Numerical Simulations for the Design of Novel Integrated Parallel Reception, Excitation, and Shimming (iPRES) Coil Arrays

by

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Thesis submitted in partial fulfillment of the requirements for the degree of Master of Science in the Medical Physics Program Duke Kunshan University and Duke University

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ABSTRACT

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Abstract

In magnetic resonance imaging, static magnetic field ($B_0$) inhomogeneity can result in image artifacts including signal loss, image blurring and distortions, which may lead to decreased diagnostic accuracy. Conventional spherical harmonic shimming, which uses a set of coils placed far away from the subject, cannot shim localized $B_0$ inhomogeneities. The multi-coil shimming method, which uses a separate shim coil array placed near the subject, is more effective, but is limited by a tradeoff between signal-to-noise ratio and shimming performance because of the interference between the radiofrequency (RF) and shim coil arrays. To address these issues, a new method termed integrated parallel reception, excitation and shimming (iPRES) was developed in which RF currents and direct currents (DC) are applied in the same coils to enable excitation/reception and localized $B_0$ shimming with a single coil array. In the iPRES(N) method, each RF coil element is further split into N smaller DC loops to enable the shimming of more localized $B_0$ inhomogeneities. However, this method requires N times more power supplies. In the switched iPRES(N) method, switches are further introduced into each RF coil element to direct the DC current generated by a single power supply to the appropriate DC loops, thereby reducing the total number of power supplies required while maintaining a high shimming flexibility. In this thesis, numerical simulations were performed to investigate the shimming performance of different iPRES(N) and switched
iPRES(N) body coil arrays. The objective of this work was to offer guidelines to optimize the design and improve the performance of iPRES coil arrays.

To investigate the shimming performance of iPRES coil arrays, a $B_0$ map was acquired in the abdomen of a human subject on an MRI scanner. Thereafter, different 8-channel iPRES(N) coil arrays were numerically modeled. Each RF coil was split into smaller DC loops in the $x$, $y$ or $z$ directions to provide extra degrees of freedom for shimming. Additionally, simulations were performed for different coil designs with a fixed number of DC loops as well as for different numbers of DC loops with a fixed coil design to find the optimal coil design and number of DC loops, respectively. Furthermore, simulations were performed for a switched iPRES(N) coil array to investigate whether the number of power supplies could be reduced while still providing a similar shimming performance as that of equivalent iPRES(N) coil arrays.

The results demonstrate that the shimming performance of an iPRES(N) coil array increased with the number of DC loops per RF coil. Furthermore, splitting the RF coils in the $z$ direction tended to be more effective at reducing the field inhomogeneity than splitting the RF coils in the $x$ and $y$ directions. Moreover, the shimming performance of an iPRES(N) coil array gradually reached a saturation level when the number of DC loops per RF coil was large enough. Finally, when switches were numerically implemented in an iPRES(4) coil array, the number of power supplies could be reduced from 32 to 8 while keeping the shimming performance similar to that of an equivalent iPRES(4) coil array.
and better than that of an equivalent iPRES(1) coil array. This thesis work can offer valuable guidelines to improve the design and shimming performance of iPRES coil arrays.
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1. Introduction

In magnetic resonance imaging (MRI), changes in magnetic susceptibility at tissue interfaces result in static magnetic field ($B_0$) inhomogeneities, which deteriorate the MRI image quality (Truong et al. (2014)). Magnetic susceptibility variations are common due to the complex structures not just in the brain but also in the abdomen. For example, $B_0$ inhomogeneity in the abdomen leads to various image artifacts, including signal loss, image distortions, and blurring. It can also lead to difficulties in water-fat signal separation and shortened relaxation time which can result in blurred images (Stockmann et al. (2015)). These image artifacts can induce loss of information, or even misdiagnosis. Furthermore, the $B_0$ inhomogeneity and resulting image artifacts become even more pronounced at higher field strength (Han et al. (2013)). Therefore, different methods have been developed to reduce the $B_0$ field inhomogeneity in different structures and to improve the image quality.

1.1 Abdominal MR Imaging

$B_0$ inhomogeneity reduction is important not only in brain imaging but also in abdominal imaging. Abdominal pain related diseases are frequent complaints in many emergency departments (Hei et al. (2016)). MRI, as one of the radiation-free diagnostic methods, is effective in the diagnoses of many abdominal diseases. MRI can play an important role in diseases like acute pancreatitis, choledocholithiasis and acute
appendicitis, which is particularly true for pregnant women diseases because MRI is radiation-free (Hei et al. (2016)).

1.2 Susceptibility-induced $B_0$ Inhomogeneity

Magnetic susceptibility changes result in a different magnetization in different tissues. The relation between the magnetization $M$ and the applied magnetic field $H$ is described in the following equation,

$$ M = \chi H $$

where $\chi$ is the magnetic susceptibility. Therefore, since the tissues in the human body have different susceptibilities, the resultant local magnetic field strength will fluctuate accordingly. For example, the interface between a paramagnetic material and a diamagnetic material induces a change in magnetic field. Because the Larmor precession frequency $\omega$ is related to the field strength $B$ by

$$ \omega = \frac{\gamma}{2\pi} B $$

where $\gamma$ is the gyromagnetic ratio, a magnetic field change causes a shift of the Larmor precession frequency of the spins.

The magnetic resonance signal is defined as:

$$ s(t) = \iint m(x, y) e^{-i\omega_x(x,y)t} e^{-i2\pi[k_x(t)x + k_y(t)y]ot} dx dy $$

where $k$ is the wave number, $m$ is the transverse magnetization, and $\omega_x$ is the inhomogeneity-induced frequency shift. The wave number $k$ is proportional to the magnetic field gradient, meaning that magnetic field inhomogeneity can change the phase
of signal in k-space. This phase error may induce image distortions if no phase correction is made.

1.3 $B_0$ Field Shimming Methods

The techniques used to reduce the $B_0$ field inhomogeneity include passive shimming and active shimming. Both of these methods are intended to generate a magnetic field that is opposite to the main magnetic field ($B_0$) inhomogeneity. Passive shimming involves many permanent magnetic materials or some ferromagnetic metals which can change the distribution of the main magnetic field (You et al. (2013)). Compared with other shimming methods, passive shimming is economical due to the low cost of the required magnetic materials. However, because it relies on the optimal arrangement of such materials, its implementation requires many iterations to measure the field and adjust the position of these materials, which is much more time consuming than active shimming (Han et al. (2013)).

Active shimming involves the application of direct current (DC) in different coils placed around the subject, which can produce a magnetic field to reduce $B_0$ inhomogeneity. By adjusting the current in each coil, the static magnetic field inhomogeneity can be minimized (Guo et al. (2013)). Active shimming is more flexible than passive shimming due to the ability to easily adjust the currents in different coils to shim different subjects.
In the spherical harmonic (SH) shimming method, the $B_0$ inhomogeneity can be written as a linear combination of different SH terms and a specific coil is used to generate the magnetic field corresponding to a specific SH term. The magnetic fields corresponding to the first-order terms ($x, y, \text{ and } z$) are generated by the gradient coils. However, for the second-order terms ($xy, xz, yz, x^2 - y^2, \text{ and } z^2$) and higher-order terms, an additional shim coil is used to generate the magnetic field corresponding to each additional SH term.

In theory, the SH terms can be infinite and their amplitude can be any value, meaning that the SH shimming method can generate any magnetic field. SH shimming is popular due to its many advantages. First, the coils are easy to design and implement, which also provides a good shimming flexibility. Second, the SH shim fields generated by SH shim coils can be diverse compared with passive shim fields due to the orthogonal coil arrangements. Third, with a higher order SH shimming, the shimmed magnetic field is less inhomogeneous. However, in practice, the most commonly used orders are only up to the third order. Higher orders are limited by the diameter of the magnet bore. Therefore, localized field inhomogeneity is hard to shim using SH shimming due to the limited number of coils that can be used (Han et al. (2013)).

To address this limitation, the multi-coil (MC) shimming method, which uses many small local shim coils placed near the subject, has been proposed. MC arrays have been implemented for both mouse brain and human brain imaging at 7T (Juchem et al. (2011a, 2011b)). Compared with the conventional low order SH shimming method, MC
shimming provides a better shimming performance. However, in the MC shimming method, the shim coil array is located between the RF coil array and the subject, which degrades the signal-to-noise ratio (SNR) of the radiofrequency (RF) coils because of the shielding effect from the shim coils. To achieve a better SNR, a compromise was made by inserting a gap in the shim coil array, which decreases the shimming performance.

A new method termed integrated parallel reception, excitation, and shimming (iPRES) was developed to resolve the tradeoff between shimming efficiency and SNR (Han et al. (2013)). This method relies on a new coil design that allows an RF current and a DC current to flow in the same coil simultaneously, thereby enabling parallel RF excitation/reception and localized $B_0$ shimming with a single coil array. Such an integrated RF/shim coil array can thus be placed close to the subject to maximize both the shimming performance and SNR. In addition, this combination saves space in the magnet bore compared to the MC shimming method, which requires two separate RF and shim coil arrays. A 16-channel iPRES coil array was developed for brain imaging and was shown to provide a much more effective shimming of localized $B_0$ inhomogeneities than SH shimming (Truong et al. (2014)). The iPRES method was further extended to the iPRES(N) method (Darnell et al. (2015)) and switched iPRES(N) (Darnell, et al. (2016)). In the iPRES(N) method, the RF coil elements are split into N smaller DC loops to enable the shimming of more localized $B_0$ inhomogeneities, which requires more power supplies. In the switched iPRES(N) method, switches are further introduced into each RF coil element
to direct the DC current generated by a single power supply to the appropriate DC loops, which consequently reduces the total number of power supplies required while maintaining a high shimming flexibility.

By improving the design of iPRES coil arrays, the shimming performance can potentially be further improved. However, building and testing a large number of iPRES coil arrays with different geometries would be very time consuming, costly, and impractical. In this thesis, numerical simulations are performed to compare the shimming performance of a variety of novel iPRES(N) and switched iPRES(N) coil designs. The results of this work can then be used to guide the design and improve the performance of real iPRES coil arrays.
2. Methodology

Figure 2.1 shows a flowchart of the overall methodology used for the numerical simulations. First, a $B_0$ map was acquired in a subject's abdomen to measure the $B_0$ inhomogeneities. Second, the Biot-Savart law, which describes the relation between the current applied in an iPRES coil and the resulting magnetic field, was used to simulate a set of basis $B_0$ maps, representing the $B_0$ field generated by a current of 1 A separately applied in each DC loop of an iPRES coil array. Third, a shimming optimization was performed by minimizing the root-mean-square error (RMSE) between the in vivo $B_0$ map (acquired in step 1) and a linear combination of the basis $B_0$ maps (simulated in step 2) to determine the optimal currents to apply in the iPRES coil array for shimming. Fourth, a
$B_0$ map after shimming was calculated by adding the in vivo $B_0$ map before shimming (from step 1) and the $B_0$ map generated by the iPRES coil array with optimal currents (from step 3). Fifth, the RMSE of the final $B_0$ map (from step 4) was calculated to measure the residual $B_0$ inhomogeneity. Steps 2-5 were repeated for different iPRES(N) and switched iPRES(N) coil designs to compare their shimming performance.

### 2.1 Assumptions

Due to the complexity of correcting the field inhomogeneity using a complex shimming system, some reasonable assumptions have been made. First, the field inhomogeneity was assumed to remain constant over time, meaning that the subject was assumed to be immobile during the scan. Otherwise, dynamic and real time shimming should be applied, which is much more challenging. Second, the DC currents applied in the coils were assumed to be steady, meaning that the generated magnetic field could be modeled by the Biot-Savart law. If the applied current changes with time, an electromagnetic wave will be generated and the Biot-Savart law will not be able to describe the generated magnetic field. In addition, eddy currents resulting from the changing magnetic field will in turn influence the distribution of the magnetic field.

### 2.2 $B_0$ Inhomogeneity Measurement

A $B_0$ map of the abdomen of a healthy volunteer was acquired on a 3T MRI scanner at Duke University. The matrix size was $64 \times 64 \times 30$ and the voxel size was
7.5 mm × 7.5 mm × 7.5 mm. The RMSE of this $B_0$ map was calculated to quantify the $B_0$ inhomogeneity before shimming, as shown in the following equation,

$$RMSE = \sqrt{\frac{1}{n} \sum_{i=1}^{n} (B_i)^2}$$

where $n$ is the number of voxels, and $B_i$ is the magnetic field in the $i^{th}$ voxel. A larger RMSE means that the field is more inhomogeneous.

The area over which the shimming is performed can also influence the shimming performance of iPRES coil arrays. In an MRI image, the voxels within the body generate an MR signal and provide useful information while the voxels in air do not. Without a shimming boundary, the field inhomogeneity in air will also be included in the shim optimization. To eliminate this influence, a boundary needs to be defined to restrict the shim optimization to the body area. An algorithm was developed to contour the body boundaries. All of the voxels inside the boundaries were included in the shim optimization while the area outside the boundaries was excluded.

![Body boundary contouring process.](image)

**Figure 2.2: Body boundary contouring process.**
Theoretically, the boundaries should match the edges of the body perfectly. In reality, if the $B_0$ map is simply thresholded to define the boundaries, the contoured area may be slightly smaller than the body area. To solve this problem, a three-step method was applied (Figure 2.2). First, a rough contour was defined based on the absolute value of the $B_0$ map using an initial threshold. Any value in the $B_0$ map that was above the threshold was set to 1 and any value that was below the chosen threshold was set to 0. As is shown in figure 2.3, the resulting edges may be inside the body because the pixel values in the body fluctuate around the chosen threshold. To eliminate these edges inside the body area, the rough contour was updated depending on its pixel values. For example, if the summation of all four surrounding pixels was larger than 3, the center pixel value was

![Graph](image)

**Figure 2.3:** Threshold based edge contour, the edges appear in the body area because the pixel values were smaller than the threshold.
changed to 1. Then the profile of the MRI anatomical image was compared with the profile of the rough contour. In theory, the boundaries of anatomical structures should match the boundaries of the contour, meaning that their size and shape should be similar. If these two profiles matched, the threshold was adopted. Otherwise, the threshold was updated and tested again. Figure 2.4 illustrates the boundary information extracted from $B_0$ map.

### 2.3 Basis $B_0$ Maps

Before shimming the $B_0$ inhomogeneities, the basis $B_0$ maps of different iPRES coil arrays were simulated using the Biot-Savart law. The magnetic field $B$ generated by a coil is related to the DC current $I$ by the Biot-Savart law,

$$B(r, t) = \frac{\mu_0 I(t)}{4\pi} \int \frac{dI \times (r - r')}{|r - r'|}$$

where $\mu_0$ is the permeability of vacuum and $\mu_0 = 4\pi \times 10^{-7} \text{H/m}$, $dl$ is the current element, $r'$ is the point of measurement off the current element and $r$ is the vector of the current.
element. The Biot-Savart law is suitable for the calculation of a magnetic field generated by a static current (Ruppeiner et al. (1995)). Given one coil with a DC current, the magnetic field at a point \( r \) off the coil can be calculated by integrating all of the current elements in the coil. The precision of the integration depends on the integral length. The integral length used in this thesis work was 0.5 cm. By reducing the integral length, the integrated field will be more precise.

### 2.4 Shimming Optimization

In this work, the dynamic shimming method was employed, meaning that the currents to apply in the shim coils were optimized for every slice separately (Sengupta et al. (2011)). Furthermore, to minimize the \( B_0 \) gradient in the \( z \) direction in addition to the \( B_0 \) gradients in the \( x \) and \( y \) directions, the slices above and below the slice of interest were also included in the shim optimizations instead of using only one slice.

To shim the field inhomogeneity, an optimization method that minimizes the RMSE between the \( B_0 \) map acquired in vivo and a linear combination of the simulated basis \( B_0 \) maps (Truong et al. (2014)) was used. An objective function was constructed to minimize the field inhomogeneity. Specifically, the inputs of the objective function were the \( B_0 \) map acquired in vivo and the simulated basis \( B_0 \) maps, the variables were the DC currents applied in the DC loops of an iPRES coil array and the output was the RMSE of the shimmed magnetic field. The following equations describe the objective function for this numerical simulation,

\[
\text{RMSE} = \sqrt{\sum (B_{0_{\text{actual}}} - B_{0_{\text{simulated}}})^2}
\]

where \( B_{0_{\text{actual}}} \) is the actual \( B_0 \) map, and \( B_{0_{\text{simulated}}} \) is the simulated \( B_0 \) map.
\[
B^i_{\text{coil}} = \frac{\mu_0 I_i}{4\pi} \sum_{j=1}^{k} \frac{dI \times (r_j - r)}{|r_j - r|}
\]

\[
RMSE' = \sqrt{\frac{1}{n} \sum (B_0 + \sum_{i=1}^{m} B^i_{\text{coil}})^2}
\]

where \(B^i_{\text{coil}}\) is the magnetic field generated by the \(i^{th}\) coil, \(RMSE'\) is the RMSE of the shimmed magnetic field, each RF coil is divided into \(k\) elements, and \(m\) coils are used in one coil array. The input currents can be optimized by minimizing the RMSE of the shimmed magnetic field using a gradient-based method. For example, \(RMSE_1\) can be calculated if the current \(I\) is given,

\[
I = \begin{pmatrix}
   I_{\text{coil1}} \\
   I_{\text{coil2}} \\
   I_{\text{coil3}} \\
   I_{\text{coil4}} \\
   I_{\text{coil5}} \\
   \vdots
\end{pmatrix} \Rightarrow RMSE_1
\]

where \(I_{\text{coil}}\) is the current applied in \(i^{th}\) DC loop. Thereafter, the currents can be shifted by a small step, which will produce a new shimmed field with an inhomogeneity of \(RMSE_2\),

\[
I + \Delta I = \begin{pmatrix}
   I_{\text{coil1}} \\
   I_{\text{coil2}} \\
   I_{\text{coil3}} \\
   I_{\text{coil4}} \\
   I_{\text{coil5}} \\
   \vdots
\end{pmatrix} + \begin{pmatrix}
   \Delta I_1 \\
   \Delta I_2 \\
   \Delta I_3 \\
   \Delta I_4 \\
   \Delta I_5 \\
   \vdots
\end{pmatrix} \Rightarrow RMSE_2
\]

where \(\Delta I_i\) is the current increment in the \(i^{th}\) DC loop. Then the gradient can be calculated using \(RMSE_1\) and \(RMSE_2\).
\[
g = \begin{bmatrix}
\frac{\partial \text{RMSE}}{\partial I_{c11}} \\
\frac{\partial \text{RMSE}}{\partial I_{c12}} \\
\frac{\partial \text{RMSE}}{\partial I_{c13}} \\
\frac{\partial \text{RMSE}}{\partial I_{c14}} \\
\frac{\partial \text{RMSE}}{\partial I_{c15}} \\
\end{bmatrix} = \begin{bmatrix}
\Delta I_1 \\
\Delta I_2 \\
\Delta I_3 \\
\Delta I_4 \\
\Delta I_5 \\
\end{bmatrix}
\]

where \( g \) is the gradient. In order to minimize RMSE, the current vector is changed along the steepest direction based on the calculated gradient, as shown in the following equation,

\[
l_{\text{new}} = l + k \cdot g
\]

where \( k \) is a constant for adjusting the step length. Depending on the predefined constraints on the iteration, the solver stops automatically if the required precision is achieved or if the maximum number of iteration is reached.

Initially, a random current that follows a uniform distribution is generated for each DC loop. The optimization starts from these currents and optimizes the currents step by step to minimize the field RMSE. However, the solver may be trapped in local minima if the currents are not properly initialized. Especially when the number of independent DC loops is large, the number of local minima may increase due to the increase in possible combinations. To address this problem, each optimization is repeated several times with different initial values. The returned RMSEs from these repeated optimizations are compared to get the smallest one, which is then treated as the optimized solution.
2.5 iPRES Coil Designs

In this work, numerical simulations were performed for an 8-channel body coil array. Each RF coil was an iPRES(N) or switched iPRES(N) coil with different designs.

2.5.1 iPRES(N) Coil Designs

Each RF coil was modeled as a $14 \times 20$ cm rectangle while the DC loops were modeled as subdivisions of this rectangle. Based on the number of DC loops in one RF coil, the iPRES coils are named differently. For example, if an RF coil is split into two smaller DC loops, the name of this coil is iPRES(2). Similarly, if an RF coil is split into three DC loops, the name is iPRES(3). Figure 2.5 shows some examples of iPRES(N) coils.

The RF coils can be split in many different ways. Depending on the needs, the number of degrees of freedom for shimming can be increased either in the transverse

![Diagram of iPRES(N) coil designs](image)

Figure 2.5: Categories of iPRES(N) coil designs.
plane or along the z direction. For example, if an iPRES(2) coil is split in the x and y directions, the shimming performance will be improved along these directions but not along the z direction. Therefore, more divisions in all directions are preferred to improve the shimming performance. However, while the shimming performance should increase with the number of independent DC loops, the complexity of the numerical simulation will also be increased. The number of possible iPRES(N) coil designs increases with N at a non-linear rate, which consequently increases the difficulty of optimizing the currents. The solver can be trapped in local minima instead of finding the globally best solution. In addition, the shimming performance may improve more slowly when the RF coils are split into smaller and smaller DC loops. To address this tradeoff, the relationship between the number of DC loops (N) and the shimming performance was investigated by simulating iPRES(1) to iPRES(5) coils, as shown in figure 2.6, in order to find the optimal number of divisions.

![Figure 2.6](image)

**Figure 2.6:** iPRES(1) to iPRES(5).
The 8-channel iPRES coil array used in the simulations was modeled according to the experimental 8-channel iPRES body coil array implemented at Duke University. The arrangement of the iPRES coils can influence the shimming performance. Modelling the iPRES coils requires detailed information about the position of the coil array relative to the body. Ideally, the iPRES coil array should be placed around the body as close as possible to maximize the SNR. However, the size and shape of the abdomen vary from person to person. As a consequence, the arrangement of the shimming coils would change with the shape of the human body. In addition, the abdomen is deformable, which further increases the difficulty of localizing the coils. Subject movement during the MRI scan would also change the shimming performance. To address these issues, a fixed semi-circular frame was employed to support and immobilize the iPRES coils while still keeping them relatively close to the subject. The size of the frame can be adjusted according to the size of different subjects. As shown in figure 2.7, the frame used in the

![Figure 2.7: A schematic of the frame and a water phantom. The radius of the half circle is 25 cm and the width of the frame is 20 cm.](image)
simulations was the same frame as used for the experimental iPRES coil array. Figure 2.7a shows the computer modeled frame with the 8-channel iPRES coil array attached to it and figure 2.7b shows the same frame with a water phantom inside. Figure 2.8 shows the designs of iPRES(1) to iPRES(4) coil arrays. The iPRES(3) coil array was designed to match the experimental iPRES(3) coil array.

The $B_0$ map acquired in vivo was aligned such that the subject’s abdomen was in the center of these iPRES coil arrays with the bottom part of the body laying on the frame. To register the MRI images with the virtual iPRES coil arrays, a registration algorithm was developed which employed the body contours from previous steps (section 2.2). The binary data in the contour matrices were added in both the x and y directions to get two one-dimension arrays. The generated arrays were analyzed in terms of symmetry. The

![iPRES 1, iPRES 2, iPRES 3, iPRES 4](image)

**Figure 2.8:** A schematic graph of 8-channel iPRES coil arrays. According to the number of DC loops in one RF coil, these coil arrays are named iPRES(1) - iPRES(4) respectively.
point of intersection of two axes of symmetry was regarded as the center of the body. Thereafter, as a preparation for the shim optimization, the body center was registered with the center of the coil arrays.

2.5.2 Switched iPRES(N) Coil Designs

Numerical simulations were also performed for a switched iPRES(4) coil design. In this case, each switched iPRES coil is only powered by one power supply and switches are used to direct the current into the appropriate DC loops, depending on the $B_0$ inhomogeneity pattern. For example, a switched iPRES(2) coil can have four different current flow patterns, as shown in figure 2.9. Since different coils in the coil array can use different patterns, the shim optimization needs to determine not only the optimal DC currents, but also the optimal patterns to use in each coil. For an 8-channel switched iPRES(2) coil array, the shim optimization would need to be repeated $4^8 = 65536$ times, each time with a different set of 8 basis $B_0$ maps. Since a switched iPRES(4) coil can have 40 different current flow patterns, the shim optimization for an 8-channel switched iPRES(4) coil array would need to be repeated $40^8 = 6.5536 \times 10^{12}$ times, which is completely impractical. Therefore, a modified three-step shim optimization method was developed. In the first step, the shim optimization was run for an 8-channel iPRES(4) coil array with no switches (i.e., with 32 basis $B_0$ maps). In the second step and for each of the 8 coils, the RMSE was calculated between the $B_0$ map generated by the optimal current obtained in step 1 and the basis $B_0$ maps corresponding to each possible pattern of the
switched iPRES(4) coil. The pattern with the smallest RMSE was then chosen as the optimal pattern. In the third step, the shim optimization was run again, but this time using only the 8 basis $B_0$ maps corresponding to the 8 optimal patterns obtained in the previous step in order to determine the optimal DC currents. The rationale for doing this is that the iPRES(4) coil array with no switches should provide the best solution, which can be used to determine the optimal patterns for the switched iPRES(4) coil array. Finally, the field inhomogeneities were shimmed using the magnetic field generated by the optimal patterns and optimal DC currents calculated in the two previous steps and the RMSE of the shimmed field was calculated to quantify the residual field inhomogeneity.

![Four possible current flow patterns in a switched iPRES(2) coil.](image)

Figure 2.9: Four possible current flow patterns in a switched iPRES(2) coil.
3. Results

3.1 $B_0$ Inhomogeneity Analysis

An inhomogeneity analysis was performed on the $B_0$ map acquired in the subject. Figure 3.1 shows the inhomogeneity distributions in 4 slices and figure 3.2 shows the corresponding anatomical images. As expected, the $B_0$ inhomogeneity varied in different parts of the abdomen because of the susceptibility effect.

Representative profiles of field inhomogeneity are shown in figure 3.3. These profiles illustrate that the inhomogeneity is primarily located close to the body-air interfaces, implying that if the shimming coils can be placed closer to the body, the shimming performance should be better.

![Figure 3.1: $B_0$ field inhomogeneity without shimming (Hz).](image)
Figure 3.2: Anatomical images in the same slices as in Figure 3.1.

Figure 3.3: $B_0$ inhomogeneity profiles.
To quantify the $B_0$ inhomogeneity, the RMSEs of all slices were calculated and are shown in figure 3.4 and listed in Table 1 in the appendix. The RMSEs of the $B_0$ field without shimming range from 38.79 Hz to 57.12 Hz.

![Figure 3.4: RMSE of the $B_0$ field without shimming.](image)

**3.2 Algorithm Parameter Settings**

On the one hand, if the number of function evaluations in the shim optimization increases, the time required to finish running the program also increases. On the other hand, a larger number of function evaluations can potentially provide a better optimization. To address this tradeoff, this algorithm parameter was optimized before the numerical simulations. Four 8-channel iPRES coil arrays, namely iPRES(1) to iPRES(4), were used to find the optimal parameter. The number of function evaluations was increased from 1000 to 28000. The relation between the improvement in RMSE and the number of function evaluations is shown in figure 3.5. The RMSE decreased when the
number of function evaluations increased. However, the RMSE decreased at a slower speed when the number of function evaluations increased to 15000, implying that the optimal parameter is around 15000. Therefore, the number of function evaluations was set to 20000 for iPRES(N) when $N \leq 4$. However, this number was further increased to 30000 when $N > 4$.

![Figure 3.5: Relation between number of function evaluations and relative RMSE.](image)

### 3.3 Comparison between Experimental and Simulated Basis $B_0$ maps

The simulated basis $B_0$ maps for the 8-channel iPRES(3) coil array shown in figure 2.8 were compared to basis $B_0$ maps experimentally acquired with an equivalent iPRES coil array at Duke University on a phantom filled with water. The experimental and simulated basis $B_0$ maps are shown in figures 3.6a and 3.6b respectively. These basis $B_0$ maps
maps show that the simulated result and experimental measurement are similar. The experimental $B_0$ maps only show the $B_0$ field inside the phantom while the simulated $B_0$ maps show the whole $B_0$ field.

![Figure 3.6: Four representative basis $B_0$ maps for an 8-channel iPRES(3) coil array experimentally acquired on a phantom (a) or numerically simulated (b).](image)

### 3.4 Eight-channel iPRES(1) to iPRES(4) Coil Designs

The absolute and relative RMSE of the shimmmed field obtained using the iPRES(1) to iPRES(4) coil designs shown in figure 2.8 are illustrated in figure 3.7 and 3.8. The relative RMSE is the ratio of the RMSE after shimming to the RMSE before shimming. Representative $B_0$ maps with and without shimming are shown in figure 3.9. The iPRES(4) coil array can reduce the RMSE by more than 40%. A maximum difference of 22.2% can be observed between the iPRES(4) and iPRES(1) RMSE.
Figure 3.8: Absolute RMSE improvement.

Figure 3.7: Percentage RMSE improvement.
The shimming performance of a given iPRES coil array can be influenced by both the slice position (relative to the coils) and the $B_0$ inhomogeneity in this slice. To investigate the relation between shimming performance and slice position independently of the $B_0$ inhomogeneity, the same slice in the $B_0$ map was replicated into 15 different slice positions (specifically 1 to 15). The resulting dataset was then shimmed using the same iPRES(1) to iPRES(4) coil arrays. Due to the symmetry of the iPRES coil arrays in the z direction, the results for slice positions 16 to 30 are identical to the results for slice position 15 to 1, respectively. Tables 2 to 5 in the appendix show the field RMSE before and after shimming. Figure 3.10 demonstrates the RMSE improvement of 6 slice replications (specifically, slices 5, 10, 15, 20, 25, and 30) using different coil arrays. The RMSEs in all 6 graphs show a similar trend and an obvious improvement can be seen between iPRES(1)-
(2) and iPRES(3)-(4), especially for slice positions 1 to 7. However, the difference gradually diminishes for slice positions 8 to 15. The average RMSE improvement from iPRES(1) to iPRES(4) was 20% in slice positions 1-7 (figure 3.11).

![Graphs showing relative RMSE improvement for different slice positions](image)

**Figure 3.10: Relative RMSE improvement of 6 different slice replications.**

In all of the previous simulations, 40 different sets of initial DC currents were used in the shim optimization for the purpose of finding the global minima. In order to evaluate
the influence of these initial DC currents on the results, figure 3.12 illustrates the variations in the optimized RMSE obtained when the program was initiated with different currents. The different initial currents introduced the large variations in the shimming performance of different iPRES(N) coil arrays, which demonstrates the need for repeating the shim optimization multiple times with different initial currents.

Figure 3.12: Optimization variation with different initial currents.
3.5 Additional 8-channel iPRES(4) Coil Designs

A coil can be split into four DC loops in different ways. Depending on the $B_0$ field inhomogeneity patterns, these different designs can result in different shimming performances. However, the abdomen is generally similar across different subjects in terms of the inner structures and tissue elements. Therefore, different iPRES(4) coil designs were simulated to find a suitable design for the abdomen. Figure 3.13 shows four ways to split an iPRES (4) coil array. In plan 1, the RF coils are split in the z direction. In plan 2, they are split in the x and y directions. In plan 3, they are split along the diagonals. In plan 4, they are split into four quadrants, as before.

![Image of coil designs]

Figure 3.13: Four 8-channel iPRES(4) coil array designs.
Figures 3.14 and 3.15 illustrate the RMSE improvement obtained with the four iPRES(4) coil arrays. Plan 1 generally resulted in the best shimming performance and plan
2 generally resulted in the worst shimming performance.

### 3.6 Eight-channel iPRES(1) to iPRES(5) Coil Designs

The simulation results of section 3.4 show that increasing the number of DC loops generally resulted in a better shimming performance. However, if the DC loops are small enough, further dividing the RF coils may not be necessary. Therefore, eight-channel iPRES(1) to iPRES(5) coil arrays were simulated to investigate the relation between shimming performance and number of DC loops per RF coil and to find the optimal size of the DC loops. The divisions were made in the z direction, as shown in figure 3.16. The simulation results are shown in figure 3.17, which was the average RMSE over 30 slices. The RMSE decreased if the number of DC loops increased in the z direction. However, it only decreased slightly from iPRES(4) to iPRES(5).

![Figure 3.16: iPRES(1) to iPRES(5) coil arrays.](image)
3.7 Eight-channel Switched iPRES(4) Coil Design

In theory, a switched iPRES(4) coil array should have a better shimming performance than an equivalent iPRES(1) coil array because it has more degrees of freedom for shimming. On the other hand, it cannot have a better shimming performance than an equivalent iPRES(4) coil array with no switches but with the same geometry because it only has one power supply per RF coil rather than four and hence only enables one DC current per RF coil. Therefore, the shimming performance of the switched iPRES(4) coil array is expected to be between those of the iPRES(1) and iPRES(4) coil.
arrays, but it is not known exactly where in-between. Simulations were thus performed on these three coil arrays to investigate whether the switched iPRES(4) coil array could still achieve a shimming performance close to that of the iPRES(4) coil array, while using the same number of power supplies as the iPRES(1) coil array, which would demonstrate the advantages of the switched iPRES(N) method. Figures 3.18 and 3.19 illustrate the simulation results of the switched iPRES(4) coil array. Table 6 in the appendix shows that the RMSE improvement of switched iPRES(4) is comparable to that of iPRES(3) in the previous simulation. The average RMSE improvement over 30 slices was 24.5%, 30.77%, 33.84%, and 32.3% for the iPRES(1), iPRES(3), iPRES(4), and switched iPRES(4) coil arrays, respectively. For the iPRES(1) coil array, the maximum and minimum RMSE reduction

![Figure 3.18: Absolute RMSE improvement of iPRES coil arrays.](image)

34
were 37.5% and 2.5% respectively. For the switched iPRES(4) coil array, the maximum and minimum RMSE reduction was 42.9% and 15.7%, respectively. The shimming performances of iPRES(3) and switched iPRES(4) are compared in figure 3.20.

Figure 3.19: Relative RMSE improvement of iPRES coil arrays.
Figure 3.20: Shimming performance of iPRES(3) and switched iPRES(4)
4. Discussion

In this research work, the shimming performance of different iPRES coil arrays was investigated by performing simulations based on a $B_0$ map acquired in the abdomen. The RMSE of the shimmed field was significantly reduced when using different iPRES(N) coil designs. Furthermore, a reduction in the number of power supplies was achieved while maintaining a similar shimming performance when using a switched iPRES(N) coil array. These simulations were performed based on one subject data, meaning that these simulation results may be different if the subjects’ $B_0$ maps are significantly different. In addition, these simulations were performed by shimming a region including the whole body in each slice, but the results may be different and the shimming performance likely better if simulations were performed by shimming only a smaller region-of-interest (e.g., a specific organ), which may be more relevant in some applications.

4.1 Local Minima

The gradient-based optimization method has one limitation that the solver may be trapped in local minima. Figure 3.12 shows that the RMSE of the shimmed $B_0$ field fluctuated when different initial currents were used. The variation in RMSE could be up to 20% with different initial currents. To reduce this variation, the simulations in this thesis work were repeated 40 times with different initial DC currents. The smallest RMSE among the resulting 40 RMSEs was recorded as the optimal solution. With more initial
currents, the optimizations tended to produce a better shimming performance but the time required to finish running the program was longer.

**4.2 Shimming Performance of iPRES(1) to iPRES(4) Coil Designs**

The average reduction in $B_0$ inhomogeneity over all slices was 23.5%, 25.6%, 30.8% and 33.8% for iPRES(1) to iPRES(4), respectively. As shown in figures 3.7 and 3.8, the RMSE was smaller for iPRES(3) and iPRES(4) than for iPRES(1) and iPRES(2) because the RF coils were split in the $z$ direction, which provided more degrees of freedom for shimming in that direction. The RMSE improvement from iPRES(1) to iPRES(2) and from iPRES(3) to iPRES(4) shows that splitting the RF coils in the $x$ and $y$ directions could also increase the number of degrees of freedom for shimming in these directions. Furthermore, the percentage RMSE improvement was larger for slices 7 to 12 and slices 20 to 25 for all the simulated iPRES coil arrays. The iPRES(1) and iPRES(2) coil arrays could not shim the central slices because they cannot generate any magnetic field in these slices. In contrast, the iPRES(3) and iPRES(4) coil arrays, which were split in the $z$ direction, were able to improve the shimming performance in the central slices. Though the RMSE improvement from iPRES(1) to iPRES(4) was 10.3% on average, the shimming performance varied with different slice positions. The maximum RMSE difference between iPRES(1) and iPRES(4) was 30.5% and the smallest RMSE difference was only 2.5%.
4.3 Additional iPRES(4) Coil Designs

The RF coils can be split in many different ways. Splitting the RF coils in a particular direction (x, y, or z) increases the number of degrees of freedom for shimming in that direction. Among the four iPRES(4) coil arrays, plan 1 offered the best shimming performance while plan 2 was less effective, which shows that divisions in the z direction are generally more effective than divisions in the x and y directions. However, this conclusion depends on the subject data. For example, plan 3 and plan 4 may be more suitable to shim $B_0$ inhomogeneities that are more isotropic because the divisions of the RF coils are in all three directions. Furthermore, although each RF coil in the iPRES coil arrays simulated in this work had the same design, it would also be possible to split each coil differently based on the inhomogeneity distribution, which is particularly useful when the $B_0$ inhomogeneity can be predicted.

4.4 iPRES(1) to iPRES(5) Coil Designs

Figure 3.17 illustrates the relation between the RMSE improvement and the number of divisions in the z direction. When the RF coils were only split in the z direction, resulting in iPRES(1) to iPRES(5) coil arrays, an evident decrease could be observed from iPRES(1) to iPRES(3) as well as an additional reduction of 0.5 Hz from iPRES(3) to iPRES(4). However, the average shimming performance of iPRES (4) was nearly the same as that of iPRES(5), suggesting that the optimal number of divisions for this coil design is 4. However, this optimal number of divisions may change with the design of the iPRES
coil arrays. Specifically, if the size and shape of iPRES coil arrays are changed in order to shim different field inhomogeneities, the optimal number of divisions may be different.

4.5 Eight-channel Switched iPRES(4) Coil Design

Compared with the 8-channel iPRES(1) coil array, the 8-channel switched iPRES(4) coil array could evidently reduce the $B_0$ inhomogeneity. For switched iPRES(4), the maximum RMSE improvement was 37.8% relative to iPRES(1) while the number of power supplies was the same for both coil arrays. The shimming performance of switched iPRES(4) was similar to that of iPRES(3). However, iPRES(3) coil array requires 24 independent power supplies while the switched iPRES(4) coil array only needs 8 independent power supplies. These results show that the cost of an iPRES coil array can be reduced by reducing the number of DC power supplies while maintaining a similar shimming performance.
5. Conclusions

The simulations of iPRES(1) to iPRES(4) coil arrays confirmed that iPRES(N+1) has a better shimming performance than iPRES(N), as expected, but they also provided useful information on how the different ways of splitting the RF coils can influence the shimming performance. Specifically, splitting them in the z direction tended to be more effective than splitting them in the x and y directions. Additionally, these simulations were also useful to determine the optimal coil design for a given number of DC loops per RF coil or the optimal number of DC loops per RF coil for a given coil design. Finally, the simulations of the switched iPRES(4) coil array showed that it could achieve a similar shimming performance as that of an iPRES(3) coil array while only using the same number of power supplies as an iPRES(1) coil array. Moreover, this work can be applied to improve the shimming performance of other types of iPRES coil arrays such as for brain imaging. Further work in optimizing the iPRES coil designs and improving the shim optimization algorithm is needed to further improve the shimming performance of iPRES coil arrays. In addition, simulations based on multiple subjects’ data are needed to fully evaluate the iPRES coil arrays.
Appendix

Table 1: B₀ inhomogeneity before shimming.

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Table 2: B₀ inhomogeneity before shimming and after dynamic shimming with the 8-channel iPRES(1) coil array

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RMSE_r is the RMSE of field without shimming, RMSE_s is the RMSE of shimmmed field, and η is the percentage improvement.
Table 3: $B_0$ inhomogeneity before shimming and after dynamic shimming with the 8-channel iPRES(2) coil array

<table>
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<tr>
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</thead>
<tbody>
<tr>
<td>RMSE$_r$ (Hz)</td>
<td>53.50</td>
<td>51.42</td>
<td>50.27</td>
<td>49.06</td>
<td>47.66</td>
<td>47.76</td>
<td>47.03</td>
<td>46.24</td>
<td>45.83</td>
<td>44.52</td>
</tr>
<tr>
<td>RMSE$_s$ (Hz)</td>
<td>46.19</td>
<td>45.52</td>
<td>44.52</td>
<td>41.11</td>
<td>37.97</td>
<td>36.42</td>
<td>34.11</td>
<td>31.44</td>
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<td>11.5%</td>
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<td>34.8%</td>
<td>35.4%</td>
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<td>38.79</td>
<td>39.74</td>
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<td>42.24</td>
<td>42.94</td>
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<td>33.7%</td>
<td>30.4%</td>
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<td>6.0%</td>
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<td>16.0%</td>
<td>24.7%</td>
<td>28.5%</td>
<td>31.3%</td>
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<td>56.55</td>
<td>54.74</td>
<td>55.85</td>
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<td>RMSE$_s$ (Hz)</td>
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<td>32.67</td>
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<td>36.48</td>
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<td>34.44</td>
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<td>40.2%</td>
<td>38.6%</td>
<td>36.2%</td>
<td>31.2%</td>
<td>25.4%</td>
<td>19.6%</td>
<td>18.3%</td>
<td>22.0%</td>
<td>37.5%</td>
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</table>

RMSE$_r$ is the RMSE of field without shimming, RMSE$_s$ is the RMSE of shimmed field, and $\eta$ is the percentage improvement.
Table 4: $B_0$ inhomogeneity before shimming and after dynamic shimming with the 8-channel iPRES(3) coil array

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<tbody>
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<td>53.50</td>
<td>51.42</td>
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<td>47.76</td>
<td>47.03</td>
<td>46.24</td>
<td>45.83</td>
<td>44.52</td>
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<td>RMSE$_s$ (Hz)</td>
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<td>37.77</td>
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<td>24.9%</td>
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<td>35.4%</td>
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<td>36.0%</td>
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<tbody>
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<td>39.85</td>
<td>38.79</td>
<td>39.74</td>
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<td>42.24</td>
<td>42.94</td>
<td>43.40</td>
<td>46.10</td>
<td>48.65</td>
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<td>31.06</td>
<td>31.33</td>
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<td>34.2%</td>
<td>33.7%</td>
<td>28.2%</td>
<td>11.1%</td>
<td>5.2%</td>
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<td>26.9%</td>
<td>32.6%</td>
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<tbody>
<tr>
<td>RMSE$_r$ (Hz)</td>
<td>52.40</td>
<td>55.08</td>
<td>56.26</td>
<td>53.20</td>
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<td>53.01</td>
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<td>55.85</td>
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<td>31.40</td>
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<td>37.8%</td>
<td>39.9%</td>
<td>39.8%</td>
<td>38.6%</td>
<td>37.1%</td>
<td>34.0%</td>
<td>29.7%</td>
<td>27.5%</td>
<td>27.9%</td>
</tr>
</tbody>
</table>

RMSE$_r$ is the RMSE of field without shimming, RMSE$_s$ is the RMSE of shimmed field, and $\eta$ is the percentage improvement.
Table 5: B₀ inhomogeneity before shimming and after dynamic shimming with the 8-channel iPRES(4) coil array

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<th>10</th>
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</thead>
<tbody>
<tr>
<td>RMSE_r (Hz)</td>
<td>53.50</td>
<td>51.42</td>
<td>50.27</td>
<td>49.06</td>
<td>47.66</td>
<td>47.76</td>
<td>47.03</td>
<td>46.24</td>
<td>45.83</td>
<td>44.52</td>
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<tr>
<td>RMSE_s (Hz)</td>
<td>39.91</td>
<td>37.04</td>
<td>34.10</td>
<td>31.49</td>
<td>29.31</td>
<td>29.34</td>
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<td>28.78</td>
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<td>28.0%</td>
<td>32.2%</td>
<td>35.8%</td>
<td>38.5%</td>
<td>38.6%</td>
<td>39.7%</td>
<td>37.8%</td>
<td>36.4%</td>
<td>36.7%</td>
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<tbody>
<tr>
<td>RMSE_r (Hz)</td>
<td>42.17</td>
<td>39.85</td>
<td>38.79</td>
<td>39.74</td>
<td>41.50</td>
<td>42.24</td>
<td>42.94</td>
<td>43.40</td>
<td>46.10</td>
<td>48.65</td>
</tr>
<tr>
<td>RMSE_s (Hz)</td>
<td>26.83</td>
<td>25.44</td>
<td>23.76</td>
<td>26.56</td>
<td>34.90</td>
<td>37.55</td>
<td>32.38</td>
<td>28.97</td>
<td>29.91</td>
<td>31.20</td>
</tr>
<tr>
<td>η</td>
<td>36.4%</td>
<td>36.2%</td>
<td>38.7%</td>
<td>33.2%</td>
<td>15.9%</td>
<td>11.1%</td>
<td>24.6%</td>
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<th>30</th>
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<tbody>
<tr>
<td>RMSE_r (Hz)</td>
<td>52.40</td>
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<td>56.26</td>
<td>53.20</td>
<td>51.16</td>
<td>53.01</td>
<td>57.12</td>
<td>56.55</td>
<td>54.74</td>
<td>55.85</td>
</tr>
<tr>
<td>RMSE_s (Hz)</td>
<td>33.26</td>
<td>34.29</td>
<td>33.70</td>
<td>31.81</td>
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<td>31.41</td>
<td>35.69</td>
<td>37.79</td>
<td>38.41</td>
<td>39.56</td>
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<td>37.7%</td>
<td>40.1%</td>
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<td>40.8%</td>
<td>40.7%</td>
<td>37.5%</td>
<td>33.2%</td>
<td>29.8%</td>
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</tr>
</tbody>
</table>

RMSE_r is the RMSE of field without shimming, RMSE_s is the RMSE of shimmmed field, and η is the percentage improvement.
Table 6: $B_0$ inhomogeneity of 6 slice replications before and after dynamic shimming with the iPRES(1) to iPRES(4) coil arrays

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<td>44.51</td>
<td>44.51</td>
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<td>44.51</td>
<td>44.51</td>
<td>44.51</td>
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<tr>
<td>RMSE1 (Hz)</td>
<td>36.03</td>
<td>37.76</td>
<td>37.81</td>
<td>36.36</td>
<td>35.04</td>
<td>32.85</td>
<td>31.05</td>
<td>30.18</td>
</tr>
<tr>
<td>RMSE2 (Hz)</td>
<td>34.52</td>
<td>36.42</td>
<td>36.38</td>
<td>35.04</td>
<td>33.33</td>
<td>31.29</td>
<td>29.59</td>
<td>28.65</td>
</tr>
<tr>
<td>RMSE3 (Hz)</td>
<td>30.97</td>
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<td>30.63</td>
<td>29.78</td>
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<td>27.74</td>
<td>27.18</td>
<td>27.53</td>
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<td>RMSE4 (Hz)</td>
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<td>26.91</td>
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<td>44.51</td>
<td>44.51</td>
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<td>30.47</td>
<td>31.01</td>
<td>32.62</td>
<td>35.25</td>
<td>40.84</td>
</tr>
<tr>
<td>RMSE2 (Hz)</td>
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<td>32.29</td>
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<td>28.69</td>
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<td>26.81</td>
<td>28.17</td>
<td>31.44</td>
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</table>

RMSE1 is the iPRES(1) shimmmed field RMSE, RMSE2 is the iPRES(2) shimmmed field RMSE, RMSE3 is the iPRES(3) shimmmed field RMSE, RMSE4 is the iPRES(4) shimmmed field RMSE.
Table 7: $B_0$ inhomogeneity after dynamic shimming with the iPRES(1), switched iPRES(4), and iPRES(4) coil arrays

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<td>38.53</td>
<td>35.53</td>
<td>32.30</td>
<td>30.07</td>
<td>28.77</td>
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<td>38.29</td>
<td>34.44</td>
<td>31.74</td>
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<td>31.28</td>
<td>29.41</td>
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<td>39.62</td>
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<td>32.76</td>
<td>33.53</td>
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<td>29.26</td>
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<td>29.48</td>
<td>31.47</td>
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<td>42.47</td>
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<td>46.21</td>
<td>44.63</td>
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<td>32.97</td>
<td>31.49</td>
<td>32.07</td>
<td>37.01</td>
<td>39.32</td>
<td>40.36</td>
<td>41.89</td>
</tr>
</tbody>
</table>

RMSE_1 is the RMSE of iPRES(1) shimmmed field, RMSE_s4 is the RMSE of switched iPRES (4) shimmmed field, and RMSE_4 is the RMSE of iPRES (4) shimmmed field.
References


