Comparison of different wireless coils for 1.5 T bilateral breast MRI

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Abstract.
This work compares three types of transceive wireless coils: Helmholtz-type coil, metamaterial-inspired coil, and quadrature coil based on their combination. Each transceive coil is electromagnetically coupled to a body "birdcage" type coil of a 1.5 T magnetic resonance scanner and improves bilateral breast imaging performance. While Helmholtz-type coil and metamaterial-inspired coil based on coupled split loop resonators are linearly polarized, their combination allows to couple with both linear components of the radiofrequency magnetic field, providing a more significant effect of a local boosting of the body coil’s transmit efficiency and radiofrequency safety in comparison with birdcage coil only.

1. Introduction
Magnetic resonance imaging (MRI) is becoming an increasingly common and affordable method. The main advantages of this method are safety, noninvasiveness, and the diagnostic value of the examination. The most common systems in clinics with a field strength of 1.5 T generally use the transceive radiofrequency (RF) body coil built into the magnetic resonance (MR) scanner to excite electromagnetic signal and dedicated local coils to receive MR signal from the area of interest. However, this approach does not fully match the modern requirements for the quality of the obtained MR images for specific tasks, e.g., for early diagnosis of breast cancer. Therefore, specialized MRI of individual organs is now gaining popularity when the signal is transmitted and received only in a certain area where the human organ of interest is located. Nevertheless, mostly local transceive coils for head and extremities imaging are developed and used at clinical field strength (i.e., 1.5 and 3 T). These coils have proven themselves to investigate fine structures, i.e., knee cartilage; however, their design restricts their application for other parts of the body. Moreover, its construction includes RF cables positioned close to the patient and can potentially breach the RF safety of the MRI study during high-current transmission mode.

Recently, the targeted MRI concept was proposed [1]. This concept includes a passive focusing of the body coil’s RF magnetic field to maximize its efficiency for small areas using the resonant structure...
located around the area of interest. In our previous work [2] it was demonstrated that the metamaterial-inspired resonant structure could be realized by using a system of coupled split-loop resonators, so-called metasolenoid coil (M-coil) [3], that could improve the transmit and receive efficiency of the 1.5 T body birdcage coil more than 4 times. However, this wireless coil supports linear polarization only, interacting only with the $H_y$-component of the transverse RF magnetic field produced by the body coil, which may limit its efficiency by a factor of $\sqrt{2}$ [4].

In this work, we propose a new quadrature bilateral breast coil design based on the combination of a Helmholtz pair and metasolenoid pair. Also, via electromagnetic simulation with a realistic female voxel model, we compare its performance in terms of transmit efficiency and RF safety with the linearly polarized Helmholtz pair and metasolenoid pair, respectively.

2. Methods

All electromagnetic simulations were performed in CST Studio Suite 2020. Fig. 1(a) demonstrates designs of the considered wireless coils: Helmholtz pair (HH-coil), metasolenoid pair (MM-coil), and quadrature coil based on their combination - HHMM-coil. Each Helmholtz coil consists of connected pair of three turns of copper wire forming three loops with the following dimensions: $124 \times 105 \text{ mm}^2$, $114 \times 95 \text{ mm}^2$, and $104 \times 85 \text{ mm}^2$. A distance of 134 mm separates two pairs of turns. These dimensions lead to tuning an HH-coil to 63.68 MHz - a Larmor frequency of 1.5 T MR scanner. Each of the coils of the MM-system consisted of 10 split-loop resonators containing two parallel telescopic brass tubes (to perform resonant frequency fine-tuning to a specific load) connected by two PCB capacitors at their ends. The capacitors are implemented as the copper strips printed on the top and bottom layers of the dielectric substrate (Rogers RO4250B material) with $\varepsilon = 3.5$, $\tan \delta = 0.0004$ at 64 MHz with the following dimensions $164 \times 114 \times 1 \text{ mm}^3$. The outer strips dimensions are $9 \times 160 \text{ mm}^2$ and the inner ones $9 \times 58 \text{ mm}^2$. These dimensions were also chosen to tune MM-coil’s fundamental eigenmode with homogeneous field distribution to the Larmor frequency. A quadrature HHMM-coil was implemented as MM-coil and HH-coil combination, which do not electrically connect. Each of the four wireless coils was tuned to the same resonant frequency.

Due to the electromagnetic coupling between two resonant structures in each pair, these systems have several eigenmodes in the operating frequency range. The set of the modes for each design was evaluated using a non-resonant loop probe [schematically shown with blue lines in Fig. 1(b)] placed above one of the M-coils (in $xz$-plane) and between Helmholtz coils (in $yz$-plane). Each setup was loaded to the breast voxel model. We have used a realistic female body model based on Ella from the Virtual Family [5] and modified it by adding breast phantom no. 1 from the UWCEM Numerical Breast Phantom Repository [6].

To estimate the transmit efficiency and RF safety, we used a shielded high-pass body birdcage coil with 16 legs, 350 mm inner diameter, and 650 mm length. We compared four setups: (1) a voxel model placed inside the body coil without any wireless coil; the same system with HH-coil (2), MM-coil (3), and HHMM-coil (4) placed around the breast. The calculated $B_1^+$-field and specific absorption rate (SAR) averaged over 10g of tissue mass distributions were normalized to 1 W of total accepted power. Transmit efficiency, and RF safety was evaluated as the root mean squared value (RMS) of the transmit magnetic field – $|B_1^+|_{\text{RMS}}$ in the breast area per 1 W of accepted power, and $|B_1^+|_{\text{RMS}}$ in the area of interest per square root of maximum local SAR ($\text{maxSAR}_{av,10g}$), respectively. Transmit efficiency was calculated in the breast tissues limited by the volume of each wireless coil, respectively.

3. Results

Figure 1b shows the numerically obtained spectra of the magnetic field in the range of 40-80 MHz. The HH-coil has a single peak in this range, which corresponds to a co-directional magnetic field in both coils polarized in the $x$ direction. This is a symmetric fundamental mode. It is associated with the greatest uniformity of the magnetic field within the coil volume. The anti-symmetric mode and the higher-order modes are outside the frequency range. As for MM-coil, two maxima correspond to the excitation of anti-symmetric and symmetric modes within the operating frequency range. They differ in
Figure 1. a) Schematic view of the Helmholtz pair (HH-coil), metasolenoid pair (MM-coil), and quadrature coil based on their combination - HHMM-coil. b) Numerically calculated spectra of the magnetic field in the middle of the left breast of the female voxel model on the frequency for each coil type correspondingly. The insets schematically demonstrate the magnetic field orientation within each coil: $H_x$-component for HH-coil; $H_y$-component for MM-coil, the opposite direction of red arrows indicates anti-symmetric and symmetric modes; circular arrow shows interaction with both $H_x$- and $H_y$- components for HHMM-coil. Blue lines depicted the position of the loop coils used for excitation.

The symmetric mode is optimal because it is characterized by a more uniform transverse RF magnetic field distribution in the area of interest. The third spectrum, for the quadrature HHMM-coil, shows three modes. These are the symmetric and anti-symmetric MM-coil modes and the symmetric HH-coil mode. Due to the magnetic fields’ orthogonality created by different coils, the electromagnetic coupling is minimal. Therefore, there is no additional mode splitting. When each type of coil’s symmetric mode is adjusted to the operating frequency of 1.5 T MRI, a magnetic field with circular polarization is induced in the volume.

Since the HHMM-coil interacts with both components of the body coil’s RF magnetic field, a greater transmit efficiency gain is observed than HH-coil and MM-coil. Fig. 2 demonstrates simulated $|B_1^+|_{RMS}$ and SAR$_{av.10g}$ maps inside the body coil without and with the proposed wireless coils designs placed around the breasts. Strong localization of the $|B_1^+|_{field}$ in the breast area in the presence of the wireless coil leads to a 3.7, 4.3, and 9.7-fold enhancement in body coil’s transmit efficiency for HH-, MM- and HHMM-coils, respectively. Although the magnetic field distribution in all cases is almost completely symmetric, SAR$_{av.10g}$ maps have an asymmetric distribution in the human body for the wireless coils that support linear polarization only. This could be explained by the complex interaction of the electric field produced by two feeding ports (with 90-degree phase shift) of the body with the HH- and MM-coils; it is not fully focused within the wireless coil’s volume, and due to the arbitrary shape of the body model has asymmetric distribution. The maximum of SAR$_{av.10g}$ (depicted as a red circle in Fig. 2b) for body coil used alone located in the muscles of the back, while in the presence of wireless coils, it occurs in the breast or abdominal muscle areas. At the same time, the maximum SAR$_{av.10g}$ in the body model for the same 1 W total accepted power is increased by 3.8, 1.8, and 3 times for HH-, MM- and HHMM-coils, correspondingly, in comparison with a reference case (body coil used alone). Table 1 summarizes the results of transmit efficiency and RF safety for all the cases. The highest RF safety improvement (5.6-fold) compared to the reference case was achieved for the HHMM-coil.
Birdcage coil Birdcage coil & HH-coil
Birdcage coil & MM-coil
Birdcage coil & HHMM-coil

Power accepted = 1 W

\(<|B_1 (+)|_{\text{ROI}} >_{\text{RMS}} = 0.3 \, \mu T >_{\text{RMS}} = 1.1 \, \mu T >_{\text{RMS}} = 1.3 \, \mu T >_{\text{RMS}} = 2.9 \, \mu T >_{\text{RMS}} = 0.3 \mu T

maxSAR_{av,10g} = 0.12 \, \text{W/kg}
maxSAR_{av,10g} = 0.46 \, \text{W/kg}
maxSAR_{av,10g} = 0.21 \, \text{W/kg}
maxSAR_{av,10g} = 0.36 \, \text{W/kg}

Figure 2. Numerically calculated $|B_1(\oplus)|_{\text{RMS}}$ maps (a) and SAR_{av,10g} maps (b) for the voxel model placed inside the birdcage coil without and with the WLC. The $|B_1(\oplus)|_{\text{RMS}}$ value was calculated only for the volume of the resonator in the breast area (ROI). SAR_{av,10g} distributions are build up through the local maximum plane. Local SAR_{av,10g} maxima are indicated with red circles.

Table 1. Numerically calculated results of transmit and SAR efficiencies for different types of wireless coils.

|                      | Body coil | HH-coil | MM-coil | HHMM-coil | Units          |
|----------------------|-----------|---------|---------|------------|----------------|
| $|B_1(\oplus)|/\sqrt{\text{P}_{\text{acc}}}$ | 0.3       | 1.1     | 1.3      | 2.9         | $[\mu T/\sqrt{\text{W}}]$ |
| maxSAR_{av,10g}     | 0.12      | 0.46    | 0.21    | 0.36       | $[\text{W/kg}]$  |
| $|B_1(\oplus)|/\sqrt{\text{maxSAR}_{av,10g}}$ | 0.86      | 1.62    | 2.84    | 4.83       | $[\mu T/\sqrt{\text{W/kg}}]$ |

4. Conclusion

In this work, we proposed and compared three designs of the wireless coil, which can be used for targeted bilateral breast imaging at 1.5 T. Two of them are linearly polarized coils based on pair of Helmholtz coils or volumetric metamaterial-inspired coils, another one is quadrature coil based on the combination of these linearly polarized coils. The combined coil interacts with the both ($B_{1x}$ and $B_{1y}$) components of the transverse RF magnetic field generated by the body coil and focuses it in the region of interest within its volume with circular polarization. Due to this fact, electromagnetic studies with the female voxel model of the quadrature HHMM-coil have shown more than a 2.5-fold increase in transmit efficiency and a 1.7-fold increase in SAR efficiency compared to the linearly polarized MM-coil and HH-coil (Table 1). Future work includes the experimental realization of the HHMM-coil, in vivo imaging and comparison coil’s receive performance with dedicated cable-connected coils.

Acknowledgments

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