The effects of asymmetric volume conductor modeling on non-invasive fetal ECG extraction

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Abstract

Objective: Non-invasive fetal electrocardiography (NI-FECG) shows promise for capturing novel physiological information that may indicate signs of fetal distress. However, significant deterioration in NI-FECG signal quality occurs during the presence of a highly non-conductive layer known as vernix caseosa which forms on the fetal body surface beginning in approximately the 28th week of gestation. This work investigates asymmetric modeling of vernix caseosa and other maternal–fetal tissues in accordance with clinical observations and assesses their impacts for NI-FECG signal processing.

Approach: We develop a process for simulating dynamic maternal–fetal abdominal ECG mixtures using a synthetic cardiac source model embedded in a finite element volume conductor. Using this process, changes in NI-FECG signal morphology are assessed in an extensive set of finite element models including spatially variable distributions of vernix caseosa.

Main results: Our simulations show that volume conductor asymmetry can result in over 70% error in the observed T/QRS ratio and significant changes to signal morphology compared to a homogeneous volume conductor model. Volume conductor effects must be considered when analyzing T/QRS ratios obtained via NI-FECG and should be considered in future algorithm benchmarks using simulated data.

Significance: This work shows that without knowledge of the influence of volume conductor effects, clinical evaluation of the T/QRS ratio derived via NI-FECG should be avoided.

1. Introduction

Non-invasive fetal electrocardiography (NI-FECG) has shown promise in recent years with several NI-FECG systems becoming commercially available for monitoring fetal heart rate (FHR) in a clinical environment (Sameni and Clifford 2010, Cohen et al 2012, Clifford et al 2014). NI-FECG demonstrates several advantages over ultrasound based cardiotocography (CTG) in that it is a passive modality suitable for long term use and may provide novel physiological information in addition to FHR (Symonds et al 2001, Clifford et al 2011, Behar et al 2016b). As the ECG waveform represents the magnitude and temporal characteristics of electrical currents within the cardiac muscle, morphological analysis of NI-FECG recordings has been proposed for identifying signs of fetal distress such as hypoxia and intrauterine growth restriction (IUGR) (Oudijk et al 2004, Behar et al 2016a, Fuchs 2016).

The process of morphological analysis can be divided into two distinct phases: (1) extraction—the process of extracting a fetal ECG waveform from noisy single or multi-channel recordings; and (2) analysis—the process of analyzing features of the extracted ECG waveform to assess physiological state. These features can be separated into the characteristics of individual physiological events (e.g. QRS complex amplitude and location) and the ratios or intervals between events. Figure 1 shows an exemplary ECG waveform segment indicating one such feature, the ratio of T wave amplitude to QRS complex amplitude (T/QRS ratio) which has been proposed for identifying fetal distress (Behar et al 2016a, Fuchs 2016). The accuracy of morphological analysis therefore depends on the accuracy of the extraction process.
However, extracting an accurate fetal ECG from a set of abdominal recordings is challenging due to interference from the maternal ECG, muscle activity and sensor noise (Stinstra and Peters 2002, Andreotti et al 2014). In addition to these disturbances, a highly non-conductive layer known as vernix caseosa forms on the fetal body surface beginning in approximately the 28th week of gestation causing significant attenuation of fetal cardiac signals (Oostendorp et al 1989, Stinstra 2001). The influence of vernix caseosa presents an important area of research for NI-FECG extraction as many signal processing approaches used in this domain (e.g. adaptive filtering (Widrow et al 1975), template subtraction (Martens et al 2007)) consider the maternal abdomen as a homogeneous volume conductor. This assumes that recordings at the skin surface contain a superimposed projection of underlying electrical sources with attenuation as a function of distance from source to sensor position.

By comparison, attenuation in an inhomogeneous volume conductor depends on the specific position and orientation of each source in the 3D domain as well as its distance to the sensor position. As the cardiac events represented by the P wave, QRS complex and T wave of the ECG waveform are oriented differently in 3D space (Nousiainen et al 1986), it is important to determine if the presence of vernix caseosa impacts NI-FECG extraction accuracy through the use of appropriate volume conductor models.

Prior work by Oostendorp et al (1989) and Stinstra and Peters (2002) investigated the effects of vernix caseosa on abdominal potentials by modeling it as a uniform layer covering the fetal body surface with holes near the mouth and umbilicus. Oostendorp et al (1989) concluded that while this model approximated some instances of clinical NI-FECG data, it did not account for all observations due to possible variations in vernix caseosa distribution. These works focused on relatively symmetric distributions of vernix caseosa and did not investigate observations presented by Akiba (1955) on a cohort of newborns (n = 623) from 28 weeks gestational age (GA) onwards which found that while 16% of the cohort had vernix caseosa covering the entire body, the remaining group had either no vernix caseosa or distributions favouring the following regions in order of decreasing prevalence: inguinal region, axillary fossa, back, buttocks, hips, thighs and neck. Preferential surface distributions of vernix caseosa have also been reported by Archana (2008) where vernix caseosa in the cohort (n = 100) was found in 78% of cases on the back, 61% in the inguinal region and 16% on the chest, and by Visscher et al (2005) where vernix caseosa in the cohort (n = 430) was significantly higher on the back than the chest. These results indicate that modeling vernix caseosa as a uniform layer does not represent a significant proportion of cases and a non-uniform, asymmetric model is required to fully understand its impacts on NI-FECG extraction. The contributions of this work are thus as follows.

- Link clinical observations of the asymmetric distribution of vernix caseosa to their relevance in the study of NI-FECG signal processing.
- Develop a novel process to simulate dynamic maternal–fetal ECG mixtures in an asymmetric volume conductor model including spatially variable distributions of vernix caseosa.
- Characterise the effects of asymmetric volume conductor models on NI-FECG extraction in terms of changes to signal morphology compared to a homogeneous volume conductor model.
- Demonstrate that NI-FECG algorithm benchmarks using a homogeneous volume conductor model do not provide representative accuracy of T/QRS ratio extraction.
- Provide recommendations for the utility of the T/QRS ratio derived via NI-FECG as a clinical tool.

![Figure 1. An exemplary ECG waveform segment indicating the P wave, QRS complex and T wave. A proposed morphological feature for identifying fetal distress is the ratio of T wave amplitude to QRS complex amplitude (T/QRS ratio).](image-url)
2. Background

To date, the accuracy of NI-FECG extraction algorithms on clinical data has primarily been assessed using two types of reference signal: (1) data recorded from an intrauterine fetal scalp electrode (FSE) (Graatsma et al 2009, Clifford et al 2011, Andreotti et al 2014) or (2) expert annotations of the NI-FECG (Hoyer et al 2017). While these approaches are appropriate for validating the temporal accuracy of individual event detections (e.g. QRS complex locations), they are unsuitable for verifying the 3D characteristics of fetal cardiac signals as the FSE records a 1D projection at the placement site and expert annotations rely on the assumption that signal morphology has not been altered by volume conductor effects prior to inspection.

In an attempt to provide a consistent reference signal against which NI-FECG extraction algorithms can be benchmarked, ongoing work in simulating dynamic maternal–fetal ECG mixtures from a synthetic cardiac source has been completed by Sameni et al (2007), Behar et al (2014) and Andreotti et al (2016) to produce the open-source fecgsyn toolbox. This MATLAB based toolbox enables the simulation of maternal–fetal ECG mixtures that incorporate beat-to-beat variability, fetal movement and realistic noise, allowing benchmarks in a range of pathological scenarios. These simulations are important to investigate NI-FECG extraction accuracy in cases where the NI-FECG cannot be visually annotated or placement of the FSE is infeasible, such as in early gestation. However, as the fecgsyn toolbox utilizes a homogeneous volume conductor model, it is important to acknowledge that such simulations may not capture all properties of the true system and could provide false confidence in extraction accuracy when applied to real data.

To simulate the propagation of cardiac electrical activity to potentials on the maternal abdomen, there are two key components that must be chosen: (1) the source model or electrical approximation of cardiac activity and (2) the volume conductor model through which this electrical activity propagates. An overview of the simulation process currently used in the fecgsyn toolbox is shown in figure 2 which describes (a) vector loop representing the time-varying source model, (b) homogeneous volume conductor model representing the maternal–fetal anatomy, and (c) resulting surface potentials where numbered squares indicate sensor positions. As this work builds upon the fecgsyn toolbox, the following sections will summarise the theory utilized to describe these components.

2.1. Source model

There are a number of source models available to approximate cardiac electrical activity, ranging from the point dipole to the equivalent double layer (EDL). For an in-depth discussion of these models and their physiological basis, the reader is referred to Malmivuo and Plonsey (1995). The source model used in the fecgsyn toolbox is the point dipole due to its straightforward volume conductor solution and ability to approximate 80%–90% of the power in observed surface potentials (Malmivuo and Plonsey 1995, Van Oosterom 2002). While the true cardiac activity is spatially distributed in 3D space, the point dipole model provides an adequate approximation in the far-field case investigated by this work. The time-varying potential vector $\phi(t)$ produced by a point dipole for a chosen set of sensor positions in the fecgsyn toolbox is given by

$$\phi(t) = H(t) \cdot R(t) \cdot d(t)$$  \hspace{1cm} (1)

where for $n$ sensor positions, $d(t)$ is a $3 \times 1$ vector representing the point dipole, $R(t)$ is a $3 \times 3$ rotation matrix and $H(t)$ is a $n \times 3$ lead field matrix. The lead field matrix $H(t)$ represents the transformation from source vector to scalar potential at each sensor position as given by the volume conductor model (see section 2.2), whereas the
\( R(t) \) matrix allows for independent rotation of the time-varying point dipole, useful for simulating physiological events such as respiration or fetal movement. Both \( H(t) \) and \( R(t) \) may be time-varying when the cardiac source is in motion. Combining the maternal and fetal cardiac sources with noise results in

\[
\phi(t) = H_m(t) \cdot \mathbf{R}_m(t) \cdot \mathbf{d}_m(t) + H_f(t) \cdot \mathbf{R}_f(t) \cdot \mathbf{d}_f(t) + \mathbf{w}(t)
\]  

(2)

where the subscripts \( m \) and \( f \) indicate the respective maternal and fetal parameters and \( \mathbf{w}(t) \) is noise.

### 2.2. Volume conductor model

To calculate the potential distribution generated by electrical sources within the human body, a model of its electrical properties must be defined. The volume conductor model used in the \textit{fecgsyn} toolbox to calculate \( \mathbf{H}(t) \) is summarized as follows with further detail available in Geselowitz (1989) and Sameni et al (2007). Considering the human body as a linear, resistive volume, the quasi-static potential \( \phi \) at a sensor position external to any current source can be described by the Poisson equation:

\[
\nabla \cdot \sigma \nabla \phi = \nabla \cdot \mathbf{J}
\]  

(3)

where \( \mathbf{J} \) is the impressed current density and \( \sigma \) is the conductivity. Assuming an infinite, homogeneous volume, a solution to (3) is given by

\[
\phi(r) - \phi_0 = \frac{1}{4\pi\sigma} \iiint_V \frac{\nabla \cdot \mathbf{J}(r')}{|r - r'|} dV
\]  

(4)

where \( r \) is the vector from source to sensor position and \( \phi_0 \) a chosen reference potential. For a single time-varying source, (4) has a solution:

\[
\phi(t) - \phi_0 = \frac{\mathbf{d}(t) \cdot \mathbf{r}(t)}{4\pi\sigma|\mathbf{r}(t)|^3}
\]  

(5)

where \( \mathbf{d}(t) \) is the time-varying vector representing the point dipole and \( \mathbf{r}(t) \) the time-varying vector from source to sensor position. Equation (5) thus expresses the solution for the elements of \( \mathbf{H}(t) \) as used in the \textit{fecgsyn} toolbox with the rotation matrix \( \mathbf{R}(t) \) used to modify the overall dipole orientation. However, this solution is only valid for a volume conductor comprised of an infinite homogeneous medium. As an inhomogeneous volume conductor affects dipole sources depending on their specific position and orientation, the proposal by Behar et al (2014) to model vernix caseosa by reducing the signal-to-noise ratio (SNR) of the fetal cardiac source will not accurately reflect morphological changes introduced by an asymmetric volume conductor.

### 3. Method

#### 3.1. Asymmetric volume conductor modeling

In this work, we extend the \textit{fecgsyn} toolbox by replacing \( \mathbf{H}(t) \) with a lead field matrix computed using the finite element method (FEM). The FEM operates by discretizing the volume conductor into smaller elements with defined electromagnetic properties, enabling the computation of electric potential and magnetic field distribution in complex geometric structures. This technique has been used previously in the field of electroencephalography (EEG) where inhomogeneities such as the highly non-conductive skull have a significant effect on EEG measurements (Chauveau et al 2004). Several open-source EEG toolboxes with volume conductor modeling capability have been released to study these effects including the Neuroelectromagnetic Forward Head Modeling Toolbox (NFT) (Acar and Makeig 2010), Brainstorm (Tadel et al 2011) and FieldTrip (Oostenveld et al 2011). In this work, we utilize the FieldTrip-Simbio FEM pipeline developed by Vorwerk et al (2018). This pipeline has been chosen due to its numerical accuracy (Vorwerk et al 2012) and straightforward integration into the MATLAB environment. For further reference on its implementation, the reader is referred to Vorwerk (2016).

To describe the maternal–fetal anatomy, we use a model containing four tissue types consisting of fetus, vernix caseosa, amniotic fluid and maternal abdomen as proposed by Stinstra and Peters (2002). We assume the maternal–fetal anatomy \( \Omega \) can be decomposed into a finite number of compartments \( \Omega_1, \Omega_2, \ldots, \Omega_n \) each containing a tissue type of a single resistive conductivity \( \sigma \). Adjacent compartments must share a common boundary and there may be disjoint compartments containing the same tissue type within the overall domain. The electrical conductivities used for each tissue type in our model are taken from Stinstra and Peters (2002) and are given in Table 1.

To generate a finite element model based on realistic maternal–fetal geometry, we utilize the publically available Fetus and Mother Numerical Models (FEMONUM) as described in Bibin et al (2010) and Dahdouh et al (2014). These models were created using a combination of 3D ultrasound and magnetic resonance imaging (MRI) data from women with pregnancies ranging from eight to 34 weeks GA. Specifically we use the Victoria-TiGroFetus-32weeks model composed of the maternal body model Victoria (provided by DAZ 3D Studio, www.
and 32 weeks GA model which discretizes the fetus, amniotic fluid and maternal body into three tri-
angulated compartments of 24 901, 382 and 11 149 nodes respectively as shown in figure 3. All units in this model
are given in millimetres where the $x$-$z$ plane represents the anatomic coronal view and the $y$-$z$ plane represents
the anatomic sagittal view.

Before use, all models were processed using the open-source MeshLab toolbox developed by Cignoni
et al (2008) to achieve the following: (1) remove self-intersections, (2) smooth and resample each compartment
to create a single closed surface and (3) compute the constrained Delaunay triangulation (CDT) of the resulting
surface.

To describe the asymmetric distribution of vernix caseosa we propose a model that delineates the fetal body
into seven regions as defined by the fetal body maps in figure 4(a). The physiological basis for this model comes
from (1) observations of vernix caseosa asymmetry in these regions as presented by Akiba (1955) and Visscher
et al (2005) and (2) results from Archana (2008) which classify vernix caseosa thickness in these regions into
five ranges ($0$ mm, $0$–$1$ mm, $1$–$2$ mm, $2$–$3$ mm, $>3$ mm) as shown in figure 4(b). The correlation between vernix
caseosa thickness in each region and breakdown by gestational age was not reported by Archana, therefore these
results should not be interpreted as the typical vernix caseosa distribution for an individual but instead serve as
an indicative range of values to be studied.

To investigate the range of observed vernix caseosa thicknesses, we manually divide the processed 32
weeks GA fetal model into the seven regions as defined in figure 4(a). Compartments of vernix caseosa are generated by
selecting a group of connected triangles $S_1$ representing a desired region and translating its inner nodes along
the vertex normals to create a new surface $S_2$. Following this, a triangulated surface $S_3$ is created joining the exterior
nodes of $S_1$ and $S_2$ to define the vernix caseosa compartment bounded by $S_1 \cup S_2 \cup S_3$. This process is repeated
until all desired vernix caseosa compartments have been created. It is important to note the true distribution of vernix caseosa may be more granular than these seven regions, however this model will help identify regions
which have a significant effect on NI-FECG extraction accuracy.

To facilitate finite element model generation from the defined surfaces, the compartments of fetus, vernix
caseosa, amniotic fluid and maternal body are combined to form a triangulated piecewise linear complex (PLC)
(Miller 1998). From the triangular PLC, a tetrahedral finite element model can be generated via constrained
Delaunay tetrahedralization. In this work, this process is achieved using the open-source TetGen tool (v1.5.1)
developed by Si (2015) invoked through the iso2mesh MATLAB toolbox (v1.8.0) developed by Fang and Boas
(2009). TetGen allows the user to set a maximum element volume for the discretization of each compartment,
important for the thin layer of vernix caseosa to ensure an accurate FEM solution. An example tetrahedral finite
element model generated using this process is shown in figure 5.
3.2. Source parameters

Each cardiac source within this model must be assigned three parameters: (1) position of the source, (2) orientation of the source representing the electrical axis of the heart and (3) source vector loop, otherwise known as the vectorcardiogram, representing the 3D path traced by the source model. For the maternal heart these parameters are affected by inter-individual variations and physiological changes in the heart during pregnancy as well as fetal presentation and fetal size due to the displacement of internal organs (Fowler and Braunstein 1951, Bacharova and Ugander 2014, Gultekin et al 2016). As the scope of investigating these parameters is beyond this
work, the default position and orientation of the maternal source in our model are determined based on the mean values for a normal adult heart as reported by Nousiainen et al. (1986) with the ability to select from nine different vectorcardiograms as provided in the fecgsyn toolbox.

For the fetal source, in addition to inter-individual variations, the ratio of right ventricular to left ventricular weight varies throughout gestation resulting in a shift in electrical axis due to the increased thickness of muscle fibre (Emery and Macdonald 1960). While a recent study by Verdurmen et al. (2016) recorded preliminary results for the normal fetal electrical axis, they are not suitable for use in this work due to their small sample size. Instead, the default position of the fetal source in our model is based on the normal position for a healthy fetus (13–40 weeks GA) as reported in Comstock (1987) with the same anatomic orientation as the maternal source adjusted for right ventricular deviation as observed in early neonates (DePasquale and Burgh 1963). A summary of the relevant fecgsyn parameters utilized in our model are shown in table 2 with visualisation of the maternal and fetal source position and orientation shown in figure 6(a).

### 3.3. Metrics

Assessment of volume conductor effects is performed by comparing lead field matrix components \((x, y, z)\) and surface potentials at selected nodes in each model using the relative difference measure \((RDM^*)\) as proposed by Meijs et al. (1989) and the logarithmic magnitude error \((lnMAG)\) as proposed by Güllmar et al. (2010):

\[
RDM^* = \frac{\|x_a - x_b\|_2}{\|x_a\|_2} \quad \text{lnMAG} = \ln\left(\frac{\|x_a\|_2}{\|x_b\|_2}\right)
\]

where for \(n\) nodes, \(x_a\) and \(x_b\) are \(n \times 1\) vectors for inputs \(a\) and \(b\) respectively. \(RDM^*\) indicates the difference in distribution patterns, bounded by 0 for identical inputs and 2 for \(x_b = -x_a\), where increasing \(RDM^*\) represents greater difference in distribution patterns. Thus, \(RDM^*\) provides a simple metric to compare the morphology of two vectors without penalizing for overall changes in magnitude. \(lnMAG\) indicates the difference in overall magnitude where a value of 0 indicates an identical norm of the inputs and positive or negative deviation indicates relative change in magnitude. \(lnMAG\) is preferable to a direct magnitude ratio as it is symmetric about \(ln(1)\), enabling one-to-one comparison between positive and negative differences. As we aim to quantify effects for sources of varying orientation, we calculate \(RDM^*\) and \(lnMAG\) for the lead field matrices as \(3 \times 1\) vectors \([RDM^*], \, [RDM^*_y], \, [RDM^*_z]\) and \([lnMAG_x], \, [lnMAG_y], \, [lnMAG_z]\) indicating the value along each coordinate axis where \(\|RDM^*\|_2\) and \(\|lnMAG\|_2\) indicate the 2-norm for each \(3 \times 1\) vector. Additionally, as the T/QRS ratio has been proposed for identifying signs of fetal distress based on its absolute value or relative value compared to a chosen baseline, we define the T/QRS ratio error in each asymmetric volume conductor model as

\[
epsilon_{TQRS} = \frac{TQRS_a - TQRS_b}{TQRS_b}
\]

where \(TQRS_a\) is the T/QRS ratio in the asymmetric volume conductor model and \(TQRS_b\) is the T/QRS ratio in the homogeneous volume conductor model. The sign and magnitude of \(\epsilon_{TQRS}\) thus specifies the relative change in T/QRS ratio introduced by asymmetric volume conductor effects where positive \(\epsilon_{TQRS}\) indicates an increase in observed T/QRS ratio and negative \(\epsilon_{TQRS}\) indicates a decrease in observed T/QRS ratio compared to a homogeneous volume conductor model.

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**Table 2. Source model parameters.**

| Parameter | Description                                      | Value  |
|-----------|--------------------------------------------------|--------|
| fectb     | Fetal ectopic beats                              | 0      |
| fheart    | Fetal source position \((x, y, z)\)              | (772, 125, 1087) |
| fhr       | Fetal heart rate                                 | 150    |
| ftraj     | Fetal movement trajectory                        | None   |
| fyypeacc  | Fetal heart rate acceleration                     | None   |
| fvcg      | Fetal vectorcardiogram                           | 1      |
| mecctb    | Maternal ectopic beats                           | 0      |
| mheart    | Maternal source position \((x, y, z)\)           | (814, 165, 1315) |
| mhr       | Maternal heart rate                              | 80     |
| mtraj     | Maternal movement trajectory                      | None   |
| mtypeacc  | Maternal heart rate acceleration                  | None   |
| mvvcg     | Maternal vectorcardiogram                         | 1      |
| posdev    | Heart position deviation                          | 0      |
| Rx        | Fetal source rotation \((\theta_x, \theta_y, \theta_z)\) | (2.8, 1.6, 3) |
| Rw        | Maternal source rotation \((\theta_x, \theta_y, \theta_z)\) | (1.5, 0, 0) |
To quantify the effects of volume conductor asymmetry on NI-FECG extraction, we compute the fetal lead matrix ($H_f$) at all maternal body surface nodes between $750 \, \text{mm} \leq z \leq 1350 \, \text{mm}$ ($n = 682$) for visualisation purposes and define a subset of nodes between $950 \, \text{mm} < z < 1200 \, \text{mm}$ ($n = 266$) as the abdominal sensors for numerical evaluation as shown in figure 6(b). To ensure minimal numerical error in the chosen finite element discretization, we perform a refinement process using a model with vernix caseosa generated at 1 mm in the Back region to determine the required TetGen parameters. This process involves repeatedly completing the TetGen and SimBio steps as shown in figure 7 and iteratively halving the maximum element volume per compartment until a defined level of convergence is obtained. For this work, refinement was performed until $\|RDM\|_2$ of the abdominal sensors lead field matrix compared with the subsequent solution was below 0.02 for two consecutive iterations as shown in figure 8. All finite element models in this work were generated using the parameters identified via this process and compared to a reference solution with halved maximum element volume per compartment showing minimal numerical error in $RDM^*$ and $lnMAG$ (results in appendix table A1).

### 3.5. Experiments

Using the developed process, this work aims to characterise morphological changes caused by volume conductor asymmetry including spatially variable distributions of vernix caseosa in the seven fetal body regions as defined in figure 4(a). Based on the data shown in figure 4(b), it would be ideal to investigate models where vernix caseosa thickness in each region takes one of four values: 0 mm, 1 mm, 2 mm or 3 mm. However, as this space represents $16^{3}$ (4³) possible combinations, we must define a subset of this space to make analysis feasible.

#### 3.5.1. Region assessment

In the first experiment, we quantify regions which have the most significant impact by analysing a subset of volume conductor models consisting of the following 16 scenarios: (1) homogeneous volume conductor model as defined by equation (5) with conductivity set to that of the maternal abdomen ($0.2 \, \text{S m}^{-1}$), (2) finite element model with no vernix caseosa and (3–16) finite element models consisting of vernix caseosa isolated in each region at 1 mm and 3 mm thickness (minimum and maximum non-zero values). Using the generated lead field matrices, we calculate pairwise $RDM^*$ and $lnMAG$ along each coordinate axis for each of the 16 volume conductor models. From these results, the three regions with the greatest $\|RDM^*\|_2$ and $\|lnMAG\|_2$ compared to the No Vernix model are selected for further analysis. The No Vernix model is used for comparison as it characterizes regional effects more precisely compared to shared differences with respect to the homogeneous model.

#### 3.5.2. Signal morphology

In the second experiment, we generate asymmetric volume conductor models in the three identified regions and assess changes to NI-FECG morphology using two methods: (1) visualizing surface potentials produced by dipoles along each coordinate axis, and (2) calculating $\epsilon_{\text{TQRS}}$ and $RDM^*$ at six sensor positions on the maternal abdomen compared to a homogeneous volume conductor model. This is achieved by simulating a 300 ms vectorcardiogram containing one fetal cardiac cycle using the parameters defined in table 2 and calculating the NI-FECG waveform at each sensor per volume conductor model. As current benchmarks using fecgsyn assume that accurately extracting this component from an abdominal mixture in a homogeneous volume conductor model indicates correct identification of the underlying source, quantifying the observed error due to volume conductor effects demonstrates the potential for estimation error in real data using this approach.
Figure 7. Developed process for generating surface potentials in an asymmetric volume conductor model. Dark gray boxes indicate data from the labeled source. Light gray arrows indicate a processing step using the labeled tool. Black arrows indicate data input to the target. (a) Indicates steps utilized for model setup and (b) indicates steps utilized for the experiments.

Figure 8. Model refinement process showing lead field matrix convergence for the Back 1 mm model as maximum element volume per compartment is halved, approximately doubling the number of tetrahedra at each step. a shows a converging relative difference measure (||RDM*||) calculated at each step compared to the subsequent solution, ▲ shows time to compute tetrahedralization via TetGen and △ shows time to compute lead field matrix via SimBio per sensor. Subsequent reference solution for the final data point is not shown.
To calculate the T/QRS ratio for each NI-FECG waveform, the QRS complex amplitude is measured as the difference between the minimum and maximum values in the QRS complex detection zone (defined as 80–120 ms) while the T wave amplitude is measured as the maximum absolute value in the T wave detection zone (defined as 180–220 ms) with respect to the isoelectric line (defined as the value at 300 ms). This process is necessary as the exact sample which represents the minimum/maximum QRS complex amplitude and maximum T wave amplitude may shift slightly depending on the chosen sensor position and volume conductor model.

4. Results

This section presents numerical results for the experiments described. All simulations were performed in MATLAB R2017a on the University of Melbourne’s High Performance Computing system (Meade et al 2017) comprised of 100 Linux computing nodes each with an eight-core 2.6Ghz Intel Xeon CPU and 62GB RAM with each simulation running on a single node.

4.1. Region assessment

Using the described volume conductor models, pairwise \textit{RDM} and \textit{lnMAG} were computed for the abdominal sensors lead field matrix as shown in figure 9. Each heat map indicates the relative change in surface potentials between models in terms of distribution patterns (\textit{RDM}) and overall magnitude (\textit{lnMAG}) as generated by components of the fetal source along each coordinate axis.

From these results, we observe that the Back models demonstrate the largest change in \textit{RDM} compared to the homogeneous model followed by the Head, Arms and Chest models. Differences in \textit{RDM} are less pronounced with the Back, Arms and Head models showing marginally greater change, while for \textit{RDM} all models show significant change compared to the homogeneous model. For all models with vernix caseosa, it can be observed that varying vernix caseosa thickness from 1 mm to 3 mm in each region has minimal effect on \textit{RDM} along all coordinate axes.

For \textit{lnMAG}, all models show a decrease compared to the homogeneous model with the Back and Chest models demonstrating the greatest change. Interestingly, while \textit{lnMAG} for the Head models is decreased compared to the homogeneous model, it is marginally increased compared to all other models. For \textit{lnMAG}, all models exhibit a large decrease compared to the homogeneous model with the Head model again showing a small increase in \textit{lnMAG}, compared to non-homogeneous models. Most models show minimal change in \textit{lnMAG} compared to the homogeneous model except for the Arms and Head models which marginally increase and the Back models which marginally decrease. Similar to \textit{RDM}, varying the thickness of vernix caseosa from 1 mm to 3 mm in each region has minimal effect on \textit{lnMAG} along all coordinate axes.

To quantify the three regions of greatest impact, we calculate $\|RDM\|_2$ and $\|lnMAG\|_2$ rankings for the 3 mm models compared to the No Vernix model as shown in figure 10. Ranking is performed using the 3 mm models as overall they demonstrate marginally greater impact compared to the 1 mm models from these rankings, the Back 3 mm model shows the greatest change in $\|RDM\|_2$ followed by the Head 3 mm and Chest 3 mm models. For $\|lnMAG\|_2$, the Back 3 mm model shows the greatest change followed by the Chest 3 mm and Arms 3 mm models. As this work is primarily concerned with morphological changes indicated by greater $RDM$, the three models selected for further analysis are the Back 3 mm, Chest 3 mm and Head 3 mm models.

4.2. Signal morphology

To visualize the effects of volume conductor asymmetry in the three selected models, surface potentials produced by dipoles along each coordinate axis for the homogeneous, No Vernix, Back 3 mm, Chest 3 mm and Head 3 mm models are shown in figure 11. As indicated by the Region Assessment results, these visualizations show: (1) large change in distribution patterns and greatly reduced magnitude for $x$ axis dipoles in the Back 3 mm model, (2) reduced magnitude for $x$ axis dipoles in the Chest 3 mm model, (3) large change in distribution patterns for $z$ axis dipoles in all non-homogeneous models, and (4) reduced magnitude for $y$ axis dipoles in all non-homogeneous models.

Following these observations, we simulate NI-FECG waveforms in volume conductor models comprising the following nine scenarios: (1) homogeneous volume conductor model as previously defined and (2–9) finite element volume conductor models with binary combinations of vernix caseosa at 3 mm thickness in the Back, Chest and Head regions where the presence/absence of vernix caseosa in each region is indicated by the model name (e.g. BackChest 3 mm model contains vernix caseosa at 3 mm thickness in the Back and Chest region, but not the Head region). The six sensor positions as shown in figure 12 are chosen to be approximately equally spaced over the abdominal surface above and below the fetal source and aligned to the closest available nodes in the model discretization. Each sensor is measured in respect to a reference node at the center of the maternal back as per the default \textit{fegsyn} sensor configuration (see Andreotti et al (2016) for detail). Figure 12 presents simulated NI-FECG waveforms for the nine volume conductor models as described with $\textit{v}_{\text{TQRS}}$ and $\textit{RDM}$ reported at each.
sensor position in table 3. To enable simple visual comparison between models, each signal’s isoelectric line in figure 12 is aligned to 0 V and the fetal vectorcardiogram amplitude is linearly scaled by a factor of $10^{-5}$ to a peak source strength of approximately 18 $\mu$Am.

As shown in table 3, a maximum $e_{TQRS}$ of $-0.77$ is observed in the BackChest 3 mm model for sensor 5, indicating 77% relative error in the observed T/QRS ratio. The next three highest $e_{TQRS}$ values are present in the BackChestHead 3 mm ($-0.73$), BackHead 3 mm ($-0.67$) and Back 3 mm ($-0.66$) models indicating that volume conductor asymmetry in the Back region has a large effect on the observed T/QRS ratio. Furthermore, in all models with vernix caseosa in the Back region, $e_{TQRS}$ is negative at all sensor positions except for sensor 6 in the BackHead 3 mm model, indicating a reduced T/QRS ratio compared to a homogeneous model. The No Vernix model also shows large variation in $e_{TQRS}$ with a minimum of $-0.26$ in sensor 3 and maximum of 0.37 in sensor 5 indicating that volume conductor effects have a large influence on signal morphology even in the absence of vernix caseosa.

$RDM^*$ compared to a homogeneous model also demonstrates considerable variation across all models with a maximum $RDM^*$ of 0.80 observed in sensor 1 for the BackChest 3 mm model followed by the BackChestHead 3 mm (0.74), BackHead 3 mm (0.61) and Back 3 mm (0.49) models with all other models having at least one sensor with an $RDM^*$ of 0.25 or greater.
Figure 11. Coronal cross-section (r > 127 mm) of selected volume conductor models and resulting surface potentials produced by dipoles along each coordinate axis with rows representing. (a) Homogeneous, (b) No Vernix, (c) Back 3 mm, (d) Chest 3 mm and (e) Head 3 mm volume conductor models. ● indicates position of the fetal source and tissue types are color coded as follows: ■ = fetus, □ = vernix caseosa, △ = amniotic fluid, □ = maternal body. Note: The color scale is asymmetric around 0 V (Am)^{-1}, indicating maximum negative potentials are approximately twice as large as maximum positive potentials.

Figure 12. Simulated NI-FECG waveforms for the homogeneous and asymmetric volume conductor models (a) shows maternal body model with six sensor positions, ●●●●●●, reference node O and fetal source position ● (b) shows potentials observed at each sensor position with respect to the reference node for the homogeneous and asymmetric volume conductor models: No Vernix — Back 3 mm — Chest 3 mm — Head 3 mm — BackChest 3 mm — BackHead 3 mm — ChestHead 3 mm — BackChestHead 3 mm —. ▼ indicates the QRS complex detection zone and ▲ indicates the T wave detection zone.
Additionally, varying the thickness of vernix caseosa isolated in each region from 1 mm to 3 mm did not greatly affect RDM as asymmetric volume conductor models including spatially variable distributions of vernix caseosa. Using this process, we classified the three fetal body regions with the greatest impact in terms of changes to RDM. To achieve this, a process was developed to compute lead field matrices and NI-FECG waveforms in a set of RDM models as shown in figure 11 demonstrated large changes in potential distribution and magnitude compared to a homogeneous volume conductor model. Specifically, we observed that volume conductor asymmetry in the Back region leads to the greatest changes in terms of both eTQRS and lnMAG and assessed the impact of varying vernix caseosa thickness in each region. Based on these results, surface potentials produced by dipoles along each coordinate axis and NI-FECG waveforms at six sensor positions on the maternal abdomen were assessed in a range of asymmetric volume conductor models demonstrating significant changes to T/QRS ratio error and overall signal morphology compared to a homogeneous volume conductor model.

### 5. Discussion

In this study, we aimed to characterise the effects of volume conductor asymmetry on NI-FECG extraction. To achieve this, a process was developed to compute lead field matrices and NI-FECG waveforms in a set of asymmetric volume conductor models including spatially variable distributions of vernix caseosa. Using this process, we classified the three fetal body regions with the greatest impact in terms of changes to RDM and lnMAG and assessed the impact of varying vernix caseosa thickness in each region. Based on these results, surface potentials produced by dipoles along each coordinate axis and NI-FECG waveforms at six sensor positions on the maternal abdomen were assessed in a range of asymmetric volume conductor models demonstrating significant changes to T/QRS ratio error and overall signal morphology compared to a homogeneous volume conductor model.

#### 5.1. Region assessment

We observed that volume conductor asymmetry in the Back region leads to the greatest changes in terms of both RDM and lnMAG, which can be attributed to its large coverage area and close proximity to the fetal source. Additionally, varying the thickness of vernix caseosa isolated in each region from 1 mm to 3 mm did not greatly affect RDM or lnMAG values along all coordinate axes. This can be attributed to the fact that as each region in isolation does not fully enclose the fetus, the path of least resistance to the maternal abdomen is only slightly modified by an increase in vernix caseosa thickness. From these studies, the three models of greatest impact were identified as the Back 3 mm, Chest 3 mm and Head 3 mm models.

#### 5.2. Signal morphology

Visualisation of surface potentials produced by dipoles along each coordinate axis for the identified three models as shown in figure 11 demonstrated large changes in potential distribution and magnitude compared to a homogeneous volume conductor model. Specifically, we observed that volume conductor asymmetry in the Back region resulted in greatly reduced magnitude for x axis dipoles, typically leading to reduced T wave amplitude as shown in figure 12. Following this observation, we note that difficulties in detecting T-waves via NI-FECG have been reported throughout the literature as summarised in Wacker-Gussmann et al (2017). While other causes may likely contribute to this phenomenon, the high level of vernix caseosa asymmetry reported in the back region by Akiba (1955), Visscher et al (2005) and Archana (2008) presents a novel explanation for this effect. In addition, signal morphology and T/QRS ratio error were significantly altered in the No Vernix model, indicating that volume conductor effects play an important role even when vernix caseosa is not present.

Based on these observations, we conclude NI-FECG algorithm benchmarks utilizing a homogeneous volume conductor model do not indicate representative accuracy of T/QRS ratio extraction in real data and volume conductor effects should be considered in future benchmarks using simulated data. Furthermore, it should be noted that even in a homogeneous volume conductor model, the T/QRS ratio will change with sensor location

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Table 3. Signal morphology results indicating eTQRS and RDM* at each sensor position per model. Worst case per row in bold.

| Model               | Metric | 1   | 2   | 3   | 4   | 5   | 6   |
|---------------------|--------|-----|-----|-----|-----|-----|-----|
| No Vernix           | eTQRS  | −0.12 | 0.12 | −0.26 | 0.08 | **0.37** | 0.07 |
|                     | RDM*   | 0.10  | **0.25** | 0.18  | 0.03 | 0.18  | 0.17 |
| Back 3 mm           | eTQRS  | −0.49 | −0.45 | −0.35 | −0.01 | **−0.66** | −0.20 |
|                     | RDM*   | **0.49** | 0.45  | 0.27  | 0.14  | 0.36  | 0.13 |
| Chest 3 mm          | eTQRS  | −0.27 | −0.08 | −**0.33** | 0.06 | −0.23 | −0.23 |
|                     | RDM*   | 0.22  | **0.26** | 0.24  | 0.05  | 0.14  | 0.16 |
| Head 3 mm           | eTQRS  | −0.11 | 0.30  | −0.27 | 0.12  | **0.51** | 0.44 |
|                     | RDM*   | 0.08  | 0.23  | 0.18  | 0.07  | **0.25** | 0.23 |
| BackChest 3 mm      | eTQRS  | −0.71 | −0.52 | −0.46 | −0.14 | **−0.77** | −0.55 |
|                     | RDM*   | **0.80** | 0.52  | 0.39  | 0.20  | 0.56  | 0.45 |
| BackHead 3 mm       | eTQRS  | −0.30 | −0.65 | −0.25 | −0.04 | **−0.67** | 0.04 |
|                     | RDM*   | 0.23  | **0.61** | 0.20  | 0.13  | 0.59  | 0.07 |
| ChestHead 3 mm      | eTQRS  | −0.27 | −0.16 | −0.44 | 0.11  | −0.26 | **0.48** |
|                     | RDM*   | 0.21  | 0.29  | **0.36** | 0.15 | 0.24  | 0.15 |
| BackChestHead 3 mm  | eTQRS  | −0.55 | −**0.73** | −0.61 | −0.26 | −0.52 | −0.39 |
|                     | RDM*   | 0.51  | **0.74** | 0.51  | 0.29  | **0.74** | 0.32 |

* Indicates multiple columns with equal value.
and fetal orientation due to variations in the projected components of the underlying cardiac dipole. As such, without knowledge of the influence of volume conductor effects, clinical evaluation of the T/QRS ratio derived via NI-FECG should be avoided.

5.3. Future work
In the present study, we considered a single fetal model with a fixed set of source parameters. While these parameters were carefully selected based on clinical measurements, further analysis should be conducted using a range of anatomic and vectorcardiogram models to determine the typical distribution of error within a larger pregnancy cohort. Further clinical studies are also warranted to determine the likelihood of the investigated vernix caseosa distributions across different gestational ages and validate the predictions of our model against observed abdominal potentials. Regardless of these factors, the presented results demonstrate the significant potential for estimation error in real data and confirm that volume conductor effects represent an important topic in the study of NI-FECG extraction accuracy.

In addition to the 32 weeks GA model utilized in this work, the developed process allows for the simulation of NI-FECG waveforms using any triangular PLC, enabling the study of volume conductor effects over a range of gestational ages and fetal positions with the possibility to include additional structures in the modeling process such as the placenta, umbilical cord and internal organs. As part of further studies, the impact of including additional compartments as well as model simplifications, such as finite element models of a constant conductivity could be investigated. Additionally, as the surface potentials in our process are generated according to a linear mixture, it can be utilized to simulate abdominal ECG mixtures for multiple fetuses (e.g. twins) via the inclusion of additional cardiac sources within an appropriate volume conductor model. Further expansion may also investigate alternative cardiac source models such as a set of spatially distributed dipoles.

Finally, our process can be easily extended to the field of fetal magnetocardiography, allowing study of the effects of volume conductor asymmetry on magnetic field distribution. As our process has been developed using MATLAB and a set of open-source tools, a future development goal is to release the developed code under an open-source license to enable rapid NI-FECG benchmarks incorporating volume conductor effects within the research community.

6. Conclusion
This work demonstrates that volume conductor effects have a significant impact on the signal morphology derived via NI-FECG. Simulation studies using finite element models of the maternal–fetal anatomy revealed that volume conductor asymmetry can result in over 70% error in the observed T/QRS ratio and significant morphological changes compared to a homogeneous volume conductor model. Future NI-FECG algorithm benchmarks using simulated data should incorporate volume conductor effects to provide better indication of NI-FECG extraction accuracy using real data.

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Appendix
Model refinement results are shown in table A1 for the abdominal sensors lead field matrix for all utilized finite element models. For each model, table A1 describes RDM* and lnMAG along each coordinate axis compared to a reference solution with halved maximum element volume in each compartment, tetrahedralization computation time via TetGen, lead field matrix computation time via SimBio and number of nodes and tetrahedra.
Table A1. Model refinement results showing RDM* and lnMAG along each coordinate axis compared to reference solution, tetrahedralization computation time (TetGen), lead field matrix computation time (SimBio) and number of nodes and tetrahedra per model. Worst case per column in bold for single region and multiple regions separately.

| Type          | Model         | RDM* | lnMAG | TetGen (s) | SimBio (s) per sensor (×10⁶) | # of nodes | # of tetrahedra (×10⁶) |
|---------------|---------------|------|-------|------------|-----------------------------|-----------|------------------------|
|               |               | x    | y     | z          |                              |           |                        |
| Single region | No Vernix     | 0.0036| 0.0066| 0.0061     | 0.0011                       | −0.0008   | 0.0008                 | 91.65 | 119.86 | 0.7221 | 4.3394 |
|               | Arms 1 mm     | 0.0063| 0.0108| 0.0037     | 0.0000                       | −0.0044   | −0.0024                | 113.67| 132.46 | 0.7968 | 4.7945 |
|               | Arms 3 mm     | 0.0029| 0.0069| 0.0052     | −0.0006                      | −0.0038   | −0.0031                | 111.02| 144.86 | 0.8094 | 4.8716 |
|               | Back 1 mm     | 0.0088| 0.0070| 0.0059     | 0.0004                       | −0.0206   | 0.0016                 | 115.87| 135.40 | 0.8103 | 4.8764 |
|               | Back 3 mm     | 0.0036| 0.0048| 0.0034     | 0.0006                       | −0.0001   | −0.0003                | 109.47| 139.09 | 0.8280 | 4.9858 |
|               | Buttocks 1 mm | 0.0062| 0.0063| 0.0026     | 0.0009                       | 0.0292    | 0.0009                 | 95.77 | 119.38 | 0.7341 | 4.4133 |
|               | Buttocks 3 mm | 0.0038| 0.0067| 0.0036     | 0.0018                       | 0.0038    | 0.0016                 | 96.60 | 115.88 | 0.7367 | 4.4290 |
|               | Chest 1 mm    | 0.0068| 0.0040| 0.0053     | 0.0043                       | 0.0024    | 0.0033                 | 104.68| 124.57 | 0.7363 | 4.4264 |
|               | Chest 3 mm    | 0.0133| 0.0087| 0.0039     | −0.0029                      | −0.0028   | −0.0035                | 100.37| 116.08 | 0.7384 | 4.4387 |
|               | Head 1 mm     | 0.0053| 0.0096| 0.0042     | −0.0002                      | −0.0003   | 0.0008                 | 104.89| 132.79 | 0.8219 | 4.9478 |
|               | Head 3 mm     | 0.0071| 0.0110| 0.0032     | −0.0018                      | 0.0023    | −0.0005                | 109.39| 156.95 | 0.8449 | 5.0884 |
|               | Inguinal 1 mm | 0.0084| 0.0115| 0.0094     | 0.0022                       | 0.0080    | 0.0073                 | 108.30| 110.73 | 0.7275 | 4.3716 |
|               | Inguinal 3 mm | 0.0071| 0.0128| 0.0052     | 0.0009                       | 0.0048    | 0.0002                 | 96.29 | 116.43 | 0.7291 | 4.3824 |
|               | Legs 1 mm     | 0.0069| 0.0020| 0.0035     | −0.0017                      | 0.0015    | −0.0026                | 113.86| 135.69 | 0.8056 | 4.8471 |
|               | Legs 3 mm     | 0.0041| 0.0077| 0.0043     | 0.0004                       | −0.0034   | 0.0008                 | 108.24| 151.92 | 0.8192 | 4.9303 |
| Multiple regions | BackChest 3 mm | 0.0164| 0.0038| 0.0069     | 0.0117                       | 0.0006    | 0.0021                 | 113.63| 150.34 | 0.8435 | 5.0795 |
|               | BackHead 3 mm | 0.0073| 0.0045| 0.0068     | 0.0080                       | −0.0029   | −0.0005                | 135.31| 173.86 | 0.9521 | 5.7427 |
|               | BackChestHead 3 mm | 0.0011| 0.0035| 0.0056     | −0.0022                      | −0.0020   | 0.0028                 | 125.54| 159.00 | 0.8606 | 5.1845 |
|               | BackChestHead 3 mm | 0.0204| 0.0053| 0.0041     | 0.0028                       | 0.0034    | 0.0060                 | 139.47| 185.91 | 0.9672 | 5.8341 |

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