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Effects of volume conductor and source configuration on simulated magnetogastrograms

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Abstract

Recordings of the magnetic fields (MFs) arising from gastric electrical activity (GEA) have been shown to be able to distinguish between normal and certain abnormal GEA. Mathematical models provide a powerful tool for revealing the relationship between the underlying GEA and the resultant magnetogastrograms (MGGs). However, it remains uncertain the relative contributions that different volume conductor and dipole source models have on the resultant MFs. In this study, four volume conductor models (free space, sphere, half space and an anatomically realistic torso) and two dipole source configurations (containing 320 moving dipole sources and a single equivalent moving dipole source) were used to simulate the external MFs. The effects of different volume conductor models and dipole source configurations on the MF simulations were examined. The half space model provided the best approximation of the MFs produced by the torso model in the direction normal to the coronal plane. This was despite the fact that the half space model does not produce secondary sources, which have been shown to contribute up to 50% of the total MFs when an anatomically realistic torso model was used. We conclude that a realistic representation of the volume conductor and a detailed dipole source model are likely to be necessary when using a model-based approach for interpreting MGGs.

1. Introduction

Slow waves are omnipresent period electrical events known to mediate the contractile activity of the stomach. In humans, normal gastric slow wave activity is known to initiate on the greater curvature around the mid-corpus of the stomach and propagate organo-axially towards the pylorus at a frequency of approximately 3 cycles/min. Dysrhythmic gastric electrical activity (GEA) has been described in relation to a number of common clinical disorders, such as gastroparesis and functional dyspepsia (Chen *et al* 1996, Lin and Chen 2001). The ability to non-invasively characterize both normal and abnormal activity would be highly desirable as a diagnostic method for gastrointestinal disorders.

The use of cutaneous electrodes to measure GEA in the form of electrogastrograms (EGGs) was first reported by Alvarez (1922). The ability to record magnetogastrograms (MGGs) using a superconducting quantum interference device (SQUID) has subsequently been reported (Bradshaw *et al* 2006). The MGG has the advantage of being non-contact and its signals are theoretically less attenuated and smoothed out by the muscle and fat layers in the abdominal wall than EGG signals. However, despite significant interest, neither of these methods have achieved widespread clinical acceptance. One major reason for this is the difficulty in interpreting the measured signals and accurately relating them back to the underlying GEA. In the cardiac field the use of mathematical models for interpreting and elucidating body surface recordings is now providing reliable information that can be used for clinical diagnosis and for understanding the underlying electrical activity (Ghosh *et al* 2008, Ramanathan *et al* 2004, Tilg *et al* 2002). We hypothesize that the use of accurate, biophysically based mathematical models of GEA activity may be a valuable aid in the interpretation of MGGs.

The majority of numerical simulations of MGGs to date have used idealized and simplified representations of the torso geometry (Allescher *et al* 1998, Bradshaw *et al* 2001, Irimia and Bradshaw 2005). The underlying GEA has also often been modelled either by a single dipole or by a limited number of dipoles arranged in an annular ring (Bradshaw *et al* 2003, Mintchev and Bowes 1997). It remains uncertain the degree to which these simplifications affect the resultant simulated magnetic fields (MFs). Recently, the use of an anatomically realistic volume conductor model along with more realistic source models derived from biophysically based slow wave simulations has been presented (Buist *et al* 2006, Komuro *et al* 2008). Although these simulations are believed to be more realistic, they also required more effort, time and computational resource to produce and it remains uncertain if this degree of sophistication was warranted.

In this study we aim to address these questions by comparing the differences in the MFs calculated using different volume conductor models and different dipole source configurations. The volume conductor models used here consisted of three simplified analytic representations of the volume conductor (an unbounded free space, a sphere and a half space) as well as an anatomically realistic torso. To represent the GEA, two dipole source configurations that contained different number of dipole sources representing a single slow wave event were used. The patterns of the resulting MFs were compared and the errors between two different sets of the MFs were computed. The degree of contribution of the magnetic secondary sources was also evaluated.

2. Methods

The MFs arising from one slow wave event were simulated using four different volume conductor models and two different dipole source configurations. The MFs were compared at the locations placed according to the multichannel gastrointestinal SQUID magnetometer located at Vanderbilt University (model 637i, Tristan Inc., San Diego, CA) (Bradshaw *et al* 2006). A total of 19 SQUID sensors were arranged in a hexagonal pattern in a single plane as illustrated by the yellow spheres in figure 1. The sensor at the centre of the SQUID array was centred over the stomach and just above the skin surface. The coordinate system

referred to in this paper is also illustrated in figure 1 with the positive y-direction orientated into the page and normal to the coronal plane.

2.1. Computation of magnetic fields

The MF due to a current dipole embedded in a conducting medium is composed of two components. One component, called the primary source, is the MF due to an impressed current that arises from the electric current dipoles (Mosher *et al* 1999, Plonsey 1982, Sarvas 1987). The second component, called the secondary source, is the MF due to a fictitious current that arises from the interface between the regions of different electrical conductivities (Mosher *et al* 1999, Plonsey 1982). The primary and the secondary sources are denoted by $\mathbf{B}_d$ and $\mathbf{B}_\nu$, respectively, and the total MF $\mathbf{B}$ is given by (Geselowitz 1970, Sarvas 1987)

$$\mathbf{B} = \mathbf{B}_d + \mathbf{B}_\nu \quad (1)$$

The primary source is also known as the free space model. At the field point $\mathbf{r}_f$ where the MF is computed, the primary source due to one dipole can be calculated by (Sarvas 1987)

$$\mathbf{B}_d(\mathbf{r}_f) = \mu_0 \frac{\boldsymbol{\rho} \times \mathbf{r}}{4\pi r^3} \quad (2)$$

Where $\boldsymbol{\rho}$ is the dipole vector, $\mathbf{r}$ is the displacement vector from the dipole to the field point, the corresponding distance is denoted by r , μ_0 the permeability of the free space and $\times$ denotes a cross product. The secondary sources are given by (Geselowitz 1970, Sarvas 1987)

$$\mathbf{B}_\nu(\mathbf{r}_f) = -\frac{\mu_0}{4\pi} \sum_{j=1}^N \sigma_j \int_G \nabla \phi(\mathbf{r}_d) \times \frac{\mathbf{r}}{r^3} d\Omega_j \quad (3)$$

where N is the total number of regions, Ω_j ($1 \leq j \leq N$) is a surface enclosing a region with electrical conductivity σ_j , G is the conductor and $\phi(\mathbf{r}_d)$ is the electric potential due to the dipole source located at $\mathbf{r}_d$.

2.2. Volume conductor models

In this study, the resulting MFs from four volume conductor models were compared. The MFs were calculated from the dipole sources enclosed by volume conductors represented by an unbounded free space, a sphere, a half space and an anatomically realistic torso model. To enable direct comparison between each model, the location of the dipole sources relative to the SQUID sensors were the same for all simulations.

For the sphere, half plane and torso models a distance of 15 mm was maintained between the plane of the SQUID sensors and the anterior surface of the model. This approximated the separation that was maintained experimentally to prevent the skin surface touching the sensor arrays due to movement during respiration. The radius of the sphere was 130 mm and was centred 145 mm below the plane of sensors. This size was chosen to approximate the

size of the anatomical model which had a distance between anterior and posterior surface of 260 mm. Each of the volume conductor models is shown schematically in figure 2.

The external MF $\mathbf{B}(\mathbf{r}_f)$ due to a dipole located inside a sphere is given by (Sarvas 1987)

$$\mathbf{B}(\mathbf{r}_f) = \frac{\mu_0}{4\pi M^2} (M_\rho \times \mathbf{r}_d - \rho \times \mathbf{r}_d \bullet \mathbf{r}_f \nabla M) \quad (4)$$

where $M = r(r_f \cdot r + r_f^2 - \mathbf{r}_d \bullet \mathbf{r}_f)$,

$\nabla M = \left[r^2 \cdot r_f^{-1} + (\mathbf{r} \bullet \mathbf{r}_f) \cdot r^{-1} + 2r + 2r_f \right] \cdot r_f - \left[r + 2r_f + (\mathbf{r} \bullet \mathbf{r}_f) \cdot r^{-1} \right] \cdot r_d$, r_f is the distance between the origin and the field point $\mathbf{r}_f$ and $\bullet$ denotes an inner product.

The MF $\mathbf{B}(\mathbf{r}_f)$ on one side of an infinite plane due to a dipole located on the other side of the plane is given by (5). This equation has been derived from (4) where the limit of the radius of the sphere has been extended to infinity (Sarvas 1987):

$$\mathbf{B}(\mathbf{r}_f) = \frac{\mu_0}{4\pi K^2} (\rho \times \mathbf{r} \bullet \mathbf{n} \nabla K - K \mathbf{n} \times \rho) \quad (5)$$

Where $K = r(r + \mathbf{r} \bullet \mathbf{n})$, $\nabla K = [2 + (\mathbf{r} \bullet \mathbf{n}) \cdot r^{-1}] \cdot r + r \cdot \mathbf{n}$ and $\mathbf{n} = [0 -1 0]^T$ is a unit vector pointing in the direction outward and normal to the boundary of the half space. In the actual computation, the half space was transformed by the average thickness of the torso in the y-direction (260 mm) so that the half space would be located around the same place as the anterior body.

The anatomically realistic torso geometry used in this study was derived from CT images of a male volunteer. The CT images were first digitized and a bicubic Hermite boundary element surface mesh defined by 254 nodes and 264 elements was created from the digitized data points using an iterative fitting procedure (Bradley *et al* 1997, Komuro *et al* 2008). The average element edge length was 52 mm and was provided a mesh resolution capable of representing the relatively smooth potential fields resulting from the dipole sources. The anatomically realistic torso model was assumed to be homogeneous and is shown with the SQUID sensor locations in figure 1. The MFs due to dipole sources inside the torso model were calculated using (1)-(3) with the secondary source computed by numerically integrating over the volume conductor surface.

2.3. Source models

A previously described numerical simulation of GEA was employed to derive the dipole sources used in this study (Komuro *et al* 2008). For a detailed description of the slow wave simulation, readers should refer to the results from subject A in Komuro *et al* (2008). In summary, the stomach geometry was constructed using the same CT dataset and methods as employed to construct the torso model. The continuum-based monodomain equation defined in (6) was used to simulate the propagation of a slow wave:

$$\nabla \cdot (\sigma \nabla V_m) = A_m \left(C_m \frac{\partial V_m}{\partial t} + I_{ion} \right), \quad (6)$$

where σ is the tissue conductivity, V_m is the transmembrane potential of a continuum cell, A_m is the surface to volume ratio of the membrane, C_m is the membrane capacitance and I_{ion} is the ionic current. Approximately 250 000 solution points representing continuum cells were embedded within the stomach musculature. The number of solution points was chosen such that there was an average grid point spacing of 1.1 mm. This resolution has previously been shown to provide a numerically converged resolution (Buist *et al* 2004). The Aliev cell model (Aliev *et al* 2000) was used at each solution point to calculate the ionic current I_{ion} that represents the subcellular ionic activities of the smooth muscle cells and pacemaker interstitial cells of Cajal.

To compute the resultant potential and MFs, dipole sources J_s were derived from the transmembrane potential V_m (Austin *et al* 2007) at each of the solution points in the stomach wall using

$$J_s = -\sigma(\nabla V_m). \quad (7)$$

For computational efficiency, these sources were then combined to a reduced number of dipoles using a weighted average approach presented in

$$C_t = \sum_{m=1}^M W_{mt} C_{mt}, \quad (8)$$

$$\rho_t = \frac{1}{M} \sum_{m=1}^M \rho_{mt} \quad (9)$$

where

$$w_{mt} = \frac{\|\rho_{mt}\|_2}{(\sum_{l=1}^M \|\rho_{lt}\|_2)}, \quad \sum_{m=1}^M w_{mt} = 1,$$

C_{mt} and ρ_{mt} are the location and direction of dipole m at time t , respectively, M is the number of reduced dipole sources used to represent the slow wave activity and $\|\cdot\|_2$ is the Euclidean norm.

To create a single dipole source representing all the slow wave activity from the entire stomach activity, M was set to equal all the solution points present in the stomach (in this case approximately 250,000) in (8) and (9). To create a source model that reflected the distributed nature of the electrical wave front, the sources in each of the 320 host finite elements was combined for each element p . The combined dipoles were computed by (8)

and (9) with $M = L_p$, where L_p is the number of sources in the p th element. Thus $\sum_{p=1}^{320} L_p$ was the number of the total number of solution points in the stomach. As each dipole was effectively constrained to be located within an element, the dipoles would only be non-zero when the wave front was located within a given element. When there was no transmembrane gradient present within an element, the dipole sources had negligible magnitude.

The single dipole source was located close to the centre of the principal axis of the stomach, while the 320 dipole sources were largely constrained to be located within the stomach musculature. As such, in all cases, the distance between the dipole sources and the boundary element skin surface were relatively large, so no numerical issues were encountered.

2.4. Methods of comparison

The effects of the different volume conductors and dipole source configurations on the simulated MFs were evaluated by comparing the strength and direction of the MFs at the 19 SQUID sensor locations as well as errors between the MFs with two different configurations. In addition, the relative contribution of the secondary sources on the MFs was also examined.

To compare the patterns of the simulated MFs, spatial maps were generated by interpolating the MFs at 19 sensor locations using a triangle-based linear interpolation algorithm.

The recorded MFs are composed of both primary and secondary sources of MFs (Mosher *et al* 1999). Therefore, it is important to examine the effects of changing the inputs of the MF simulation on both the primary and secondary sources. However, there are certain situations where there is no contribution from the secondary source: such as the free space, sphere and half space models when the MF is measured normal to the surface. As shown in (1), the secondary sources of the sphere, half space and anatomically realistic models were calculated by subtracting the primary source from the total MFs given by (4) and (5). The primary sources for those three models are equivalent to the MFs produced by the dipole(s) in the free space.

At time t and for SQUID sensor n ($1 \leq n \leq 19$), the contribution of the secondary source in each of the three directions was computed as the ratio of the magnitude of the secondary source to the sum of the magnitudes of the primary and secondary sources:

$$\left(\frac{|\mathbf{B}_{\nu}^{(x,n,t)}|}{|\mathbf{B}_d^{(x,n,t)}| + |\mathbf{B}_{\nu}^{(x,n,t)}|}, \frac{|\mathbf{B}_{\nu}^{(y,n,t)}|}{|\mathbf{B}_d^{(y,n,t)}| + |\mathbf{B}_{\nu}^{(y,n,t)}|}, \frac{|\mathbf{B}_{\nu}^{(z,n,t)}|}{|\mathbf{B}_d^{(z,n,t)}| + |\mathbf{B}_{\nu}^{(z,n,t)}|} \right), \quad (10)$$

where the primary source is given by $\mathbf{B}_d^{(n,t)} = (\mathbf{B}_d^{(x,n,t)}, \mathbf{B}_d^{(y,n,t)}, \mathbf{B}_d^{(z,n,t)})$ and the secondary source is given by $\mathbf{B}_{\nu}^{(n,t)} = (\mathbf{B}_{\nu}^{(x,n,t)}, \mathbf{B}_{\nu}^{(y,n,t)}, \mathbf{B}_{\nu}^{(z,n,t)})$.

The total MF at time t and for SQUID sensor n is given by $\mathbf{B}_d^{(n,t)} + \mathbf{B}_{\nu}^{(n,t)}$. The symbol $|\cdot|$ denotes the absolute value. Also the contribution of the secondary source for the magnitude of MFs was computed in the same manner:

$$\frac{\|\mathbf{B}_{\nu}^{(n,t)}\|_2}{\|\mathbf{B}_d^{(n,t)}\|_2 + \|\mathbf{B}_{\nu}^{(n,t)}\|_2}. \quad (11)$$

3. Results

3.1. Comparison of the volume conductor models

The MFs simulated using the multiple and single dipole sources with the different volume conductor models are shown in figure 3. The maps for each MF component and the MF magnitude are presented for each volume conductor model at three time points of the slow wave cycle (at $t = 0, 10$ and 20 s).

For both dipole configurations, the free space model resulted in the largest MFs, except in the y-direction. In the y-direction, the strongest MFs were computed with the sphere model. The MFs in the y-direction simulated with the half space model were equal to those with the free space model as there was no secondary source contribution in the direction normal to the half plane. The half space model produced the most similar MF distribution in the y-direction to that of the anatomically realistic torso model. The multiple dipole source produced larger MFs for all volume conductor models than when a single dipole source was used.

Using the law of cosines

$$a \bullet b = \|a\|_2 \cdot \|b\|_2 \cdot \cos \theta, \quad (12)$$

where θ is the angle between vectors a and b , the differences in the direction of the MFs simulated using the three simplified volume conductor models and the anatomically realistic torso model were computed (figure 4). On average, the direction of the MFs simulated with the half space model was the closest to those with the anatomically realistic torso model as shown by the means and standard errors in the graphs; however, these results show that none of the simplified volume conductor models were able to consistently produce MFs within 25° of those of the anatomically realistic torso model. The difference in MF direction produced by the multiple and single dipole sources for all models varied by less than 15° . Figure 5 shows the root mean square errors (RMSEs) between the MFs of the simplified volume conductor models and those of the anatomically realistic torso model. It should be noted that the errors for the free space and half space models were equal in the y-direction because the MFs simulated with those two models were equivalent in that direction. In addition, the MFs were almost similar to the torso model, although differences of up to 1 pT were observed. The free space model produced the largest differences in MFs compared to the torso model except in the y-direction, where the sphere model had the largest difference in MF values. For the single dipole source, the half space and sphere models both produced MFs that had a similar RMSE difference to the realistic torso model (less than 0.5 pT for all components and the magnitude).

3.2. Comparison of dipole source configurations

In order to qualify the difference between single and multiple dipole configurations, RMSEs between the MFs simulated with 320 dipoles and an equivalent single dipole were computed (figure 6). The MFs of the free space model were the most sensitive to the dipole source configuration in all directions. The sphere, half space and torso models produced MFs with similar RMSEs. The variation between each of these three models was less than 0.5 pT for

all components and the magnitude. This indicated that all models had a similar change in MF magnitude due to the different source models.

3.3. Contributions of secondary sources

The contribution of the secondary source to the total MFs is shown in figure 7; the percentages were computed using (10) for the x -, y -, z -directions and (11) for the magnitudes. It should be noted that the free space model did not yield a secondary source contribution in any of the three directions, and neither did the half space model in the y -direction.

The secondary sources generally contributed approximately 50% of the total MFs. However, the sign of the primary and secondary sources were generally opposite; meaning the resultant MF was typically half the size of the free-space model. Notable exceptions were in the y -direction where the half space model which had no volume conductor contribution, meaning the MF would be overestimated and have the same result as the free-space model and the torso model which had a secondary source contribution of approximately 30%. However, the MFs produced in the y -direction are known to be relatively weaker than the other two despite a smaller resistive effect from the secondary source.

4. Discussion

It is important to accurately measure and simulate MFs generated from GEA as they are known to be weak in nature (Allescher *et al* 1998, Bradshaw *et al* 2007, 2009, Richards *et al* 1996, Saijyo *et al* 1995). In addition, only the MFs in the direction normal to the coronal plane are typically measured experimentally, and the simulation results have shown that the MFs in this direction are generally much smaller than those in the other two directions (Komuro *et al* 2008). We have used two different dipole source configurations representing a single gastric slow wave event and four different volume conductor models to simulate the resultant MFs as measured by a multichannel SQUID magnetometer. The models representing the volume conductor were a free space, sphere, half space and an anatomically realistic model derived from CT images of a human. The source models consisted of 320 moving dipole sources derived from a simulation of GEA and the equivalent single dipole source. To our knowledge, this is the first time that the simulated gastric MFs of an anatomically realistic volume conductor model and simplified volume conductor models have been quantitatively compared. The effects of the volume conductor geometries and dipole source configurations on simulations of MFs generated from GEA were examined in terms of the strength, direction and contribution of the secondary sources.

The MFs resulting from the primary source were the intrinsic property of the dipole sources and were the same for all the volume conductors used in our simulations. As a consequence, the differences in the MFs produced by the different volume conductor models were solely due to the resistive impact of the volume conductor boundaries. Since the free space model did not have a volume conductor boundary, it therefore produced the largest total MFs in all the principle directions compared to the other volume conductor models. The only exception was the sphere model which produced larger MFs in the y -direction. This was because the volume conductor effects using this model had an additive effect rather than a resistive

effect shown by the other models. Significant differences in MFs orientation and magnitudes were observed between the anatomically realistic and simplified models. The direction of the MFs was greater than 25° and the difference in magnitude was approximately 1 pT. Therefore, caution should be taken if simplified models are to be used for interpretation of the experimentally recorded MFs.

We have shown that the half space model produced MFs most similar to the anatomically realistic torso model in the x - and y -directions when compared to the other simplified models despite the fact that the half space model does not yield any secondary source contributions in the y -direction. Therefore, if the MFs in only the y -direction were considered, the free space model would provide the same solutions as the half space model.

As the y -direction was perpendicular to the boundary of the half space model, the MFs in this direction could be thought as an infinite sphere case (Sarvas 1987) and the half space model did not produce secondary sources in the y -direction. However, it should be noted that the abdominal surface of a torso is not an absolute infinite plane; thus the secondary sources would still contribute to the total MFs based upon the enclosed curved surface of the torso model.

Simplified geometries, such as concentric homogeneous spherical shells, have been successfully used to approximate the realistic head shapes in magnetoencephalography source localization (de Munck 1988, Mosher *et al* 1999, Zhang 1995). This is because the contribution of the secondary source on the total MF is known to be zero when the SQUID sensor is orientated normal to an exact spherical volume conductor surface. However, with more complex geometries such as a torso, an accurate representation of the secondary sources becomes important (Hamalainen and Sarvas 1989). We found that with the use of a torso volume conductor, the secondary source had up to 30% contribution to the total MF in the y -direction. In the x - and z -directions, the sphere, half space and torso models all had secondary source contributions of between 45 and 55%. However, in all cases the standard errors were less than 3% indicating that the total secondary source contributions were relatively constant irrespective of the dipole source parameters. The relatively high contribution of the secondary sources indicated that an anatomically realistic volume conductor should be strongly considered for all numerical simulations and model-based interpretation of MF recordings.

The type of dipole source configuration had the greatest impact on the MFs for the free space model. The use of multiple dipole sources consistently produced larger MFs compared to the single dipole source. The MFs produced by the multiple dipole sources were approximately 1–2 pT larger than the single dipole source when the sphere, half space and realistic torso models were used with similar changes in magnitudes were observed for all three volume conductor models. In addition, the different dipole models altered the MF direction by approximately 10° . However, it should be noted that this change was relatively minor compared to the type of volume conductor model used. Both a single dipole and limited number of dipoles have previously been used to simulate the underlying GEA (Bradshaw *et al* 2003, Mintchev and Bowes 1997). Thus, additional investigations are required to determine the appropriate dipole configuration to represent the underlying GEA.

Several assumptions have been made in our simulations, and these need to be taken into account when drawing conclusions. The electric current dipoles used in our simulations have been derived from a simulation of a single slow wave. We have assumed that the simulation is realistic and also that the dipoles provide an accurate representation of the underlying slow wave activity. The stomach model used for the simulation of GEA had a constant thickness (Komuro *et al* 2008) and approximated tissue structure with five alternating layers (two layers of interstitial cells of Cajal and three layers of smooth muscle). Secondary sources arise from both cellular level and tissue level in a physiology preparation (Plonsey 1982). In this study, the variation of electrical activities at the cellular level was not modelled. As such the torso model used in this study was represented by a homogenous volume conductor. Since secondary sources can arise from the interfaces between regions with different electrical conductivities, the simulated secondary sources from the anatomically realistic torso model may have been underestimated. Komuro *et al* (2008) have previously shown the importance of stomach anatomy in MF simulations. However, in this study, only one anatomical model was considered. The shape, direction and position of the stomach are known to be highly variable between individuals; thus the relative effects of volume conductor models and dipole source configurations on resulting MFs will vary depending on the anatomical stomach model. Furthermore, the MF solutions simulated with simplified volume conductor geometries were only compared to those simulated with one torso geometry. The results have shown the importance of the volume conductor geometry in determining the secondary sources and the total MFs; however, it is possible that different torso geometries may result in different trends.

A mathematical model of the MFs generated from the underlying GEA provides a tool to help interpret MF measurements obtained using a multichannel SQUID magnetometer. Although our representation of the simulated MFs has yet to be carefully validated with experimental recordings, the large differences observed between the simplified and realistic volume conductor models infer that realistic models are necessary for interpretation of MFs due to GEA. If simplified models are to be used, then the half space model produces the most similar MFs, even though the model does not have a secondary source contribution in the direction normal to the boundary of the half space. In addition, if only the MFs normal to the coronal plane are analysed, then the free space model provides the same results as the half space model does. The use of realistic models has been largely omitted in previous studies due to the additional requirement of accurate anatomical data and the computational costs involved in model construction and simulation. However, with the availability of medical imaging modalities such as MRI, CT and 3D ultrasound, methods for rapid construction of subject specific models (Cheng *et al* 2005) should strongly be considered in future simulation studies.

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Figure 1.

Relative positions of the torso and sensor locations used in the MF simulations. The pink surface illustrates the position of the stomach within the torso model. The yellow spheres show the distribution of the 19 SQUID sensors located in the coronal plane just above the skin surface. The SQUID sensors were arranged in a hexagonal pattern corresponding to the gastrointestinal SQUID array (model 637i Tristan Inc., San Diego, CA). The purple ring indicates the extent of the SQUID magnetometer relative to the body.

Figure 2.

Schematic diagrams for the (a) free space, (b) sphere, (c) half space and (d) anatomically realistic torso models as viewed in the sagittal plane. The dotted outline of the stomach was not explicitly included in the volume conductor models and has been included here for illustrative purposes. The conductor is denoted by G , and σ_G and σ_A are the electrical conductivities in the conductive and non-conductive regions, respectively. The locations of the dipole and field points from the origin are denoted by $\mathbf{r}_o$ and $\mathbf{r}_f$, respectively. The vector $\mathbf{r}_{obs}$ is the displacement vector from $\mathbf{r}_o$ to $\mathbf{r}_f$, and its length is denoted by r_{obs} . Vector $\mathbf{p}$ is the dipole source located in G .

Figure 3.

Spatial maps of the simulated MFs in the x-, y-, z-directions and their magnitudes for one slow wave. The MFs were simulated with (a) 320 dipoles and (b) an equivalent single dipole. Each row corresponds to a different volume conductor, and the spatial maps at time 0, 10 and 20 s corresponding to the beginning, middle and end of the slow wave are presented for each case. The colour bars at the top show the ranges of the MFs in pT.

Figure 4.

Spatially and temporally averaged differences in direction between the MFs simulated with the three simplified volume conductor models and the anatomically realistic torso model over one slow wave cycle. Each bar graph is presented as the mean $\pm$ standard error.

(a) RMSEs for 320 dipoles

(b) RMSEs for a single dipole

Figure 5.

Spatially and temporally averaged root mean square errors (RMSEs) between the MFs of the three simplified volume conductor models and the anatomically realistic model over one slow wave cycle. The results are shown for the MFs in the x -, y -, z -directions as well as for their magnitudes for both (a) 320 dipoles and (b) a single dipole. Each bar graph is presented as the mean $\pm$ standard error.

Figure 6.

Spatially and temporally averaged root mean square errors (RMSEs) between the MFs simulated with 320 dipoles and a single dipole over one slow wave cycle for each of the four volume conductor models. The results are shown for the MFs in the x -, y -, z -directions as well as for their magnitudes. Each bar graph is presented as the mean $\pm$ standard error.

Figure 7.

Contribution of the secondary source to the full MF using a multiple dipole source and the sphere, half space and torso models. Each bar graph is presented as the mean $\pm$ standard error and results have been temporally averaged.