Published in final edited form as:

IEEE J Biomed Health Inform. 2018 May ; 22(3): 758–765. doi:10.1109/JBHI.2017.2690879.

Tracking Fetal Movement Through Source Localization From Multisensor Magnetocardiographic Recordings

Recep Avci, PhD [student],

Bioinformatics program at University of Arkansas Little Rock (UALR)

James D. Wilson, PhD,

SARA Research Center at University of Arkansas for Medical Sciences (UAMS)

Diana Escalona-Vargas, PhD [Member, IEEE], and

SARA Research Center at University of Arkansas for Medical Sciences (UAMS)

Hari Eswaran, PhD [Member, IEEE]

SARA Research Center at University of Arkansas for Medical Sciences (UAMS)

Abstract

Due to its high spatial and temporal resolution, fetal magnetocardiography (fMCG) measurements have been used for fetal movement (FM) detection in several studies which considered the changes in the amplitude and/or morphology of measured fMCG signals. Using source localization for fMCG measurements, we propose a novel method to fit a magnetic dipole moment to fetal heart signals and investigate the positional changes of magnetic dipole in order to detect FMs. We first split each fMCG recording into 6-second time windows. Then the magnetic dipole location and orientation for each time window are estimated using our inverse solution model. Finally, the distance between magnetic dipole positions in adjacent time windows is computed. Also, we calculate the dot products of the normalized magnetic dipoles to monitor the orientational changes. We analyzed 28 fMCG measurements from 23 subjects to investigate accuracy of the dipole fitting results. For each dipole fit, our model described the measured data with a goodness of fit value over 97% and with a fitting error of less than 2%. We observed that magnetic dipole positions significantly moved for some time windows. The time points at which the significant movement was observed were correlated with the heart rate acceleration as well. In addition to identifying the time points of the movement, our method is capable of observing rotational movement checking orientation of the dipoles.

Index Terms

Fetal Movement; Inverse Solution; Fetal Magnetocardiography

I. Introduction

Fetal movement (FM) is an important sign of fetal wellbeing and reduced FM may be an indicator of fetal anomalies. Although maternal perception is the oldest and most widely used method for monitoring FM, it is highly subjective and can be misleading [1].

Ultrasound investigations, on the other hand, provide an objective assessment but the

integration of ultrasound with other fetal assessment tools like fetal magnetoencephalography (fMEG) or fetal magnetocardiography (fMCG) is not practical. Therefore fetal evaluation by ultrasound generally limits the observer to visual assessment of a single parameter at a time, either heart rate or movement. By using Doppler Ultrasound, it was shown that fetal heart rate (fHR) acceleration was highly correlated with FM after 24 weeks of gestation [2]. Therefore, changes in fHR has been employed as an indicator of FM. The changes in fHR was classified into four different patterns by Nijhuis using ultrasound studies [3]. These patterns define fetal behavioral states and corresponding fHR activities. Nevertheless, this technique alone is not sufficient to identify the characteristics of FM.

Due to its high spatial and temporal resolution, fMCG has been used to investigate changes in fHR in order to enhance the interpretation of FMs [4], [6]. Several studies have investigated FM by examining the changes in fMCG signals [4], [5], [6], [7]. These variations in fMCG can be due to either FM-related alterations in heart activity or changes in the amplitude of the recorded heart activity caused by movement of the signal source or the combination of both. In one approach, the variation of the amplitude of R wave over time from a single sensor is used to identify FM [4]. In another study, instead of a single R wave, the changes in the morphology of QRS complexes from multiple sensors are used to indicate movement intervals [5]. The maximum and minimum values of each sensor for the entire QRS complexes are identified for a moving window of 20 beats and then standard deviations are computed for all maxima and minima. The summation of standard deviations are used to quantify the FM. Thus, in this approach more than one sensor is used to identify FM. Govindan et al. [6] developed a method in their study in which the magnetic field strength and spatial location from all sensors of the fMCG signal were used to quantify the FM and then correlated it with fHR. Lutter et al. [7] used statistical detection and formulated indices of FM to improve the sensitivity using information from all sensors. All these methods which use fMCG recordings for FM detection take into account the changes in fMCG signals measured at the sensors. Even though those methods can detect intervals of the FM, they do not provide an exact measurement. They can only provide FM intervals and only [6] can quantify FM using center of gravity of magnetic fields at the sensors. Changes in center of gravity cannot identify the exact measurement of movement in X, Y, and Z directions and therefore provide two dimensional information about the FM. Especially, to identify exact movements in depth is not possible using current methods. One approach to overcome this problem is to identify and monitor source positions of electro-physiological activities within fetus that periodically repeat e.g. heart signals.

In order to identify the characteristics of the sources responsible for the electrical activity in fetal heart, inverse problem needs to be solved from the measured fMCG signals. Several studies used the equivalent current dipole (ECD) model for cardiac source reconstruction [8], [9], [10], [11]. These studies made assumptions about the volume conductor and considered them as either spherically symmetric or uniformly half-space. These assumptions enable to execute forward computations and reconstruct magnetic signals from one or multiple signal sources. Stinstra et al. [12] showed that the assumptions on physical conditions of volume conductor affect the results of source localizations. Therefore, a forward model that is independent from physical properties of volume conductor provides more robust results. Stinstra [13] also showed that the equivalent magnetic dipole (EMD)

model which is not defined using the physical properties of the volume conductor is not affected by volume conditions and can describe fMCG recordings with a very low explained variance.

Using EMD model in source localization, we propose a novel approach to detect FMs by tracing the source of fMCG signal. Given fMEG recordings which include maternal heart (mMCG), fetal heart (fMCG), environmental noise and other biological interferences, the task is to approximate the position and amplitude of the source that has generated the fMCG signal. After the fMCG is isolated from the fMEG recordings, the magnetic fields generated by the electrical activity in fetal heart can be described by an EMD. Identifying the EMD and its position for an fMCG signal provides the spatial information of heart as well. Tracing EMD positions for different time instances reveals any occurring FM. Our method can identify EMD position for a measured heart signal and detect the positional changes during the FM. Hence, it provides three dimensional information about FM. Also, it is capable of identifying all characteristics its magnitude and orientation as well as time intervals of movement.

II. Materials and Methods

A. Cardiac Depolarization

Cardiac tissues have the capability of producing their own electrical impulse as part of the cardiac conduction system. The components of the cardiac conduction system include the sinoatrial (SA) node, the atrioventricular (AV) node, the His bundles and branches, and the Purkinje fibers [14]. When the SA node and the remainder of the conduction system are at rest, the myocardial cells do not conduct any electric impulses. When the SA node initiates the action potential, cardiac depolarization begins and the electric impulses move across atria. This action potential in atria leads atrial contraction.

After the electrical activity reaches the AV node, there is a delay of approximately 100 ms that allows the atria to complete pumping blood [14]. Following the delay, the impulse moves through the His bundles and Purkinje fibers and the impulse spreads to the contractile tissues of the ventricle. Then ventricular contraction occurs.

Fig. 1 shows the normal electrical pattern followed by contraction of the heart which is called the sinus rhythm [15]. The P wave in Fig. 1 is the outcome of the depolarization of the atria. The QRS complex represents the depolarization of the ventricles, which requires a much stronger electrical signal because of the larger size of the ventricular cardiac muscle. The ventricles begin to contract as the QRS reaches the peak of the R wave. Lastly, the T wave represents the repolarization of the ventricles. The repolarization of the atria occurs during the QRS complex, which is not observed in the sinus rhythm.

Cardiac depolarization is a periodic activity and repeats with every heart beat. Heart rate is usually identified using the time periods between two R waves which is called RR interval. R wave and RR intervals are also used to average MCG signals to increase the signal-to-noise (SNR) ratio.

B. Data Collection

Data were recorded at a sampling rate of 312.5 Hz using the SARA (SQUID Array for Reproductive Assessment) non-invasive system installed at the University of Arkansas for Medical Sciences (Little Rock, AR). SARA is a 151-sensor array system that is curved to fit the mother's abdomen. The array and channel arrangement are shown in Fig. 2. Sensor array of SARA covers an area greater than 850 cm² spanning the maternal abdomen longitudinally from the symphysis pubis to the uterine fundus and a similar distance laterally. It is used to detect magnetic field changes generated by electrophysiological activity of mother and fetus.

For this study, we used 28 fMCG measurement between 28 and 38 weeks of gestation age (GA) from 23 pregnant women. One subject has three different measurements and three subjects have two different measurements recorded at different gestational ages. Subjects were picked from a set of measurements prior to the analyses among the *normal* subjects such that mothers are not diagnosed with any medical conditions and fetuses are not at risk of congenital diseases.

For obtaining the fMCG signals, dataset were first bandpass filtered between 1 Hz and 50 Hz using a zero phase Butter-worth filter. Next, mMCG was attenuated using Orthogonal Projection algorithm [16], [17], [18]. Then, fetal R peaks were automatically identified using an adaptive Hilbert transform approach [19], [20]. R peaks were used to obtain time-averaged signals to increase the SNR and also used to obtain fHRs.

C. Forward Model Using Magnetic Dipole Moment

Activity of fetal cardiac tissues is not only associated with electric currents that flow in cardiac cells but also associated with a closed volume conductor that is formed together by the currents flowing within the cardiac muscle cells and the currents spreading through the body. The net electrical vector of the individual cells in the volume that act as bioelectric sources is clinically known as cardiac vector and represented by an ECD [21]. The cardiac vector, ECD, indicates the direction of the depolarization in time. Therefore, for different time instances during heart depolarization, ECD holds a certain magnitude and position within the heart.

In addition to electrical activity, the closed volume conductor also generates a magnetic field outside the maternal abdomen measured as fMCG. Similar to the ECD, the current that loops in this volume conductor also behaves like a magnetic dipole moment, $\vec{M}$ which is represented by an EMD. Note that the EMD and the ECD are different vectors and their positions in the conducting system are not identical but they are very similar. As the ECD, the EMD is directly related with cardiac depolarization and holds a certain magnitude and orientation for different time points during the depolarization. We use EMD in our analyses since it is a simpler model than ECD and more importantly EMD modelling is not dependent on the volume conductor. On the other hand, modelling ECD requires knowledge of the geometry of the volume conductor. Moreover, EMD is adequate for observing any significant change in position of heart signals in a time series measurement.

Any system possessing a magnetic dipole (EMD in our case) generates a magnetic field (fMCG in our case) that can be measured outside of the system. If the measurement is done

at a position distant from the system, higher-order multipole components of the magnetic field are not observed since they drop-off with distance more rapidly. Since the distances to the channels that perform fMCG recording are relatively large, it is expected that EMD would extensively explain the measured magnetic field due to cardiac activities of fetus (fMCG). If position, orientation and magnitude of EMD are known, the magnetic field $\vec{B}$ at position P can be mathematically computed which is called *forward computation*. For a magnetic dipole located in a system, the magnetic vector potential, $\vec{A}$, reads

$$\vec{A}(\vec{r}) = \frac{\mu_0}{4\pi} \frac{\vec{M} \times \vec{r}}{|\vec{r}|^3} \quad (1)$$

where $\vec{M}$ is the magnetic dipole moment, $\vec{r}$ is the distance from EMD to the observation point P, μ_0 is the permeability constant also known as the magnetic constant, and $\times$ is the cross-product operation. Then the magnetic flux density, $\vec{B}(\vec{r})$, can be obtained by the gradient, $\vec{\nabla}$, of the magnetic vector potential as follows:

$$\vec{B}(\vec{r}) = \vec{\nabla} \times \vec{A}(\vec{r}) = \frac{\mu_0}{4\pi} \left[\frac{-\vec{M}}{r^3} + \frac{3(\vec{M} \cdot \vec{r})\vec{r}}{r^5} \right] \quad (2)$$

D. Inverse Solution

For a measured signal, the procedure of localizing the source position is known as the inverse problem. So, identifying the source location and magnitude of magnetic dipole moment from fMCG measurements requires solving the inverse problem. Solution of the inverse problem is obtained by iteratively performing the forward computations for different possible source points until the fitting/modelling error is minimized. In our model, we use (2) as the mathematical model for forward computations, and then inverse problem is solved by minimizing the least-square error (*lse*) defined as

$$lse = \sum_{i=1}^N (Bm_i - Bf_i)^2 \quad (3)$$

where Bm_i is the measured magnetic field and Bf_i is the fitted magnetic field for $i = 1, \dots, N$ channels.

Using the least-square error approximation, our model converges to a point which satisfies the least error. We first select a certain time sample of the time-averaged fMCG signal which is obtained from the magnetic field measured by the SARA array. For the determination of the time sample to use, we compute the variances for all time samples during the range of QRS complex of the time-averaged fMCG signal. The sample with highest variance is selected to use in the inverse solution because it corresponds to the R peak which also corresponds to the time point of the largest EMD observed in cardiac depolarization. $\vec{B}$ of (2), is a vector of $[1 \times N]$ which is fMCG data for one time sample.

Fig. 3 illustrates the steps of inverse solution for one data set. The approach for source localization is as follows:

- Source localization of the fMCG signal requires an initial position to start the computation of the forward problem. In our case we compute the center of gravity of the measured data and use as the initial guess of the magnetic dipole (EMD) position.

Forward computation of $\vec{B}$ from an investigated point requires that we know the parameters of $\vec{M}$ and $\vec{r}$. The parameter, $\vec{r}$ is the distance from the investigated point to the channels and can easily be computed for each channel on SARA device. However, the magnitude and orientation of $\vec{M}$ are unknown, therefore we position three unit vectors along with X, Y, and Z axes on the investigated point and use them as $\vec{M}$ in forward computations. Then, the magnetic field values at each channel due to these unit vectors are computed. Note that for each SARA channel we compute a magnetic field value and Bu_x , Bu_y and Bu_z are the arrays holding these magnetic field values due to the unit vectors on the X, Y, and Z axes, respectively. Considering each of these magnetic field arrays as a component of the forward-computed magnetic field data, i.e. $Bu = [Bu_x, Bu_y, Bu_z]$, we obtain a forward-computed magnetic field array with the size of $[3 \times N]$ due to the unit vectors.

Once Bu is computed, the next step in the inverse problem is to find the magnitude and orientation of the best-fitted magnetic dipole moment, $\vec{M}$. Since we have a measured data, Bm with the size of $[1 \times N]$ and a forward-computed data using unit vectors, Bu with the size of $[3 \times N]$, the best-fitted $\vec{M}$ gives $Bm \approx \vec{M}Bu$. We perform a least square solution for this linear system and obtain $\vec{M}$ which has a size of $[1 \times 3]$ such that $\vec{M} = [M_x, M_y, M_z]$ where M_x , M_y , and M_z are the X, Y and Z components of $\vec{M}$. Then, the forward-computed magnetic field data, Bf due to $\vec{M}$ reads

$$Bf = MBu = M_x Bu_x + M_y Bu_y + M_z Bu_z \quad (4)$$

- Thus, we obtain the magnitude and orientation of the best-fitted EMD at the position investigated and the magnetic field values at SARA channels due to this identified magnetic dipole. Finally the lse is computed using (3) and the model moves to another possible source position until lse value is minimized. We used Nelder-Mead simplex method to minimize the lse [22], [23]. Simplex method has been used on several reports that solve the bio-magnetic inverse solution [10], [24], [25], [26] because of its simplicity and low computational effort.
- Once the position with the minimum error is identified, the goodness of fit (GOF) value is computed for the fitted magnetic field (Bf) forward computed from the identified position. Then the point satisfying the minimum error and maximum GOF is returned as the source (EMD) position for that particular time sample. GOF value for source localization [13] is computed as

$$GOF = 1 - \frac{\sum_{i=1}^N (Bm_i - Bf_i)^2}{\sum_{i=1}^N Bm_i^2} \quad (5)$$

GOF value expresses how well the fitted magnetic dipole (EMD) explains the measured magnetic field (fMCG). Even though the least error gives the highest GOF for one time sample, we need GOF value as a comparative criteria because the *lse* is dependent on the amplitudes. In other words, even if the error for one dataset is smaller than the other, it does not necessarily give a higher GOF value than the other dataset because of the amplitudes. Therefore *lse* is not useful to compare fitting results of different datasets.

E. Fetal Movement Monitoring

Source localization for each time sample of fMCG provides spatio-temporal information of the best-fitted magnetic dipole which is directly related with fetal heart depolarization. In other words, EMD holds a different positions and magnitudes for different instances of cardiac depolarization. Since cardiac depolarization is a periodic activity, EMD position for the same instance of different cycles of cardiac depolarization should be the same as long as the fetus holds its position stable. Otherwise the EMD positions for the same instance show differences which identifies the FM. Thus, we can easily track the movement of fetus by solving the inverse problem for a specific time sample of heart's waveform for different depolarization cycles. For this purpose, the recordings were split into time windows with no overlapping. Then, using R peak positions detected previously, time-averaged fMCG signal for each time window was obtained separately. Averaging was carried out on all the channels. Then time samples on each averaged data are used for inverse solutions and source positions are identified for each time window. Finally, the position of the identified sources are compared by computing the Euclidean distance. For each time window, the distance is computed with respect to the position in the previous time window. Also, the dot product of normalized magnetic dipoles in adjacent time windows are computed to observe the changes in orientation.

F. Fetal Movement - Heart Rate Acceleration Coupling

Some studies investigated the relation between FM and fHR changes by calculating coupling index values between significant FM and fHR accelerations [27], [28]. Coupling index (CI) is an important expression of the maturation of coordinated behavior in the fetal central nervous system. In our study, we compute the fHR (bpm) by using the identified R peaks of the entire recording. We used the number of R peaks in each 6-second time windows to obtain the fHR for that corresponding time window. Thus for each time window we obtain a fHR value as well as a FM amount obtained from the inverse solution approach.

Once heart rates and movements are obtained, we interpolate them using spline function to convert them into continuously sampled data with a sample rate of 312.5 Hz. Then, we performed a baseline correction for each subject. Substantial FM was identified when source displacement attained 0.2 cm over baseline and excursions in fHR 5 bpm over baseline

were considered as fHR acceleration. FM and fHR accelerations are defined as coupled if their intervals overlap at least 5 seconds. CI is defined as the ratio of the number of coupled movements to the total number of identified movements and gets a value between 0 and 1.

III. Experimental Results

A. Accuracy

We first tested the accuracy of the forward model in explaining the measured data. Hence, for each subject we compared B_m with B_f obtained from the forward computation using the identified EMD and source position. Fig. 4 illustrates how good our model can approximate the measured data for a subject. In Fig. 4(a), the averaged fMCG data is illustrated where the vertical red line shows the time sample used for source localization. The averaged fMCG in the figure was obtained using the entire measurement which gives the highest SNR. In Fig. 4(b), the recorded magnetic field values (blue), B_m and the fitted data values (red), B_f on the channels are shown. Fig. 4(c) and Fig. 4(d) show the magnetic field contour map over SARA sensors for the measured (B_m) and the fitted (B_f) data, respectively. Fig. 4(d), shows the source position identified by the model with a red point. As the figures illustrate, for this data set the recorded data and the fitted data by our model are almost identical. The GOF is 99.89% for this data set. We have seen that for an averaged fMCG signal our model can identify a source position and a magnetic dipole that describes the measured data with a GOF value over 99% and with an error less than %1.

Then, we tested the performance of our model for time windows with different lengths. For all subjects, R peaks in QRS complexes attained the highest GOF values. When we used 5-second windows, the GOF values for R peaks in some time windows were as low as 93%. On the other hand, for 6-second windows the GOF values for R peaks were not less than 97% but most of the values were above 99%. Therefore, we decided to split the recordings in 6-second windows to avoid low SNR ratio which will affect GOF values. Fig. 5(a) shows the averaged fMCG and Fig. 5(b) shows the GOF values for each time sample for a subject. For the P wave region and QRS complex of the fMCG signal, GOF values are significantly larger than other regions. The GOF value is over 97% for most of the P wave region and over 98% for most of the QRS complex. Since P wave and QRS complex are outcomes of atrial and ventricular depolarization, we were expecting to get higher GOF values for those regions. In other words, the EMD can describe cardiac signals much better when cardiac tissues are depolarized which creates a significant heart vector. On the other hand, if there is no electrical activity in cardiac tissues e.g. during the time period between P wave and QRS region, there is no observable cardiac vector and therefore our model cannot identify an EMD which can describe the measured data.

The total number of 6-second time windows and GOF values for a subject vary depending on the duration of recording. We analyzed total of 12373 6-second time windows from 28 datasets. We found that the magnetic dipole model explained the recorded data with a value of GOF over 97% for all subjects and for all time windows. Fig. 6 shows the boxplots for GOF values of each subject. Only 301 out of 12373 GOF values are less than 99% but above 97%. These results suggest that the forward model for fitting magnetic dipoles can be used to identify source location for fMCG signals especially for QRS complex. We also observed

that GA (last number on subject ID) does not have any effect on dipole fitting performance. GOF values for smaller GAs like 28 weeks are as high as GOF values of larger GAs.

B. Fetal Movement

Our EMD fitting results suggest that QRS complex and specifically R peak is the best time sample to use in source localization and for monitoring FM. To accomplish this, we split the measured fMCG data of each subject into 6-second time windows. These time windows are mutually disjoint and correspond to 12–15 heart beats depending of the fHR. In order to increase the SNR, we averaged the data of each time window. For each time window, we first determined the time sample to use in source localization. To identify the instance of R peak in QRS complex, we first computed the variances of each time sample instead of checking the highest amplitude. We selected the time sample with the highest variance in QRS complex which is a more robust way to identify R peak rather than depending on a single sensor input. Once the time sample of R peak was selected, the inverse problem was solved for this time sample. Thus we obtained the position of EMD for this time window. This process was repeated for all time windows and we obtained source positions for every 6 seconds.

Fig. 7 shows how the source positions of 9 time windows are located in SARA space for a subject (S23.1.37) which was one of the most active fetuses of the sample group. Fig. 7(a) display the source positions with respect to the SARA array and a scaled version of source positions (blue dots) is displayed in Fig. 7(b). The identified source positions i.e. heart positions change in time which reveals that the fetus moved during the recording.

Once all the source positions were identified for all time windows, displacements were computed. For each time window, the Euclidean distances with respect to the previous time window were computed. Thus, we could see if there is any FM observed since the previous time window i.e. during 6 seconds. Fig. 8(a) shows the displacement with respect to previous position in each time window for the subject S23.1.37. We observed that the displacement is around or lower than 0.2 cm for most of the time windows but for other windows 2 cm displacements were found.

In addition to displacements we also analyzed if there was any change in the orientation. For this purpose, we compared EMD identified in one time window with the next one. We first normalized the magnetic dipole vectors of each time window and then computed the dot products which is displayed in Fig. 8(b). Similar to the displacement results, we observe that most of the time the dot product is over 0.99 while it is significantly lower for some time points. Note that lower dot products were obtained whenever higher displacement values were observed. Therefore, displacement values and dot products suggest that there are some time points at which the fetus was actively moving. Moreover, in Fig. 8(a), we can observe that the fHR increases whenever we observe higher displacements which is also correlated with lower dot products.

We compared our results of FM with a previous approach reported by Leeuwen et al. [9] by calculating Pearson correlation coefficients between the two FM traces obtained from each method. We obtained a mean value of correlation coefficient of 0.48 with [0.41, 0.55] 95%

confidence limits across subjects. We also calculated Pearson correlation coefficients between fHR and displacement and dot products. In addition to correlation scores, CIs of FM and fHR acceleration intervals were computed as well. For 7 out of 28 measurements no significant FM interval was identified. Total of 96 FM intervals were detected in the remaining 21 measurements. 76 of them (72.9%) were coupled with fHR acceleration. In addition, 9 measurements had a CI of over 0.9 and only 2 measurements had a CI less than 0.5. Moreover, 9 out of 21 measurements had a correlation coefficients 0.4 or greater for fHR and displacement while 14 out of 21 measurements had a correlation coefficient -0.4 or less for fHR and dot products. Fig. 9 displays the histogram of correlation scores and CIs for 21 data sets.

IV. Discussion

FM is a sign of fetal well-being and therefore monitoring FM is important to assess fetal health. It has been shown that MEG recordings during pregnancy can provide wide range of information e.g. fMCG about physiological activities of fetus. fMCG is a reliable measurement that gives insights of fetal cardiac activity throughout the pregnancy. It can also be helpful to locate fetal heart and monitor its position during a recording. Our study propose to track FM by source localization with fMCG signals.

We analyzed 28 fMCG recordings to measure the performance of the forward model for describing the measured data. We found that the model attained GOF values over 97% for all subjects when using 6-second time windows. The main factor that can affect the fitting performance is low SNR values. Averaging data over a certain time window is the only way to overcome this problem. We have seen that 6-second was the optimum length that produced an acceptable value of GOF for all recordings in this sample group. However, we have seen that we could use shorter time windows for certain subjects. So, while working on a single subject in future studies, the length of time windows can be adjusted depending on the GOF values. For some recordings and with the help of other pre-processing techniques, it is even possible to work on 1-second or shorter time windows depending on the noise conditions which allows us to get more precise FM detection. For this study, the deterministic Nelder-Mead simplex method found the optimum values with few evaluations, thus the computational cost was very low. Probabilistic and evolutionary optimization strategies may improve the source localization when low SNR conditions are present or high-dimensional search space (two or more dipole sources). Overall, our study showed that EMD was robust to describe the fMCG data and can be used for FM detection.

Our results show that FM intervals were consistent with a previous method used to estimate FM based on amplitude of the fMCG, but our study provides exact movement amounts and directions. Previous reports show that fHR accelerations coincided with or preceded FMs nearly all the time [4]. Correlation coefficients showed good agreement between fHR with displacement and orientation changes of localized sources of the fMCG signals. CI for interval of 28–38 GA were in the range of previous studies using Doppler-Ultrasound measurements and fMCG [28], [29].

Monitoring FM is important for other studies investigating fetal heart and brain activities. The FM is one of the factors that interfere with the fetal neurological studies. Having the position of the fetal head is very critical in brain studies since brain signals are significantly smaller than all other biological signals. The positions of fetal heart and head are identified before fMEG recordings using ultrasound. However, if a fetus moves during the recording, the new spatial information for fetal heart and head needs to be identified. We cannot monitor and quantify the FM by using simultaneous ultrasound measurement with fMEG as it would obscure the fetal brain data. Therefore detecting only the FM intervals is not enough for precise computations. All characteristics of FM need be identified as well so that the new position of head and heart can be obtained.

The current methods do not address this problem. Those methods can determine the FM intervals considering only the morphological changes of the fMCG signals which lacks of the capability to identify the amount of the FM for X, Y, and Z coordinates. Only [6] quantifies the FM using center of gravity of magnetic fields at the MEG sensors but it does not provide the exact amount and direction of the FM either. Especially, identifying the characteristics of FM in depth is not possible with current methods. On the other hand, our proposed method can track FM on 3 dimensions by obtaining spatial information of the source position of the fMCG signal and identifying exact displacement and direction of source positions. Thus, our method provides all characteristics of FM including the time interval, direction and amount.

V. Conclusion

We have developed an inverse solution model to identify source position of fetal heart signals. We showed that our model can identify a source position that describes the measured data with a GOF value over a 97%. Then we also showed that source localization can be used to track the FM. We saw that fHR acceleration and displacement of source positions were correlated with movement intervals. We observed the same pattern with heart rate acceleration and orientation changes as well. We have seen and shown that our method can identify the amount and the direction of the movement, as well as rotational movements by comparing the orientation of the dipoles. Our fetal movement method can help to improve other studies especially the ones related with fetal brain signals since it is now possible to monitor fetal head position as well as heart location through the recording.

Acknowledgments

The authors would like to thank Pamela Murphy and Eric Siegel for their support and contributions to the study. Research supported by a grant from NIH/NIBIB R01EB007826-01A1 PI: Hari Eswaran.

References

1. Hertogs K, Roberts AB, Cooper D, Griffin DR, Campbell S. Maternal perception of fetal motor activity. *British Medical Journal*. 1979; 2.6199:11831185.
2. Tatsumura M. Studies on features of fetal movement and development of human fetus with use of fetal actogram. *Nihon Sanka Fujinka Gakkai Zasshi*. 1991; 43:864873.
3. Nijhuis, JG. *Ultrasound in Obstetrics and Gynecology*. Boston, MA: Brown & Co; 1993. Fetal behavioral states; p. 447455

4. Zhao H, Wakai RT. Simultaneity of fetal heart rate acceleration and fetal trunk movement determined by fetal magnetocardiogram actocardiography. *Physics in Medicine and Biology*. 2002; 47:839846.
5. Geue D, Leeuwen V, Lange S, Gronemeyer D. Automatic detection of gross fetal movements using actography based on magnetocardiography. *Biomed Tech*. 2007; 52
6. Govindan RB, Vairavan S, Ulsar UD, Wilson JD, McKelvey SS, Preissl H, Eswaran H. A novel approach to track fetal movement using multi-sensor magnetocardiographic recordings. *Annals of biomedical engineering*. 2011; 39(3):964972.
7. Lutter WJ, Wakai RT. Indices and Detectors for Fetal MCG Actography. *IEEE Transactions on Biomedical Engineering*. 2011; 58(6):18741880.
8. Lewis MJ. Review of electromagnetic source investigations of the fetal heart. *Medical engineering & physics*. 2003; 25(10):801–810. [PubMed: 14630467]
9. Leeuwen, PV., Beuvink, Y., Lange, S., Klein, A., Geue, D., Gronemeyer, D. Proc. 14th Int. Conf. on Biomagnetism. Boston, MA, USA: 2004. Assessment of fetal growth on the basis of signal strength in fetal magnetocardiography; p. 327-8.
10. Popescu M, Popescu E, Fitzgerald-Gustafson K, Drake WB, Lewine JD. Reconstruction of fetal cardiac vectors from multichannel fMCG data using recursively applied and projected multiple signal classification. *IEEE Transactions on Biomedical Engineering*. 2006; 53:256476.
11. Horigome H, Shiono J, Shigemitsu S, Asaka M, Matsui A, Kandori A, Miyashita T, Tsukada K. Detection of cardiac hypertrophy in the fetus by approximation of the current dipole using magnetocardiography. *Ped Res*. 2001; 50:2425.
12. Stinstra JG, Peters M. The influence of fetoabdominal tissues on fetal ECGs and MCGs. *Archives of physiology and biochemistry*. 2002; 110(3):165–176. [PubMed: 12221516]
13. Stinstra, JG. Ph.D. dissertation. University of Twente, Enschede; The Netherlands: 2001. Reliability of the fetal magnetocardiogram.
14. van Weerd JH, Christoffels VM. The formation and function of the cardiac conduction system. *Development*. 2016; 143:197–210. [PubMed: 26786210]
15. Marriott, HJL. Practical Electrocardiography. Sixth. Waverly Press, Inc; Baltimore, MD: 1981.
16. Vrba J, Robinson SE, McCubbin J, Lowery CL, Eswaran H, Wilson JD, Murphy P, Preissl H. Fetal MEG redistribution by projection operators. *IEEE Transactions on Biomedical Engineering*. 2004; 51:120718.
17. McCubbin J, Robinson SE, Cropp R, Moiseev A, Vrba J, Murphy P, Preissl H, Eswaran H. Optimal reduction of MCG in fetal MEG recordings. *IEEE Transactions on Biomedical Engineering*. 2006; 53(8):1720–4. [PubMed: 16916111]
18. Sriram, B., Wilson, JD., Govindan, RB., Lowery, CL., Preissl, H., Eswaran, H. The effect of applying orthogonal projection technique in short window segments to obtain fetal magnetocardiogram; Annual International Conference of the IEEE Engineering in Medicine and Biology Society IEEE Engineering in Medicine and Biology Society Annual Conference; 2013. p. 421424
19. Wilson JD, Govindan RB, Hatton JO, Lowery CL, Preissl H. Integrated approach for fetal QRS detection. *IEEE Transactions on BioMedical Engineering*. 2008; 55:21902197.
20. Ulsar, UD., Govindan, RB., Wilson, JD., Lowery, CL., Preissl, H., Eswaran, H. Adaptive Rule Based Fetal QRS Complex Detection Using Hilbert Transform; Annual International Conference of the IEEE Engineering in Medicine and Biology Society IEEE Engineering in Medicine and Biology Society Conference; 2009. p. 46664669
21. Karp, P. Biomagnetism, Proceedings of the 3rd Biomagnetism Conf. Berlin: 1980. Cardiomagnetism; p. 210258
22. Nelder JA, Mead R. A simplex method for function minimization. *The computer journal*. 1965; 7(4):308–313.
23. Mathews JH, Fink KK. Numerical Methods Using Matlab, 4th Edition. Pearson. 2004
24. Achim A, Richer F, Saint-Hilarie JM. Methods for separating temporally overlapping sources of neuroelectric data. *Brain topography*. 1998; 1(1):22–28.

25. Mosher JC, Lewis PS, Leahy RM. Multiple dipole modeling and localization from spatio-temporal MEG data. *IEEE Transactions on Biomedical Engineering*. 1992; 39(6):541–557. [PubMed: 1601435]
26. Uutela K, Hamalainen M, Salmelin R. Global optimization in the localization of neuromagnetic sources. *IEEE Transactions on Biomedical Engineering*. 1998; 45(6):716–723. [PubMed: 9609936]
27. Baser I, Johnson TR, Pain LL. Coupling of fetal movement and fetal heart rate accelerations as an indicator of fetal health. *Obstet. Gynecol.* 1992; 80(1):6266.
28. DiPietro JA, Hodgson DM, Costigan KA, Hilton SC, Johnson TR. Development of fetal movement fetal heart rate coupling from 20 weeks through term. *Early Hum. Dev.* 1996; 44(2):139151.
29. Schmidt A, Schneider U, Witte OW, Schleussner E, Hoyer D. Developing fetal motor-cardiovascular coordination analyzed from multichannel magnetocardiography. *Physiol. Meas.* 2014; 35:1943–1959. [PubMed: 25229562]

Fig. 1.

Normal Sinus Rhythm: Normal MCG signals due to one cycle of cardiac depolarization consist of three main components. The P wave is the outcome of the depolarization of the atria that triggers atrial contraction. The QRS complex represents the depolarization of the ventricles. The T wave represents the repolarization of the ventricles. Repolarization of atria occurs during ventricle depolarization therefore it is not observed in cardiac signals.

Fig. 2.
SARA Device and Channel Layout: SARA system is curved to fit the mother's abdomen and used to detect magnetic fields generated by electrophysiological activity of mother and fetus.

Fig. 3.

Flowchart of source localization for data of one time sample: The process starts with fMCG extraction from SARA measurements. Next, data is split into 6-second time windows and fMCG in each time window is averaged using R peaks. After selecting time sample for source localization, Nelder-Mead simplex method finds the best-fitted EMD position and orientation for the dataset. The source localization process is terminated after error is minimized. Finally, GOF is computed to assess the dipole fitting performance.

Fig. 4.

Magnetic dipole fit accuracy for S23.1.37: (a) Time-averaged fMCG and R peak position used in source localization. Source localization was performed on the 58th time sample of the averaged fMCG. (b) Recorded magnetic signals on the SARA channels are shown with blue color. Fitted data (computed from the identified source position i.e. EMD position) across channels is shown with red color. (c) Magnetic field contour map for the recorded data on SARA space. (d) Contour map for the fitted data on SARA space. Red point is the EMD source position identified by the model.

Fig. 5.

GOF values for S23.1.37-Time Window 1: (a) Averaged fMCG data in the first 6-second time window. (b) Source localization was performed for all time samples of time-averaged fMCG signal. GOF values in the P wave region and QRS complex are significantly higher than other time samples.

Fig. 6.

Boxplots for GOF values of 28 datasets. Subjects were ordered by GA. The first number in the labels corresponds to subject ID while the second number indicates the measurement ID for the corresponding subject. The last number in the labels corresponds to the GA of fetuses. For 28 datasets, source localization is performed on one time sample selected from the QRS complex of each time window.

Fig. 7.

SARA array and EMD source locations for S23.1.37. (a) SARA array and the location of its sensors (red points) are shown in the upper part. Source positions with respect to SARA array for 9 time windows (TW) identified by our model are displayed in the lower part. (b) A scaled version of source positions is shown. Blue dots represent the source positions and the arrows show the direction of movements for 9 different time windows. TW1 is the initial position (position for the first 6 seconds of recording) of the fetus while TW300 is the final position during the recording. The figure shows that the fetus has some significant movements between the first and final TW.

Fig. 8.

Fetal movement with dot products for S23.1.37. (a) Displacements of the source positions with respect to previous positions are shown in the bar plot. fHR in each time window is shown by the orange plot. fHR acceleration which implies some FM correlates with higher displacements i.e. FM. (b) The dot products of normalized EMDs in adjacent time windows. In the time windows where FM is observed, the dot products significantly decrease comparing the other time windows. This shows that due to the FM the orientation of the EMD is changed as well.

Fig. 9.

Correlation coefficients and coupling scores of the measurements that include at least one identified movement interval. Correlation coefficients were computed between fHR and displacement and dot product. CI was computed for the intervals of fHR acceleration and of FM.