Head Coil for 10.5 Tesla Magnetic Resonance Imaging Human Body Scanner

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Abstract—The use of ultra-high magnetic fields (more than 7 Tesla) in magnetic resonance imaging (MRI) can provide a much higher signal-to-noise ratio (SNR) compared to commonly used 3T clinical MRI scanners. It will enable the in-vivo imaging of human anatomical details that have never been seen before with other imaging techniques. However, conventional radiofrequency (RF) coils are no longer suitable for providing homogeneous magnetic field (H) distributions. Homogeneity of the magnetic field distribution is an essential property of MRI coils. The uniform H-field distribution of the coils is important for accurate description of the interaction of the electromagnetic waves with the human body. Therefore, with the increase of the main magnetic field, novel RF technology and coil designs are needed. In this paper, we present a new human head coil design for 10.5 Tesla MRI scanner. This novel design provides homogeneous coil sensitivities in the area of interest according to the simulation results. First measurements of the fabricated prototype demonstrated strong resonance at the desired frequency.

Keywords - head coil, 10.5T human body MRI

I. INTRODUCTION

The first 10.5 Tesla magnetic resonance imaging (MRI) human scanner was first introduced by the Center for Magnetic Resonance Research in the University of Minnesota in 2014 [1]. This ultra-high (UH) magnetic field will provide a much higher signal to noise ratio (SNR) in comparison with the commonly used 3T clinical MRI scanners [2], [3], [4]. The increased resolution will enable the in-vivo imaging of human anatomical details that had never been seen before with any other imaging techniques [5]. The examination of dynamic and functional activities, e.g. brain activity, blood-oxygen-level dependent contrast imaging (BOLD) etc. will be possible with the help of UH magnetic field strength MRI scanners [6].

However, as it was in the case of the 7T and 9.4T MRI scanners before, magnet technology has gone ahead of radio-frequency (RF) technology in the evolution of MRI. Now we have created even higher magnetic field strengths, but do not have the ability to use it. With the increased strength of the main magnetic field (B₀), the transmitting and receiving RF coils have to work at a higher Larmor frequency. It means that the wavelength (\( \lambda \)) decreases as the Larmor frequency increases. When the circumference of a coil becomes electrically large (\( \geq \lambda /20 \)), the efficiency of the coil decreases notably [7]. Conventional RF coils (with the existing configurations and physical dimensions) are no longer suitable for providing homogeneous magnetic field (H) distributions (coil sensitivities). As a result, the coil can only produce a relatively weak magnetic field in certain regions. The lack of homogeneity in RF coils sensitivities significantly changes the performance of the surface coils. Therefore, when the B₀ magnetic field increases, novel RF technology, coil designs, methods and techniques are necessary. This paper presents the design of a human head coil, which gives a promise to solve the aforementioned problem.

II. DESIGN PROBLEM AND CHALLENGES

RF transmit/receive surface coils used in conventional clinical 3T MRI scanners are typically solid loop-like antennas. They are designed to operate at frequency of 128 MHz and produce strong and homogeneous magnetic field (H-field) in the area within the loop. Homogeneity of the H-field is a crucial factor in accurate description of the interaction of the electromagnetic waves with the human body and it affects the image quality significantly. However, high-field 10.5T MRI scanner mentioned in the beginning requires a coil that operates at 450 MHz. At such high frequencies traditional designs of the antenna do not work. The perimeter of the coil becomes electrically large and the phase change due to the finite propagation velocity of the signals on the antenna becomes significant and nulls the current on the loop. As a result, the H-field produced by the coil is non-homogeneous and, thus, this classical type of surface coil design cannot be applicable for higher field MRI scanners. The challenge of RF coil design is to make sure that the current is of equal magnitude and in-phase along the coil circumference, in order to produce a strong and uniform magnetic field distribution in the interrogation region [8], [9].

III. LITERATURE REVIEW

In order to solve the problem, we made a literature review on the ideas implemented for radio-frequency identification (RFID). RFID is a technology that provides wireless identification and tracking capability [10], [11]. The data transmission between the reader antenna and the tag is done by means of inductive (magnetic) coupling. RFID technology has a similar challenge, design a reader antenna...
with a large interrogation zone, i.e. producing a strong and uniform magnetic field in the region around antenna.

Ultra-high frequency (UHF) RFID systems typically work at the range of 840-960 MHz. This frequency is still higher than the desired Larmor frequency for MRI application, however, some of the ideas from the RFID field can be applied to solve our problem.

There are reports demonstrating electrically large loop antennas that are able to generate strong and even magnetic field. Particularly, segmentation of loop antenna is found to enable the design of such antennas for RFID applications [9], [12], [13], [14]. These broken loop antennas can demonstrate a desirable performance at UHF.

The studies above inspired us for designing surface coils for the emerging technology of UH magnetic field strength MRI.

IV. COIL CONFIGURATION

In order to study and justify the proposed MRI surface coil, its performance was analyzed together with the performance of a conventional one at 450 MHz. The inner area of the coils was kept the same in the both cases for the consistency.

A. The conventional coil design

A conventional surface coil is simply a copper loop made either of a cooper tape or printed on a printed circuit board (PCB). A rectangular loop coil was modeled using four copper tape stripes connected together and a matching circuit. Figure 1 (a) shows a schematic view of the traditional coil design. The inner area of the loop is 140x115mm, the width of a copper tape stripe is 5mm, and the thickness of the copper tape is 0.035mm.

B. The proposed coil design

The scheme of the proposed coil printed on a PCB is shown in Figure 1 (b) - (d). The coil comprises two loops (inner and outer) made of multiple line segments and a matching circuit. The coil is symmetrically structured with respect to the centerline (vector parallel to y-axis) and all the elements are located on the same side of a substrate sheet (dielectric thickness h=0.1mm, copper thickness t=0.0175mm, and the relative dielectric constant \( \varepsilon_r = 3.8 \)). The main geometrical parameters of the coil are the following

- Lin and Lout, the lengths of the inner and outer line segments, respectively,
- s, the gap between the inner/outer series line segments,
- sm, the separation between the inner/outer parallel lines, and
- W, the width of the lines.

Those parameters are found to be the most crucial ones, affecting the performance of the coil significantly. The impedance matching between the coil and the 50Ohm transmission line is achieved using an impedance matching circuit, which consists of tuning and matching lumped elements.

The line segments provide a very small phase delay between the neighboring sections so that the current flowing along the coil is kept in the same direction. In other words, the current distribution looks in phase. Thus, the segmented loop coil can produce a uniform magnetic field H distribution. It should be noted that even though the coil physical size is still electrically large, this segmentation help to “reduce” the effective electrical size and allows the coil to create a homogeneous H-field distribution at high Larmor frequency.

V. SIMULATION RESULTS

The conventional and the proposed coils were designed with the help of CST Studio Suit software [15]. Full 3D electro-magnetic simulations were performed for both cases using frequency and time domain solvers of CST.

The black curve at Figure 2 shows the results of the simulation for the conventional coil across the frequency range of 300-500MHz. The coil resonates barely at 450MHz with the attenuation of about -10dB. Compare this with the results for the proposed coil shown at the same figure with red color. The resonance is profound at 450MHz with the attenuation of about -35dB.

Figure 3 demonstrates a comparison in H-field distributions at 30mm away from the conventional design coil and the proposed one. The homogeneity of the field is much better in the case of the proposed coil.
Figure 4 demonstrates a comparison of current distribution on the coil’s surface for the conventional and the proposed designs. While the current flow in the (a) is not equal along the loop, the current flow on the surface of the segmented coil (b) is of equal amplitude along the coil’s circumference. This shows the importance of making the current flow equal and in-phase along the loop in order to produce a uniform magnetic field.

VI. MEASUREMENT RESULTS

Eight identical pieces of the proposed coil were further fabricated on a flexible PCB laminate sheet (ROGERS RO430B [16]). These coils were symmetrically arranged on the inner surface of a Plexiglas cylinder of 250mm outer diameter in order to construct an eight-port surface head coil. The overlapping is used in order to cover a larger area of interest inside the cylinder (imaging area). A cooper shield was placed on the outer surface of the Plexiglas cylinder in order to prevent the interaction of the coil array with the surrounding objects and to reduce the noise. After the arrangement all eight coils were tuned and matched separately to resonate at 450MHz. A two-port Rohde&Schwarz ZVH8 Cable and Antenna Analyzer [17] was used to measure the S-parameters of the neighboring coils.

Figure 5 and Figure 6 demonstrate the performance of one of the pairs of the neighboring coils across the frequency range of 400-500MHz. The resonance at 450 MHz is profound, with attenuation of less than -27dB. The coupling effect between the neighboring coils (S21 and S12) is sufficiently small of less than -16dB. That ensures the independence of the coils.

VII. DISCUSSION

According to the simulation results, the proposed coil has a strong resonance at the desired Larmor frequency. The segmentation of a solid loop and the addition of a double loop help to reduce the electrical size of the coil and allow fair operation at high Larmor frequency. The performance was verified and proved on a fabricated prototype coil array.
The field distributions seen in the simulations results give a promise for the even H-field distribution in the area above the coils. However, the actual H-field distributions can be obtained only in the scanner. Thus, the further step is to test the head coil in the 10.5T scanner at the University of Minnesota.

VIII. CONCLUSIONS

It is challenging to design a surface coil for UH magnetic field MRI. The main consideration is to make sure that the current along the coil’s circumference is equal in magnitude and in phase, in order to produce a strong and uniform magnetic field distribution. The proposed segmented loop surface coil has demonstrated the capability of performing at high Larmor resonance frequency and producing a relatively homogeneous H-field distribution.

REFERENCES


Figure 3. Scattering parameters (a) S21 and (b) S12 measured for two neighboring coils.

Figure 4. Scattering parameters (a) S11 and (b) S22 measured for two neighboring coils.