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On the fetal magnetocardiogram

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Abstract

Fetal magnetocardiography is a non-invasive technique for studying the electrical activity of the fetal heart. Fetal magnetocardiograms (fMCG) can be used to diagnose and classify fetal cardiac arrhythmias reliably. An averaged fMCG shows a QRS-complex, a P-wave, and a T-wave. However, it is still unknown if the currents in the tissues surrounding the fetal heart disturb these features. Furthermore, the measuring technique is not yet optimised for fMCG registrations. Simulation studies may provide guidelines for the design of an appropriate magnetometer system. Therefore, finite-element and boundary-element models were constructed in order to study the possible influence of the volume conductor. Especially, the influence of the layer of vernix caseosa, a fatty layer that covers the fetus, was investigated. The computations showed that the layer of vernix caseosa will affect the waveform of the fMCG. The signal processing procedure used is also discussed. It turned out to be difficult to deduce the onset and offset of the T-wave from the resulting averaged signals. Possibly, the QRS-complex does not provide a correct trigger to obtain a distinguishable T-wave in the averaged signal, because the RT-interval may be variable. © 1998 Elsevier Science S.A. All rights reserved.

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1. Introduction

Electric sources associated with the depolarisation and repolarisation processes within the fetal heart generate Ohmic currents in the surrounding tissues. Both the electric sources and the Ohmic currents give rise to a magnetic field. The registration of a component of this magnetic field in points near the abdomen of the pregnant woman is called a fetal magnetocardiogram (fMCG). The strength of this field is about 10^{-12} T, which is extremely small. For comparison, the magnetic field of the earth is 5×10^{-5} T. These small fields can be measured by means of a SQUID magnetometer. A SQUID is a sensor for magnetic flux that has to be superconducting and consequently needs to be cooled. At this moment the coolant is liquid helium boiling at 4.2 K, i.e., -269°C. In order to cancel disturbing fields the fetal signal is coupled into a SQUID by means of a flux transformer, which can be constructed in such a way that it is, for instance, insensitive to a homogeneous field or a field having a homogeneous gradient. Due to the flux transformer, the magnetometer is more sensitive to fields from sources in the fetal heart than to sources at a greater distance than the fetal heart, such as the field from a passing car or from the heart of the mother. In spite of this, the signal-to-noise ratio of the fetal MCG is still low. This is why most research groups carry out their measurements in a shielded room, although this is not a necessity [1].

During the second and third trimesters of pregnancy, registrations of the fetal QRS-complexes can also be obtained by means of fetal electrocardiography (fECG). However, the recording of an fECG, measured using electrodes attached at the maternal abdomen, is a troublesome process and it proves difficult to get information about the P- and T-waves. The fECG may be measurable from about the 20th week of pregnancy onward. Fetal vectorcardiograms, which reflect the rotation and strength of the electrical heart vector during the cardiac cycle, recorded before 28 weeks of pregnancy show open loops. The heart vector is a term used to designate all of the electrical activity. The magnitude and direction changes during a cardiac cycle. At distances that are large compared to the dimensions of the

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heart, the heart vector can be described by an equivalent current dipole. After about 28 weeks of gestation the fECG signal usually decreases, and at 30 weeks it is often immeasurable. This drop in signal strength is ascribed to the insulating effect of the vernix caseosa, a pasty covering that protects the skin of the fetus. It consists mainly of dead cells and sebaceous secretions. As the conductivity of this layer is about a factor 10^6 lower than that of the surrounding tissue, only a small electric current flows to the surface of the maternal abdomen, and hence no potential differences can be measured there. After a few more weeks, the fECG signal reappears. However, vectorcardiograms computed from fECGs measured after 32 weeks of gestation show almost straight lines. In other words, the electrical vectorcardiogram does not represent the electrical heart vector anymore, as it is known that this vector changes in strength and orientation during the cardiac cycle. Presumably, the uncharacteristic shape of the vectorcardiogram can be ascribed to preferred pathways in the insulating layer of vernix caseosa. Oostendorp tried to simulate the findings described above using the boundary-element method (BEM) [2]. The model of the volume conductor consisted of the fetus, and the maternal abdomen. To simulate the situation of current escaping from two holes in an electrically insulating layer, a source description of two monopoles was used, having equal strength and opposite polarities. This model was applied to the data for late pregnancies. For some of these data sets the positions of the localised monopoles seemed to agree with the presumption that the current escapes the fetus through the umbilical cord and the mouth. However, for other data sets the monopoles were not localised near the fetal mouth and umbilical cord.

In most leads of the adult ECG the T-wave is somewhat stronger than the P-wave. However, in averaged transabdominal fetal ECGs the T-wave is practically always smaller. This can be explained by the fact that the volume conductor acts as a high-pass filter. Normally, the volume conductor problem is solved by using the Maxwell equations in a static approximation. Whether this simplification is justified depends on the frequencies of the analysed signals and the properties of the media, e.g., the conductivity σ, the permittivity ε, and the magnetic permeability μ. For human tissue at frequencies below 1000 Hz induction and propagation terms are negligible [3]. The absolute value of the complex conductivity $|\sigma_c|$ was obtained from Oostendorp et al. [4], who measured the impedance of a layer of vernix caseosa as a function of frequency. Typical values found are $|\sigma_c| = 10^{-7}$ S/m for 1 Hz and $|\sigma_c| = 10^{-6}$ S/m for 100 Hz. They observed a phase shift of $\pi/2$ in the interval of 1–100 Hz. It has been shown for a three-sphere model that the layer of vernix acts as a temporal filter. This model, using the values of the complex conductivity quoted above, described the fetus, a layer of vernix and the maternal abdomen. As a result, the features of the signal were changed considerably; the P-wave was reduced, the T-wave had disappeared, and the shape of the QRS-complex was changed [5].

The fMCG can be measured from the 13th week of gestation onward [6]. Also between the 28th and 32nd week of gestation it remains measurable. Furthermore, the vectorcardiogram computed from the fMCG shows an open loop during the entire period of pregnancy [7]. The influence of the volume conduction on the signals is unknown. Oostendorp and van Oosterom performed model studies of the fMCG [8]. As a source, they took a current dipole, with a time course as obtained from inverse calculations based on fECGs, measured before the vernix caseosa had formed. Two models of the volume conductor model were studied; firstly, a model consisting of the fetus, the placenta, the uterus, and the maternal abdomen and secondly, a homogeneous model with the shape of the fetus. In the latter case no currents were assumed to flow outside the isolating layer. Forward computations were carried out using both models. It was found that the resulting magnetic fields differed substantially. No measurements were available to decide which model was best.

It has been shown that the fMCG can be used to diagnose and classify cardiac arrhythmias [9]. However, the diagnostic applicability of fMCG is still limited, because the onset and strength of the T-wave cannot be determined reliably. Moreover, the magnetometer systems that are used at present are very expensive and a skilled technician is necessary to perform measurements. Therefore, we want to design a fetal magnetocardiograph that is user-friendly, uses nitrogen as a coolant, and can be used in a clinical (unshielded) environment [10]. Magnetometer systems cooled by liquid nitrogen that can be used to measure the adult MCG in unshielded surroundings have already been constructed [11]. To optimise the design, the influence of the volume conductor on measured fMCGs is discussed, because simulations may be of help (a) to determine whether features of the fMCG are severely disturbed by the volume conduction; (b) to optimise the flux transformer; and (c) to decide where and in what direction to measure the magnetic field. Effects of the volume conductor on fMCGs are studied by means of simulations using the BEM or a combination of the finite element method (FEM) and the application of the Biot–Savart law.

2. Methods

A fetal MCG signal, even when it is measured in a magnetically shielded room, has to be processed to acquire a signal in which the different stages of the depolarisation and repolarisation of the heart (PQRST-wave) are clearly visible. This is carried out by averaging the fetal heart signal over a number of heartbeats. To enable signal averaging the individual heartbeats have to be detected and aligned. Sometimes the fMCG is severely contaminated by signals from the mother's heart. To be able to distinguish between maternal and fetal heart signals, an ECG of the mother is measured simultaneously. The signal processing procedure is illustrated in Fig. 1. Fig. 1a shows an example of a raw signal measured in a magnetically shielded room. The component of the field in the vertical direction was measured with the mother in supine position. The flux transformer consisted of two coaxial coils connected in anti-series, thus forming a so-called first-order gradiometer [12]. Both coils had five turns with a diameter of 40 mm. The distance between the centres of the coils (i.e., the baseline) was 50 mm. The mother's ECG was measured via electrodes attached to the right wrist and the left ankle. The time instants of the R-wave peaks obtained from the maternal ECGs were used for coherently averaging the raw data. The averaged maternal signal found in the raw data of the fetal MCG is shown in Fig. 1b. From the averaged maternal signal (20 beats) a template was taken describing the mother's QRS-complex (see Fig. 1c). The template was subsequently subtracted from the raw data and digital filters were applied (1–70 Hz band-pass and 50 Hz notch). The result is pictured in Fig. 1d. Next, the fetal R-peaks were detected using a threshold. The time instants of these R-peaks were used for coherently averaging the original raw data. The averaged signal, obtained from 20 fetal heartbeats, is shown in Fig. 1e. As demonstrated, averaging improves the signal-to-noise ratio such that filtering and subtraction of the maternal signal are not needed. The signal, as shown in Fig. 1d, allows us to determine the duration of the PR-interval and RR-interval. The latter can also be used to calculate the fetal heart rate as a function of time. The fetal heart rate is given in Fig. 1f.

3. Models

In order to design a flux transformer that is optimal for fMCG measurements, the sensitivity for the fetal magnetic field has to be estimated in relation with the field generated by other sources such as the mother's heart. The ratio of the contributions of the fetal heart and the maternal heart to the magnetic field was estimated using a homogeneous model with the shape of the maternal abdomen. The realistic shape at a gestational age of about 30 weeks was obtained from an anatomical atlas. The BEM was used for the computations. The triangulated model is shown in Fig. 2. The electric sources were modelled by two current dipoles positioned in the centres of the hearts. The projections of these positions are indicated in the figure. The dipole strength representing the electrical heart activity of

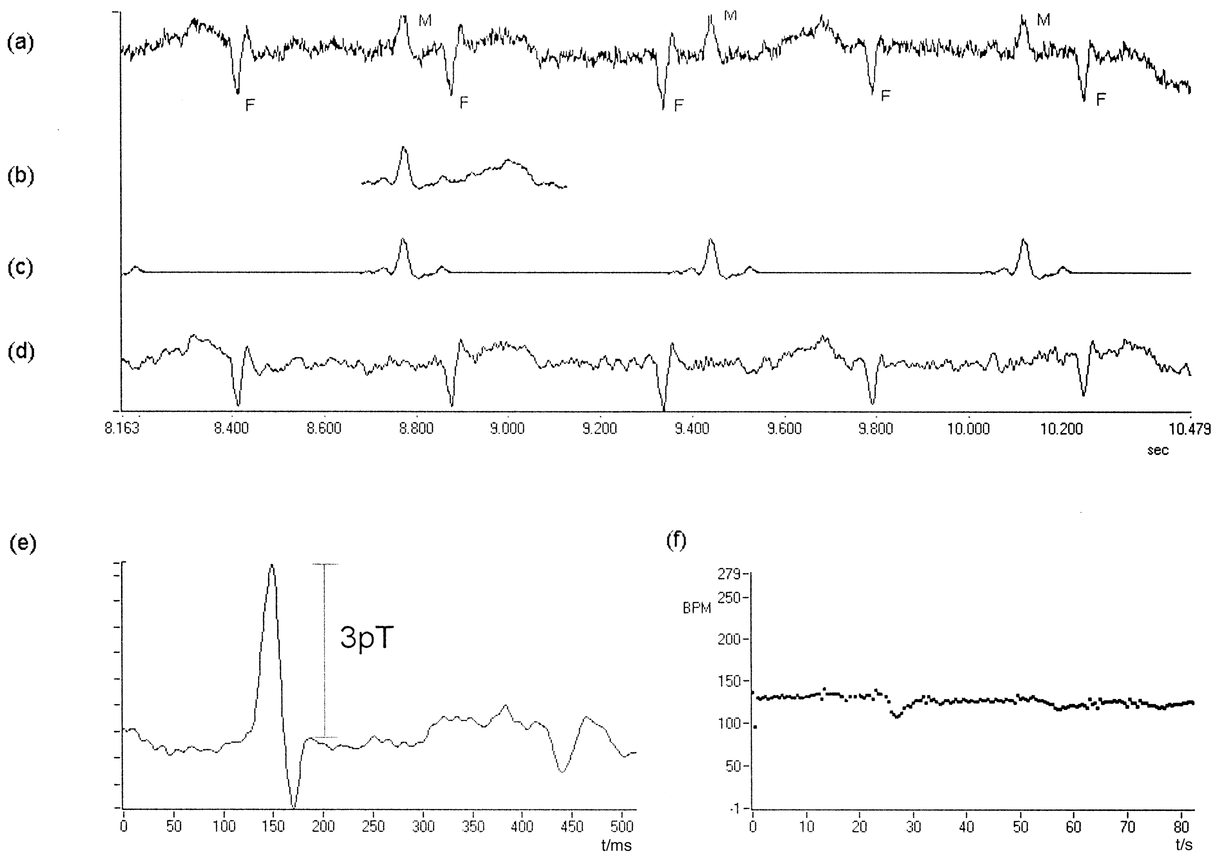

Fig. 1. The signal processing procedure: (a) the raw signal; (b) the averaged maternal ECG as found in the raw fMCG signal; (c) the template which is subtracted from the raw data; (d) the fMCG after filtering and subtracting the QRS-complex due to the mother's heart; (e) the averaged raw fMCG signal; (f) the fetal heart rate as a function of time in beats per minute (bpm).

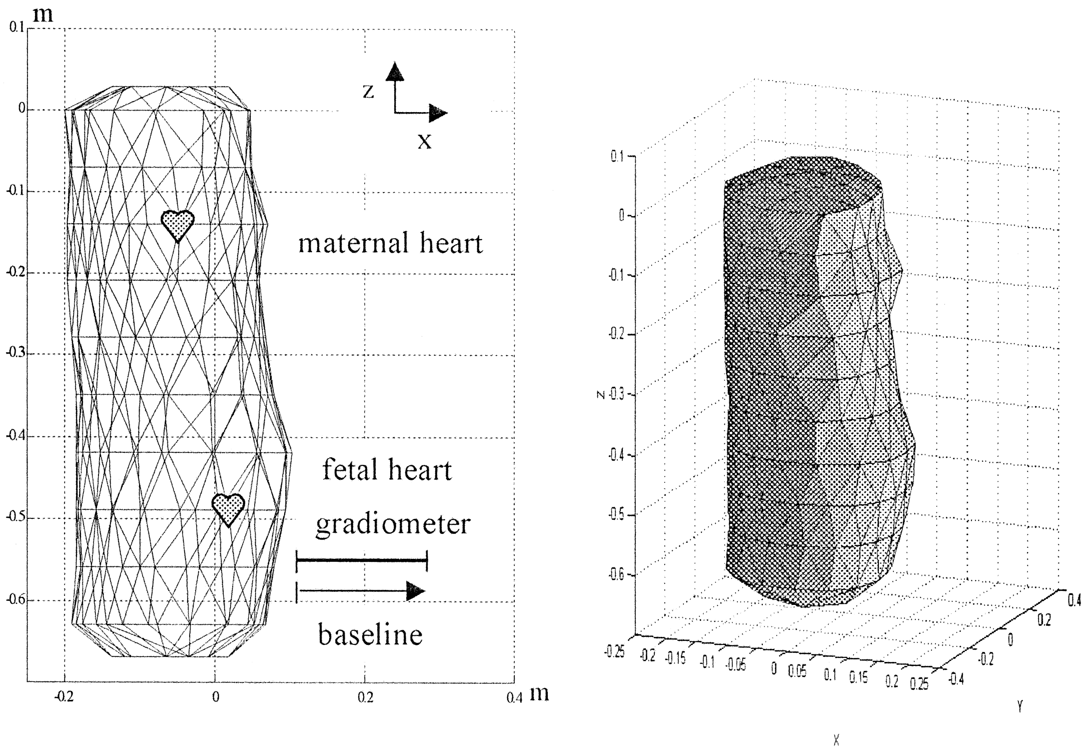

Fig. 2. The homogeneous model describing the maternal abdomen used to estimate the influence of the flux transformer.

the fetus was chosen to be 0.5 μA m and that of the mother 8 μA m. These values were deduced from measurements. Apart from these values, the ratio depends on the flux transformer and on the measurement position. For the latter, the position was taken where the field component has its maximum. The flux transformer was supposed to be a gradiometer consisting of two small loops that are equidistant and wound in opposite directions.

To estimate the frequency dependent effect of the layer of vernix caseosa, the volume conductor was described by three homogeneous and isotropic compartments. The Maxwell equations were used in a quasi-static approximation. In this approximation, the fields generated by a time-varying source contain terms up to and including the first power in the radian frequency ω. If the time varying fields are written in phasor form and the complex conductivity is used, the Poisson equations for the electrical potential and the magnetic fields and the appropriate boundary conditions are similar to the ones used in the case that the Maxwell equations are used in a static approximation. Hence, the electrical potential and the magnetic field for each frequency can be calculated in the same way as in the non-capacitive case [13]. The model is depicted in Fig. 3. Two confocal spheroids described the

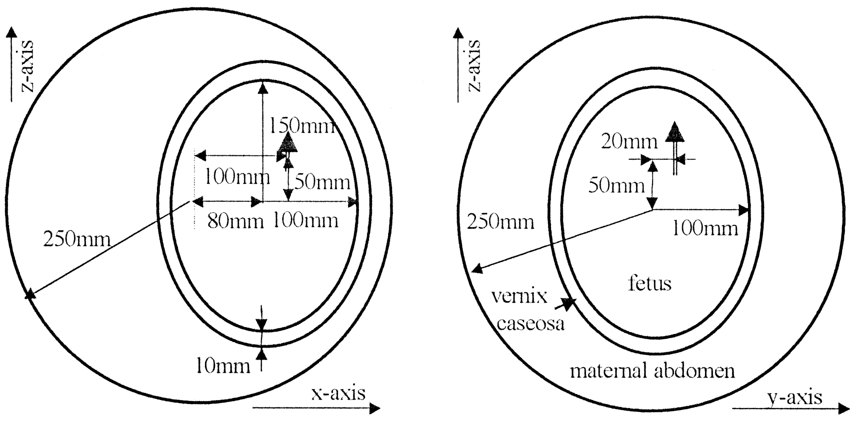

Fig. 3. The three-compartment model used to simulate the capacitive effect of the layer of vernix caseosa. The arrow depicts the current dipole acting as the source.

fetus covered by a layer of vernix caseosa. The outer compartment, representing the maternal abdomen, was a sphere with a radius of 250 mm. The compartments describing the fetus and maternal abdomen were given a conductivity of 0.22 S/m and a relative dielectric constant of 5×10^5. The source was a current dipole embedded in the fetal compartment, orientated along the z-axis. The dipole moment varied sinusoidally with time (radian frequency ω). The radial component of the magnetic field was calculated for various frequencies in the range of 0.1–1000 Hz by means of the BEM. Each surface was described by 638 triangles. The magnetic field was computed in 319 points on a sphere, which was concentric with the sphere describing the mother's abdomen. Because the BEM is known to have numerical problems with thin layers, both the dielectric constant and the thickness of the vernix caseosa layer were enhanced by a factor 5. For a spherical model, this did not influence the potential distribution at the outer surface.

The influence of gaps appearing in the layer of vernix caseosa, such as the mouth opening and the umbilical cord, on the fMCG was studied by means of the FEM and the application of the Biot–Savart law [14]. The model consisted of three spherical compartments, representing the fetus, the layer of vernix caseosa and the maternal abdomen. The spheres describing the fetus and the layer of vernix caseosa were concentric. The radii were respectively 150 and 160 mm. The sphere representing the maternal abdomen was eccentric, and had a radius of 300 mm. The distance between the centres of the inner two spheres and the outer sphere was 120 mm. Normalised conductivities of 1, 10^{-6} and 1 S/m were used for the fetus, the vernix caseosa, and the maternal abdomen, respectively. A cross-section of the model is pictured in Fig. 4. Two holes were modelled in the layer of vernix caseosa, at the top and bottom end of the vertical axis of the concentric spheres. A current dipole at the centre of the fetus compartment represented the fetal heart activity. The orientation of this dipole was varied in the xz-plane, see Fig. 4. The normal component of the magnetic field was

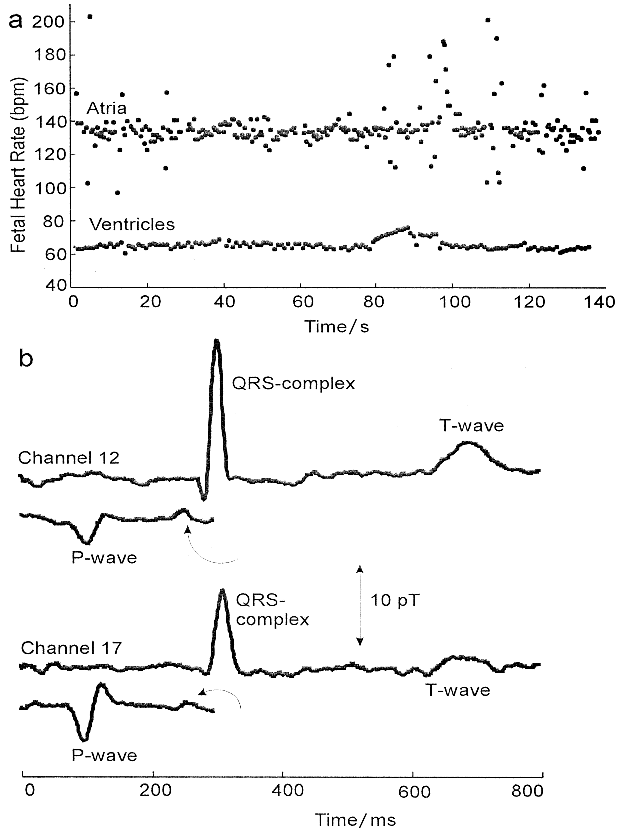

Fig. 5. fMCGs of a fetus with a complete AV-block. (a) Upper trace the fetal heart rate of the atria, lower trace the fetal heart rate of the ventricles; (b) averaged fMCG signals. Both signals were measured simultaneously. In the upper traces the R-wave was used as a trigger and 60 QRS-complexes were used to average. In the lower traces the P-wave was used as a trigger and 120 P-waves were used to average. Note the repolarisation of the atria in the lower traces.

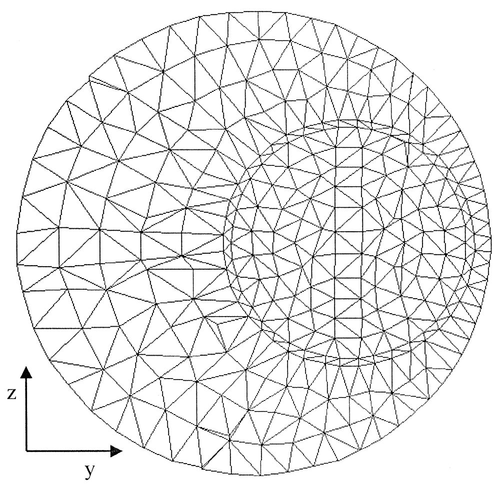

Fig. 4. A cross-section of the mesh used to simulate gaps in the vernix layer covering the fetus.

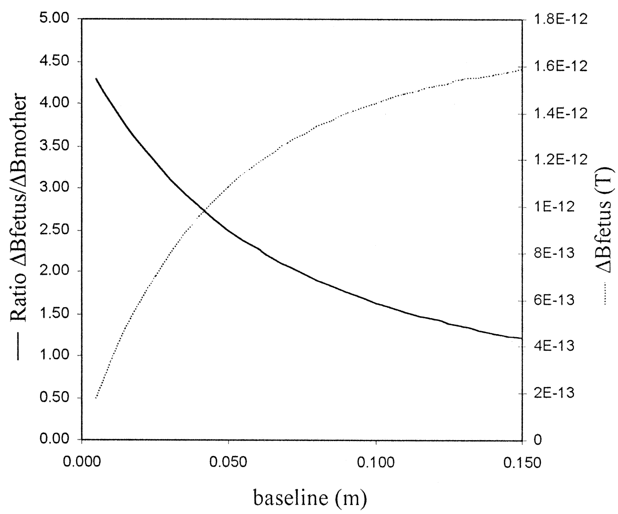

Fig. 6. The ratio of the contribution of the heart of the fetus and that of the mother as a function of the baseline of the (first-order) flux transformer. The sensitivity of the fetal heart is also shown.

computed in 76 points on the hemisphere indicated in the figure.

4. Results

4.1. Measurements

The fetal MCG was registered for twenty healthy women in various stages of pregnancy. It was always possible to obtain the fetal heart rate. The averaged fMCG showed P-waves, QRS-complexes, and T-waves. However, it turned out to be difficult to determine the onset and ending of the T-wave and the duration of the QT-interval. This may partly be explained by the fact that the averaging procedure used the QRS-complex as a trigger. If the RT-duration varies from beat-to-beat, this procedure will result in a lowering and spreading out of the T-wave. fMCGs were also measured with a 19-channel magnetometer system [15]. It was found that the onset of the T-waves in the various channels differed. The differences in time could be as large as 50 ms.

As example, averaged fMCGs of a patient with an arrhythmia are shown in Fig. 5. It was found that the PP- and RR-intervals were completely independent. The time instants of both the P-waves and those of the R-peaks could be determined reliably. Using these time instants separately we could average with two different triggers. Both types of averages are shown. When the P-waves were used as a trigger, the repolarisation of the atria can be seen in the averaged signal. The heart rate could be determined beat-to-beat for both the atria and the ventricles. It was found that the contracting rate of the ventricles was ± 70

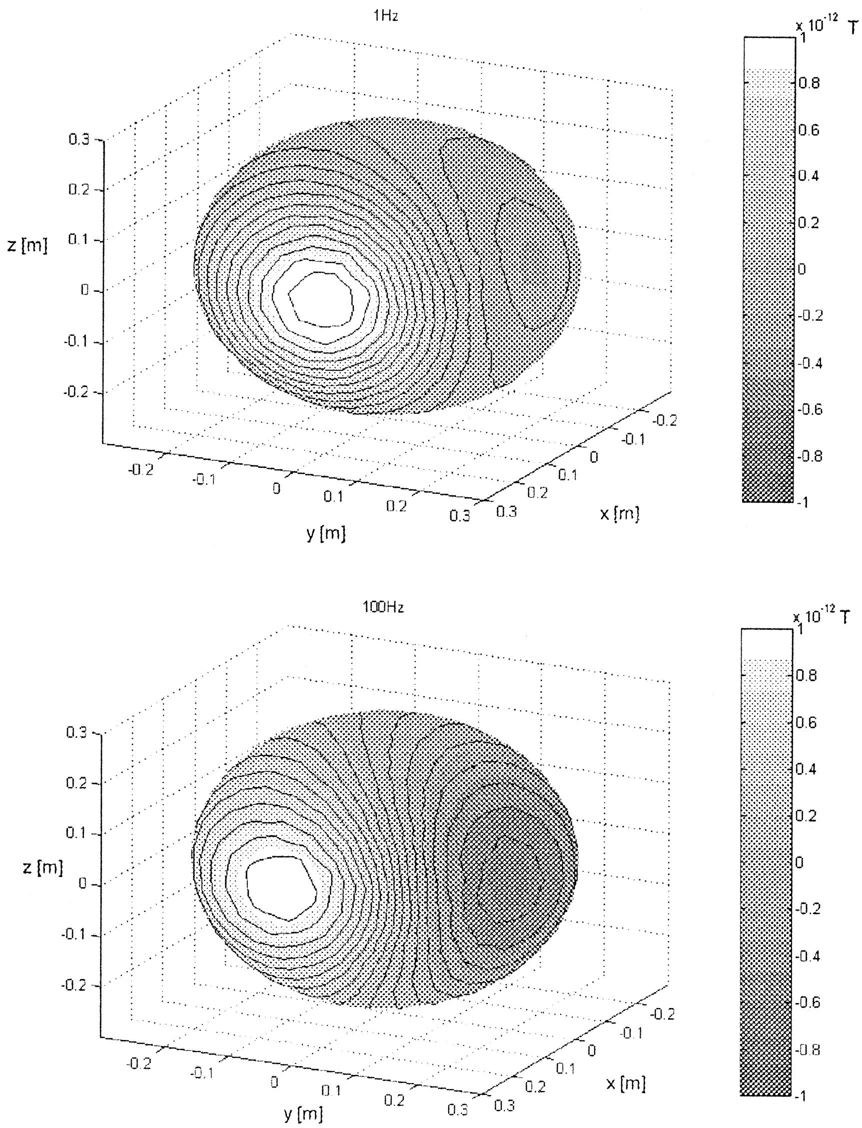

Fig. 7. The radial component of the magnetic field for a dipole varying sinusoidally in time for two frequencies. The isofunction plots shown were computed at the time instant when the strength of the dipole was maximal. The dipole was orientated along the axis of the spheroids depicted in Fig. 3.

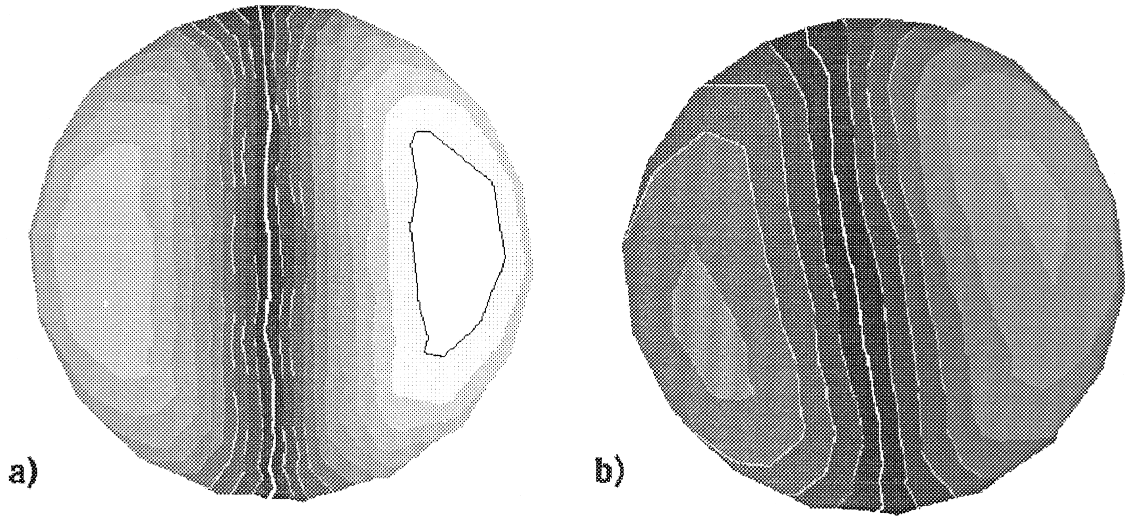

Fig. 8. The radial component of the magnetic field generated by a dipole in the centre of the model shown in Fig. 4. (a) The dipole points at an angle of 15° with the z-axis and (b) at an angle of 60°. The coding of the scale is the same in both plots.

and that of the atria ±140 beats per minute, indicating a complete AV block.

4.2. Calculations

The quotient of the contribution of the fetal heart and that of the mother was calculated, using the model depicted in Fig. 2. The result for the case that both the current dipoles were orientated in the negative y-direction is shown in Fig. 6. As shown, the ratio between both contributions decreases with an increasing baseline of the flux transformer. On the other hand, the sensitivity for the fetal heart increases with the baseline.

Simulations were carried out to study the frequency dependent effect of the vernix caseosa, which acts as a lossy capacitor [13]. If the frequency of the source is zero (i.e., a constant current source), this layer acts as a pure resistor and the currents at the other side of the layer are in phase with the currents in the fetal compartment. For high values of the frequency, the layer acts as a pure capacitor and in that case the currents in the maternal compartment will have a phase shift of 90° with respect to the phase of the source. The magnetic field observed outside the body is composed of contributions of the currents in all three compartments. The currents at both sides of the layer do not have the same phases and the same strengths for different frequencies. Thus, a frequency-dependent behaviour of the magnetic field has to be expected. The radial component of the magnetic field is calculated as a function of the frequency. As an example, the instantaneous value of the radial component is shown in Fig. 7 in two cases. In both cases, the isofunction maps shown are plotted for the time instant when the dipole strength reaches its maximum. In the first example, the current dipole oscillated at 1 Hz and in the second at 100 Hz. It was found that for a frequency of 1 Hz, the phase shift did not exceed an angle of 5°. In case of 100 Hz, the phase shift could reach up to an angle of 25°. The actual current dipole describing the heart activity is composed of signals with different frequencies. As can be deduced from the results shown in Fig. 7, the layer of vernix acts as a spatial temporal filter.

The distribution of the radial component of the magnetic field was calculated using the finite-element model shown in Fig. 4. The gaps in the layer of vernix were at the north and south poles of the sphere describing the fetus. If the dipole was pointing in the z-direction, the field was a factor 4 stronger than if it was pointing in the x-direction. This is because much more current escapes the fetus when the source is pointing towards a hole. For both a z- and an x-dipole, the zero line of the magnetic field pattern had the same orientation as the source. The pattern of the x-dipole was undistorted. However, the pattern for the z-dipole was stretched in the z-direction. An arbitrary dipole in the xz-plane can be decomposed into a dipole in the z-direction and one in the x-direction. The field is the superposition of the fields of both dipoles. In Fig. 8 two examples of these superimposed field patterns are shown. In the first case, the dipole was pointing in a direction at an angle of 15° with the line connecting the holes. In the second case, the dipole was pointing in a direction at an angle of 60° with the line connecting the holes. In both cases, the zero line of the pattern is not the same as the orientation of the source. This is explained by the fact that the field of the x-component of the source is attenuated by a factor 4.

5. Discussion

The diagnosis of cardiac arrhythmias can be obtained from ultrasound, fECG, and fMCG. Because of the low resolution, registration of the fetal heart rate with the first method can lead to a misreading in case of frequent fetal extra systole (i.e., irregular contractions of the heart). The failure rate of transabdominal fECG, which is reported to be approximately 30%, is mostly due to the large interference of the maternal ECG and the insulating and capacitive effects of the surrounding tissues [16]. With fMCG the identification and classification of arrhythmias is possible because the duration of the PR-interval, the QRS-complex and the RR-interval can reliably be deduced from fMCGs. Hence, fMCG can be useful in the diagnosis of fetal cardiac disorders, especially for the classification of congenital conducting disturbances and fetal arrhythmia [17]. It is much more complicated to deduce the QRS-complex, P- and T-waves, and the duration of the RT-interval.

Almost all biomagnetometer systems are designed to measure magnetoencephalograms or adult MCGs. For a good design of a magnetometer system that is to be used for fMCG measurements thought must be given to the number of channels, the signal processing, the flux transformer, the measurement positions, etc. First, the flux transformer should be designed such that the contribution from the fetal heart to the measured field component is large compared to that from the mother's heart. This asks for a flux transformer with a relatively short baseline, as can be concluded from the result given in Fig. 6. On the other hand, the magnetometer has to be sensitive to the contribution of the fetal heart. This requires a long baseline. In other words, a reasonable compromise has to be found. In order to decide what is reasonable, measurements using different types of flux transformers need to be carried out under identical conditions. Especially, the relation between the flux transformer and the strength of the T-wave has to be studied.

Simulations showed that the disappearance of the low frequency T-wave in the fECG might probably be ascribed to a capacitive effect [7]. The fMCG is generated by currents within the fetus as well as by currents in the maternal abdomen. Capacitive effects also disturb the fMCG, but this effect is much smaller than in fECG. The simulations presented above showed that the fMCG will also be distorted by holes in the layer of vernix caseosa, although less than the fECG.

The fact that the onset of the T-wave cannot be determined reliably from averaged fMCGs may be explained by the following. First, the duration of the RT-interval may vary and consequently the T-wave is lowered and spread out. Second, the flux transformer acts as a spatial filter and may have a larger cancelling effect on the T-wave than on the other waves. Third, the contribution of the mother's heart contaminates the fMCG. The maternal signal may have a duration of 700 ms, whereas the duration of the fetal cycle is usually at least a factor two smaller. Variation in the contribution of the maternal T-wave may obscure the T-waves of the fetus. Fourth, the volume conductor is also not constant in time. The electrical currents are modified by, for instance, cardiac, respiratory and gastric activity. Fifth, maybe the measurement position is not optimal to show the T-wave in the chosen component of the magnetic field. For an adult it is found that the amplitude of the T-wave varies considerably over the torso [18]. Values of QT-intervals in 15 patients at a gestational age of 27–42 weeks as reported by Quinn et al. [19] varied between 132 and 297 ms. More detailed model studies have to be carried out in order to conclude to what extent the T-wave in the fMCG is influenced by the tissues and whether its amplitude and the duration of the ST-interval can give medical information as it does in adult ECG recordings.

Fetal magnetocardiography is an elegant and non-invasive technique for studying the electrical activity of the fetal heart. Applications in prospective studies such as the classification of arrhythmias are promising. The results encourage further refinement of the fMCG technique. It is premature to draw general conclusions from the model studies, although it has been shown that the layer of vernix caseosa may effect the shape of the wave forms. Detailed studies and a more sophisticated signal processing procedure are needed to decide which features of the waveform of a fMCG are reliable and how to improve the fMCG.

References

[1] Y.E. Zhuravlev, D. Rassi, S.J. Emery, Clinical assessment of fetal magnetocardiography, in: C. Baumgartner et al. (Eds.), Biomagnetism: Fundamental Research and Clinical Applications, Elsevier, IOS Press, 1995, pp. 700–703.

[2] T.F. Oostendorp, Modeling the fetal ECG. PhD thesis, Catholic University of Nijmegen, the Netherlands, 1989.

[3] R. Plonsey, D.B. Heppner, Considerations of quasi-stationarity in electrophysiological systems, Bulletin of Mathematical Biophysics 29 (1967) 657–664.

[4] T.F. Oostendorp, A. van Oosterom, H.W. Jongsma, Electrical properties of tissues involved in the conduction of foetal ECG. Med. Biol. Eng. Comp. (1989) 322–324.

[5] Y. Seppenwoolde, S.P. van den Broek, M.J. Peters, Volume conduction effects: the fetal magnetocardiogram, Biomed. Tech. 42 (1) (1997) 261–264.

[6] Z. Dunajski, M.J. Peters, M, Development of the fetal magnetocardiogram from the 13th week of gestation onward, in: C. Baumgartner et al. (Eds.), Biomagnetism: Fundamental Research and Clinical Applications, IOS Press, 1995, pp. 704–708.

[7] D. Rassi, Y.E. Zhuravlev, M.J. Lewis, S.J. Emery, Vector fetal magnetocardiography, in: 10th Int. Conf. on Biomagnetism, Biomag '96, Santa Fe, USA, 16–21 February 1996, Abstracts, p. 237.

[8] T.F. Oostendorp, A. van Oosterom, Modelling the fetal magnetocardiogram, Clin. Phys. Physiol. 12A (1991) 15–18.

[9] J.A.F. Huirne, Y. Seppenwoolde, H.W.P. Quartero, M.J. Peters, Antenatal magnetocardiographic diagnosis of fetal arrhythmias, submitted to the Brit. J. Obstet. Gynaecol.

[10] A.P. Rijpma, Y. Seppenwoolde, H.J.M. ter Brake, M.J. Peters, H. Rogalla, Application of SQUID magnetometers in fetal magnetocardiography, in: H. Rogalla (Ed.), Applied superconductivity, Inst. Physics Conf. series, IOP, 1997, pp. 771–774.

[11] J. Borgmann, P. David, G. Ockenfuss, R. Otto, J. Schubert, W.

Zander, A.I. Braginski, Electronic high-temperature radio frequency superconducting quantum interference device gradiometers for unshielded environment, Rev. Sci. Instr. 68 (7) (1997) 2730–2734.

[12] J. Vrba, SQUID gradiometers in real environments, in: H. Weinstock (Ed.), SQUID Sensors: Fundamentals, Fabrication and Applications, NATO ASI series E, Vol. 239, 1996, pp. 117–178.

[13] J.G. Stinstra, M.J. Peters, The volume conductor may act as a temporal filter on ECG and EEG, Med. Biol. Eng. Comput. 36 (1998) 1–6.

[14] S.P. Van den Broek, H. Zhou, M.J. Peters, Computation of neuromagnetic fields using finite element method and Biot–Savart law, Med. Biol. Eng. Comput. 34 (1996) 21–26.

[15] H.J.M. ter Brake, J. Flokstra, W. Jaszczuk, R. Stammis, G.K. van Ancum, A. Martinez, H. Rogalla, The UT 19-channel DC SQUID based neuromagnetometer, Clin. Phys. Physiol. Meas. 12B (1990) 45–50.

[16] W.J. van Wijngaarden, D.K. James, E.M. Symonds, The fetal electrocardiogram, Bailliere's Clinical Obstetrics and Gynaecology 10 (1996) 273–294.

[17] S. Achenbach, W. Moshage, T. Menéndez, E. Beinder, A. Bittl, M. Hofbeck, A. Kollert, H. Singer, K. Bachmann, Investigation of fetal cardiac arrhythmias by magnetocardiography, Biomed. Tech. 42 (1) (1997) 253–256.

[18] H. Eskola, On the properties of vector magnetocardiographic leads, PhD thesis, Tampere University of Technology, Finland, 1983.

[19] A. Quinn, A.J. Weir, U. Shahani, R. Bain, P. Maas, G. Donaldson, Antenatal fetal magnetocardiography: a new method for fetal surveillance?, Brit. J. Obstet. Gynaecol. 101 (1994) 866–870.