

FORMATION OF A HOMOGENEOUS FIELD IN LOW-FIELD PM-BASED MRI DEVICE

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Abstract

A method to optimize the magnet system for a 70mT salient-pole open MRI device is presented which is based on FEM simulations. The proposed method gives near tenfold reduction in the shim mass and offers more efficient solution the compared to the conventional passive shimming techniques.

For an H-shaped magnetic circuit the desired field homogeneity of 137 ppm in 240 mm diameter spherical volume can be achieved through 2-3 iterations.

Key words

low-field MRI, design optimization, FEM, field homogeneity, shimming, dipole, permanent magnets.

1 Introduction

Magnetic Resonance Imaging (MRI) is one of the most demanded clinical diagnostics due to its ability to produce detailed non-invasive pictures of tissue structures [Currie et al., 2012; Foltz and Jaffray, 2012; Vaughan,

1989]. High image quality requires a highly homogeneous static magnetic field within the imaging volume to provide uniform spin polarization.

In recent years, significant interest in the development of affordable and mobile tomographic systems has shifted toward systems based on highly-coercive permanent magnets (HCPM) used in 0.05 - 0.3 T MRI for emergency detection of brain strokes [Morgan et al., 1999; Chun and Devine, 2005; Liu et al., 2009; Miyamoto et al., 1990].

This approach is based on the positive experience gained in the development of precision magnetic systems [Arslanova et al., 2023; Arslanova et al., 2024; Grosheva et al., 2026; Sirota and Ovsyannikov, 2025].

In global practice, three main approaches have emerged for generating a homogeneous field in permanent magnet systems:

1. use of Halbach arrays of highly coercive permanent magnets [Halbach, 1990] which provide high magnetic flux density but impose strict requirements on manufacturing and assembly accuracy;

2. systems with a return magnetic circuit (yoke), in which HCPMs serve as a source of magnetomotive force, and homogeneity is achieved by shaping the steel pole pieces [Tadic and Fallone, 1990];
3. in systems with a return magnetic circuit, HCPMs perform a dual function: they not only generate the magnetomotive force but also, due to their special configuration, ensure the formation of a homogeneous magnetic field [Poole et al., 2022];

Each of the methods has its own advantages and disadvantages. This paper considers an approach to selecting the profile of a steel pole piece based on a patented method [Amoskov et al., 2025].

The research is aimed at developing a magnetic system that provides a specified magnitude and homogeneity of the field with minimal weight and dimensions. It should be noted that the mass of the pole piece constitutes a significant part of the total magnet mass.

2 Calculation of magnetic field

All necessary magnetic field calculations in this work were performed using finite element modeling [Volakis et al., 1998; Pryor, 2011].

In the FEM problem setting, we proceed from Maxwell's equations for regions free from currents [Tamm, 1979]:

$$\begin{aligned}\nabla \times \mathbf{H} &= 0, \\ \nabla \cdot \mathbf{B} &= 0, \\ \mathbf{B} &= \mu_0(\mathbf{H} + \mathbf{M}),\end{aligned}$$

where $\mathbf{M}$ is magnetization, $\mathbf{H}$ is the field strength, $\mathbf{B}$ and $\mu_0 = 4\pi \cdot 10^{-7}$ H/m is the magnetic permeability of vacuum.

To ensure high accuracy of the FEM approximation we employ tetrahedral elements.

3 Magnet configuration

The MRI system to be optimized is designed as a bi-planar H-type magnet. The magnet comprises a PM assembly, a yoke forming a flux-return path and reducing the stray field, pole pieces and shimming rings. The imaging volume is 240 mm diameter sphere to accommodate for a patient head. The pole gap is ≥ 280 mm to provide enough space for the patient, gradient coils, and RF coils. An open magnet configuration with four yoke columns shown in Fig. 1 is analysed.

The static field of 60 mT is targeted in the imaging volume. Shaped pole pieces installed on PMs are employed to provide a homogeneous field as good as 300 ppm. To suppress local inhomogeneities caused by the yoke, shimming rings are positioned on the pole pieces. The following materials are used:

1. Permanent magnets: Nd-Fe-B with coercive $H_c = 750$ kA/m, relative permeability $\mu = 1.08$, remanent magnetization $B_r = 1.02$ T;
2. Yoke: steel ST10;

3. Pole pieces: ARMCO steel;
4. Shimming rings: ARMCO steel.

The optimization was aimed to minimize the size and weight of the magnet without loss in the field homogeneity in the imaging volume. Realistic magnetic properties of structural steels were incorporated in the numerical model according to B-H curves presented in Fig. 2. The mesh included about $2 \cdot 10^6$ finite elements. This enabled the FE approximation to converge with the error below 30 ppm.

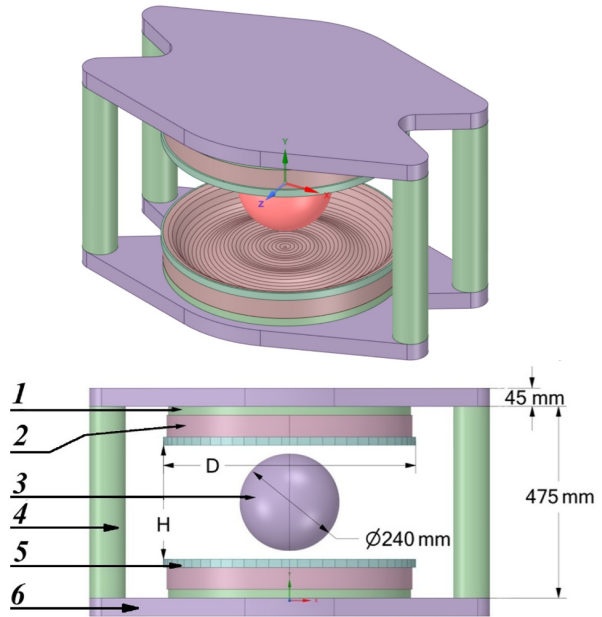


Figure 1. Schematics of MRI magnet structure: 1 - permanent magnet; 2 - pole piece; 3 - imaging volume; 4 - yoke column; 5 - shimming ring; 6 - yoke plate.

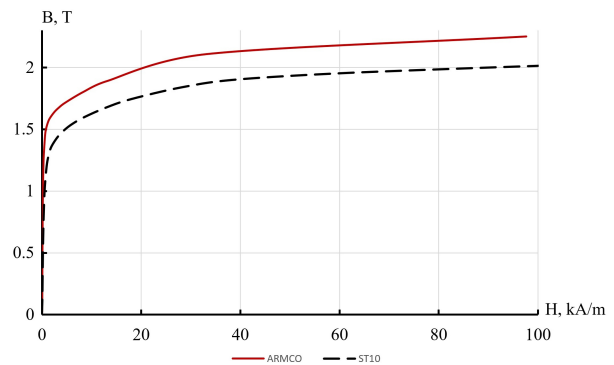


Figure 2. B-H curves for ARMCO and ST10.

4 Optimization of pole pieces and shimming rings

Generally, the MRI magnet optimization includes several phases because of the design complexity. To minimize the computational cost, we divided the optimization into two phases: first we optimized the shape of the pole pieces using the 2D axisymmetric approximation. Then we applied a 3D FE model to establish accurate field distribution and select the sizes and positions of the shimming rings.

4.1 Optimization of pole piece shape

In the 2D approximation the pole pieces were assumed to have the axial symmetry along the axis Z, and the return magnetic flux to close at infinity, see Fig.3. Due to the symmetry, the calculated region was reduced to a 1/2 of the magnet cross section.

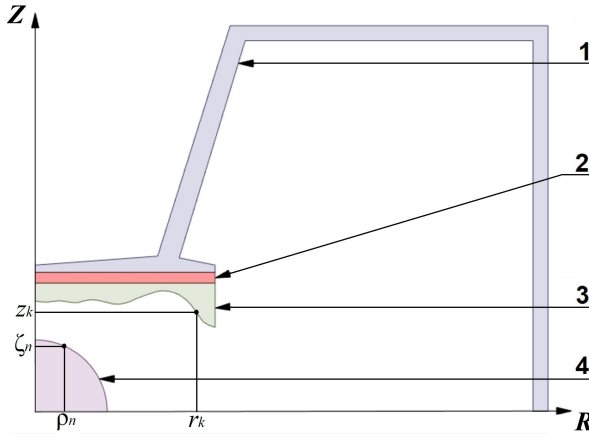


Figure 3. 2D axisymmetric model for pole piece optimization: 1 - steel yoke; 2 - NdFeB PM; 3 - ARMCO pole piece; 4 - imaging volume.

The optimization process was aimed to achieve the minimum of residual functional S reflecting the field homogeneity. The functional is found as a sum of squared deviations of calculated field B_n^{calc} from the target magnetic field B_{target} at N points at the boundary of the imaging volume:

$$S = \sum_{n=1}^N (B_n - B_{target})^2 = \|F\|^2 \rightarrow \min, \quad (1)$$

$$B_{target} = \frac{1}{N} \sum_{n=1}^N B_n.$$

The pole piece shape $z_k = z(r_k)$ is described through a piecewise-linear approximation at $k = 1 \dots K$ points with fixed values r_k . Field $B_n^{(i)}$ at boundary points n can be represented in terms of the current pole piece coordinates $z_k^{(i)}$ as $B_n^{(i)} = B_n^{calc}(z_1^{(i)}, z_2^{(i)}, \dots, z_K^{(i)})$.

Any change of a pole piece coordinate z_k by δz_k produces a change δB_n in the field B_n at the boundary point n , which can be predicted as

$$\delta B_n = B_n^{(i+1)} - B_n^{(i)} \approx \sum_{k=1}^K \frac{\Delta B_{n,k}}{\Delta z} \cdot \delta z_k = \sum_{k=1}^K \gamma_{n,k} \cdot \delta z_k,$$

where

$$\Delta B_{n,k} = B_n(z_1^{(i)} \dots z_k^{(i)} + \Delta z \dots z_k^{(i)}) - B_n(z_1^{(i)} \dots z_k^{(i)} \dots z_K^{(i)})$$

is field disturbance at point n due to small fixed deviation Δz in the only pole profile point $\delta z_k^{(i)}$, $\gamma_{n,k} = \frac{\Delta B_{n,k}}{\Delta z}$ is the sensitivity matrix, which recalculated at certain iterations.

The field value at the $i + 1$ iteration can be predicted as

$$B_n^{(i+1)} = B_n^{calc(i)} + \sum_{k=1}^K \gamma_{n,k} \cdot \delta z_k^{(i+1)}.$$

Then the profile optimization problem (1) at the next iteration can be formulated as follows:

$$\sum_{n=1}^N \left(B_n^{calc(i)} + \sum_{k=1}^K \gamma_{n,k} \cdot \delta z_k^{(i+1)} - B_{target}^{(i)} \right)^2 \rightarrow \min \left(\delta z_k^{(i+1)} \right),$$

where $B_n^{calc(i)}$ is the field calculated at the previous iteration.

The optimized pole piece shape at the next iteration is determined as $z_k^{(i+1)} = z_k^{(i)} + \delta z_k^{(i+1)}$.

Then the field with the pole profile $z_k^{(i+1)}$ is calculated. The optimization process is repeated until δz_k becomes negligibly low.

The optimal shape of the pole piece was obtained with MathCad using the Levenberg-Marquardt method [Levenberg, 1944; Marquardt, 1963]. The field distribution was calculated in the FEM simulation with COMSOL Multiphysics at every iteration of the optimization process. The retrieved residual vector $\mathbf{F}$ was then applied to adjust the pole piece shape. The optimized shape was obtained when the maximal field uniformity $\Delta B/B_0$ was reached at the given pole piece radius R . Overall, the optimization process is presented schematically in Fig. 4.

Table 1 shows the results of the optimization obtained for a set of the pole piece diameters. As expected, an increase in the pole piece diameter would improve the field homogeneity.

The pole pieces give a substantial contribution to the total mass of the magnet. Reasoning from the lightweight requirement, the minimum acceptable diameter $D = 600$ mm was chosen (Fig. 5a) for fabrication of the portable low-field MRI. With such pole pieces the field homogeneity is expected to be much lower the required threshold of 150 ppm. The field obtained with the optimized pole piece of 600 mm in dia is shown in Fig.5b.

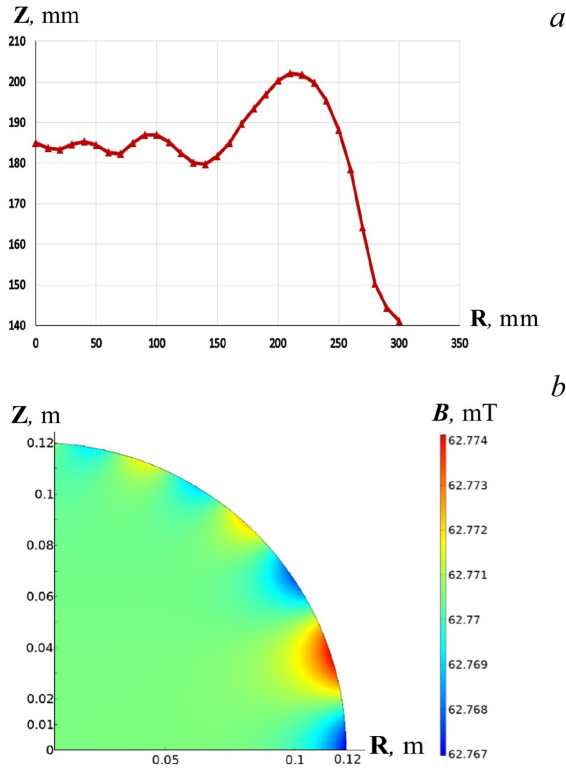


Figure 5. Pole piece shape (a) and field distribution over imaging volume after pole piece optimization (b).

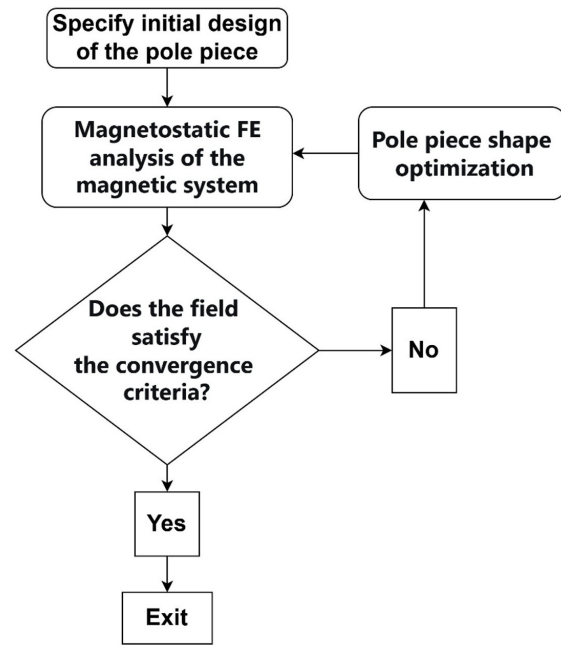


Figure 4. Optimization of pole piece shape.

Table 1. Field homogeneity $\Delta B/B_0$ vs pole piece diameter D with optimized shape

D , mm	560	580	600	620	640
$\Delta B/B_0$, ppm	468	207	100	70	57

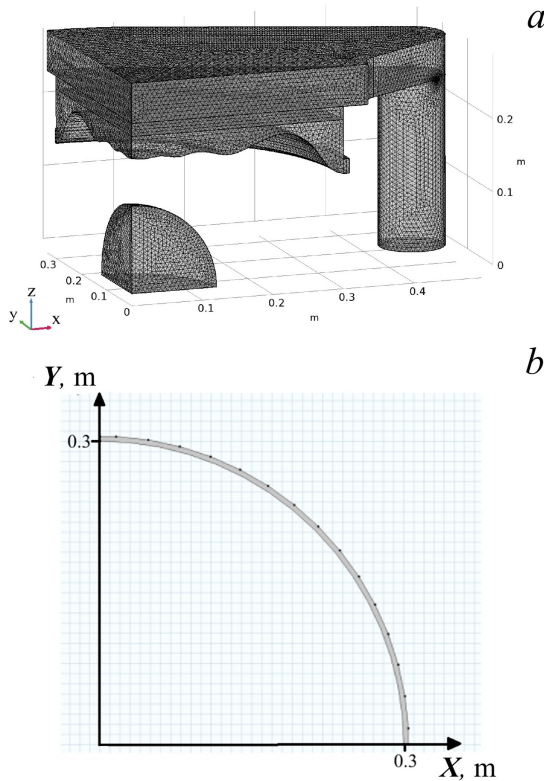


Figure 6. 3D FEM model of pole piece with shimming ring (a) and initial parametric model for 1/4 of shimming ring (b).

4.2 Shimming ring optimization

The 2D approximation of the pole pieces used in the first optimization step gave an idealized field distribution for axisymmetric magnet configuration. The actual magnet configuration is not axisymmetric, and the open yoke will introduce significant field disturbances. To correct the field distribution and provide the required field quality, it is sufficient to place light shimming rings only on the pole lateral surfaces.

The preliminary optimized pole shape allows us to minimize the shimming mass necessary to repair the field disturbance associated with manufacture tolerances and imperfect material specifications.

Practically, it is sufficient to place several shimming rings only on the pole surfaces to tune the field.

The sizes and positions of the shimming rings are optimized with 3D FEM simulations.

In the 3D model the magnet sizes and the optimized pole piece shape ($D = 600$ mm) were prescribed from the 2D simulation. Due to the magnet symmetry with respect to the planes $x=0$, $y=0$, $z=0$, the calculated region was reduced to a 1/8 of the model as shown in Fig. 6a. Each steel shimming ring was modelled parametrically

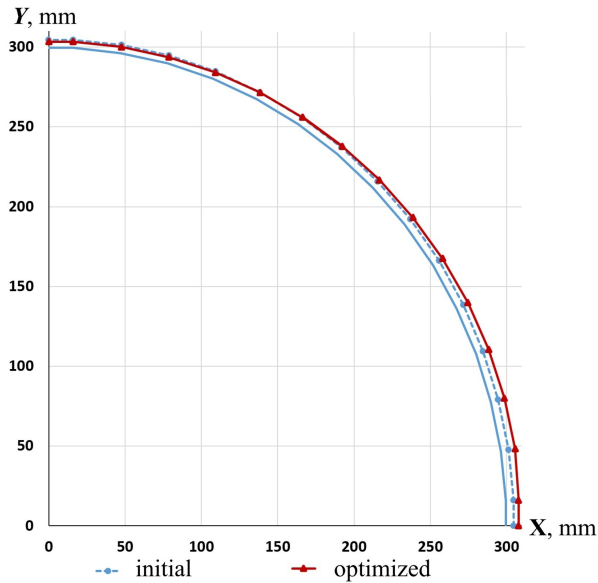


Figure 7. Shimming ring thickness.

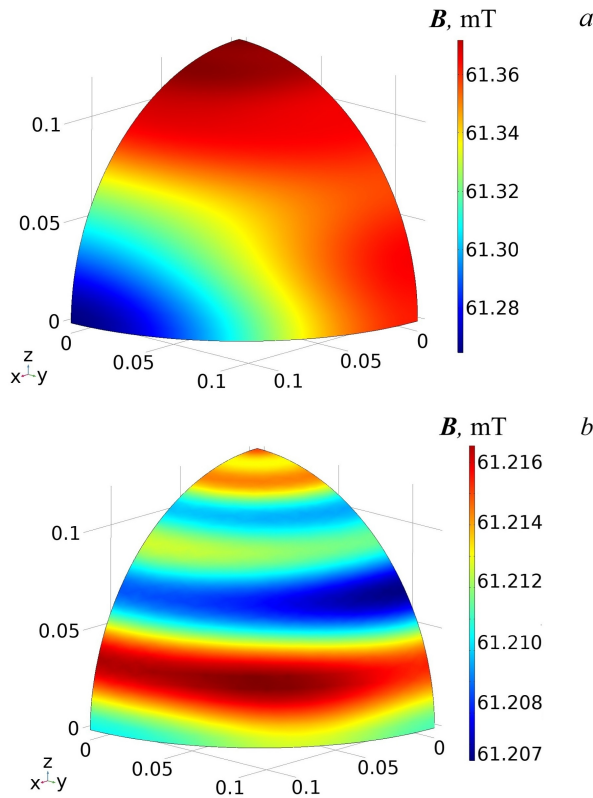


Figure 8. Field distribution before (a) and after optimization (b).

as a ring quarter divided into 15 angular sectors with the step of 6° , see Fig. 6b).

Such representation enabled changes in the ring thickness without update of the FE meshing thus minimizing the computational error. The initial shimming ring thickness was taken as 5 mm.

The optimal thickness of a shimming ring was sought

through the mean square minimization of the residual functional for the field component B_z in $N = 60$ points over 3 reference arcs with the radius of 120 mm at the boundary of the imaging volume.

$$S(\mathbf{r}) = \sum_{i=1}^N (B_i(\Delta\mathbf{r}) - B_{target,0})^2 = \|F(\mathbf{r})\|^2.$$

The optimization process employed the Levenberg-Marquardt.

Table 2 illustrates this optimization process. The initial inhomogeneity introduced by the iron yoke was 1866 ppm. The first correction of the shimming ring thickness reduced the inhomogeneity down to 342 ppm. Further optimization tuned the field distribution to 169 ppm at the 2nd iteration and 165 ppm at the 3^d iteration. The calculation corresponds to the optimized pole piece diameter of 600 mm.

Table 2. Optimization of field homogeneity with shimming rings

Iteration	$\Delta B/B_0$, ppm
0	1866
1	342
2	169
3	165

The optimized shimming rings improved the field homogeneity to 165 ppm that is almost twice lower than the admissible threshold (below 300 ppm). The third iteration demonstrates only small decrease in the field disturbance as the limit of accuracy is approached.

Figures 7, 8 show the optimized shimming ring and the final field distribution in the imaging volume.

5 Conclusions

The two-step numerical optimization method has been developed for the 70 mT magnet of an MRI scanner. The method combines 2D and 3D simulations that allows significant reduction in the computational cost without deterioration of accuracy. The simulation with the simplified axisymmetrical 2D model makes it possible to select the minimal diameter of the pole pieces capable to provide the desired field homogeneity. The field distortions anticipated due to manufacturing and assembly tolerances in practical situations are then compensated by passive shimming. The realistic 3D model, where a pair of shimming rings is added, provides selection of the optimal parameters to tune the field distribution in the imaging volume.

The results of simulation demonstrate that simple ring shims with a varied radial thickness ensure the field P2P

homogeneity of 165 ppm. The second step of the optimization involves only 2-3 iterations to reach the optimal solution. This reflects high sensitivity and efficiency of the proposed method.

The use of the pole pieces with the shimming rings allows us to localize correction efforts and therefore reduce the shimming mass necessary to improve the field homogeneity. The optimized shape of the pole pieces provides quite good field distribution, and it is sufficient to position the shimming rings only on the pole lateral surfaces to tune the field. This would substantially reduce the steel weight and facilitate the shimming.

As demonstrated, the open MRI device with the 600 mm dia pole pieces and shimming rings of the optimized thickness will have the target field of 165 ppm homogeneity in the in 240 mm diameter spherical volume. This completely satisfies the requirement of the field homogeneity of 300 ppm or better.

The proposed method can be applied for the design and optimization of portable, lightweight MRI systems to advance in the engineering of medical diagnostics, particularly, brain scanning. Further efforts are expected to be focused on prediction of possible field distortions due to manufacture tolerances and validation of simulated results with measurements on a prototype.

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