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Simulation of MRI RF Coil

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Abstract

In this paper, four-element coil operating at 7T is simulated. It can be used as a volume coil for reception. MRI imaging at ultra-high field strength requires frequencies in the ultra-high frequency range. The resonant frequency of each element was tuned at 300 MHz (the proton (^{1}H) Larmor frequency of the 7T magnet). This study showed that the intensity of the magnetic field is higher in the region where the elements were closest to the phantom. The one element coil is useful for detecting signal near the surface of the region of interest. It was noted that a surface coil covers a small area that results in a small field of view (FOV). Simulations have shown that, adding more than one coil increased the field of view. A full image is acquired using a volume coil and it almost completely covered the phantom region, compared to a single coil of the same dimensions and, hence, excellent magnetic fields were achieved.

المستخلص

تم في هذه الورقه محاكاة ملف من أربع عناصر عند 7 تسلا والذى يمكن إستعماله كملف حجمي يستقبل الإشارة. يتطلب التصوير بواسطة الرنين المغناطيسي مجال عال. لقد تم ضبط تردد الرنين عند 300 ميغا هرتز (تردد لارمور للبرتون ^{1}H عند 7 تسلا). أظهرت هذه الدراسة أن شدة المجال المغناطيسى تكون عالية في المنطقة التى تكون فيها العناصر أقرب ما يمكن للفانتوم. لوحظ أن ملف واحد يغطى منطقة صغيرة مما أنتج مجال رؤية (FOV) صغير. وقد أظهرت المحاكاة أن زيادة أكثر من ملف زاد من مجال الرؤية. لقد تم الحصول على صورة كاملة باستعمال ملف حجمي والتى تكاد تغطى منطقة الفانتوم مقارنة بالملف الفردي الذي له نفس الإبعاد كما تم الحصول على مجال معناطيسي ممتاز.

Key words: B_1 field; coupling; field of view.

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Introduction

Two options exist for imaging a large region with surface coils: (a) use of a single large coil; (b) use of an array coil of relatively small coils, known as the phase array coil. In recent years, the study of the interaction of radiofrequency fields with human bodies has become a lively topic. The finite difference time domain (FDTD) approach is rapidly becoming one of the most widely used computational methods in electromagnetic studies. The FDTD is capable of calculating electromagnetic interaction for problem geometries that are very difficult to analyze by other methods. The FDTD method was used to simulate the time varying magnetic and electric fields. This method is used to solve Maxwell's curl equations [2]:

$$\frac{\partial \mathbf{B}}{\partial \mathrm{t}} + \nabla \times \ \mathbf{E} = 0, \qquad (1)$$

$$\frac{\partial \mathbf{D}}{\partial \mathrm{t}} - \nabla \times \ \mathbf{H} = \mathrm{J}, \qquad (2)$$

$$\mathbf{B} = \mu \mathbf{H}, \qquad (3)$$

$$\mathbf{D} = \varepsilon \ \mathbf{E}, \qquad (4)$$

Where μ is the permeability, ε is the permittivity and σ is the conductivity of the medium. We used MATLAB 7.10 (R2010a) to calculate the RF magnetic B_1^- field and generate the images in the *x*-*y* plane by using the following equation [3]:

$$B_1^- = \frac{1}{2}(B_x - i\ B_y) \qquad (5)$$

The RF coil is the main device in the MRI scanner. It has two main functions; excitation of nuclear spins and detection of the signals. The transmitter generates RF pulses with suitable frequencies, amplitude and phase in order to excite nuclei**.** The coil can be divided into three types: 1) transmit and receive, 2) transmit only and 3) receive only [4]. Using high field such as 7T faces technical challenges in obtaining good RF penetration and a large field of view. The problem of the limited FOV of surface coil can be solved by using many coils known in an array [5]. The coil needs to be constructed very carefully so that the individual coils should not interact with each other. A volume probe was modelled to be placed inside a tissue of interest to fit such that the size of the probe produces a transverse field that optimally covers the region of interest. In this study, four elements were modelled as a volume shape. The images can be generated from the individual coil and then they will combine together to generate an image with large FOV.

Method

RF coil suitable for multichannel receiver operating at 300 MHz has been simulated. The optimization of the RF magnetic resonance imaging (MRI) coil requires studying high frequency effects such as multiple coils interaction. Four element geometries were considered with the same size. All four coils had identical dimensions (12 cm × 12 cm). The coils were loaded with the muscle tissue phantom filled with a dielectric of human tissue. The distance between the surface of each element and the surface of phantom being 8.39 *mm*. A homogenous phantom was preferred because the character of the electromagnetic field can be observed without perturbation by tissue heterogeneities and was placed between the four elements (Fig. 1). The phantom had properties (conductivity and relative permittivity) of human muscle tissue. Dielectric properties were computed at 300 MHz, which was reported by the Italian National Research Council [6].

Each resonate element of the coil was made of a Perfect Electric Conductor (PEC). The PECs were split into many parts with small gaps for the capacitors which were distributed at four equally spaced intervals around the square loop. Discrete capacitors utilized in this coil network should be chosen carefully in order to acquire the best performance from the coil. This method has the advantage of reducing the electrical coupling (parasitic capacitance) that occurs between the coil and the dielectric of the sample [7]. This also reduces the electric field generated by the coil and frequency shifts between the unloaded and loaded coils are minimized. Therefore, the sensitivity of the receiver should be improved. Moreover, the RF feeding is usually done by connecting the main coil with the source. In this study, a small search loop was modelled in the simulation by driving the coil with a Gaussian pulse to determine the resonance frequency of each element. It was noted that there is no variation in resonance frequency from element to element because the four elements have identical element geometries.

The volume coil should be tuned to the same frequency, as when using only one element. The coupling between the two resonators can be characterized by the coupling coefficient k. A coupling coefficient k with value between 0 and 1 is indicator of flux linkage between the two coils. This means that the mutual inductance of the elements would have been diminished. For MRI coil can be approximated as [5,8]:

$$k = L_{12}/(L_1 L_2)^{1/2} \tag{6}$$

Where the mutual inductance between the two coils, L_{12}, can be computed as:

$$L_{12} = \int B_{1,coil1} . B_{1,coil2} dV \tag{7}$$

If significant resonant splitting is observed between the two elements, i.e. they cannot successfully be tuned to the same frequency. Better decoupling is necessary and small adjustment in the coil geometry may be adequate. But, other approaches can be used such as adding decoupling capacitors, adding inductors between the elements or connecting each element to low input impedance pre-amplifier. An important problem is how to generate a current in each coil. One way is to use the loop coil that produces electromagnetic fields to propagate inside the phantoms. A small search loop coil was used in this study to excite four elements as shown in Figure 2.

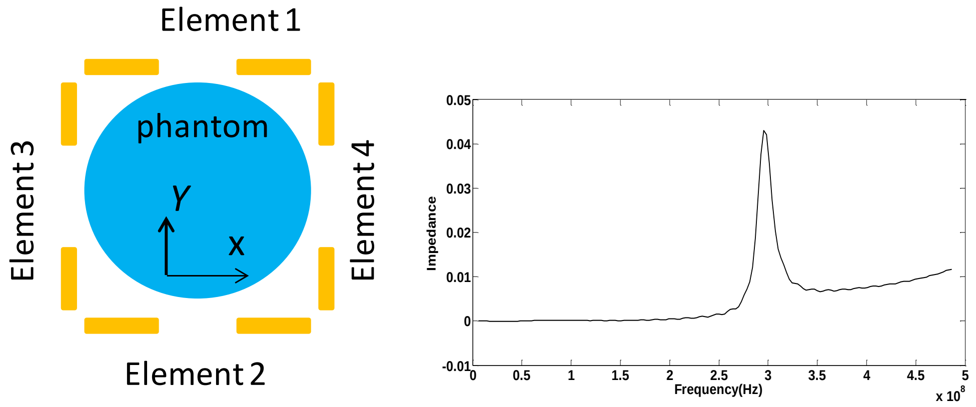

Figure 1. Schematic representation of the four-element coil (left) tuned approximately at 300 MHz (right).

The net input power driving used to generate the electric and magnetic fields was about 2.07×10^{-2} W. The loop transfers electrical energy to the receiver. It becomes an equivalent to the signals that are detected from the spins to induce *emf* in each element. Faraday's law provides the *emf* induced in one circuit by current I flowing in the second:

$$\xi_1 = L_{12} \frac{dI_2}{dt} \tag{8}$$

Where L_{12} is the mutual inductance between the circuits.

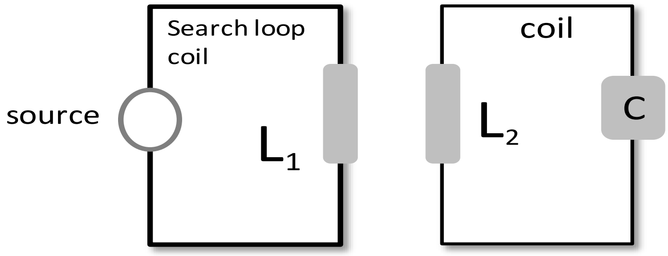

Figure 2. Circuit diagram representing the search loop coil.

This voltage has the same frequency as the Larmor frequency in each element when all elements are tuned at the same frequency (Fig. 1). Two comparison studies were performed as follows: the B_1^- generated from an individual element was calculated by adding two elements together (combining the element 1 and element 2 after that combining element 3 and element 4). Using this method means that the probe is decoupled numerically. Coupling effect is due to currents being induced between the probes and changing the *emf* induced field. In order to decouple elements in the simulation, the first element is tuned and the second element acted as open circuit. High impedance generated in the second element, prevented the current flow in the element. Furthermore, it is possible to tune each element at the proton Larmor frequency and then calculate the negative transverse magnetic fields.

Results and Discussion

Calculations were performed at 300 MHz with current source amplitude equal to 1 Amps. Fields are generated within each element due to *emf* and currents induced in the elements by the RF signals emitted from the phantom. The B_1^- reached a peak of 0.309 μT close to the surface of the receiver and then the B_1^- dropped away and reached a minimum value of 0.063 μT close to the second element. This comparison indicates that the magnetic field was stronger near the first element, but weaker near the second element and vice versa as shown in figures 3 and 4.

This study has shown that it is possible to simulate a volume probe consisting of four-elements. We simulated a four-channel non overlapping probe useful for imaging at 7T. The result is shown in figure 1. Adding capacitors as lumped elements, the coils would resonate at specific frequency. Therefore, each element is modelled independently. This technique should lead to accuracy, because the mutual inductance between the elements is reduced. It was found that this method is the key to improve the FOV. The image intensity varied as a function of position of the elements even for a homogeneous phantom. It can be seen that when adding four elements, the image FOV was increased and completely covered the entire phantom. Simulation results showed that an increase in the volume of homogenous B_1^- can be obtained by adding four-elements loaded with a cylindrical phantom.

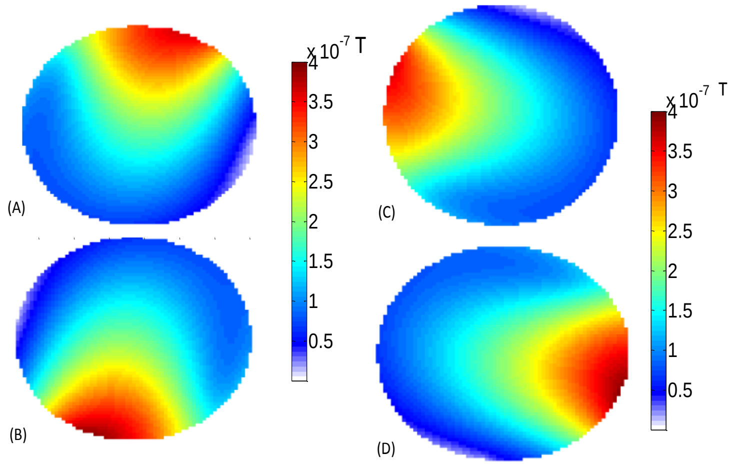

Figure 3. The B_1^- mapping generated: A) the first element, B) second element, C) third element and D) fourth element. The images were obtained from a homogeneous cylinder phantom (10 cm in diameter and 8 cm in length).

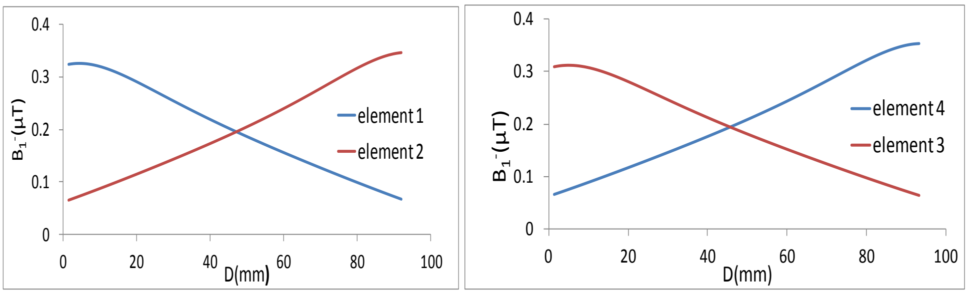

Figure 4. Line profiles across the B_1^- mapping, computed for comparison at the same line generated by the four elements.

All line profiles were computed numerically in the same depth, so the numerical method presented in this study offers a simple approach to make a comparison between the coils. Therefore, the line profiles generated by the four elements appeared approximately identical as shown in figure 4. The four elements model could be used as a fully decoupled pair of elements for multi-channel data acquisition. Accordingly, during reception each element detects the magnetic field produced by processing sample magnetisation as it relaxes back to the equilibrium state. Therefore, each coil will function independently and detect signals from a single region; it is bright close to each element.

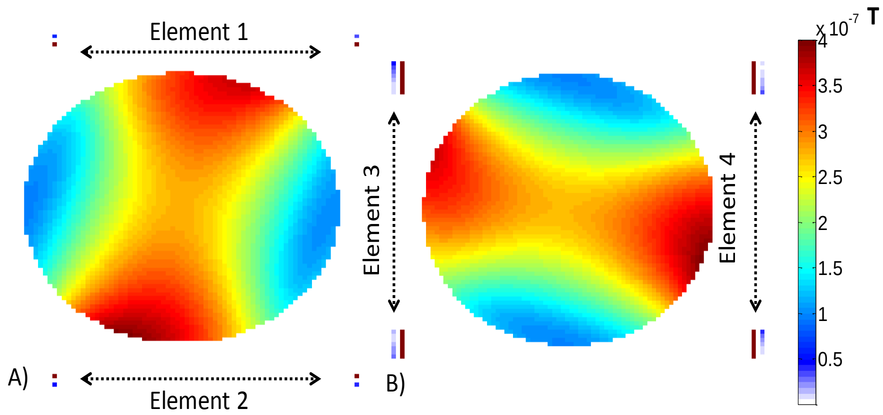

Figure 5. The B_1^- mapping generated by combining two elements: A) element1 and element 2 and B) element 3 and element 4.

Conclusion

It can be concluded that we can now compare the results of the simulations. A one element coil detects signals only in very close proximity to the sample while increasing the elements, the coil can pick up signals from a wider area surrounding the region of interest (ROI). Each element coil effectively images only a small region within the field of view. Individual images generated by an array of coils can be combined together to obtain images of a large area with SNR higher than that of a single large element coil. This probe is the type that can be used in magnetic resonance imaging.

References

[1] Ray, F.L., Giaquinto, O. and Hardy, C. (2002). Coupling and decoupling theory and its application to the MRI phased array, Magnetic Resonance in Medicine. **48**, 203-213.

[2] Yee, K. S. (1966). Numerical solution of initial boundary value problems involving Maxwell's equations in isotropic media, IEEE Transactions on Antennas and Propagation. **14**, 302-307.
[3] Christopher, M. C. and Smith, M. (2001). Calculations of B_1 distribution, SNR, and SAR for a surface coil adjacent to an anatomically-accurate human body model, Magnetic Resonance in Medicine. **45**, 692-699.
[4] Donald, W M., Elizabeth, A. M., Graves, J. G. and Martin, R. P. (2007). MRI from Picture to Proton, 2rd edition, Cambridge, 168p.
[5] Roemer, P. B., Edelstein, W. A., Hayes, C. E., Souza, P.S. and Mueller O.M. (1990). The NMR phased array, Magnetic Resonance in Medicine. **16**, 192-225.
[6] Gabriel, C., Gabriel, S. and Corthout, E. (1996). The dielectric properties of biological tissues, Phys. Med. Biol. **41**, 2231-2249.
[7] Kwok, W.E., and Seo, G. (2003). A four element phased array coil for high resolution and parallel MR imaging of the knee, Magnetic Resonance imaging. **21**, 961-967.
[8] Wei, J., Yuan, J., and Shen, G. X. (2006). Effect of tuning capacitor placement on mutual coupling for MRI array coils, Magnetic Resonance Engineering. **29B**, 50-54.