electrical activity. Komuro *et al* (2010) previously showed that the use of a realistic model was necessary for accurate interpretation of magnetic fields and discussed the possibility of the different trends depending on the representation of the volume conductor. The results obtained from realistic torso models with different geometries in our study will help to improve the understanding of the relationships between the underlying gastric slow wave activity and the resultant far-field electric and magnetic fields.

2. Methods

Anatomically realistic torso models were used with dipole source models to simulate resultant electric and magnetic fields measured external to the torso. Two source models were used to represent the slow wave activity: a single dipole representing one gastric slow wave and multiple dipoles that represented up to three simultaneous slow wave events. Magnetic fields were calculated at SQUID sensors located regularly on a plane just above the skin surface and potentials were calculated at evenly spaced electrodes located on the anterior surface of the torso. The resultant fields were calculated on the different volume conductor models and the changes in amplitude and field patterns compared.

2.1 Torso Models

The default model (Model 01) was derived from the Visible Human dataset (Buist *et al* 2004, Spitzer *et al* 1996). The model included boundary element surfaces representing the stomach, the abdominal muscle and the fat and skin surfaces with conductivities of 0.135,

0.22 and 0.04 mS mm⁻¹ respectively. These surfaces were interpolated using cubic Hermite basis functions. The default model had an average fat thickness of 18.2 mm in the area covered by the SQUID and electrodes. Additional models were created with thicker fat layers by projecting the appropriate surfaces inwards or outwards. Model 02 had the fat layer increased by a 5 mm projection of the inner fat surface inwards (skin surface remained the same). Although the changes imposed on Model 02 are unlikely to be observed in reality, it provides a scenario whereby the fat layer was increased, yet the distance between the sources and the sensors remained constant. Models 03, 04 and 05 had the fat layers increased by 5, 10 and 30 mm, respectively, by projecting the skin surface outwards. Figure 1 illustrates model 01 and model 05.

2.2 Dipole Sources

The dipole source configurations used in the simulations included a single and multiple dipole(s) in the stomach. The single dipole had been derived from a simulation of slow wave activity on the visible human stomach as described previously (Hinder *et al* 1977). The multiple dipoles were constructed by overlapping the single dipole source such that up to 3 simultaneous waves were always present as has been shown in recent high-resolution mapping studies (Egbuji *et al* 2010, Lammers *et al* 2009, O'Grady *et al* 2010). In the multiple dipole sequence, a new source initiated in the mid-corpus area prior to the previous source terminating at the pylorus (i.e., when a dipole was approximately a third of the way down the length of the stomach, a subsequent source initiated in the pacemaker region).

2.3 Magnetic and Potential Calculations

Both magnetic fields and potentials resulting from dipole sources in a volume conductor have two components – one is due to dipole in free space and the other due to volume current in the body (Sarvas 1987). When dipoles are in a homogeneous volume conductor, magnetic fields and potentials are calculated by (1) and (2), respectively.

$$B(r) = \frac{\mu_0}{4\pi} \left(\int_G J'(r') \times \frac{r-r'}{|r-r'|^3} dv' \right) \quad (1)$$

$$V(r) = \frac{1}{4\pi\sigma} \left(\int_G J'(r') \cdot \frac{r-r'}{|r-r'|^3} dv' \right) \quad (2)$$

where $B(r)$ is the magnetic field and $V(r)$ is the electric potential at a point r ; $J'(r')$ is a bioelectric source due to a dipole located at r' ; G is the region containing the source J' ; σ is conductivity, and μ_0 is the permeability of free space.

These equations show that both field values decrease reciprocally when the distance ($|r-r'|$) between the sensors/electrodes and the dipole increases. The potential is affected by conductivity of the medium while the magnetic field is governed by the magnetic permeability. The magnetic permeability of tissue is nearly the same as that of free space.

When dipoles are located in an inhomogeneous volume conductor, (3) and (4) show that both the distance of the sensors/electrodes from the dipole and the difference in conductivity between different regions are involved with the calculation of both fields. However, the term $(\sigma'_k + \sigma''_k)$ in (4) indicates that the potential has more influence from the inner and outer conductivity of the layer where potentials are calculated than magnetic fields.

$$B(r) = \frac{\mu_0}{4\pi} \left(\int_G J'(r') \times \frac{r-r'}{|r-r'|^3} dv' - \sum_{j=1}^n (\sigma'_j - \sigma''_j) \int_{S_j} V(r') n(r') \times \frac{r-r'}{|r-r'|^3} dS_j \right) \quad (3)$$

$$V(r) = \frac{1}{2\pi(\sigma'_k + \sigma''_k)} \left(\int_G J'(r') \cdot \frac{r-r'}{|r-r'|^3} dv' - \sum_{j=1}^n (\sigma'_j - \sigma''_j) \int_{S_j} V(r') n(r') \cdot \frac{r-r'}{|r-r'|^3} dS_j \right) \quad (4)$$

where n is the total number of surfaces surrounding regions of different conductivity; j identifies a surface of a region with constant conductivity σ_j and boundary defined by S_j and σ'_j and σ''_j are the conductivities on either side of surface S_j .

2.4 Squid Sensors and Electrodes

Magnetic fields were simulated on a hypothetical SQUID sensor array. The absolute magnetic field values were calculated at 110 SQUID sensors located in a regular grid with a spacing of 25 mm. In accordance with typical experimental procedures, the sensors were located in the coronal plane just above the anterior surface of the skin. Only the magnetic fields normal to the coronal plane were considered. Electric potentials were simulated at 110 electrodes located on the skin surface positioned to cover approximately the same area as the SQUID sensors.

2.5 Comparison Methods

Two simulation studies were performed to investigate the effect of the fat layer. In the first study, the thickness of the fat layer was altered while maintaining the same conductivity. When the fat layer was enlarged by projecting the skin surface “outwards” (e.g., models 03, 04 and 05), electrodes and SQUID sensors were moved correspondingly. In the second study, the conductivity of the fat layer was altered between 0.005 and 0.22 mS mm⁻¹ from with the default fat conductivity value of 0.04 mS mm⁻¹ while maintaining the same anatomy. This setup enabled the location of the electrodes/sensors to remain constant. Previous experimental studies in literature have shown that fat conductivities vary between 0.01 and 1 mS mm⁻¹ (Gabriel *et al* 1996 and Foster and Schwan 1989). Model 01, with a fat conductivity 0.04 mS mm⁻¹, was used as a reference model for all simulations.

Three metrics were used to compare simulations. Amplitudes of signals were averaged over 110 electrodes/sensors and the averaged amplitudes were normalized by the amplitude of reference model. The pattern of fields was investigated by examining 2 parameters: the midpoint location and the relative orientation between maximum and minimum points (shown in figure 2). The first metric provides a global measure of field amplitude, while the last two metrics characterize key differences in the field patterns.

In addition for the magnetic fields SCD was calculated and compared for model 01 and model 05. The SCD method has previously been used to localize dipole positions using magnetic fields (Kim *et al* 2010). The potential and magnetic values were interpolated at 110 points using a modified Kriging interpolation method (Cressie 1989, van Beers and Kleijnen 2003). Differences between each model and the reference model were determined.

Both the single and multiple dipole configurations were used for comparison since the influence of dipole directions or locations in deep stomach area on fat thickness or conductivity should be investigated even though the multiple dipole source provided a more realistic representation of underlying electrical events.

3. Results

3.1 Changes in Bioelectro-Magnetic Fields due to Increased Fat Thickness

3.1.1 Single dipole—Figure 3(a) illustrates that when the fat thickness was increased outwards the amplitude decreased for both magnetic and potential fields. However, the magnetic field decreased at a faster rate. For example, the magnetic field amplitude had a 10% greater reduction than the potential when a model with an increase in fat layer of 30 mm was used. However, when the fat increased inwards, there was a minimal change in the amplitude of the magnetic field while the amplitude of the potential increased.

The relationship between potential or magnetic field value at one electrode/sensor and the distance (r) of electrode/sensor from the dipole at 56 s was shown in figures 4(a) or 4(c). Results showed that the potential fall-off was approximated by r^{-1} with up to 10 mm additional fat thickness but shifted to r^{-2} when the fat layer was increased from 10 to 30 mm. However, the magnetic field had a consistent fall-off rate of $r^{-2.5}$, suggesting that magnetic field value decreased faster than potential with larger fat thickness.

Figures 5(a) and 5(c) shows that the changes in midpoint locations and orientations increased for both type of fields when the fat thickness was increased, but changes in the potential were larger than in the magnetic field, especially when the fat thickness increased outwards. When the additional fat thickness was 30 mm, the midpoint location changes were 15.8 and 5.1 mm, and the angle changes were 5.5 and 4.4 degrees for potential and magnetic field, respectively.

3.1.2 Multiple dipoles—Figures 3(b), 5(b) and 5(d) show the amplitude and pattern changes when using multiple dipoles. The trends between magnetic and potential when using multiple dipoles were similar to those produced by the single dipole source. The field fall-off rates were also similar to those when using a single dipole in both fields (shown in figures 4(b) and 4(d)). In general, the pattern changes were smaller mostly when using multiple dipoles than when using a single dipole for both magnetic and potential fields.

3.2 Changes in Bioelectro-Magnetic Fields due to Changes in Fat Conductivity

3.2.1 Single dipole—Figure 6(a) shows that when the distance between source and the SQUID sensors/potential electrodes remained constant and fat conductivities were changed, the amplitude of the magnetic field remained fairly constant while the amplitude of the potential fields decreased as fat conductivities increased.

With the change of fat conductivities, Figures 7(a) and 7(c) indicated that changes in midpoint locations and orientations increased slightly for both fields when the fat conductivity was close to the muscle conductivity. However, when the conductivity was reduced, potential field patterns were changed far more than magnetic field patterns.

3.2.2 Multiple dipoles—The trends observed between magnetic fields and potentials when using multiple dipoles were similar to those produced by a single dipole source (shown in figures 6(b), 7(b) and 7(d)). In general, the pattern differences were smaller when using multiple dipoles than when using a single dipole for both magnetic and potential fields.

3.3 Changes in Surface Current Density due to Changes in Fat Thickness

A recent study showed that the surface current density (SCD) method could be efficiently related to slow wave parameters using magnetic field measurements (Kim *et al* 2010). It was shown that when multiple sources were present, the SCD method was only able to localize

the most distal source. In order to see the effects of the fat thickness on magnetic field and on the SCD method, we calculated SCD values from the magnetic fields of models 01 and 05.

The blue and red traces in figure 8 represent the maximum SCD values for model 01 and model 05, respectively, when multiple dipoles were present in the stomach. The figure shows that the red trace in both horizontal and vertical axes almost matched the blue trace throughout time. The difference of the average position changes to the dipoles between model 01 and model 05 was 1.4 mm. This result suggested that maximum SCD values were not significantly affected by the additional 30 mm of fat.

4. Discussion

We have investigated the influence of fat thickness and conductivity on resultant MGG and EGG fields using anatomically realistic torso models. The underlying slow wave was represented using both a single and multiple dipole source(s). The resultant magnetic and potentials were then calculated at the 110 sensors/electrodes located on the anterior surface of the torso and the changes in field patterns and amplitudes were compared. Results have shown that magnetic field patterns were less affected by the fat layer and conductivity than potentials when either source model was used.

When the size of the model was increased due to an increase in fat layer thickness (model 03–05) or when the size of the model remained the same with an increase of fat thickness (model 02), the average field values decreased faster in the magnetic field than in the potential. In fact, the potential fall-off changed from r^{-1} to r^{-2} with the increased fat thickness. This is consistent with the fat layer introducing additional signal reduction. However the magnetic field fall-off was consistently $r^{-2.5}$ for the different models. This resulted in a faster decrease of the magnetic field amplitude than the potential amplitude. However, magnetic field patterns were less affected by the change in fat thickness when compared to the potential fields. In addition, when the location of dipole sources were reconstructed from magnetic fields by tracking the position of the maximum surface current density, there was a change of less than 2 mm when the fat thickness was increased by 30 mm. This suggests that magnetic field patterns were preserved well even though amplitudes were reduced by 60%. This result indicates that the body habitus of the patient affects the potential fields more than magnetic fields.

When the conductivity of the fat layer was altered, but the distance between source and the SQUID sensors/potential electrodes remained constant, field values and amplitudes changed little in magnetic fields but more in potentials, especially when the conductivity was reduced. Changes in patterns increased slightly for both fields when the fat conductivity was similar to the muscle conductivity, but when the conductivity was reduced, potential patterns had much greater changes than magnetic fields. Note that the distance between dipole location and torso surface was constant for all models in these simulations. Therefore, magnetic fields were affected by only the differences of adjacent layer conductivities, but potentials were affected by the reciprocal of the fat conductivity as well as the differences of adjacent layer conductivities, resulting in the large changes when the conductivity was reduced. Since magnetic fields were minimally affected by conductivity, inverse solutions to determine gastric electrical activity from magnetic field measurements may not require a detailed description of inhomogenities of a torso model.

The trends observed for magnetic fields and potentials were similar when either using a single or multiple dipole(s). In general, less change in both potential and magnetic field patterns was observed with multiple dipole sources. This may be due to the fact that with this source model, a dipole source was always near the skin surface and may dominate

potential and magnetic field patterns. The change of field patterns between when using a single and multiple dipole(s) was larger in potential than in magnetic fields, suggesting that the distance of a dipole from electrodes affected to the pattern changes larger in potentials than in magnetic fields.

The simulations have included number of assumptions and simplifications to reality. We have assumed that each gastric slow wave cycle was represented as a single dipolar source and stomach motility and movement were ignored. The torso models were represented by three boundary element surfaces (representing the stomach, muscle, and fat regions). Organs such as the lungs and liver, which are in close proximity to the stomach, were not explicitly represented. The inclusion of additional inhomogeneities may exaggerate the changes in both magnetic fields and potentials; however, it is expected that similar trends would be observed. Finally, the amount of fat was increased by projecting the surface either inwards or outwards resulting in a uniform increase. However, in practice, excess fat tends to be distributed mainly around the hips and belly regions.

In conclusion, this study quantified the influence of fat thickness or conductivity on gastric bioelectric and biomagnetic fields using realistic torso models and has shown that magnetic fields preserved the information of gastric electrical activity better than potential fields in the change of fat thickness or conductivity. These results may provide insights into the properties of electric potential and magnetic fields and will help the interpretation of future experimental MGG and EGG recordings.

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References

- Alvarez WC. The electrogastrogram and what it shows. *J Am Med Assoc.* 1922; 78:1116–19.
- Bradshaw LA, Ladipo JK, Staton DJ, Wikswo JP Jr, Richards WO. The human vector magnetogastrogram and magnetoenterogram. *IEEETrans Biomed Eng.* 1999; 46:959–70.
- Bradshaw LA, Richards WO, Wikswo JP. Volume conductor effects on the spatial resolution of magnetic fields and electric potentials from gastrointestinal electrical activity. *Medical and Biological Engineering and Computing.* 2001; 39:35–43. [PubMed: 11214271]
- Bradshaw LA, Myers AG, Redmond A, Wikswo JP, Richards WO. Biomagnetic detection of gastric electrical activity in normal and vagotomized rabbits. *Neurogastroenterol Motil.* 2003; 15:475–82. [PubMed: 14507349]
- Bradshaw LA, Sims JA, Richards WO. Noninvasive assessment of the effects of glucagon on the gastric slow wave. *Am J Physiol Gastrointest Liver Physiol.* 2007; 293:G1029–38. [PubMed: 17884978]
- Bradshaw LA, Irimia A, Sims JA, Richards WO. Biomagnetic signatures of uncoupled gastric musculature. *Neurogastroenterol Motil.* 2009; 21:778–88. [PubMed: 19222760]
- Bradshaw LA, Cheng LK, Richards WO, Pullan AJ. Surface current density mapping for identification of gastric slow wave propagation. *IEEE Trans Biomed Eng.* 2009b; 56:2131–9. [PubMed: 19403355]
- Buist ML, Cheng LK, Yassi R, Bradshaw LA, Richards WO, Pullan AJ. An anatomical model of the gastric system for producing bioelectric and biomagnetic fields. *Physiological Measurement.* 1989; 25:849–61. [PubMed: 15382826]
- Cressie N. The Origins of Kriging. *Mathematical Geology.* 1989; 22:239–52.
- Du P, O'Grady G, Cheng LK, Pullan AJ. A multiscale model of the electrophysiological basis of the human electrogastrogram. *Biophysical J.* 2010; 99:2784–92.

- Egbuji JU, O'Grady G, Du P, Lammers WJ, Cheng LK, Windsor JA, Pullan AJ. Origin, propagation and regional characteristics of gastric slow wave activity determined by high-resolution mapping. *Neurogastroenterology & Motility*. 2010; 22:292–300.
- Farrugia G. Interstitial cells of Cajal in health and disease. *Neurogastroenterol Motil*. 2008; 20(Suppl 1):54–63. [PubMed: 18402642]
- Foster KR, Schwan HP. Dielectric properties of tissue and biological materials: A critical review. *Crit Rev Biomed Eng*. 1989; 17:25–104. [PubMed: 2651001]
- Gabriel C, Gabriel S, Corthout E. The dielectric properties of biological tissues: I. Literature survey. *Phys Med Biol*. 1996; 41:2231–49. [PubMed: 8938024]
- Hinder RA, Kelly KA. Human gastric pacesetter potential. Site of origin, spread, and response to gastric transection and proximal gastric vagotomy. *Am J Surg*. 1977; 133:29–33. [PubMed: 835775]
- Kim JHK, Bradshaw LA, Pullan AJ, Cheng LK. Characterization of gastric electrical activity using magnetic field measurements: A simulation study. *Annals Biomed Eng*. 2010; 38:177–86.
- Koch KL, Stern RM, Stewart WR, Vasey M, Sullivan ML. Gastric emptying and gastric myoelectric activity in patients with symptomatic diabetic gastroparesis: Effects of long-term domperidone treatment. *Am J Gastroenterol*. 1989; 84:1069–75.
- Komuro R, Qiao W, Pullan AJ, Cheng LK. Effects of volume conductor and source configuration on simulated magnetogastrogams. *Phys Med Biol*. 2010; 55:6881–95. [PubMed: 21048291]
- Lammers WJ, Ver Donck L, Stephen B, Smets D, Schuurkes JA. Origin and propagation of the slow wave in the canine stomach: the outlines of a gastric conduction system. *Am J Physiol Gastrointest Liver Physiol*. 2009; 296:G1200–10. [PubMed: 19359425]
- Lin Z, Eaker EY, Sarosiek I, McCallum RW. Gastric myoelectrical activity and gastric emptying in patients with functional dyspepsia. *Am J Gastroenterol*. 1999; 94:2384–89. [PubMed: 10483996]
- Obioha, C.; Erickson, JE.; Somarajan, S.; Hajri, T.; Chung, E.; Richards, WO.; Bradshaw, LA. Effect of body mass index on the magnetogastrogram. 2011. manuscript submitted
- O'Grady G, Du P, Cheng LK, Egbuji JU, Lammers WJ, Windsor JA, Pullan AJ. The origin and propagation of human gastric slow wave activity defined by high-resolution mapping. *Am J Physiol Gastrointest Liver Physiol*. 2010; 299:G585–92. [PubMed: 20595620]
- Sarvas J. Basic mathematical and electromagnetic concepts of the biomagnetic inverse problem. *Phys Med Biol*. 1987; 32:11–22. [PubMed: 3823129]
- Spitzer V, Ackerman MJ, Scherzinger AL, Whitlock D. The visible human male: a technical report. *J Am Med Inform Assoc*. 1996; 3:118–30. [PubMed: 8653448]
- van Beers WCM, Kleijnen JPC. Kriging for interpolation in random simulation. *J Oper Res Soc*. 2003; 54:255–262.
- Verhagen MA, Van Schelven LJ, Samsom M, Smout AJ. Pitfalls in the analysis of electrogastrographic recordings. *Gastroenterology*. 1999; 117:453–60. [PubMed: 10419928]

Figure 1. Anatomically realistic torso model constructed from Visible Human data. Shown are (a) the default model and (b) the same model with the fat layer increased by 30 mm. The torso models have three boundary element surfaces representing the stomach (red), muscle (brown) and fat and skin (grey) surfaces.

Figure 2.

Contour plots of a (a) potential or (b) magnetic field, with red contours and blue contours representing positive and negative values, respectively. The location of the maximum field value is indicated by the red circle, the minimum field value by the blue triangle and the midpoint between these two locations by the green square. The orientation of the maxima and minima relative to the coordinate system was represented by the black lines.

Figure 3.

The normalized amplitudes averaged over 110 electrodes/sensors for the models with varying fat conductivity when using (a) a single dipole or (b) multiple dipoles. The * denotes the model with the default level of fat thickness.

Figure 4.

The comparison between the field (potential and magnetic) fall-off and power of the distance from a dipole to a sensor/electrode at $t=56$ s when using (a), (c) a single dipole or (b), (d) multiple dipoles.

Figure 5.

Changes in field parameters (mid-point locations and orientations) due to changes in fat layer thickness when using (a),(c) a single dipole and (b),(d) multiple dipoles. The * denotes the model with the default level of fat thickness.

Figure 6.

The normalized amplitudes averaged over 110 electrodes/sensors for the models with varying fat thickness when using (a) a single dipole or (b) multiple dipoles. The * denotes the model with the default fat conductivity value.

Figure 7. Changes in field parameters (mid-point location and orientation) due to changes in conductivities using (a),(c) a single dipole source or (b),(d) multiple dipole sources. The * denotes the model with the default fat conductivity value.

Figure 8.

SCD estimates of the (a) horizontal and (b) vertical positions of the underlying sources. The black lines represent the dipole locations. The blue and red (dot) lines represent the maximum SCD approximations of the positions for model 01 and model 05, respectively. The average difference between the two models was 1.4 mm.