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Integrated Parallel Reception, Excitation, and Shimming (iPRES) with multiple shim loops per RF coil element for improved B_0 shimming

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Abstract

Purpose—Integrated parallel reception, excitation, and shimming (iPRES) coil arrays allow radio-frequency (RF) currents and direct currents (DC) to flow in the same coils, which enables excitation/reception and localized B_0 shimming with a single coil array. The purpose of this work was to improve their shimming performance by adding the capability to shim higher-order local B_0 inhomogeneities that are smaller than the RF coil elements.

Methods—A novel design was proposed in which each RF/shim coil element is divided into multiple DC loops, each using an independent DC current, to increase the number of magnetic fields available for shimming while maintaining the signal-to-noise ratio (SNR) of the coil. This new design is termed iPRES(N), where N represents the number of DC loops per RF coil element. Proof-of-concept phantom and human experiments were performed with an 8-channel body coil array to demonstrate its advantages over the original iPRES(1) design.

Results—The average B_0 homogeneity in various organs before shimming and after shimming with the iPRES(1) or iPRES(3) coil arrays was 0.24, 0.11, and 0.05 ppm, respectively. iPRES(3) thus reduced the B_0 inhomogeneity by 53% and further reduced distortions in echo-planar images of the abdomen when compared to iPRES(1).

Conclusion—iPRES(N) can correct for localized B_0 inhomogeneities more effectively than iPRES(1) with no SNR loss, resulting in a significant improvement in image quality.

Keywords

RF coil; Shim coil; B_0 shimming; Coil array; Abdomen

Introduction

MRI image quality is dependent on the homogeneity of the main magnetic field, B_0 . Traditionally, B_0 inhomogeneities are corrected by using whole-body spherical harmonic (SH) shim coils. However, such coils are typically distant from the subject and the magnetic fields used for shimming are often limited to the 2nd or 3rd order, thus hampering their

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ability to correct for localized, higher-order B_0 inhomogeneities, such as those induced by susceptibility differences at air/tissue interfaces.

A multi-coil shimming solution with a larger number of small localized shim coils was shown to be more effective at correcting for local B_0 inhomogeneities in the mouse (1) and human (2) brain. This design employed two separate coil arrays, a direct current (DC) coil array used for localized B_0 shimming and a second radio-frequency (RF) coil array outside (2), or alternatively inside (1), the shim coil array used for RF signal reception. With the shim coil array closest to the subject, the multi-coil array improved the shimming at the expense of a reduced signal-to-noise ratio (SNR) because the DC shim coil array acts as an RF shield between the RF coil array and the subject. A large 10 cm gap was introduced into this design to allow RF penetration and partially mitigate the SNR loss, but at the expense of shimming performance. Conversely, with the shim coil array outside the RF coil array, the SNR was maintained, but the shimming efficiency was reduced. As such, the multi-coil design has an inherent trade-off between SNR and shimming performance.

A novel technique termed integrated parallel reception, excitation, and shimming (iPRES) that provides localized B_0 shimming while maintaining SNR has recently been proposed (3–5) to address this limitation. This design allows both an RF and a DC current to flow on the same coil simultaneously, which enables parallel RF excitation/reception and localized B_0 shimming with a single coil array. An integrated RF/shim coil array can be placed close to the subject to maximize both the SNR and shimming performance while saving valuable space in the magnet bore. In vivo human experiments have demonstrated that an iPRES head coil array can provide a much more effective local B_0 shimming and distortion correction than conventional SH shim coils, particularly in the frontal brain region, which suffers from large susceptibility-induced B_0 inhomogeneities (4).

However, a limitation of the current iPRES implementation is that each DC shim loop is constrained to have essentially the same size and geometry as the RF coil element onto which it is integrated. As a result, local, higher-order B_0 inhomogeneities that are spatially smaller than the RF coil element cannot be fully corrected. While this limitation was not an apparent issue for brain imaging with a 32-channel iPRES head coil array comprised of small coil elements (4), it is problematic for body imaging where localized B_0 inhomogeneities within larger RF coil elements are more challenging to shim. In the present study, we propose an improved iPRES coil design that is capable of correcting for local B_0 inhomogeneities that are smaller than the RF coil elements. Preliminary results have been presented in abstract form (6).

Methods

iPRES(N) Coil Design

To address the limitation of the iPRES coil design mentioned above, we propose a solution that increases the number of independent magnetic fields available for B_0 shimming within each RF coil element of an iPRES coil array. Specifically, each RF/shim coil element is divided into multiple smaller DC loops, each of which uses an independent DC current to generate an additional magnetic field for B_0 shimming, while maintaining the SNR of the

coil. The improved spatial resolution for B_0 shimming offered by the additional DC loops can provide a more effective correction of local B_0 inhomogeneities than the original iPRES design. This modified iPRES coil design with additional DC loops is termed iPRES(N), where N represents the number of DC loops, or degrees of freedom for B_0 shimming, within each RF coil element. Accordingly, the original iPRES coil design with a single DC loop per RF coil element is termed iPRES(1). Schematics for iPRES(1), iPRES(2), and iPRES(3) coil elements are shown in figure 1.

As for the iPRES(1) coil design, inductors, L , are used to bypass any capacitor and allow an independent DC current to flow in each DC loop to generate an additional magnetic field for B_0 shimming. The SNR of the coil is maintained by using inductors, L_{choke} , to isolate the RF currents from losses incurred to the DC power supplies, as well as inductors, L_{loop} , to isolate the RF currents from the interior DC traces. Each inductance value is chosen specifically to be in self-resonance with the capacitance of the inductor pads, which produces a series resonator at the Larmor frequency. Attention must be paid to the manner in which these inductors are integrated onto a RF coil to ensure that the self-resonance frequency (SRF) does not shift lower in frequency by adding capacitance to the equivalent circuit.

Single Coil SNR Measurements

To demonstrate that the SNR of an RF coil is not degraded by the addition of DC shim loops within its perimeter, a proof-of-concept experiment was performed for which a single RF coil was constructed and SNR maps were acquired with the coil in three different configurations: baseline (no iPRES integration) (Fig. 2a), iPRES(1) (Fig. 2b), and iPRES(2) (Fig. 2c). The coil RF parameters, such as impedance and return loss, were carefully monitored during the iPRES(1) and iPRES(2) integration to guarantee that any changes in SNR would be strictly attributed to the addition of the DC shim loops. This single coil experiment removes extraneous integration factors, such as cable routing and layout, to clearly assess the effect of DC shim loop integration to the interior of the RF coil.

Specifically, a 10×10 cm single-turn RF coil was constructed with copper traces on an FR4 substrate. The coil was tuned and matched to resonate at 127.7 MHz using a four-port vector network analyzer (VNA) (ZNB4, Rhode & Schwarz, Munich, Germany) on a square water phantom before iPRES integration. The coil was connected to a pre-amplifier via an sub-miniature, type A (SMA) RF connector and a 50Ω transmission line. A PIN diode was used to detune the coil and protect the pre-amplifier during the scanner transmit period.

All experiments were performed on a 3T MR750 scanner (GE Healthcare, Milwaukee, WI). SNR maps of the phantom were acquired using a gradient-echo sequence (repetition time (TR) = 500 ms, echo time (TE) = 1.5 ms, flip angle = 12° , field-of-view (FOV) = 25.6×25.6 cm, matrix size = 64×64 , slice thickness = 4 mm). First, an SNR map was acquired with the coil in the baseline configuration. Next, the RF coil was modified into an iPRES(1) coil by adding an inductor $L = 800$ nH to bypass the capacitors in the circuit, which allowed a DC current to flow in the coil and enabled simultaneous RF reception and local B_0 shimming. Additional inductors $L_{\text{choke}} = 800$ nH were added to isolate the DC power supply from the RF coil. All the inductors used for iPRES integration in this work have a minimum

isolation of -25 dB and a maximum DC resistance of 0.02Ω . The isolation and high impedance provided by the SRF of a typical inductor are shown in Supporting Figure S1. The return loss, S_{11} , and the impedance of the coil were monitored during the iPRES(1) modification to evaluate changes in the quality factor and mismatch loss, which impact the SNR. An SNR map was acquired with the iPRES(1) coil and with a current of 1 A applied to the DC shim loop. Next, the coil was modified into an iPRES(2) coil by then inserting RF-isolated DC traces that split the coil in half in the left/right (L/R) direction, creating two independent DC shim loops. The return loss and mismatch loss of the coil were again evaluated on the VNA during the iPRES(2) modification. Finally, an SNR map was acquired with the iPRES(2) coil with a current of 1 A applied in both DC shim loops.

8-Channel Coil Array SNR and Flip Angle Measurements

To demonstrate that the integration of multiple DC shim loops can be extended to all the elements of an RF coil array without degrading its SNR or transmit efficiency, SNR and flip angle maps were acquired with an 8-channel RF body coil array in two different configurations: baseline (no iPRES integration) and iPRES(3) (Fig. 3). These maps were acquired with the conventional double angle method (7) using a gradient-echo sequence ($TR = 15$ s, $TE = 3$ ms, flip angle = $60^\circ/120^\circ$, $FOV = 48 \times 24$ cm, matrix size = 128×64 , slice thickness = 7.5 mm) on a 20-cm spherical water phantom ($T_1 \sim 3$ s).

Initially, SNR and flip angle maps were acquired with the baseline 8-channel coil array. Similarly to the previous single coil experiment, each element of the coil array was then modified into an iPRES(3) coil element by adding inductors $L = 800$ nH to bypass the capacitors, thereby enabling a DC current to flow, and $L_{choke} = 800$ nH to isolate the DC power supplies from the RF coil elements. Additional inductors, $L_{loop} = 800$ nH, were added to isolate the three interior DC loops from each of the RF coil elements onto which they were integrated (Fig. 1c). Finally, SNR and flip angle maps were acquired with the 8-channel iPRES(3) coil array using the same transmit and receive gains as for the baseline coil array.

In addition, a three-port bench-top isolation measurement was performed between two adjacent coil elements of the 8-channel iPRES(3) coil array and an interior DC loop of one of these two coil elements to demonstrate the effectiveness of the inductors (Supporting Figure S2). This isolation measurement quantifies the fractional amount of RF current flowing in the interior DC loop relative to each of the coil elements. The input power provided by the VNA at the RF ports was fixed at 0 dBm. The DC loop inductors, L_{loop} , provided more than -50 dB of isolation between the DC loop and both of the RF coil elements, thereby maintaining the SNR of the unmodified coil array. For reference, the overlapped RF coil design of the two coil elements provided -20 dB of isolation between them.

Shim Optimization

The shim optimization procedure used for iPRES(N) is the same as that used for iPRES(1) (4). First, a set of m basis B_0 maps (B_0^{basis}), representing the B_0 field generated by a current of 1 A separately applied in each DC loop, is acquired in a phantom. Second, a B_0 map of

the B_0 inhomogeneities to shim ($B_0^{\text{to shim}}$) is acquired. Then, the optimal DC currents (I) to apply in the DC loops for shimming are determined by minimizing a cost function equal to the root-mean-square-error (RMSE) between $B_0^{\text{to shim}}$ and a linear combination of the B_0^{basis} , weighted by the currents I :

$$RMSE = \sqrt{\frac{1}{n} \sum_{j=1}^n (B_0^{\text{to shim}}(j) + \sum_{k=1}^m I(k) * B_0^{\text{basis}}(j, k))^2}, \quad [1]$$

where n is the number of voxels over which the shim optimization is performed.

In this work, the shim optimization was performed for each slice separately. The voxels outside the phantom or subject, which were defined as those with a signal intensity smaller than 2% of the maximum signal intensity, were excluded from the shim optimization and, for each slice being shimmed, the two adjacent slices were also included in the shim optimization to minimize through-plane B_0 gradients in addition to in-plane B_0 gradients (2).

The shim optimization was performed with the *fmincon* function from the Matlab Optimization Toolbox, because it can find the minimum of a constrained nonlinear multivariable function. The DC currents were constrained within a range of ± 2.5 A, which was the maximum current rating for the inductors. The maximum number of iterations was set to 2000 and the termination tolerance of the function value was set to 10^{-9} .

Phantom Experiments

To demonstrate that the iPRES(N) coil design can provide a more effective shimming of local B_0 inhomogeneities than the iPRES(1) coil design, an initial proof-of-concept phantom experiment was conducted with the 8-channel body coil array. Specifically, one of the RF coil elements was modified into an iPRES(1) coil and subsequently into an iPRES(2) coil, as described in the single coil SNR experiment, while the remaining coil elements were unmodified. Local B_0 inhomogeneities with a spatial variation smaller than the RF coil element were introduced into a uniform square water phantom by placing a DC perturbation loop (Fig. 4a) in the coronal plane between the phantom and the iPRES coil. Applying a DC current to the perturbation loop generated local, asymmetric B_0 inhomogeneities in the L/R direction inside the iPRES coil (Fig. 4b,d). The perturbation loop was RF-isolated with two inductors, $L_{\text{choke}} = 800$ nH, to prevent SNR loss from coupling between itself and the iPRES coil.

The iPRES coils and the perturbation loop were driven by a 5 A, 16 V 32-channel modular power supply (W-IE-NE-R, Plein & Baus Corp., Springfield, OH). The ground reference of the power supply was filtered to minimize common mode noise between the power supply and the scanner, which would otherwise result in image artifacts. Additionally, approximately -50 dB of RF-isolation per channel was provided by adding filters at the penetration panel between the machine room and the scanner room to minimize SNR loss. Axial B_0 maps were acquired using an 8-echo gradient-echo sequence ($TR = 500$ ms, $TE =$

1.5 ms, ..., 11.2 ms, flip angle = 12° , FOV = 25.6×25.6 cm, matrix size = 64×64 , slice thickness = 4 mm), computed with a linear regression of the phase images acquired at different TEs, and smoothed by using the *smooth3* Matlab function with a $3 \times 3 \times 3$ boxcar convolution kernel.

Starting with the iPRES(1) coil, a B_0 map was acquired with a 1 A current applied to the perturbation loop and no current applied to the iPRES(1) shim loop. Subsequently, a basis B_0 map was acquired with 1 A applied to the iPRES(1) shim loop, but no perturbation applied. The optimal DC current to apply to the iPRES(1) shim loop was then determined by minimizing the RMSE between the perturbation B_0 map and the basis B_0 map as described above. A B_0 map was then acquired with this optimal current applied to the iPRES(1) shim loop and with the perturbation applied.

Next, the coil was modified into an iPRES(2) coil by adding two RF-isolated traces to create two independent DC shim loops (Fig. 1b). After modification, a B_0 map was acquired with the same 1 A current applied to the perturbation loop. Subsequently, two basis B_0 maps were acquired with 1 A applied to each iPRES(2) shim loop separately, but no perturbation applied. The optimal DC currents to apply to each of the iPRES(2) shim loops were determined by minimizing the RMSE between the perturbation B_0 map and a linear combination of the two basis B_0 maps. Finally, a B_0 map was acquired with these optimal currents applied to the iPRES(2) shim loops and with the perturbation applied.

Human Experiments

In vivo experiments were carried out on four healthy volunteers who gave written informed consent to participate in this study under a protocol approved by our Institutional Review Board. The subjects were all male and were between 25 and 39 years old. Linear shims, but no higher-order SH shims, were used in these experiments. All 8 RF coil elements of the aforementioned body coil array were modified into iPRES(3) coil elements (Fig. 3) to better shim the local B_0 inhomogeneities in vivo, as described previously for the 8-channel coil array SNR measurements.

A calibration was performed to determine the basis B_0 maps for each of the 24 iPRES(3) shim loops. Since the body coil array was flexible, wrapping it tightly around the body would have required the calibration to be performed in vivo for each subject, at the expense of scan time. In this work, the iPRES(3) coil array was mounted to a rigid acrylic frame so that its position in the scanner was reproducible, thereby allowing the calibration to be performed only once in a phantom. The shape and size of the frame and phantom were chosen such that they had a similar cross-section as that of an average subject's abdomen to ensure that the coils were still close to the subject and hence avoid any reduction in SNR or shimming performance. Specifically, the phantom was an ellipsoid with dimensions of $40 \times 24 \times 36$ cm in the right/left, anterior/posterior, and superior/inferior directions, respectively. It was made of Acrylonitrile Butadiene Styrene plastic, with a dielectric constant and loss tangent at the Larmor frequency of 2.0 and 0.005, respectively, and was filled with tap water ($T_1 \sim 2$ s).

With a DC current of 1 A applied separately to each shim loop in the coil array, 24 B_0 maps were acquired on the phantom using an 8-echo gradient-echo sequence (TR = 500 ms, TE = 1.1 ms, ..., 8.9 ms, flip angle = 12° , FOV = 48×48 cm, matrix size = 64×64 , slice thickness = 7.5 mm, number of slices = 30, slice gap = 0 mm). Next, a baseline B_0 map was acquired on the phantom without DC currents applied to the shim loops and was subtracted from each of these 24 B_0 maps to remove any residual B_0 inhomogeneities from imperfect shimming. Finally, a baseline B_0 map of the subject's abdomen was acquired without applying DC currents using the same pulse sequence and scan parameters. To avoid off-resonance effects due to the presence of lipids in the abdomen, the odd and even TEs were 180° out-of-phase and the B_0 maps were reconstructed with a multipoint Dixon fat-water separation method (8).

In contrast to the phantom experiments, modifying the iPRES(3) coil array into an iPRES(1) coil array to compare their B_0 shimming performance in the same slices was not feasible in these in vivo experiments. Instead, we applied DC currents with the same magnitude and polarity to each of the three DC loops inside the iPRES(3) coil elements to produce magnetic fields equivalent to those of iPRES(1) coil elements, since the magnetic fields generated by the DC traces interior to the RF coil elements cancel out with one another, leaving only the magnetic fields generated by the traces shared with the original RF coil elements. Experimentally, the cancellation may not be perfect due to the small gaps between the DC traces (Fig. 3), but this small discrepancy should not have a significant impact on the comparison between iPRES(3) and iPRES(1).

Accordingly, B_0 maps were acquired with a set of DC currents optimized to minimize the RMSE for the iPRES(3) body coil array operated in the iPRES(1) configuration (i.e., using 8 basis B_0 maps). The 8 basis B_0 maps used for iPRES(1) shimming were generated by adding the three basis B_0 maps corresponding to each RF coil element from the 24 basis B_0 maps acquired in the phantom. Similarly, in the same session and without any coil modification, B_0 maps were acquired with a second set of DC currents optimized to minimize the RMSE for the body coil array operated in the iPRES(3) configuration (i.e., using all 24 basis B_0 maps) to compare the shimming performance of the iPRES(1) and iPRES(3) coil designs. All computations were performed in Matlab vR2014a (The MathWorks, Natick, MA) on a Sun Grid Engine Linux cluster, but using only one CPU core with 6 GB of RAM. The computation time for each optimization was 15 s. The optimized currents were then automatically uploaded to the DC power supply via the scanner internal LAN network.

Additionally, to compare the ability of iPRES(1) and iPRES(3) to correct for geometric distortions, a baseline echo-planar imaging (EPI) dataset was acquired without DC currents followed by additional EPI datasets acquired with the optimal DC currents used for the iPRES(1) and iPRES(3) configurations. The images were acquired using a single-shot spin-echo EPI sequence (TR = 2000 ms, TE = 29 ms, FOV = 48×48 cm, matrix size = 256×256 (no interpolation), slice thickness = 7.5 mm, number of slices = 30, slice gap = 0 mm, readout duration = 81 ms, bandwidth/pixel = 1953 Hz, partial Fourier factor = 62.5%, frequency encoding direction = L/R, SENSE acceleration factor = 2). The standard EPI reference scan and pre-emphasis implemented by the scanner manufacturer were used to correct for Nyquist ghosting and to compensate for eddy currents, respectively.

In addition, undistorted T_2 -weighted images were also acquired without DC currents with a single-shot fast-spin echo (FSE) sequence (TR = 960 ms, TE = 46 ms, FOV = 48×28.8 cm, matrix size = 256×77 (interpolated to 256×154), slice thickness = 7.5 mm, number of slices = 30, slice gap = 0 mm, echo train length = 77, SENSE acceleration factor = 2) for anatomical reference. The subjects were instructed to remain still throughout the entire study, all EPI and FSE scans were acquired during an end-expiration breath-hold and, for each slice, the EPI scans with and without shimming, as well as the FSE scan, were acquired back-to-back to minimize potential motion artifacts. The total scan times for the EPI and FSE scans were 2 s and 11 s, respectively. Contour lines of the anatomy were extracted from the FSE images and overlaid onto the EPI images.

Results

Single Coil SNR Measurements

Adding multiple DC shim loops to the RF coil did not significantly affect its SNR. The SNR maps of a representative slice for the baseline RF coil (Fig. 2d), the iPRES(1) coil with 1 A applied to the DC shim loop (Fig. 2e), and the iPRES(2) coil with 1 A applied to both DC shim loops (Fig. 2f) displayed no appreciable change after the DC shim loops were added. The average SNR within that slice for each of these configurations was 91.7, 88.4, and 89.7, respectively. Further, the SNR histograms for each of these configurations (Fig. 2g) have very similar distributions before and after iPRES integration, indicating that the addition of inductors and RF-isolated traces to the interior of the coil had minimal impact on the SNR.

8-Channel Coil Array SNR and Flip Angle Measurements

As in the single coil SNR experiment, adding multiple DC shim loops to all coil elements of the 8-channel coil array did not significantly affect the SNR or flip angle. The SNR maps of a representative slice for the baseline (Fig. 5a) and iPRES(3) (Fig. 5b) coil arrays showed no appreciable change after the DC shim loops were added. Further, the associated SNR histograms computed over the whole phantom (Fig. 5c) are very similar before and after the iPRES(3) integration. Likewise, flip angle maps of the same slice for the baseline (Fig. 5d) and iPRES(3) (Fig. 5e) coil arrays, as well as the associated histograms (Fig. 5f), remained similar, which shows that the B_1 transmit efficiency was not compromised and that the specific absorption rate was not increased after the iPRES(3) integration.

Phantom Experiments

The DC current applied to the perturbation loop introduced asymmetric B_0 inhomogeneities within the perimeter of the coil (Fig. 4a). After shimming with the iPRES(1) coil, large B_0 inhomogeneities remained (Fig. 4b) because it did not have a sufficient number of magnetic fields available to shim these asymmetric B_0 inhomogeneities. After the coil was modified from an iPRES(1) to an iPRES(2) design, the B_0 inhomogeneities in the phantom (Fig. 4c) were drastically reduced (Fig. 4d), thereby demonstrating the usefulness of having additional magnetic fields for shimming. The B_0 RMSE in the representative slice shown in Fig. 4 was reduced by 4.9% and 70.2% for iPRES(1) and iPRES(2), respectively. Some residual B_0 inhomogeneities remained with the iPRES(2) implementation, which can be further reduced

by using a higher-order iPRES(N) implementation that has additional magnetic fields available for shimming.

Human Experiments

Susceptibility differences at tissue interfaces induced large and localized B_0 inhomogeneities throughout the abdomen (Figs. 6a and 7a, magenta arrows), which resulted in severe geometric distortions in the EPI images (Figs. 6b and 7b, magenta arrows). Even after shimming each slice with iPRES(1), B_0 inhomogeneities remained within the abdomen (Figs. 6c and 7c, magenta arrows) leaving geometric distortions in the EPI images (Figs. 6d and 7d, magenta arrows). However, shimming with iPRES(3) was able to improve the B_0 homogeneity by generating local magnetic fields for shimming that more accurately matched the B_0 inhomogeneities (Figs. 6e and 7e, green arrows), which further reduced the geometric distortions in the EPI images (Figs. 6f and 7f, green arrows). By splitting each iPRES(1) DC loop into three smaller iPRES(3) DC loops, more localized magnetic fields could be generated to effectively shim B_0 inhomogeneities smaller than the RF coil elements (Fig. 8a). In contrast, the magnetic fields generated by the iPRES(1) DC loops were distributed over the entire area of the coil elements, which was not as effective in regions with highly localized B_0 inhomogeneities (Fig. 8b).

The B_0 RMSE for the two representative slices shown in Figs. 6 and 7 was reduced by 59.9% and 31.4% with iPRES(1) and by 64.2% and 43.8% with iPRES(3), respectively. In specific regions-of-interest (ROIs) corresponding to various organs with large susceptibility artifacts, iPRES(3) resulted in an even greater improvement in B_0 homogeneity relative to iPRES(1). For example, the RMSE in the subject's right kidney (Fig. 6c,e, blue ROI) was only reduced by 47.3% with iPRES(1), but was reduced by 70.8% with iPRES(3). The B_0 inhomogeneities that remained after shimming with iPRES(1) (Fig. 6c, both ROIs, magenta arrows) caused misregistration of the subject's right kidney (Fig. 6d, blue ROI, magenta arrow) and intestine (Fig. 6d, orange ROI, magenta arrow) with respect to the anatomical contours overlaid on the EPI image. In contrast, shimming with iPRES(3) reduced the B_0 inhomogeneities in these ROIs (Fig. 6e), which resulted in a dramatic reduction of the geometric distortions in the kidney (Fig. 6f, blue ROI, green arrow) and intestine (Fig. 6f, orange ROI, green arrow).

For a different slice, the B_0 RMSE in the intestine (Fig. 7c,e, blue ROI) was only reduced by 15.1% with iPRES(1), but was reduced by 74.9% with iPRES(3). This substantial improvement in B_0 homogeneity resulted in an improved registration between the EPI and FSE images (Fig. 7f, blue ROI, green arrows). Likewise, the RMSE around the anterior abdominal wall (Fig. 7c,e, orange ROI) was only reduced by 48.4% with iPRES(1), but was reduced by 75.5% with iPRES(3). Although neither iPRES(1) nor iPRES(3) was able to fully shim this ROI, iPRES(3) still resulted in a better EPI image registration than iPRES(1) (Fig. 7d,f, orange ROI). Additionally, the average RMSE between the B_0 maps predicted from the shim optimization and those measured in vivo in these four ROIs were only 0.79 and 0.78 Hz for iPRES(1) and iPRES(3), respectively, demonstrating an excellent agreement between the two.

Discussion and Conclusions

The experimental phantom and in vivo results of this study demonstrate that the proposed iPRES(N) RF/shim coil design can significantly improve the shimming of local B_0 inhomogeneities that are spatially smaller than the RF coil elements by increasing the number of independent magnetic fields available for B_0 shimming relative to the original iPRES(1) coil design. This new iPRES(N) coil design will be highly valuable for applications that require a high B_0 field homogeneity in regions of the anatomy, like the abdomen, that suffer from local susceptibility-induced artifacts. Additionally, the iPRES(N) design retains all the advantages of the iPRES(1) design since it still uses a single integrated RF/shim coil array, which can be placed close to the subject to maximize both the SNR and shimming performance while saving precious space in the scanner bore to accommodate larger patients, in contrast to other shimming methods that require separate RF and DC shim coil arrays (1, 2, 9).

Although the iPRES(3) coil array with 24 DC shim loops used in this proof-of-concept study left residual B_0 inhomogeneities and geometric distortions in the EPI images in a few regions of the abdomen, these residual artifacts can be further reduced by using a higher-order iPRES(N) coil array, which provides additional magnetic fields for B_0 shimming. To this end, the RF circuit boards can be moved away from the coil array to free up room for the routing of additional DC traces within the perimeter of each RF coil element and produce more DC loops for B_0 shimming. In addition, both the RF coil design and the shim trace layout (e.g. size, geometry, and location of the shim loops within each RF coil element) can be simultaneously optimized using simulations to maximize both the RF sensitivity and the B_0 shimming performance of the coil array during the coil design process. Finally, since our implementation did not include dynamic shimming, separate B_0 maps and EPI datasets were acquired with different sets of DC currents optimized to shim different slices. However, dynamic shimming can be implemented by enabling an interface between the programable power supply and the scanner, which has been previously demonstrated for SH shimming (10–12) and a multi-coil design (1, 2).

In conclusion, our new iPRES(N) design can achieve a superior B_0 homogeneity than the original iPRES(1) design by providing additional degrees of freedom for B_0 shimming within each RF coil element without compromising the SNR. It is anticipated that this added advantage will be valuable for a wide variety of research and clinical MR applications (13) that require a high B_0 homogeneity over a large region affected by localized B_0 inhomogeneities (e.g., body diffusion MRI (14)).

Supplementary Material

Refer to Web version on PubMed Central for supplementary material.

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Figure 1.
iPRES(1) (a), iPRES(2) (b), and iPRES(4) (c) coil elements. The independent shim loops are highlighted in blue, magenta, and orange.

Figure 2.

Pictures (a–c), SNR maps (d–f), and SNR histograms (g) for the RF coil in three configurations: baseline, iPRES(1), and iPRES(2), showing no appreciable change in the SNR distribution after DC shim loop integration. The artifact in the upper left hand corner of each SNR map is a byproduct of losses due to the RF connector.

Figure 3. Four representative RF coil elements of an 8-channel body coil array that have been modified to iPRES(3) coil elements with three independent DC shim loops within the perimeter of each RF coil element.

Figure 4.

Diagram of the perturbation loop, which was placed in the coronal plane between the phantom and the iPRES(1) or iPRES(2) coil, to generate asymmetric B_0 inhomogeneities (a). The blue line represents the perimeter of the iPRES(1) or iPRES(2) coil. Axial B_0 maps with the perturbation applied to the phantom (b) after iPRES(1) modification without shimming, (c) after iPRES(1) modification with shimming, (d) after iPRES(2) modification without shimming, and (e) after iPRES(2) modification with shimming.

Figure 5. Axial SNR maps (a–b), flip angle maps (d–e), and corresponding histograms (c and f) showing no appreciable change in the SNR or flip angle distributions after iPRES(3) integration into an 8-channel coil array.

Figure 6.

In vivo B_0 maps (a,c,e) and EPI images (b,d,f) of a representative slice acquired in a 39-year old male with superimposed contour lines from the anatomical image (g) comparing the iPRES(1) and iPRES(3) shimming performance. The magenta arrows highlight susceptibility-induced B_0 inhomogeneities, which result in geometric distortions in the adjacent EPI images. The green arrows highlight the corrected B_0 inhomogeneities and geometric distortions after shimming with iPRES(1) or iPRES(3). The red lines surrounding the B_0 maps indicate the positions of the eight RF coils relative to the subject. Additionally,

ROIs magnifying the subject's right kidney (blue ROI) and intestine (orange ROI) emphasize the improved shimming performance of iPRES(3) relative to iPRES(1).

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Figure 7.

In vivo B_0 maps (a,c,e) and EPI images (b,d,f) of a representative slice acquired in a 39-year old male with superimposed contour lines from the anatomical image (g) comparing the iPRES(1) and iPRES(3) shimming performance. The magenta arrows highlight susceptibility-induced B_0 inhomogeneities, which result in geometric distortions in the adjacent EPI images. The green arrows highlight the corrected B_0 inhomogeneities and geometric distortions after shimming with iPRES(1) or iPRES(3). The red lines surrounding the B_0 maps indicate the positions of the eight RF coils relative to the subject. Additionally,

ROIs magnifying the subject's intestine (blue ROI) and anterior abdominal wall (orange ROI) emphasize the improved shimming performance of iPRES(3) relative to iPRES(1).

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Figure 8.

Representative basis B_0 maps used for shimming with the iPRES(3) (a) and iPRES(1) (b) coil array configurations, showing that iPRES(3) offers more degrees of freedom for shimming relative to iPRES(1). The remaining basis B_0 maps (not shown) are symmetrical in the R/L direction. The dashed lines in the axial and coronal B_0 maps indicate the positions of the coronal and axial B_0 maps, respectively.