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Phased Array Receiving Coils for Low Field Lungs MRI: Design and Optimization

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Recent techniques of radiofrequency (RF) probes and preamplifiers in Magnetic Resonance Imaging (MRI) developments almost reached the physical limits of signal to noise ratio (SNR). More improvements in speed accelerations of data acquisition are very difficult to achieve. One exception, called RF phased array coils, is recently being developed very progressively. The approach is conceptually similar to phased array used in radar techniques; hence it is usually called MRI phased array coils. It is necessary to ensure independence of the individual coil channels in the array by the coil and preamp decoupling and the coil geometry optimization to get maximum benefits from this technique. Thus, the qualitative design and method for optimization of geometric properties of the coil elements in phased arrays, with aim to increase SNR, minimize the G-factor and to limit noise correlation, are proposed in this paper. By the finite element method (FEM) simulations, we obtained the sensitivity maps and inductances of the coils. The introduced program primarily calculates the Sensitivity Encoding (SENSE) G- factor along with other parameters that can be derived from sensitivity maps. By the proposed optimization algorithm, the program is capable to calculate the optimal values of the geometric coil parameters in a relatively small number of iterations.

Keywords: magnetic resonance imaging, phased array, field calculation, optimization

1. INTRODUCTION

USUALLY, the MR data are acquired sequentially by applying different magnetic field gradients. Parallel imaging technique in MRI uses spatial information about the origin of the detected MR signal from sensitivity maps of every particular RF receiver. This information may be used for the image reconstruction. As the coil sensitivity map does not depend on the examined object, it can be obtained prior to the measurement and just once for each coil. Thus, it can significantly shorten the time of image acquisition, e.g., by using fewer phase encoding steps (without losing the image quality), or increase the image quality in the normal acquisition time. Careful design of a suitable phased coil array is essential for optimal parallel imaging [1], [2]. Optimization of the four RF conductor coil design using genetic algorithms was described in [3].

Mostly, phased arrays are used as receivers, and a separate homogenous volume coil is used for transmission. Receiving coils in array are connected to independent preamplifiers, amplifier and afterwards are separately digitalized.

Resulting data are combined in an optimal way, with focus on the origin of the signal, by the reconstruction algorithm, e.g., sum of squares, Sensitivity Encoding (SENSE) reconstruction algorithm [2], Simultaneous Acquisition of Spatial Harmonics (SMASH) [4], Generalized Auto-calibrating Partially Parallel Acquisitions (GRAPPA) [5], [6], or Partially Parallel Imaging with Localized Sensitivities (PILS) [7]. Using these methods, it is possible to acquire SNR of local coil with a field-of-view (FOV) typical for a volume receiver, or/and speed up image acquisition [8], [9], [10].

Noise in Parallel Imaging - Geometry Factor

A serious limitation of all techniques in MRI is the level of noise or more specifically, signal to noise ratio (SNR). Phased array techniques and parallel imaging are not an exception. A significant part of the noise is generated by the sample, thus the use of an array of numerous local resonators - receivers instead of one global – was proposed. This way, it was easier to ensure the detection of the signal (and noise) only from the examined part of the sample. Applicability of the array for parallel imaging is mostly dependent on the geometry of the resonator elements as well as their position in array. Therefore, array coils are usually described by the parameter called geometry factor (G-factor), and can be defined by the formula of Pruessmann [2]:

$$g = \frac{SNR^{full}}{SNR^{red}\sqrt{R}} \tag{1}$$

Where g is a geometry factor and it is always equal or higher than one, SNR^{full} is SNR in full encoding acquisition, SNR^{red} denotes SNR in sample-reduced acquisition and R is a factor by which the number of samples is reduced in comparison to full acquisition. For SENSE reconstruction the following formula was derived:

$$g_\rho = \sqrt{\left[\left(S^H\Psi^{-1}S\right)^{-1}\right]_{\rho,\rho}\left(S^H\Psi^{-1}S\right)_{\rho,\rho}} \tag{2}$$

Where

- g_ρ is a local geometry factor,
- ρ denotes the index of the voxel,

- S^H transposed sensitivity matrix,
- ψ receiver noise matrix.

For the GRAPPA reconstructions a different formula for quantitative G-factor must be used [6].

Decoupling

Mutual inductances and parasitic capacitances may cause coupling - an undesired transfer of signal and also an additive noise between the coils that may cause the so called noise correlations. This causes loss of the spatial information and also decreases the SNR. Therefore, the reduction of these unwanted interactions between coils with overlapping FOVs is critical in phased array techniques. That might be performed in several ways (or their combinations):

- By the mutual position of the coils in array. By overlapping of the neighboring coils it is possible to compensate mutual inductions almost to zero [8]. On the other hand, the overlapped coils will receive the signal (and noise) from the overlapped regions so it will decrease the advantage of the local coils in array and decrease the SNR. Mutual inductances can be decreased down to the reasonable level by distance and mutual position between the coils (gap design). It is rather impossible to get the noise correlation to the functional level this way, thus this technique is usually combined with other decoupling techniques to decrease the level of coupling to the optimal level [11].
- By the so called "preamplifier decoupling" based on using either high impedance or very low impedance input preamplifiers. The aim is to minimize the mutual inductions by decreasing the current in the coils or voltage on the input of the preamplifier.[12]
- By the so called "coil decoupling", which uses lumped elements (capacitors or inductances). There are several ways how to decouple coils from their immediate neighbor; such as adding contra-mutual inductance to the coils mutual inductance, or using a decoupling capacitor connected in series in between neighboring coils so that the voltage across the capacitor counter balances the inducted electromotive force [13].
- By coil shielding (passive or active).

In this paper, phased array coil design and optimization method is proposed. We propose an effective method for optimization of phased array resonator design using simulations of electromagnetic fields by finite element methods (FEM). Description of the coil array and setting of the simulation parameters is based on common interface of the Matlab and Comsol. By means of this method, the four channel coil array for ESAOTE G-scan for thorax imaging was designed and optimized respecting the SENSE G-factor. Designed coils were constructed, tuned by capacitors, connected to the preamplifier and first images were obtained.

2. Subject & Methods

Analytical precalculations

For the study of the butterfly coil behavior, analytical calculations in Wolfram Mathematica derived from Biot-Savart law have been used:

$$H(r) = \frac{I}{4\pi}\oint_l \frac{dl \times r}{\left|r^2\right|} \tag{3}$$

Vector of magnetic field ***H(r)*** is calculated in a simplified model made of four infinite long strip conductors fed by current *I*. Thickness of the strips can be neglected. For one single strip, formulas (4) and (5) have been derived. The x-component of magnetic field H_x can be written as follows:

$$H_x = \frac{15}{2\pi}\left[\begin{array}{l} \tan^{-1}\left(\dfrac{-a-x+f}{y+b}\right) \\ -\tan^{-1}\left(\dfrac{a-x+f}{y+b}\right) \end{array}\right], \tag{4}$$

For the y–component of the magnetic field H_y the following expression [14] was written:

$$H_y = \frac{15}{4\pi}\left[\begin{array}{l} Log\left\{\left(-a-x+f\right)^2+\left(y+b\right)^2\right\} \\ -Log\left\{\left(a-x+f\right)^2+\left(y+b\right)^2\right\} \end{array}\right]. \tag{5}$$

Where *a* is width of the strip, *x* and *y* are coordinates in two dimensional Cartesian coordinate system, *f* is a shift of the strip on *x* axis, *b* is a shift of the strip on *y* axis.

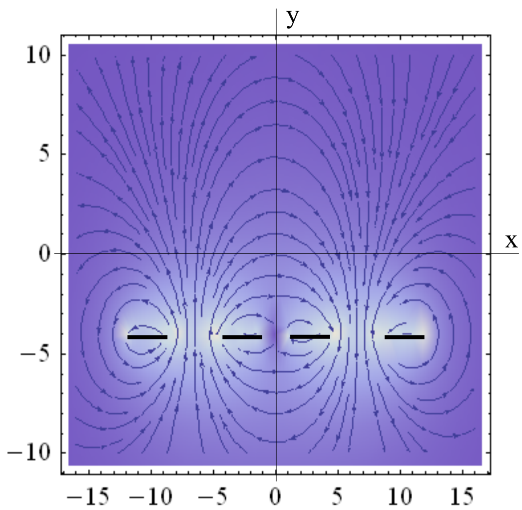

Fig.1. Magnetic field $H_{x,y}(x,y)$ of the one horizontal strip-wire butterfly coil calculated from equations (4), (5), relative dimensions.

Design of the coil

The proposed coil array was developed for MR imaging of lungs using hyperpolarized ^{3}He for ESAOE G-Scan 0.25T open bore MR system. The aim of the design was to pick up the signal from the whole thorax (lungs) with maximum sensitivity and minimum overlapping of FOVs (overlapping FOVs in low sensitive regions), to decrease mutualcoupling. The first concept originates in the idea drawn in Fig.2.

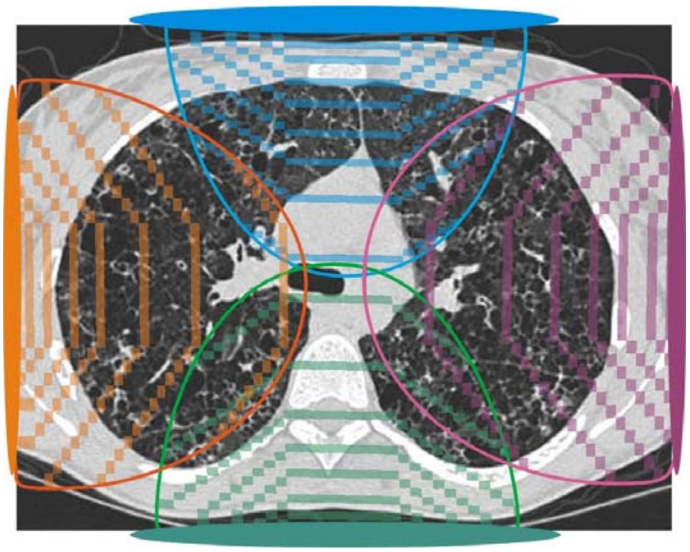

Fig.2. Proton MR image of the lungs with proposed FOV's of each resonator in the array. Sagittal slice, illustrative picture.

Small bore of the G-Scan together with B_0 direction perpendicular to scanner cavity does not allow using the typical design of several identical surface coil loops (circular, square or hexagonal) placed symmetrically or semi-symmetrically around the examined object for full thorax measurement. Hence, a design combined from two square loops and two so called *butterfly coils* was proposed.

The idea of this setup is based on the combination of good sensitivity of simple surface coil and the advantage of butterfly coils with significant sensitivity for the longitudinal magnetic field components (this means field perpendicular to the face of the coil). See Fig.1 and 5.

FEM Analysis – Comsol multiphysics

Analytical calculations using the Biot-Savart law at our wavelengths and dimensions of the coils are very precise. Also phase changes do not play a significant role. The mathematical description and consecutive integration, however, is very complicated already with an undersized difficulty of the shape of the coil. Thus, FEM analysis was used for optimization and more complex calculations.

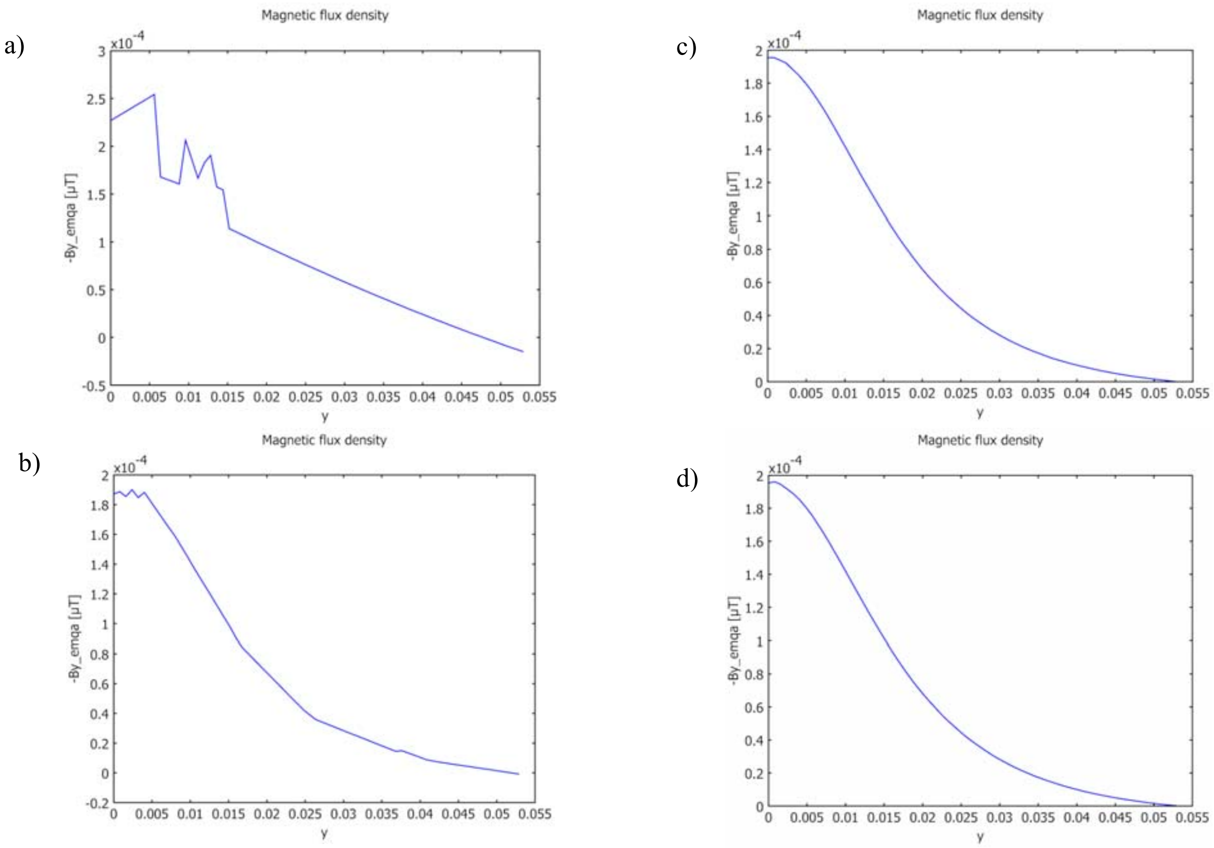

Fig.3. Calculated B_y component of magnetic flux density in the midline (place with smallest mesh density) depicts the "quality" of the final FEM analysis in dependence on mesh density. Mesh consists of: a) 406 elements, b) 2 110 elements, c) 55 450 elements, d) 213 280 elements and FEM analysis was done in 0.3s, 1.5s, 16s and 72s, respectively.

For FEM analysis, Comsol Multiphysics (Comsol AB. Stockholm, Sweden) package with an interface for Matlab (Mathworks, Natic MA, USA) was used. This allows preparation of all models by a source code written in Matlab and also running all Comsol simulations from Matlab. Data extracted from the simulations are processed and parameters (G-factor, for instance) can be calculated. According to the calculated parameter, the new dimensions of the coil are proposed and calculated.

As a relatively large number of parameters of the coil are adjusted, a fast converging optimization algorithm is necessary; otherwise the calculations take an unreasonably long time.

For the proposed array design an iterative approach was performed. It means that the limits of all adjusted dimensions were specified. According to the limit dimensions, the size of boundaries and mesh density were swept and simulations to find the least possible mesh density and least possible boundary condition were calculated. This has been done to decrease the number *s* (the degree of freedom of FEM analysis), to get reasonably precise results and to keep the time of the simulation at minimum.

Midline of the coil is the most important space and usually the smallest mesh density is calculated there. In Fig.3, the dependence of FEM analysis on the mesh density is depicted.

Optimization of coil arrays configuration, optimization algorithm

The high number of variables together with the relatively time consuming FEM analysis might cause unreasonably long optimization time and therefore has high requirements on computational endurance of the computer.

There is no reason to search for a perfect result, because the coil parameters slightly change with the loading and surroundings, the model will always slightly differ from the reality. So, the optimization convergence threshold can be set quite high. Also from knowledge of the behavior of the coil and arrays, it is possible to improve the optimization by reasonable initial conditions.

It is clear that from the number of variable dimensions it is not possible to use a *brute force* to calculate not even five steps for every variable in every combination of variables, because the time needed would be unbearable.

From the physical nature of the G-factor and its dependence on sensitivity maps we can predict that our unknown function is continuous and smooth.

This method is based on assumption of a high probability that the optimum lies not far from the initial conditions. Four analyses for every variable were calculated with a defined step around the initial point (one variable is changing while the rest is in the zero point). The results of the FEM analysis are G-factors (also other parameters could be calculated or extracted from analysis). The best result was chosen as the initial point for the next step. Around the new initial point, four analyses for every variable were calculated. These steps were repeated until the minimum was found.

The flow chart of the optimization procedure is depicted in Fig.4.

The algorithm stops if the difference between the last two calculated values is smaller than the defined value or if two selected consecutive initial points are the same. The optimization step of the algorithm is fully adjustable, but for our purpose just a simple setting was sufficient. The following computational system was used: Intel I7 - quad

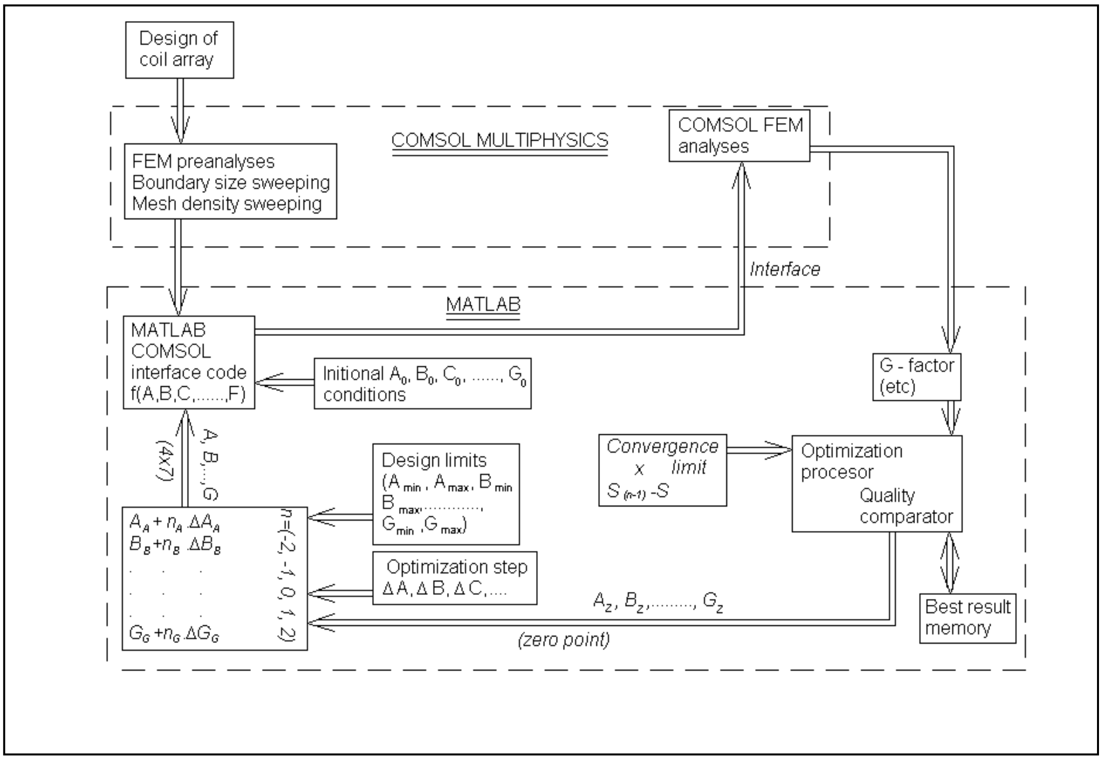

Fig.4. Flow chart of the optimization procedure

core (2.9GHz); 12GB DDR III Windows Vista Business 64bit; Comsol v3.5a; Matlab 2008a- 64bit.

3. RESULTS

The method for optimization of phased array receivers uses simulations of electromagnetic fields by finite element methods available in Comsol Multiphysics. Optimization algorithm was implemented as a routine written in Matlab. Description of the coil array and setting of the parameters of the Comsol simulation was based on common interface of the Matlab and Comsol.

The lung phased array for ESAOTE G-scan (ESAOTE Genoa Italy) system was optimized for SENSE - G-factor by adjusting the seven dimensional parameters. The results were obtained by the proposed method and algorithm in three hours on our computational system.

Designed coils were constructed, tuned by fix capacitors, connected to the preamplifier and first images were successfully obtained.

Tab.1. Results of the optimization steps of the 4-channel phased array for 3He lung imaging in ESAOTE G-Scan. Dimensions (according Fig.4) of the coils Optimized on SENSE G-factor calculated by (2) with R=4. A, B, C, D, E, F, G are dimensions of the coil, see Fig.4.

Step	**0**	**1th**	**2nd**	**3rd**	**4th**	**5th**	**6th**	**7th**
A	19	24	24	24	24	24	24	24
B	24	20	26	20	20	20	23	26.5
C	1.5	1.0	0.5	0.5	0,5	0.5	0.5	0.5
D	8.0	5.5	5.5	6.5	5.5	5.5	5.5	5.5
E	30	25.5	33	38	32.5	40	40	40
F	1.5	1.0	1.5	1.0	1.0	0.5	0.5	0.5
G	5.0	4.5	3.0	4.0	3.0	4.0	3.0	3.0
G-Fac.	**3.02**	**2.19**	**2.04**	**1.96**	**1.84**	**1.73**	**1.67**	**1.67**

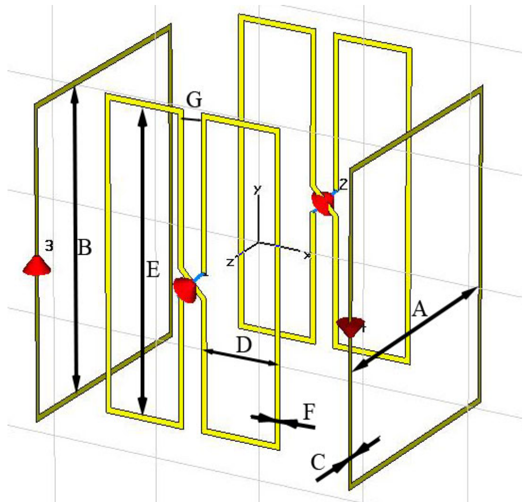

Fig.5. Model of the 4 channel phased array coil system designed for ESAOTE G-Scan MR tomograph

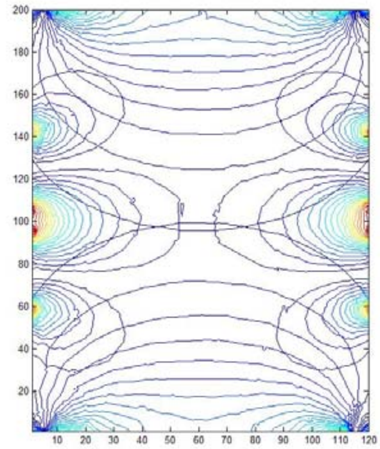

Fig.6. Mid plain sagittal slice coil sensitivity maps of the array from Fig.5 calculated in Comsol Multiphysics, real dimensions 240 x 400 mm.

4. DISCUSSION/CONCLUSIONS

The paper proposes a method for optimization of phased array resonators by using simulations of electromagnetic fields by finite element methods in Comsol Multiphysics and optimization routine written in Matlab. Description of the coil array and setting of the parameters of the simulation is based on common interface of the Matlab and Comsol. By using this method, the four channel coil array for ESAOTE G-scan for thorax imaging was designed and optimized for SENSE G-factor. Currently designed coils were manufactured, tuned by fix capacitors, connected to the preamplifier and first images were obtained.

Naturally, the correct coil tuning and decoupling are unavoidable to achieve the best results.

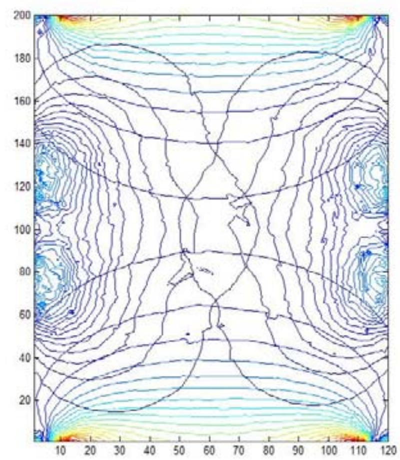

Fig.7. Border plain sagittal slice coil sensitivity maps of the array from Fig.5 calculated in Comsol Multiphysics, real dimensions 240 x 400 mm.

The advantage of the proposed method lies in the versatility of its usage, in effective and fast calculations and in its relative simplicity.

In order to reduce the computational time, it was necessary to decrease the degree of freedom to the minimum. Therefore, the real thickness of the copper strips was not considered (the real small thickness would locally excessively increase the mesh density). The skin effect may cause that the effective copper section will not be really increased. Also the used thickness is still much smaller in comparison to the other dimensions. No significant changes were found with the decrease in thickness in the FEM analysis.

We can see in Fig.6 that the *butterfly* coils have smaller penetration depth and smaller homogeneity compared to square loops, therefore FOVs are smaller. The reason is that a significant part of the sensitivity of these coils is in B_0 direction and thus has no impact on the image. But still it seems to be one of the best choices for the four channel array configuration for this purpose.

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