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# Multi-channel receive coil array for human head MPI system

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**Abstract**

Magnetic particle imaging is an emerging medical tomographic technique that employs a combination of static and alternating magnetic fields to spatially encode superparamagnetic nanoparticle-based tracers injected into a patient's bloodstream. Unlike the widely used magnetic resonance imaging, in MPI, the signal excitation and reception processes occur simultaneously, which imposes limitations on the architecture of the receive coils used in the equipment. This work presents a novel matrix receive coil concept featuring significant gradiometric characteristics, enabling the construction of a multi-channel array for a human head MPI system. The feasibility of this concept is demonstrated through numerical simulations on two coil arrays with different numbers of elements. These results show that the presented approach can serve as an efficient way to expand the number of independent receive channels, efficiently increasing the available field of view and sensitivity of the imaging system.

## I. Introduction

Magnetic particle imaging (MPI) is a medical imaging modality that is currently in the transition phase from preclinical experimental setups to certified clinical devices and trials. The essence of the method lies in the excitation of a nonlinear magnetic response of superparamagnetic nanoparticles by alternating magnetic fields and their localization using gradient fields gleich2005tomographic. Compared to the well-established magnetic resonance imaging (MRI), which relies on magnetic fields to visualize proton density in the human body, MPI is able to visualize only contrast agent injected into the blood. An even more fundamental difference between the two technologies occurs in the conceptual part, where, instead of the time-separated excitation and reception steps of measuring relaxing nuclei, it is necessary to track the immediate response of nanoparticles in the presence of an alternating high-amplitude fields (25 kHz, 6 mT for a head-sized system Thieben2024), which will inevitably interact with the receive coils, overloading sensitive electronics. In order to cope with that, many technical solutions were proposed, including powerful filters schumacher2020highly, active compensation techniques Pantke2019, and so-called gradiometric receive coils Graeser2017. The latter are one of the simplest in practice and at the same time very effective ways of suppressing feedthrough signals in MPI, making it the standard in this field.

A typical gradiometer coil, as demonstrated in Figure 1(a), consists of two coils that are connected opposite to each other. The nanoparticles should be located within the measuring coil, while the compensating coil should remain empty, with the entire coil still staying within the excitation field. To increase the area available for imaging, the compensating coil is often made more compact and shifted away from the region of interest. However, even with these countermeasures in place, gradiometric design narrows the scanner's field of view and introduces non-uniformities in the sensitivity profile between the two coils (Figure 2(a,b)).

Another limitation concerns spatial sensitivity. The gradiometric coil design can effortlessly be wound as a solenoid on a cylindrical surface, which corresponds to the receive sensitivity profile along a single spatial direction $x$. However, the implementation of such a concept for the other two orthogonal directions is technically difficult in practice. For this reason, the majority of MPI scanners presented so far are typically equipped with only one receive gradiometric coil.

In this paper, we address these limitations by introducing a novel receive coil concept based on matrix gradient coil design, originated from the field of MRI Juchem2010. By transitioning from a typical gradiometer to a matrix coil array, we aim to establish distinct, independent sensitivity profiles for all three spatial directions while also preserving important gradiometric features to minimize feedthrough. To evaluate the efficacy of this approach, we perform simulated comparisons between the proposed matrix coil architecture and traditional gradiometric coil across two distinct concepts.

This approach utilizes the reciprocity principle; by adjusting the current through independent small coil elements in Figure 1(b,c), we can control and synthesize field profiles inside the coil. For the MPI task, this allows us to establish distinct sensitivity profiles. This setup enables more independent receive channels to be used simultaneously, expanding the system's capabilities. Furthermore, since parallel receive channels can provide basic spatial encoding kretov2025spatial, this matrix architecture potentially offers a pathway to improved image resolution.

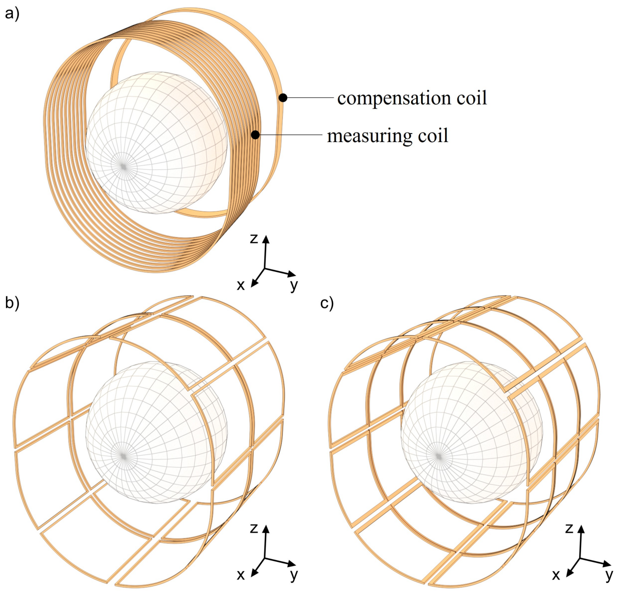


**Figure 1:** a) A first order gradiometric coil. The compensation coil has more dense windings and is shown in a simplified manner. Matrix coil array concept with b) 16 and c) 24 elements. The field optimization volume is represented by a sphere with a radius of 75 mm.

## II. Methods and materials

As a basis for the coil structure, we selected bore dimensions from Thieben2024. On this surface, we defined geometry for two arrays with 16 and 24 elements, where each element is represented by rectangular coil with a thickness of 2 mm and dimensions of 85 x 103 mm and 85 x 70 mm, respectively. Figure 1(b) shows the array of 16 elements arranged in two rings, whereas for the second array of 24 elements (Figure 1(c)) it was three rings. We designed both models in Solidworks (Dassault Systèmes, Vélizy Villacoublay, France) and exported resulted CAD models into the numerical simulation software COMSOL Multiphysics (COMSOL AB, Stockholm, Sweden) for further processing.

We designed geometry of the numerical model as cubic air block with dimensions of 400 x 400 x 400 mm containing coil structure defined as copper. The magnetic fields static solver was used with normal physics-controlled mesh grid element size. We performed 16 simulations for the first structure, and 24 simulations for the second structure, setting 1 A current consequently for all elements in both arrays. The simulation results were exported as a 3D magnetic flux data grid and used for optimization.

To find a unique winding configuration for both arrays, we defined a sphere with a radius of 75 mm, positioned in the center of the simulated data, as the region to perform optimization. As an optimization target, we used 3D field distributions corresponding to spherical harmonic (SH) functions that were derived from magnetic scalar potential, defined on the surface of the sphere. Since SH are an orthogonal set of functions, they permit us to find combinations of coil currents that correspond to linearly independent sensitivity profiles in the considered volume. Calculated coil currents could be directly converted into coil turns number, with the sign of the current indicating the winding direction.

We arranged the optimization task as a least squares problem, that reflects the difference between the target field and fields of coils, where current in each coil was optimized value. After solution we estimated root mean square (RMS) mismatch error in percent which shows how accurately we can synthesize the fields for each spherical harmonic by using the coil array.

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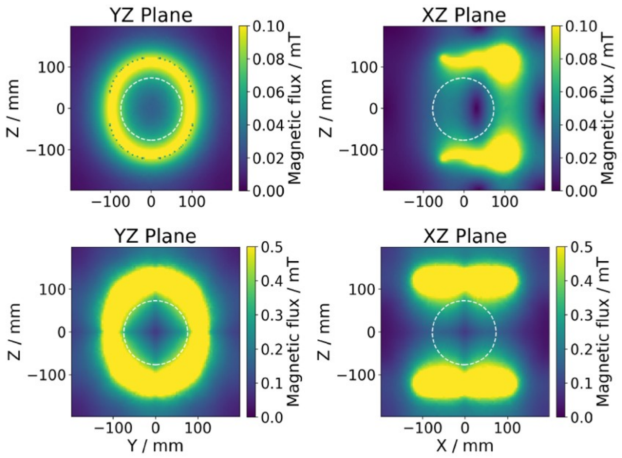


**Figure 2:** Magnetic density norm at 1 A current in axial (a, c) and sagittal (b, d) planes for first-order gradiometric coil (a,b) and matrix coil array with 16 elements summed over 6 field configurations (c, d). The circle represents the cross section of sphere with a radius of 75 mm.

**Table 1:** Error between target and calculated fields for 16 and 24 coil arrays.

| No | SH | 16 coils, error (%) | 24 coils, error (%) |
|---|---|---|---|
| 1 | SH (1,-1) | 10.66 | 7.27 |
| 2 | SH (1,0) | 46.74 | 34.08 |
| 3 | SH (1,1) | 10.24 | 7.89 |
| 4 | SH (2,-2) | 11.32 | 11.36 |
| 5 | SH (2,-1) | 33.16 | 20.03 |
| 6 | SH (2,0) | 17.37 | 6.38 |
| 7 | SH (2,1) | 36.49 | 19.06 |
| 8 | SH (2,2) | 10.88 | 10.40 |
| 9 | SH (3,-3) | 31.19 | 32.14 |
| 10 | SH (3,-2) | 29.36 | 13.04 |
| 11 | SH (3,-1) | 99.56 | 44.24 |
| 12 | SH (3,0) | 89.36 | 94.10 |
| 13 | SH (3,1) | 99.68 | 46.29 |
| 14 | SH (3,2) | 30.26 | 13.19 |
| 15 | SH (3,3) | 30.33 | 29.30 |

# III. Results and discussion

As shown in the Table 1, a majority of the calculated field configurations resulted in error rates less than 50% for both coil arrays. The 24-coil array has more degrees of freedom and can therefore establish a more accurate field patterns that matches the target field profile. Despite this advantage, this array is considered to be more complex for practical implementation.

To determine the optimal configuration for the 16-coil array, we performed a systematic exclusion of the spherical harmonics based on three primary criteria: hardware coupling, geometric feasibility, and sensitivity.

First, regarding hardware constraints, harmonics 1 and 2 from Table 1 represent homogeneous fields in the $x$ and $z$ directions, which coincide with the excitation fields of the MPI system, making receive coils strongly coupled to the transmitting coils. For this reason, these configurations could not be considered for further use.

Second, we addressed geometric limitations. High errors for third-degree spherical harmonics (numbers 11–13 in Table 1) were observed due to the restrictions of cylindrical geometry used. These field configurations require a field change in polar regions of the sphere close to the cylinder openings. The lack of a surface for currents limits control of the field profile in these regions, especially in the case of a 16-element array, resulting in large error values.

Finally, we considered signal sensitivity. Our evaluation showed that even for the third-degree harmonics (numbers 9-15 in Table 1) have field levels at least one order of magnitude lower than the second-degree harmonics near the sphere of the interest. For the receive coil, this means a dramatic drop in sensitivity, so for practical implementation these configurations also turn out to be not useful.

Considering all above, we decided to proceed with a 16-coil array and SH with number 3-8 from Table 1. For each harmonic we calculated number of turns for array elements proportional to the currents, obtained after optimization and restricted the maximum value to be 25 turns. This resulted in six coil configurations, which can be considered as separate receive channels for MPI system.

In order to check coverage and sensitivity of the new array, we once again performed numerical simulations using the optimized coil structures. We checked the magnetic density norm at 1 A, which serves as a proxy for spatial sensitivity profile of the coils. While a state-of-art first order gradiometer (Figure 2(a, b)) exhibits sensitivity non-uniformities between the measuring and compensation coil-effectively narrowing the field of view-the proposed matrix coil array offers a distinct advantage. By combination of 6 coils (Figure 2(c, d)), the array achieves higher peak sensitivity and a significantly broader field of view. It is worth noting, however, that the new coils are not intended as an alternative to the standard gradiometer coil and can be used together.

# IV. Conclusions

In this work we demonstrated an efficient method for designing a large number of gradiometric receive channels for the MPI system using matrix coil approach. For this we numerically investigated two array structures with 16 and 24 elements. While both were able to reproduce a range of SH-related 3D field profiles, The 24-element design is considered to be more complex in terms of

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winding pattern and manufacturability, making the construction more difficult to implement.

When analyzing the sensitivity profiles for SH up to the third-degree, it was found that higher-degree harmonics have a sharp decrease in sensitivity in the central region of the sphere. This is a limitation on the number of channels that can be synthesized using the proposed method for cylindrical geometry. In accordance with this, we calculated six receive coil configurations and numerically verified their sensitivity maps in comparison with the standard gradiometric receive coil.

As the next step, we plan to build designed coil configurations and implement them into the existing imaging hardware. Since the real MPI system is equipped with conducting shields that introduce eddy currents, it has non-uniform excitation fields, which may affect the feedthrough damping performance of proposed receive arrays. To deal with this, an additional compensation procedure in calculations may be required.

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## Author's statement

Conflict of interest: Authors state no conflict of interest.

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