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Characterization of electrophysiological propagation by multichannel sensors

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Abstract

Objective—The propagation of electrophysiological activity measured by multichannel devices could have significant clinical implications. Gastric slow waves normally propagate along longitudinal paths that are evident in recordings of serosal potentials and transcutaneous magnetic fields. We employed a realistic model of gastric slow wave activity to simulate the transabdominal magnetogastrogram (MGG) recorded in a multichannel biomagnetometer and to determine characteristics of electrophysiological propagation from MGG measurements.

Methods—Using MGG simulations of slow wave sources in a realistic abdomen (both superficial and deep sources) and in a horizontally-layered volume conductor, we compared two analytic methods (Second Order Blind Identification, SOBI and Surface Current Density, SCD) that allow quantitative characterization of slow wave propagation. We also evaluated the performance of the methods with simulated experimental noise. The methods were also validated in an experimental animal model.

Results—Mean square errors in position estimates were within 2 cm of the correct position, and average propagation velocities within 2 mm/s of the actual velocities. SOBI propagation analysis outperformed the SCD method for dipoles in the superficial and horizontal layer models with and without additive noise. The SCD method gave better estimates for deep sources, but did not handle additive noise as well as SOBI.

Conclusion—SOBI-MGG and SCD-MGG were used to quantify slow wave propagation in a realistic abdomen model of gastric electrical activity.

Significance—These methods could be generalized to any propagating electrophysiological activity detected by multichannel sensor arrays.

Keywords

Bioelectric phenomena; biological system modeling; electrogastrography; magnetogastrography; SQUIDs

I. Introduction

Multichannel sensor arrays record spatiotemporal data from underlying electrophysiological activity that can include information about propagation [1], and can be useful in encephalography, cardiology, and myography [2-4]. A variety of techniques for decoding or decomposing spatiotemporal information from multichannel array recordings of electrophysiological activity is available, but systematic methods to characterize physiological propagation are not well-developed. Innovative algorithms that characterize electrophysiological propagation from multichannel electrodes in the gastrointestinal (GI) serosa are available [5, 6], but the ability to characterize the propagation of electrical activity in the gastric musculature from external sensor arrays has not been well studied.

Many of the mechanical functions of the stomach depend on the integrity of underlying electrical activity generated in the muscular syncytium defined by layers of smooth muscle, interstitial cells of Cajal (ICCs) and the enteric nervous network [7]. Previous electrophysiological studies have demonstrated that a gastric pacemaker initiates activity from the upper corpus on the greater curvature. Activity then propagates in the syncytial network toward the antrum and pylorus. The frequency of this electrical activity is generally around 3 cpm in normal humans. Recent multichannel serosal electrode studies have shown that multiple slow waves can be measured simultaneously [8-12].

We have previously introduced the idea of magnetic detection methods for noninvasive monitoring of the gastric slow wave [13, 14]. The properties of Superconducting QUantum Interference Devices (SQUIDs) allow measurement of magnetic fields in the picoTesla (pT) range typical of bioelectric sources. Previous studies showed that the magnetogastrogram records signals with the same waveshape and frequency as slow waves recorded with internal serosal electrodes [15, 16]. In addition, studies show that slow wave propagation can be detected with multichannel magnetometers [17], and magnetogastrograms (MGG) propagation characteristics are altered in disease and by both mechanical and pharmacological uncoupling [18, 19].

Propagation of slow waves can be characterized from serosal or mucosal electrode potentials by measuring time-of-arrival (TOA) and subsequent velocity of particular wave characteristics [20, 21]. Similar methods can be used for MGG signals, but spatial filtering properties of the abdomen and geometrical characteristics of magnetic fields from intra-abdominal current sources cause extracorporeal magnetic field maxima to appear displaced transverse from the locations of internal sources. A robust method for the characterization of propagation is essential to the effective interpretation of noninvasive biomagnetic data.

In this paper, we evaluate the characterization of propagation of gastric slow waves recorded in multichannel biomagnetic sensors. We analyze specific methods using a realistic abdominal model of the gastric slow wave, an analytical half-layer model and by direct MGG measurement using a SQUID magnetometer.

II. Methods

A. Mathematical Modeling

Three configurations of propagating source models, all of which were based on current dipoles, were used to simulate magnetic fields. Dipolar current sources have been used extensively in electrophysiological modeling, including models of the gastric slow wave [22-25]. Our propagation pattern geometries were based on data from the visible human project [26], as shown in Figure 1. The variety of geometries was chosen to investigate the efficacy of propagation determination when the sources are (i) superficial dipoles (SD), (ii) deep dipoles (DD), or (iii) to examine the effect volume conduction in the realistic abdomen by comparing with a semi-infinite horizontal layer (HL) model in which tissue layers with different conductivities representing muscle, fat and skin are layered horizontally above the abdominal cavity that contains the source dipole.

To represent a single propagating slow wave event, a stomach model was used (Figure 1a). The solution points representing continuum cells were embedded within the stomach musculature and the Aliev cell model [27] was used at each solution point to represent the ionic current. The continuum-based bidomain equations were then solved at each time point, and dipole sources were derived from the transmembrane potential gradient at each solution point. These points were combined to produce a single dipole with varying location and orientation [28]. Lastly, the dipole sources were located in a fixed plane parallel to the sensor array so that the distance between the sources and sensors remained constant. The superficial dipole (SD) configuration was set with the dipoles 50 mm below the sensor array, and the deep dipole (DD) source configuration had dipoles at 170 mm below the sensor array (Figure 1c). Although the source-sensor distance is probably more realistic in the SD configuration, the DD configuration tests the effects of source depth on the propagation estimates.

The source models were embedded into a realistic torso volume conductor and a simplified analytic HL model. For the SD and DD configurations, we used a realistic abdominal model derived from the Visible Human dataset [26, 29]. The model (shown in Figure 1b) consisted of a boundary element surface representing the skin. A total of 110 sensors (Figure 1b) were used to simulate magnetic fields. Sensors were separated by 30 mm in the cranial-caudal direction and 25 mm in the lateral direction.

For the HL configuration, the abdomen was modeled as a horizontally-layered semi-infinite volume conductor with horizontal layers representing abdominal cavity, subcutaneous fat, muscle and skin. Analytic magnetic field solutions are available for the HL volume conductor model [28, 30]. The source model was a planar dipole progression similar to the SD configuration located 50 mm below the sensor array.

B. Propagation Analysis

We evaluated two methods for determining propagation characteristics from external magnetic field measurements.

1) Surface Current Density—The Surface Current Density (SCD) method computes an estimate of the current density in the surface of the sensor plane that would produce the observed magnetic field [18, 31]. The SCD method was initially proposed by Hosaka and Cohen [32] and Cohen and Hosaka [24]. An estimate of the curl of the magnetic field $\mathbf{B}$ is computed by calculating

$$\mathbf{J}_s = \frac{1}{\mu_0} (\nabla \times \mathbf{B}) \quad (1)$$

where the vector $\mathbf{J}_s$ represents the surface current density and μ_0 is the magnetic permeability of free space.

In fact, there is no current density at sensor locations as the curl of the magnetic field is zero. However, non-zero current density values at surfaces can be related to current running under the surface parallel to the sensor plane. The simulated magnetic fields on the SQUID array were interpolated using a Kriging method on a denser plane and SCD values were calculated from these fields. By tracking the maximum of the SCD in a sequence of time we hypothesize that the location of the underlying source can be estimated and the propagation velocity computed. It should be noted that most SQUID magnetometers are only designed to record the magnetic fields in the orientation normal to the volume conductor (denoted as B_z). In most cases, the B_x and B_y terms are therefore neglected in the estimation of the SCD calculations, and the calculations are reduced to only taking the spatial derivative of one magnetic field component.

$$\tilde{\mathbf{J}}_s = \frac{1}{\mu_0} \left(\frac{\partial B_z}{\partial y} \hat{\mathbf{x}} - \frac{\partial B_z}{\partial x} \hat{\mathbf{y}} \right) \quad (2)$$

If there are no significant contributions from x and y magnetic field components, or if these components have little spatial variation compared to that of the z component, then this expression represents the current density according to Maxwell's equations. Any spatial dependence as well as any errors that arise from approximating the partial differentials by differences could distort the accuracy of the SCD estimate as an inverse assessment of the actual current density. We have used SCD previously to estimate propagation characteristics of gastric magnetic fields with z -component fields [18, 31]. However, in our simulations, the magnetic field was not restricted to a setup determined by a SQUID magnetometer. Therefore, all three components of the magnetic field were simulated in our study.

2) Second Order Blind Identification—Second Order Blind Identification (SOBI) is a blind source separation (BSS) algorithm that has been applied to electroencephalogram

(EEG) [33, 34] and magnetoencephalogram (MEG) recordings [35, 36]. This method was also used to separate GI source signals in magnetoenterogram (MENG) recordings [37]. SOBI considers cross-correlations at multiple time delays instead of simply minimizing the instantaneous correlation with a fixed temporal delay of zero. In this way, SOBI supports the use of temporal information present in the time series for source separation and takes into account the influence of other sources after a certain time delay.

The SQUID measurements, $\mathbf{x}$, are described by

$$\mathbf{x}(t) = \mathbf{A}\mathbf{s}(t) + \mathbf{n}(t) \quad (3)$$

where SQUID data are sampled at times t , and each sensor's data is arranged in rows: $\mathbf{x}(t) = [x_1(t), x_2(t), \dots, x_m(t)]^T$, where $x_i(t)$ is the recording on SQUID channel i . Similarly, $\mathbf{s}(t) = [s_1(t), s_2(t), \dots, s_n(t)]^T$, where $s_j(t)$ is the j th underlying component. $\mathbf{A}$ is the transfer function between sources and sensors and $\mathbf{n}(t)$ represents the noise, assumed to be spatially and temporally white. Since both $\mathbf{A}$ and $\mathbf{s}$ are not known, the matrix $\mathbf{A}$ was first estimated using the second order statistics of the measurements or data [38]. The estimates of the underlying sources were then computed by

$$\mathbf{s} = \mathbf{A}^{-1}\mathbf{x} \quad (4)$$

We applied SOBI to raw magnetic field data and identified SOBI components that were primarily sinusoidal with dominant frequencies in a range normally identified with the gastric slow wave (2.5 – 4 cpm). We then reconstructed magnetic fields at each sensor location using only the gastric SOBI components and produced spatial maps of the signal strength at different time instants. These spatiotemporal SOBI maps generally contained more well-defined propagation patterns than corresponding patterns developed from noisy data.

3) Propagation Velocity Computation—The measurements consisted of N_c channels or sensors recording N_t time points. For each of the N_t time points, a spatial map was formed by plotting data from each of the N_c sensors at the 2-dimensional location of the sensor. These data could be raw magnetic field data, SCD values or SOBI-reduced magnetic field data. For the MGG and SOBI-MGG maps at each time point, the spatial location of the midpoint between the map maximum, $\mathbf{r}_{\max} = (x_{\max}, y_{\max})$, and the map minimum, $\mathbf{r}_{\min} = (x_{\min}, y_{\min})$ was computed, $\mathbf{r}_{\text{mid}} = (\mathbf{r}_{\max} + \mathbf{r}_{\min})/2$, and the location of this midpoint was tracked over time. The velocity of propagation was estimated as

$$\bar{v} = \frac{|\mathbf{r}_{\text{mid}}(t_f) - \mathbf{r}_{\text{mid}}(t_i)|}{\Delta t} \quad (5)$$

A similar procedure was followed for computing SCD propagation velocities, except that the map maximum was tracked instead of the midpoint between the maximum and minimum, since the SCD presumably represents a more direct measurement of source current

(magnetic field maxima from dipolar sources, and hence maxima in SOBI patterns necessarily occur at x, y locations offset from the dipolar source location). We also investigated the results obtained with SOBI when tracking pattern maxima instead of pattern midpoints. We report the results in terms of the mean-square error (MSE) in locations of both x and y dipole positions rather than a one-dimensional MSE since the dipolar field pattern could preferentially bias localization in a particular dimension.

Small time durations result in localized estimates of propagation velocity over a small region, while longer durations produce an estimate of the average velocity of propagation generally over a larger region of tissue. In human tissue, the velocity of slow wave propagation in the stomach is not constant, but rapidly accelerates near the antrum of the stomach [9]. Thus, one can investigate the dynamics of slow wave propagation by using smaller time intervals that allow characterization of changes in propagation velocity. For many purposes, however, the average propagation velocity suffices to characterize slow wave dynamics. Since noninvasive measurements of slow wave activity necessarily involve spatial averaging on the order of at least several millimeters if not centimeters due to volume conduction, we concentrated on estimates of the average propagation velocity.

Four different computation algorithms were developed to estimate propagation velocity. Method PV_A used the average of the first five time samples of a given data sequence as t_i and the average of the last five as t_f in equation (5). For Method PV_B, we used the average of the first half of the points in the data sequence as t_i and the average of the last half as t_f . Method PV_C used the 2nd five points in the sequence as t_i and the 2nd to the last five points as t_f and in Method PV_D, we computed the velocity at each time point and averaged the resulting instantaneous velocities to obtain the average velocity estimate. To test the performance of algorithms in the presence of noise, we also added 60% white noise to the magnetic field data using MATLAB's random number generator with a sequence of random numbers with a variance 60% that of the noise-free data.

4) Experimental Validation—To validate these methods, we performed simultaneous measurements of the multichannel MGG with multichannel serosal electrode recordings in porcine subjects ($N = 5$). Experiments were performed with the approval of the Institutional Animal Care and Use Committee at Vanderbilt University. After anesthesia and laparotomy, we attached a custom-built 48-channel serosal electrode platform (1.9 cm inter-electrode spacing in 4 rows of 12 electrodes) to the anterior gastric serosal surface. The electrode was constructed from nonmagnetic materials and demonstrated to have no effect on magnetic data by motion in its approximate location underneath the SQUID magnetometer. The laparotomy was closed and the animal was positioned with the stomach centered underneath the array of a multichannel SQUID biomagnetometer (Model 637, Tristan Technologies, San Diego, CA). Electrode signals were amplified (BioSEMI, Amsterdam) as were SQUID signals (Model 5000, Quantum Design, San Diego, CA), and the analog signals were digitized and stored on a personal computer. Signals were sampled at 3000 Hz with hardware low-pass filters set at 1 kHz at 16-bit resolution. We downsampled and filtered signals to 30 Hz for easier processing in MATLAB (bandwidth 1-120 cpm) and computed Fast Fourier Transform (FFT) spectra (using 120-s windows with Hamming windowing) to identify the dominant frequencies of signals. The SOBI algorithm was used to decompose

SQUID signals into components and noise components were eliminated [35, 39]. SOBI-reconstructed magnetic field spatiotemporal maps were composed for computation of propagation velocity. We also computed SCD maps from the filtered spatiotemporal magnetic field data.

Electrode signals were visually analyzed to detect time-of-arrival (TOA) of waveforms in adjacent electrodes in the electrode platform. Propagation velocity in the serosal electrodes was computed by dividing the electrode separation by the time-of-arrival of electrode waveforms.

III. Results

A. MGG, SOBI and SCD

Magnetic field maps from the three dipole patterns are shown in Figure 2. The SD patterns showed spatiotemporal signatures similar to the HL patterns since both configurations were from sources at a depth of 50 mm. The temporal progression was also similar with the DD patterns except that the magnetic field contours appeared larger because of the depth of the source dipoles. The time sequences of these field maps showed clear evidence of underlying propagating current sources, and characteristics of these model field maps were relatively simple to track with the typical dipolar field patterns, particularly in the SD and HL models.

Figure 3 shows SOBI-MGG maps corresponding to the magnetic field maps shown in Figure 2 at the same five time instants. These maps were reconstructed from SOBI components computed from MGG that are primarily-sinusoidal with dominant frequencies in the gastric range. The SOBI method increased the signal-to-noise ratio compared with MGG maps in Figure 2 by removing non-gastric signal components. SOBI conformational patterns were similar to magnetic field patterns. The raw MGG and SOBI pattern maxima do not correspond to the dipole location exactly because of the double-lobed nature of the dipolar pattern. The original dipole location is shown by a circle and the pattern midpoint tracked for the propagation velocity estimate is indicated by a cross. Tracking the midpoint between the pattern maximum and minimum presumably gives a better estimate of the propagation velocity since the pattern itself propagates with a similar velocity, but we tested this assumption, as shown below in the discussion of Figure 6.

SCD-MGG maps corresponding to the MGG maps in Figure 2 at the same time points are shown in Figure 4. These patterns are characterized by a single maximum that propagates across the sensor array. As in Figure 3, the original source location is indicated by a circle in each frame and the SCD estimate of the dipole location computed from the pattern maximum is shown as a cross.

B. Source Localization and Propagation Velocity

The results of the source locations computed from SCD and SOBI maps are shown in Figure 5 for the three source configurations (a, b and c) without and with noise added (Figure 5.a.i and ii, respectively). For the SCD method, we plot the x - y locations of the maximum of the surface current density computed from the magnetic field maps as squares with locations at successive time points connected by dashed lines. The midpoints of the SOBI-reduced

magnetic field patterns are plotted as circles with locations at successive time points connected by dotted lines. These SCD and SOBI locations are plotted along with the actual dipole positions, connected by the solid lines. Location estimates are most accurate for both SOBI and SCD algorithms in the SD and HL models with more spread in the DD model, as might be expected, but generally SOBI estimates perform slightly better than SCD. Noise increases the spread of dipole estimates from the actual location, particularly with deep source (DD) SCD estimates.

The mean squared error between actual and estimated x and y dipole locations for the SCD and SOBI methods is summarized in Figure 6a for each source configuration. Generally, SOBI estimates resulted in lower Mean Squared Error (MSE) than SCD estimates with the exception of the mixed results in the DD model. In SD, HL and noisy HL models, the x position dipole location estimates were more accurate than y position estimates; y estimates were more accurate in the noisy DD model and the results were mixed for the other two configuration.

For computing propagation velocity, the absolute dipole position is not as critical as the relative position over time. Figure 6b shows the correlation (R^2) from a regression analysis of x and y dipole position time sequences from SCD and SOBI compared with actual dipole location time sequences, which removes any absolute position bias. These values indicated that for SD, noisy SD and noisy HL models, SOBI location estimates are better correlated with the actual dipole location. SCD estimates correlated better in the DD and noisy DD configurations, and the methods were equivalent in the noise-free HL configuration. Both SCD and SOBI perform better with more superficial sources, but the modestly better performance of the SCD method for deep dipole source configurations suggests that the method mediates the effect of volume conduction that may distort the location of extrema in the double-lobed SOBI field patterns.

Because the two lobes of the dipolar magnetic field pattern are less well-defined with the DD source configuration, we assessed tracking the pattern maximum instead of the midpoint for propagation estimation. The result was, generally, greater error in the absolute dipole location estimation when tracking the midpoint, markedly in the x component (762% increase in MSE_x for SD; 289% increase in MSE_x for DD), and less pronounced for the y component (30% increase in MSE_y for SD; 81% increase in MSE_y for DD). For the sequence correlation, there was essentially no change in either component for the SD configurations (0% difference in both x and y correlation), but a marked increase in sequence correlation for the DD configuration (4300% increase in x correlation; 209% increase in y correlation). This improvement in sequence correlation translates to a more accurate estimation of propagation velocity when tracking pattern midpoint with deep sources where maxima of double lobes may extend outside the measurement region. MGG patterns without clearly defined double lobes may thus benefit from tracking the pattern maximum rather than the midpoint when using SOBI-propagation estimation.

As expected, the best position estimates were obtained from the SD and HL source configurations. Apparently, the source-sensor distance is most influential in determining dipole location. The DD configuration with deep sources produced the most error in dipole

location estimates. The presence of noise predictably increased the localization error and decreased the correlation between estimated and actual dipole locations; however, the SOBI routine performed substantially better in the presence of noisy forward signals than SCD. The process of selecting signal components in the SOBI algorithm necessarily removed noise components from the data and increased the signal-to-noise ratio.

Propagation velocities computed from SCD and SOBI position data using the four methods described above are presented in Figure 6c. Methods PV_B and PV_C produced the most accurate propagation velocity estimates. SCD performed slightly better than SOBI with the superficial source under noise-free conditions, but SOBI out-performed SCD otherwise. The presence of noise increased errors in propagation velocity estimates from both SCD and SOBI, but SOBI tended to produce a more accurate estimate of propagation velocity in the presence of noise, with the exception of the deep source configuration and the horizontal layered configuration using method PV_B .

C. Experimental Validation

SCD and SOBI patterns (Figure 7a and b) computed from the magnetic field data in the porcine experiment were used to compute propagation velocity. We also computed the corresponding propagation velocity from direct serosal electrode measurements using the 48-channel serosal electrode platform. A typical serosal electrode recording from adjacent channels is shown in Figure 7c. Patterns in the SCD and SOBI maps (Figure 7a and b) varied somewhat from those observed in the dipole model, but general characteristics, including single maxima in SCD and double lobes in SOBI maps, were similar.

The average propagation velocities computed by each method from serosal slow wave electromyogram (EMG), SOBI-MGG and SCD-MGG are shown in Figure 8. Propagation velocities from MGG estimates agreed with serosal EMG computations to within 1 mm/s, with the exception of MGG-SOBI using Method PV_B , which consistently resulted in a velocity estimate that was significantly larger than the velocity measured by serosal electrodes. Estimates using Method PV_C were most consistent with the propagation velocity from serosal electrodes. Using the first few and last few data points as in Method PV_A , or the average of the instantaneous velocities of nearest neighbors (Method PV_D) seemed to introduce spurious errors in the velocity estimates.

IV. Discussion

The well-coupled cellular electrical syncytium in the GI tract allows bioelectric activity originating in ICC and the enteric nervous system to propagate through the smooth musculature. Similar phenomenology has driven the field of cardiac electrophysiology and resulted in numerous clinically significant diagnostic capabilities using the electrocardiogram that also motivate treatment methodology. Concomitant advances in electrogastrography have not occurred, presumably because of difficulties in accurately relating cutaneous electrical recordings to tissue-level electrical events. The EGG is capable of evaluating temporal dynamics, but these parameters do not seem to correlate well with pathological conditions [40-44]. Spatiotemporal properties of the slow wave, however, may have a more direct correlation with disease [17, 45, 46], and the SQUID magnetometer is the

only methodology to date able to accurately assess these parameters non-invasively. Our study focused on the MGG, but a similar analysis of the multichannel EGG would be interesting, particularly in light of the different effects of volume conduction on cutaneous potentials and magnetic fields [47].

Clinically, assessment of the propagation of electrical activity in the gastric syncytium could prove critical to the distinction of specific disease processes with potentially overlapping symptomology. Functional gastric disorders like gastroparesis, functional dyspepsia and chronic unexplained nausea and vomiting often present with similar symptomology but have distinct etiologies and require specific treatment protocols. Current diagnostic methods including electrogastrography and gastric emptying do not provide criteria specific and sensitive enough to identify etiological differences [48-50].

In this work, we developed a forward model of the GI electrical activity using propagating current dipoles in a realistic abdominal volume conductor. The electric and magnetic fields corresponding to current dipoles are well understood, and the field patterns produced by our model contained characteristic dipolar signatures, with normal-component magnetic fields producing a double-lobed pattern. We used similar dipole source models in previous work [23, 31]. The question of whether a single current dipole represents a reasonable model for the gastric slow wave remains, particularly since experimental data tends to resemble that shown in Figure 7, where the double-lobed pattern is not particularly evident in magnetic field data. These patterns could be represented by a single propagating dipole if the source-sensor distance was comparatively large, as with the deep-dipole model (Figure 2b).

Although the distance between our sensors and the gastric musculature is generally not as large as the deep dipole model, the actual gastric slow wave is probably better modeled as a wavefront of adjacent dipoles rather than as one single dipole. The raw pattern maximum (and minimum) from SOBI maps is offset from the actual dipole location because of the double-lobed dipolar magnetic field pattern above the single propagating dipole in our model, but tracking the midpoint between the pattern maximum and minimum mediates this effect to some extent. Nevertheless, the positional offset of lobe extrema in SOBI maps likely explains the higher y position MSE in the SD, HL and noisy HL cases. Otherwise, the variation in x and y position MSE is likely caused by volume conduction and noise effects.

To examine the effect of a wavefront of source dipoles more fully, we also tested a model consisting of a concentric ring of dipoles propagating down the gastric body (Figure 9c). Field patterns from the ring model are compared with those from the SD and DD models in Figure 9. The concentric ring source created a field pattern with similarities to both SD and DD source models. Dipole locations and propagation velocities obtained from the concentric ring source were similar to the SD and DD configurations ($R_x^2=0.72$; $R_y^2=0.78$; $MSE_x=1.0$ cm; $MSE_y=1.0$ cm; $PV_A=7.6$ mm/s; $PV_B=5.6$ mm/s; $PV_C=4.6$ mm/s; $PV_D=8.8$ mm/s). Although a more accurate source model for gastric activity is likely even more complex, the concentric ring model shows that additional complexity alone does not invalidate these results. The single dipole model provides a reasonable basis for evaluating methods to assess gastric slow wave propagation.

The velocity of electrical activity propagating in a cellular syncytium like gastric tissue can be difficult to quantify. Currents flow in the syncytium in response to pacemaking activity by enteric neurons and/or ICC [7, 51]. Tissue anisotropy creates directional dependence for propagation velocities, and generally these dependencies are separated into transverse and longitudinal components. The transverse propagation velocity is believed to be significantly larger than the longitudinal, so that electrical activity spreads quickly in the circumferential direction around the body of the stomach and more slowly from the antrum toward the pylorus. Further, recent data have suggested that there are spatial variations in longitudinal propagation velocities that have only been measured with the use of high-density serosal electrode arrays. Propagation velocities calculated from extracorporeal magnetic field measurements in our realistic volume conductor model agreed well with the known source propagation velocity. Application of the same methods to measure propagation velocity in experimental animals led to results consistent with previous experiments and with the model results. Discrepancies between model data and experimental data SCD and SOBI patterns likely result from a simplified source model, as discussed before. Nonetheless, the propagating dipole model is useful for corroborating experimental findings and testing analysis algorithms.

V. Conclusion

Gastric slow wave propagation velocity was best assessed in most situations using the SOBI algorithm. The SCD method was more sensitive to external noise, but this increased noise sensitivity was balanced by increased accuracy in the case of deep sources. Thus, a high signal-to-noise ratio is essential to an accurate estimate of propagation velocity. Therefore, the optimal method to use in a realistic recording environment depends strongly on the source characteristics and the noise profile. This study validates the accuracy of magnetic field recording techniques to non-invasively determine gastric slow wave propagation including velocity and location of dipolar sources. Ultimately, the identification of propagation in multichannel biomagnetic recordings could be used to determine the underlying characteristics of functional GI motility disorders such as gastroparesis and functional dyspepsia. The methods we have developed would be generally applicable to a wide variety of multichannel sensor recordings of propagating activity.

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Figure 1.

Boundary element skin and stomach models and sensors. (a) The stomach model, (b) the torso model with the 110 sensors in the hypothetical magnetometer system used to evaluate magnetic field propagation, and (c) the coronal plane of sources and sensors in the torso model. The distances of SD and DD from the SQUID are 50 mm and 170 mm, respectively.

Figure 2.

Magnetic field maps from (a) the SD, (b) DD, and (c) the HL source configurations used to characterize propagation velocity plotted every four seconds as a source dipole moves across the stomach from left-to-right. The map at the top corresponds to $t = 4$ s and the bottom map corresponds to the end of the sequence at $t = 20$ s. Magnetic fields at successive 4 s intervals are shown as the dipole traverses the gastric musculature. The location of the source dipole is shown as a circle in the field maps. The corresponding electric potential computed on the

body surface from the SD configuration is also shown for comparison in (d). For (a)-(d), $c_{min}/c_{max} = [\pm 15 \text{ pT}, \pm 2 \text{ pT}, \pm 15 \text{ pT}, \pm 25 \text{ mV}]$, respectively.

Figure 3.

SOBI reconstructions of the data in Figure 2 for the (a) SD, (b) DD and (c) HL source configurations. The location of the original source dipole is represented by a circle and the SOBI estimate of the dipole location by a cross. For (a)-(c), $c_{min}/c_{max} = [\pm 5 \text{ pT}, \pm 0.2 \text{ pT}, \pm 2 \text{ pT}]$, respectively.

Figure 4.

Surface current density maps computed from magnetic fields in Figure 2 for (a) SD (b) DD and (c) HL source configurations. The location of the original dipole is represented by a circle and the SCD estimate of the dipole location by a cross. SD conformations corresponding to the double-lobed magnetic field patterns from a single dipole tend to show one predominant lobe whose maximum is centered above the dipole location. For (a)-(c), $c_{min} = 0$ A/m²; c_{max} are, respectively: [5.5, 0.2, 2.5] μ A/m²

Figure 5.

Dipole location estimates from SCD propagation method (squares) and the SOBI propagation method (circles) in (a.i) superficial dipole, (b.i) deep dipole, and (c.i) horizontal layer models without noise. Actual dipole locations are represented by the solid line. The origin and termination of the propagating dipole sequences are indicated by the time in seconds; numbers are oriented normally for the actual dipole positions, leaning right for SOBI estimates and left for SCD. The performance of the algorithms in the presence of noise is illustrated by panels a.ii, b.ii, and c.ii for the SD, DD and HL models, respectively.

Figure 6.

(a) Mean square error (MSE) between SCD and SOBI dipole location estimates and actual dipole location. (b) Correlation of SCD/SOBI dipole location estimate with actual dipole location. Squares mark SCD dipoles while circles represent SOBI estimates, with x positions represented by open symbols and y positions represented by filled symbols. Generally, SOBI estimates result in higher correlation with actual dipole positions and lower mean square error. (c) Accuracy of propagation velocity (PV) computation using the four methods in each of the different source configurations tested. Circles indicate propagation velocity estimated using SOBI and crosses indicate that the SCD method was used. The dotted line represents the actual propagation velocity. Generally, SOBI and SCD show comparable performance

for computing propagation velocity in noise-free data with methods PV_B and PV_C showing the highest accuracy. SOBI propagation calculations tend to outperform SCD with noise except in the deep dipole configurations for methods PV_A and PV_B , PV_C .

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Figure 7.

Data from porcine experiment. (a) SCD map, (b) SOBI map, and (c) electrode platform recordings. The cross in (a) and (b) show the SCD and SOBI estimate of the underlying source. All show evidence of propagation of electrical activity. For (a) $c_{min}/c_{max} = [0, 2.5]$ $\mu\text{A}/\text{m}^2$, and for (b) $c_{min}/c_{max} = \pm 0.25$ pT.

Figure 8.

Results of propagation velocities determined from porcine MGG data calculated by each of the four methods.

Figure 9.

Magnetic field patterns from (a) the SD, (b) DD and (c) ring dipole configurations at successive time instants. The spatial distribution of field patterns from the ring dipole model is intermediate between the SD and DD models. For (a)-(c), $c_{min}/c_{max} = [\pm 15, \pm 2, \pm 15]$ pT, respectively.