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A magnetic induction measurement system for adult vital sign monitoring: evaluation of capacitive and inductive effects

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Abstract. Magnetic induction (MI) measurement technique could provide an unobtrusive contactless method for continuous monitoring of vital signs such as breathing and cardiac activity in bed. In this paper, we present a magnetic induction system to evaluate the feasibility of monitoring heart and lung activity and the preliminary measurement results. The excitation and detection coils are designed to be implemented in a single printed circuit board, allowing the use of the system in a bed with coils under the mattress. The electronic system is based on a 16 bit arbitrary waveform generator (PXI-5422, National instrument) operating at a sample rate of 200 MS/s for the excitation signal and the detected amplified signal is then, sampled at 100 MS/s by a 14 bit digitizer (PXI-5122, National Instruments). The preliminary results at 10 MHz show the ability of the system to detect solutions with different conductivities. However the capacitive effect is in the same order of magnitude as the inductive effect due to eddy currents. Safety of the system has been evaluated to be in accordance with the standards of human exposure to the magnetic fields.

1. Introduction

Unobtrusive methods for monitoring vital signs are of high importance in many medical issues. Home health care applications, monitoring neonates or burn victims and patients with sleep problems are some of those areas in which unobtrusive techniques are of particular interest. Magnetic induction (MI) could provide a method for continuous monitoring of vital signs such as breathing and cardiac activity in bed. The theory on which this method is based is that, when a conductive object is placed in a time varying magnetic field (B), eddy currents are induced in the body. These currents produce a secondary magnetic field that can be detected by a properly designed receiver system. The amplitude of the eddy current is proportional to the magnetic flux density and the conductivity of the material. Based on this theory, monitoring the changes of conductivity in the thorax produced by the inflation and deflation of the lungs and the blood circulation in the trunk, enable us to monitor these vital signs. Since the conductivities of biological tissues are very low, the secondary signals to be measured are very weak which makes the measurement process difficult. For a sample of material between an excitation coil and a sensing coil [1],

$$\Delta B/_{B} \propto \omega(\omega\varepsilon_0\varepsilon_r - j\sigma) \qquad (1)$$

where σ is the conductivity of the sample, ε_r is its relative permittivity, ε_0 is the permittivity of free space, ω is the angular frequency of the excitation and $j = \sqrt{-1}$. For biological tissues, the conductivity (imaginary) component will normally be dominant [2]. From (1) it is seen that the eddy currents produce a component of the secondary signal, which is proportional to σ and ω. The total detected field ($\Delta B + B$) lags the primary field by an angle φ [3]. The objective of the presented work is to evaluate the capacitive and inductive effects in a practical MI system to be used under the mattress for monitoring vital signs. In addition, since the system is based on the exposure of an electromagnetic field, evaluation of the specific absorption rate (SAR) is of interest to be in accordance with international standards.

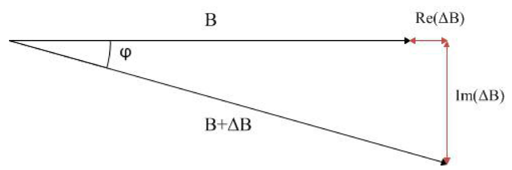

Figure 1. Phasor diagram of the primary B and secondary (ΔB) magnetic fields detected.

2. Materials and Method

2.1. SAR Simulation

As stated previously, safety is an important issue to be studied in the systems based on the exposure of the electromagnetic field. In accordance with ICNIRP recommendations [4], the maximum field strength is limited by the specific absorption ratio (SAR) which shall not exceed 2 W/Kg for head and torso. SAR simulation has been done using COMSOL Multiphysics by a simplified anatomical 3D model of a human trunk (figure 3).

2.2. Tank Simulation

As a reference for experiments, the experimental setup has been simulated using the same software. The model consists of a cylinder with a radius of 8 cm and a height of 40 cm (approximate volume of 8 liters). By using the parametric sweep, solutions with different conductivity have been simulated and the magnetic field was obtained for each of the solutions.

2.3. MI system

The system is based on a 16 bit arbitrary waveform generator (PXI-5422, National instrument) operating at a sample rate of 200 MS/s for the excitation signal. The signal detected at the detector coil was passed to a two stage low noise amplifier (AD8432) with a gain of 64. The amplified signal is then, sampled at 100 MS/s by a 14 bit digitizer (PXI-5122, National Instruments). The driving frequency for the measurements is fixed to 10 MHz. As a result of the very low conductivity of biological tissues, the secondary magnetic signals to be measured are very weak and since the secondary coil will receive the superposition of the primary and secondary fields, the dynamic range of the receiver channels would be very wide. Finding an efficient coil configuration is an important issue in the system design process in order to suppress the large primary magnetic field as much as possible. Up to now, several flux cancellation techniques have been proposed from which we plan to use a planar gradiometer to cancel the primary magnetic field [5]. In addition, it is known from the state of the art that capacitive coupling effects exist between the transmitting and receiving coils [6] and also between the object and the sensors [7]. Several methods were reported to minimize this effect. In our experiments the coils are two 8-turn coils with a diameter of 5 cm that could be placed in a common printed circuit board (PCB). The generator (PXI-5422) has been configured to produce a sine wave with a desired frequency and amplitude as excitation signal with an injected current of 22 mA. The signal at the detection coil is then passed through the AD8432 to be amplified to the input level of the digitizer. The amplifier is an AD8432 which is a dual channel, low power amplifier with an input voltage noise of 0.85 $nV\Big/\sqrt{Hz}$.

2.4. Measurement protocol
Six saline solutions, in plastic tanks of 8 liters with similar dimensions to the simulated model were made up with conductivities of 0.0005, 0.2, 1, 2, 5, 16 $[Sm^{-1}]$. The experiment steps are as follows; place the tank at a distance of 5 cm from the system, step back, wait for 3 seconds, step forward and touch the tanks for 2 seconds, step back, wait for 3 seconds more and finally take it out. For each experiment, 40 measurements have been done in 22 seconds.

3. Results
The estimation of the absorption rate at a range of frequencies at different distances of excitation coil from the body, shows that even considering the worst case for exposure to the magnetic field (a frequency of 10 MHz and minimum distance), the maximum SAR is orders of magnitude lower than the safety standards' limits (figure 2). That is, for an excitation current of 1 A in a one turn coil at a distance of 5 cm from the chest at a frequency of 10 MHz, we are more than one order of magnitude under the safety limits imposed by ICNIRP standard (Maximum calculated SAR=0.0056 W/Kg) [8]. Figure 4 shows the changes of the detected phase for all solutions following the measurement protocol. Figure 5 shows the increment of the phase angle with respect to the changes in conductivity of the solution, in experiments and simulations. Since the phase changes produced by distilled water have to be zero, this value was subtracted from all the measurements.

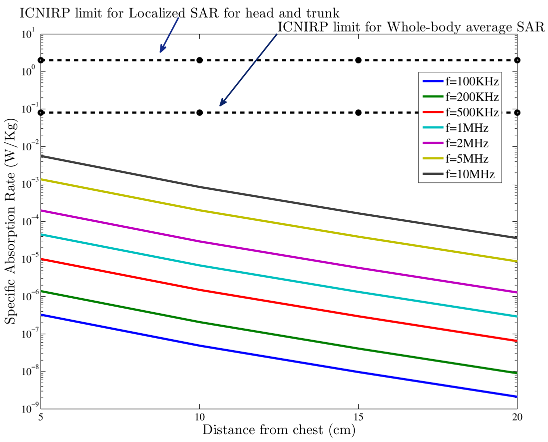

Figure 2. SAR vs. distance from chest

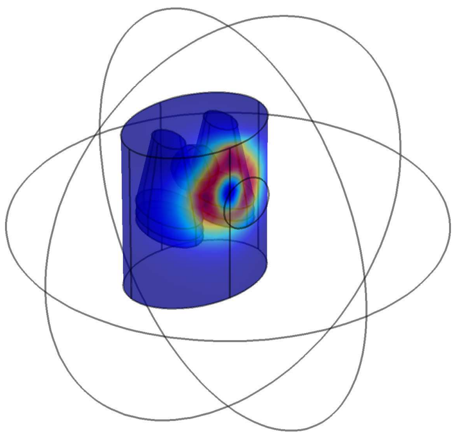

Figure 3. SAR simulation at 10 MHz

4. Discussion and Conclusions
Equation (1) predicts that the imaginary component of the secondary magnetic field (ΔB) must be proportional to σ. The equation is valid when the skin depth of the electromagnetic field is much larger than the thickness of the sample. As the conductivity increased, the skin depth became comparable to the diameter of the tank, i.e. for a conductivity of 16 $[Sm^{-1}]$ the skin depth is 3 cm in comparison with 16 cm of tank diameter. The simulation shows that the skin depth affects the measurement results as expected and it can be seen in figure 5 that, there is a change in the slope for greater values of σ.
Moreover, figure 5 shows that we have an overestimation of the phase angle. When the operator

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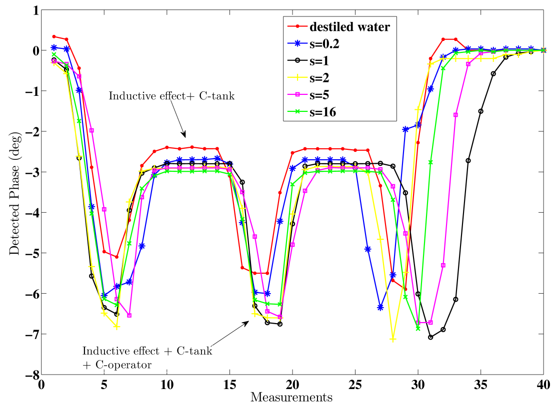

Figure 4. Phase angle of the detected signal

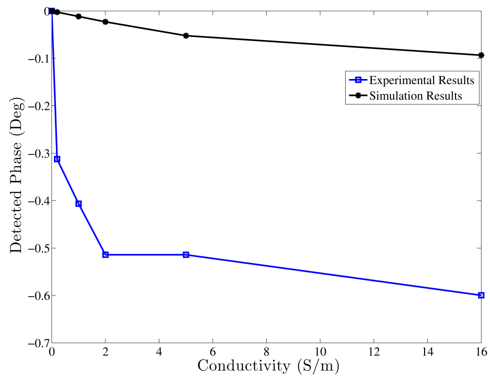

Figure 5.
Phase angle increment due to changes in σ

is touching the tank, a phase change -from free space situation- of about 6° can be seen in the graphs (figure 4) while without touching, the maximum phase change is about 3°. Although the phase was adjusted to the empty space case, these phase changes could still not be explained only by magnetic induction effects, and the capacitive coupling must play an important role that is in accordance with [3] and [6]. There are also measurement errors due to misplacement of the tank over the sensor. Experimentally, we obtained that a misplacement of 1 cm, in the same axis as cylinder's axis, produces a change of about 0.29° and a misplacement of 1 cm, in the axis perpendicular to the cylinder's axis, produces an approximate phase shift of 0.047°. These phase changes could also be explained by capacitive coupling. As capacitive effect is mainly related to the distance between the object and the coils, and magnetic induction is more sensitive to the changes in conductivity, we want to evaluate the possibility of using both magnetic and capacitive effects to monitor vital signs. The advantage of monitoring physiological signals is that they are rhythmic and the artifacts and unwanted static signals, produced by environmental factors, could be suppressed by signal processing techniques. Future studies will be focused on the design of the coils to optimize the balance between capacitive and magnetic effects. Since it is necessary to use a gradiometer as a field cancellation method to suppress the large primary magnetic field and increase the sensitivity of the magnetic induction signal, we will study different topologies of gradiometers and coils suitable for being implemented on a PCB.

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