A 32-channel multi-coil setup optimized for human brain shimming at 9.4T

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Purpose: A multi-coil shim setup is designed and optimized for human brain shimming. Here, the size and position of a set of square coils are optimized to improve the shim performance without increasing the number of local coils. Utilizing such a setup is especially beneficial at ultrahigh fields where $B_0$ inhomogeneity in the human brain is more severe.

Methods: The optimization started with a symmetric arrangement of 32 independent coils. Three parameters per coil were optimized in parallel, including angular and axial positions on a cylinder surface and size of the coil, which were constrained by cylinder size, construction consideration, and amplifiers specifications. $B_0$ maps were acquired at 9.4T in 8 healthy volunteers for use as training data. The global and dynamic shimming performance of the optimized multi-coil were compared in simulations and measurements to a symmetric design and to the scanner’s second-order shim setup, respectively.

Results: The optimized multi-coil performs better by 14.7% based on standard deviation (SD) improvement with constrained global shimming in comparison to the symmetric positioning of the coils. Global shimming performance was comparable with a symmetric 65-channel multi-coil and full fifth-order spherical harmonic shim coils. On average, an SD of 48.4 and 31.9 Hz was achieved for in vivo measurements after global and dynamic slice-wise shimming, respectively.

Conclusions: An optimized multi-coil shim setup was designed and constructed for human whole-brain shimming. Similar performance of the multi-coils with many channels can be achieved with a fewer number of channels when the coils are optimally arranged around the target.

Keywords
$B_0$ inhomogeneity, $B_0$ shimming, echo planar imaging, multi-coil, optimization, ultrahigh field
1 | INTRODUCTION

Ultrahigh field (UHF) MRI has been increasingly used by researchers in the last decade for neuroimaging of the human brain. UHF enables gaining a better understanding of brain functions by means of higher spatial resolution and signal-to-noise ratio (SNR). However, there are several significant challenges for imaging at UHF, for example, inhomogeneity in $B_0$ and $B_1$ field and specific absorption rate (SAR).\(^1\)\(^-\)\(^3\)

In particular, static field inhomogeneity is a long-standing issue from the early days of nuclear MR (NMR) imaging,\(^4\) which is more pronounced at UHF. Challenges originating from $B_0$ inhomogeneities are anticipated and reported in literature, including geometric distortion,\(^5\) spectra line broadening and weak water suppression,\(^6\) shortening $T_2^*$ relaxation time,\(^7\) or voiding the signal.\(^8\) There are a few postprocessing approaches which can partially correct the consequences of poor $B_0$ uniformity.\(^9\),\(^10\) These methods attempt to retrospectively mitigate adverse effects of the static field perturbation rather than to address the problem at its origin.

Shimming—the process of homogenizing the static magnetic field—is a routine solution provided by vendors. One can quantitatively describe the existing $B_0$ field in the magnet with the aim of Laplace’s equation, whose general solution yields a set of basis functions called spherical harmonics (SH). Most state-of-the-art UHF scanners are equipped with shim coils that can model spatial field distribution up to third-order SH. The size and specific winding pattern of such shim coils come along with a large inductance, resulting in long transient times after switching attributed to induced eddy currents and mutual inductance. This limitation makes such coils inflexible, and a pre-emphasis circuit is essential when rapid switching is required.\(^11\),\(^12\)

Another proposed method, known as multi-coil shim array, suggests using a group of small local coils to generate a local magnetic field.\(^13\) The generated local fields can contribute effectively to counteract the existing $B_0$ inhomogeneity. Small size, low inductance, and lower power consumption of such coils make them a suitable choice for dynamic slice-wise shimming.\(^14\),\(^15\) Integration into the radiofrequency (RF) receive coil,\(^16\),\(^17\) real-time correction of temporal $B_0$ alteration,\(^18\) generating spatial encoding magnetic fields,\(^19\) and novel parallel imaging methods.\(^20\)

The major part of the $B_0$ field perturbation comes from the sample or subject being imaged rather than the magnet imperfection itself.\(^21\) The human body is composed of air and water. The magnetic susceptibility difference of 9.41 ppm\(^22\) between air and water induces a magnetic field inhomogeneity nearby air-tissue boundaries.\(^23\) This effect scales with the magnetic field strength and becomes more severe at UHF.\(^1\) To overcome this issue, one can increase the number of the local coils or feed more current to the coils, which necessitates taking thermal issues into account. A simulation comparing the performance of multi-coils with a different number of coils is reported in Stockmann et al.\(^16\) Although increasing the number of coils yields better shimming, it requires more dedicated amplifiers and, consequently, would not be cost-effective. Furthermore, difficulties in maintenance and troubleshooting are expected. A successful combination of a scanner’s zeroth and first-order SH shim coils with a 16-channel multi-coil setup for dynamic shimming of the human brain at 9.4T is reported in Aghaeifar et al.,\(^15\) which allowed for more degrees of freedom (DOFs) in shimming at UHF using the scanner’s built-in hardware.

In the brain, the strongest field inhomogeneities are found in the temporal lobe (TL) and prefrontal cortex because of proximity to the ear canals and sinuses, respectively.\(^24\) In a few studies, active shim coils are placed in the mouth or over the nose to improve $B_0$ uniformity in the frontal lobe.\(^25\),\(^26\) This procedure can be considered as manual optimization of the coil positioning for local shimming. However, the subject’s safety, their comfort, and the stability of the coils are questionable in this case. Given that the overall geometry and structure of the head and skull as well as the relative location of air cavities and ear canals are similar across humans, a similar pattern of $B_0$ inhomogeneity in the human brain is expected and has been observed. Therefore, the shim array can be modified for better performance on an identified target pattern. Theoretical design of shim arrays with an irregular shape to generate low-order SH\(^27\) and to fit with C-type permanent magnet has recently been presented.\(^28\) Another study has demonstrated the application of a genetic algorithm to design a coil with an irregular shape, which is optimized only for 4 representative slices of the brain\(^29\) and extended later for optimization of position and geometry of 16 coils for 2 slices of a single brain.\(^30\)

Thus, the aim of the present work was to design and construct an optimized multi-coil shim array to target the $B_0$ inhomogeneity in the whole human brain. This will help to enhance $B_0$ shimming at UHF without the need to add more coils. Optimization was performed under specific constraints to keep the construction step simple. The performance of the optimized multi-coil is compared to the conventional symmetric coils arrangement with a different number of local coils and with different orders of SH. The efficiency of the designed and constructed setup is evaluated in vivo with several sequences susceptible to $B_0$ inhomogeneity.

2 | METHODS

2.1 | Optimizations

In this study, the optimization was performed on a 32-channel multi-coil setup to improve shimming performance for a human brain target. All coils were positioned on a cylinder with a diameter of 323 mm. This is the minimum allowable diameter that can house the utilized RF coil.\(^31\) Optimization
started from a symmetric arrangement of 32 coils in 4 rows on the surface of the cylinder (Figure 1A). All coils had an identical square shape with an equal side length of 60 mm and 25 wire turn. Because of the cylindrical skeleton, the optimization was performed in a cylindrical coordinate system with 3 DOFs for each coil as follows:

1. Size of the coil (i.e., side length of a square coil) where the shape does not change.
2. Axial coordinate (Z) or height of the center of coils in the cylinder surface.
3. Angular coordinate (θ), the angle between the reference axis on a chosen plane and a line from the origin to the projection of the coil center to that plane (Figure 1B).

The first 2 DOFs (i.e., size and height) were constrained during the optimization while the angular coordinate was unconstrained given that, theoretically, it should lay between \(-\pi\) and \(+\pi\). The lower and upper bounds for the size of the coils were [20 and 100] mm, respectively, and for the height of the coils were [−150 and +80] mm, respectively. Therefore, based on the maximum side length of the coils, the required overall length for the cylinder would be 330 mm. During the construction, the cylinder length was chosen longer to reserve space for wires going outside the multi-coil (final cylinder length was 400 mm). The initial side length of 60 mm was chosen as the middle of the upper and the lower bound of the coil size. For the chosen initial coil arrangement, there was no overlapping between the coils and the free spacing between a coil and the next closest coil was negligible.

A nonlinear constrained optimization was implemented in MATLAB (The MathWorks, Inc., Natick, MA) using the function, \texttt{fmincon}. The purpose of the nonlinear constrained optimization was to find the most effective configuration of coils. After every successful iteration, new shim basis maps were generated, and the required current per coil was calculated by a constrained linear least-squares optimization (\texttt{lsqlin} command in MATLAB). The flowchart of the optimization process is depicted in Figure 2A, where the box with a yellow outline refers to the constrained linear least-squares optimization. Given that the gradient of the objective function with respect to the position of the coils was simply not analytically attainable in a closed-form expression, the choice of solver was limited to one that can numerically estimate the gradient. Two algorithms were utilized to this end: sequential quadratic programming (SQP)\textsuperscript{32} and interior point.\textsuperscript{33} Both algorithms yielded acceptable improvement; however, SQP converged faster and was therefore selected as solver for the final optimization. To calculate shim currents, the bounds were variable according to the coil size. Given that smaller coils produce less heating, the current for the largest coils with a side length of 100 mm was constrained to 1.5 A, and it was modeled to increase linearly with decreasing coil size. The current bound was truncated to 3 A for the coils with side...
length smaller than 50 mm because of the employed amplifier specifications (e.g., the currents for coils with a size length of 100, 75, 50, and 30 mm were constrained to 1.5, 2.0, 3.0, and 3.0 A, respectively).

B₀ maps of the brain from 12 volunteers were measured at a magnet with a static field of 9.4T after applying the scanner’s second-order SH global shimming. All maps were interpolated into a standard coordinate system with a field of view (FOV) of 300 × 300 × 300 mm³ and isotropic resolution of 3.0 mm. The B₀ maps were split into 2 groups of 8 and 4 maps. The first group with 8 maps was used for optimization (training), and the second group with 4 maps was used for validation of the optimization outcome. The cost function of the optimization is described by Equation 1:

\[
\text{cost}(x) = \sum_{i=1}^{8} \sum_{j=1}^{32} (c_{ij}m_j + b_i)
\]

where \(x\) includes the size and position of the coils in the current iteration, \(b_i\) is the B₀ map of the \(i\)th brain, \(m_j\) is the basis-map of the \(j\)th coil, and \(c_{ij}\) is the current of the \(j\)th coil calculated for the \(i\)th brain through constrained linear least-squares optimization. Given that the coils may overlap partially or completely after optimization, the overlapping coils were mounted in different layers during the construction. However, it is possible to modify Equation 1 and add an additional regularization term to address coil overlapping, that is, \(\text{cost}_{\text{new}}(x) = \text{cost}(x) + kF(x)\), where \(F(x)\) is given by the summation of intersections between coils and \(k\) represents a weighting factor.

Additionally, the performance of the optimization algorithm was investigated with noisy inputs. Thus, white Gaussian noise with standard deviation (SD) between 0 and 40 Hz was added to the training B₀ maps. The coil optimization was then repeated and the performance of the obtained

**FIGURE 2** A, Flowchart of the optimization process used in this study. There are 3 inputs for the optimizations, including constraints, initial coil arrangement, and the training data. It is permitted to adjust the current constraints based on the coil size. The returned value of the cost function is the sum of the residual off-resonance after shimming in individual training B₀ maps as explained in Equation 1. The optimization will be terminated when the changes in the arrangement and size of the coils are smaller than a defined threshold. B, Impact of the overlapping regularization term on the efficiency of the final design. Increasing the weighting decreases the total overlapped areas, but degrades the final loss (loss: output of cost function defined in Equation 1). The numbers in pink represent the maximum number of the coils that do not need to be moved to outer layers. The loss values are normalized with respect to the loss of the symmetric design. C, Speed of convergence of 2 algorithms, SQP, and interior point, used in this study from symmetric initial coils arrangement and 2 representative random initial coils arrangements. SQP could converge faster and yield a smaller loss for all cases. The values are normalized to the loss obtained from symmetric design.
optimized coil was compared to the case with 0 additive noise. Given that inaccurate placement of the coils during construction is possible, stability of the optimization output was further investigated. The axial and angular value of the coil locations were shifted randomly in the range of ±10 mm and ±4°, respectively. The performance of the shimming after coil repositioning was evaluated and compared with the original optimized multi-coil.

2.2 | Construction

As mentioned in the optimization procedure, smaller coils were allowed to carry higher currents up to 3A. The generated magnetic field is proportional to \( n \times I \), where \( n \) is the number of turns and \( I \) is the passing current, and power dissipation is equal to \( R I^2 \), where \( R \) is the resistance of the coil. Therefore, the smaller coils were built with more windings instead of supplying higher current to reduce thermal losses (i.e., all currents were constrained into 1.5 A in experimental measurements because of increasing the coil’s winding). A copper wire with a thickness of 0.8 mm was used for coil winding. A 3D printed support was designed to avoid bowing of the overlapped coils and to fasten the coils to the cylinder (Supporting Information Figure S1). The support was drawn in CATIA (Dassault Systems, Suresnes, France) for each coil while considering the cylinder peripheral curve. A thin plastic pipe split into 3 branches was swirled around the coils for the water cooling of the setup in case it was needed. Later, the whole free space between the inner cylinder (diameter = 323 mm) and the outer cover (diameter = 370 mm) was filled with epoxy (Polytec EP 641; Polytec PT GmbH, Karlsbad, Germany) to prevent mechanical vibration of the coils.

2.3 | Simulations

Basis-maps of the multi-coils with a different number of coils (8, 16, 24, 32, 48, 65, and 96) were analytically calculated using Biot-Savart’s law. Individual coils of all multi-coils were symmetrically positioned on the surface of a cylinder with diameter and length of 323 and 330 mm, respectively (Figure 3). All coils were simulated with 25 wire turns and a square shape. The diameter of the coils was adjusted for each

![Figure 3](image-url)

**Figure 3** Comparison of the simulated shimming performance between optimized multi-coil, multi-coils with a different number of coils, and spherical harmonics basis set. Shimming is carried out in global and dynamic slice-wise fashion while the current is constrained and unconstrained. Only unconstrained shimming with spherical harmonics is performed given that they are calculated analytically. The shimming is performed on 14 brain \( B_0 \) maps, which were acquired at 9.4T. All maps are transformed into a standard space with an isotropic resolution of 1.5 mm. The side length of the coils is indicated above each sketched multi-coil. Two different 32-channel multi-coils are simulated, one has larger coils for full coverage of the cylinder, and the other is the one used for optimization.
multi-coil to cover the whole cylinder surface with minimal overlapping. Performance of the multi-coils was compared in terms of global and slice-wise shimming for 14 brain $B_0$ maps while the input current was constrained to 2.0 A and unconstrained. All $B_0$ maps were interpolated to a standard space with an isotropic resolution of 1.5 mm. The target volume of shimming consisted of 379,555 ± 28,740 voxels corresponding to 1282 ± 97 mL (mean ± SD). The shim currents were optimized for a 4.5-mm slab centered with respect to the slice of interest in case of slice-wise shimming. As a benchmark, unconstrained shimming was performed with SH basis sets up to sixth order as well. $\Delta B_0$ SD and root mean squares (RMS) were calculated after shimming and averaged over all volunteers for whole brain. The shimming performance was further assessed locally within a spherical volume, which had an average diameter of 4 and 2.8 cm around the prefrontal cortex (PFC) and ear canals, respectively.

### 2.4 Setup characterization

Home-built amplifiers were used to supply current for the local coils.34 Each channel could supply up to 5 A (120 A in total). The output voltage was adjustable through the user interface to control thermal loss (maximum, 24 V). A current sense resistor of 0.1 $\Omega$ in series connected to the output for real-time current monitoring, and feedback control. A 10-Hz square wave signal with an amplitude of 2 V (corresponding to 2 A) and 50-µs ramp time was applied to the current amplifier while monitoring the output current. Then, the proportional-integral-derivative controller of the amplifiers was adjusted on a channel-by-channel basis to achieve minimum settling time and overshooting for the inductive load of the individual channels. Each control term was adjusted by a digital variable resistor through the user interface. All the adjustment values were saved in the amplifiers’ memory for the future experiments. A custom-built LabVIEW program (National Instruments, Austin, TX) received the trigger signal from the scanner and then updated the output of a PXIe-6738 unit (digital to analog output module) to control the amplifiers. All the required currents were imported to the LabVIEW program before start of the scan. The LabVIEW program changes the current within a 2-ms time interval using 200 intermediate steps (where time interval and number of steps were adjustable). The currents passing into the coils were read from the current feedback signal and displayed in the LabVIEW program to discover mismatch between inputs and outputs.

Investigation of the possible interaction between overlapped coils was studied through temporal field monitoring with a field camera.35 The field camera consisted of 16 $^{19}$F-NMR sensors (Skope Magnetic Resonance Technologies, Zurich, Switzerland) tuned for a magnet with a field strength of 9.4T and distributed on the surface of a sphere with a diameter of 20 cm. A signal consisting of a sequence of alteration between [0, +2, −2, +2, 0] was applied to the amplifier input in 5 steps. The field measurement lasted 20 ms for each step, which started 4 ms before input current alteration and sampled at a temporal resolution of 1 µs. The resulting time-varying phase was extracted from the signal of the probes, unwrapped, and used for $B_0$ estimation by linear regression of the phase time course of every 1 ms. Furthermore, gradients waveform estimation with NMR probes was checked in the presence and absence of the multi-coil to test disturbance rejection of the amplifiers and influence of the setup on the produced encoding magnetic fields.

Similar to our previous design,15 several measurements, including $B_1^+$, temporal SNR (tSNR), and SNR, were carried out with and without multi-coil to evaluate the influence of the multi-coil on image quality. Thermal behavior of the setup was characterized by applying the maximum current to all channels for an hour and measuring the temperature on the surface of the setup.

### 2.5 $B_0$ shimming and imaging protocols

Five healthy volunteers participated in the study (average age, 25 ± 4 years) in accord with the local ethics protocol. A dual-echo gradient echo (GRE) sequence was used to measure reference $B_0$ maps for the subsequent calculations (flip angle [FA], 12°; TE1/TE2/TR, 2.8/7.8/15 ms; FOV, 208 × 208 × 160 mm³; isotropic resolution of 2.0 mm). The magnitude images and $B_0$ maps were reconstructed offline from the raw data. Later, a 3D brain mask was created from the magnitude image using brain extraction tools36 and spatial phase unwrapping was applied to the $B_0$ maps.37 Then, constrained least-squares optimization was performed on the shimming problem, which can be described by Equation 2:

$$\min_x \| (Ax - B) \|^2$$

where $B$ is a $v \times 1$ vector representing the unshimmed brain $B_0$ map ($v$: number of voxels after masking), $A$ is a $v \times n$ matrix of the shim basis-set ($n$: number of shim channels), and $x$ is an $n \times 1$ vector containing the unknown shim currents. The MATLAB lsqlin function was used to solve this minimization problem. Equation 2 was also used for the shimming of training $B_0$ maps in every iteration when optimizing the coil arrangement and size as explained above, with the difference that matrix $A$ was updated in every iteration. The calculation of the required shim currents was performed in MATLAB. Once the shim currents were calculated, they were saved as a table in a text file to be used by the LabVIEW program as explained previously.
The performance of global and dynamic slice-wise shimming was evaluated by using multiecho GRE, echo planar imaging (EPI), and balanced steady-state free precession (SSFP) sequences. Multi-echo GRE was not only used for assessment of the dynamic B₀ shimming, but also for T₂ characterization. All echoes were acquired with monopolar readout gradients. The acquisition protocol included FOV: 204 × 204 mm², isotropic resolution of 2.0 mm, FA: 20°, TE/TR: [5, 9.5, 15, 22, 30, 40]/80 ms, and 16 slices (slice gap = 200%). EPIs were measured with different isotropic resolutions and acceleration factors. The bandwidth (BW) was adjusted for each EPI to minimize echo-spacing and, accordingly, geometric distortions. The acquisition parameters were FOV: 204 × 204 mm², isotropic resolution of [2.0, 2.0, 1.5, 1.0] mm, FA: 60°, TE/TR: 24/2000 ms, 16 slices (slice gap = [200, 200, 300, 500] %), BW: [2132, 2131, 1838, 1442] Hz/Px, 6/8 partial Fourier, and generalized autocalibrating partially parallel acquisitions (GRAPPA) factor: [1, 2, 2, 3] (parameters between brackets “[“]” are ordered consistently with respect to each other). Images with a balanced SSFP (bSSFP) sequence were acquired using sign-alternated RF pulses. The actual image acquisition of each slice was preceded by 16 RF ramp preparation pulses followed by 100 dummy pulses to ensure steady-state conditions. The acquisition protocol included FOV: 204 × 204 mm², isotropic resolution of 1.0 mm, FA: 30°, TE/TR: 7.5/15 ms, and 16 slices (slice gap = 500%). T₁ and bSSFP data were acquired for 4 subjects. All of the employed sequences were able to send out a trigger signal before RF excitation to synchronize the sequence and the shimming hardware. The trigger signal was followed by a 2-ms delay to compensate for the ramp transition of shim currents between 2 states and any lag in the setup.

3 | RESULTS

Figure 1A demonstrates the result of optimizing the position and size of the coils. The shim coils are depicted in the scanner gradient coordinate system (+Z = feet, −Z = head, +Y = anterior, −Y = posterior, +X = left, and −X = right). Figure 1C shows arrangement of the optimized multi-coil when the cylindrical coordinate is transformed into a 2D plane. Supporting Information Figure S1 shows how the coils’ supports with a thickness of ≈7.5 mm are layered. Figure 1D displays the constructed 32-channel multi-coil setup optimized for the human brain. Coils were installed in 4 layers including 16, 11, 3, and 2 coils in the layers 1 to 4, respectively. Overall, the optimized multi-coil consisted of 8, 11, 9, and 4 coils placed in the top (anterior), right and up-right, left and up-left, and bottom (posterior) of the cylinder, respectively. The measured inductance and resistance of the coils at 1 kHz was ranging from 108 to 232 μH and from 2.1 to 5.7 Ω, respectively.

The addition of the setup (connected to the amplifier with 0 A in all channels) did not significantly affect the excitation profile (B₀ map), SNR, and tSNR (Supporting Information Figures S2, S3, and S4). Supporting Information Figure S5 shows the results of the thermal tests measured with an array of 16 temperature sensors and an infrared camera (images were acquired at the end of measurement). Given that the setup temperature can be considered safe for human measurements with the specified current bounds, no water cooling was used during the experiments.

Figure 2B shows how an increased weighting of the overlapping regularization term, k, affects the final loss, total overlapped areas, and the maximum number of the coils in the first layer. The pink numbers in the plot represent the maximum number of the coils which can be installed in the first layer (their original position). Including an additional regularization term increases the nonlinearity degree of the problem and yields a less effective coil arrangement. Investigating the effect of noisy training B₀ maps shows that adding Gaussian noise with SD below 30 Hz to training data did not affect performance of the obtained optimized coil (changes below 1% on average over 14 brain B₀ maps). However, the shimming performance decreased by 12% for the case comprises Gaussian noise with SD of 40 Hz. The evaluation of the stability of the optimization output reveals that a minor inaccuracy in the placement of the coils can degrade the performance by 1.5% and 0.4% in global and slice-wise shimming, respectively (on average over 5 sets of inaccurate positions).

Summary of the shimming performance in simulation for several symmetric coils arrangements, the 32-channel optimized multi-coil (with and without layering), and SH term is shown in Figure 3. The results correspond to the shimming performance averaged across 14 brain B₀ maps, which were not included into the training. Two different symmetric 32-channel multi-coils are simulated; one has slightly larger coils which cover the cylinder fully, and another one served as initial configuration of the coils for optimization. The performance of the 32-channel optimized coil (with SD of 40.9 and 37.4 Hz for the case of constrained and unconstrained global shimming, respectively) is comparable to the 65-channel symmetric design. In comparison to the SH terms, the performance is slightly better than full fifth-order SH for the case of unconstrained global shimming. Layering the coils resulted in a slight increase of the SD by approximately 1.1 Hz for constrained global shimming. In comparison to the initial symmetric design and based on SD of the residual off-resonance after shimming, performance improved by 14.7% and 20.8% after optimization with constrained and unconstrained global shimming, respectively (which decreased to 12.4% and 19.9% after layering). Table 1 shows the amount of performance...
improvement of the optimized multi-coil with respect to the symmetric design based on SD and RMS for different subregions, constrained and unconstrained shimming, as well as global and dynamic slice-wise shimming. The presented results are calculated before and after coil layering. RMS is a better metric for subregions given that it characterizes the off-resonance fully while SD ignores the mean of the off-resonance. Supporting Information Table S1 displays the achieved improvement for 14 individual $B_0$ maps used in the simulation (the optimized design after layering compared to the symmetric 32-channel). On average, the required shim currents for unconstrained global/slice-wise shimming are in total 47.1/5512 A and 122.5/5457 A for symmetric and optimized multi-coil, respectively.

Redundancy in the magnetic fields generated by the individual coils after optimization is quantified based on correlation coefficient and singular value decomposition. The results are depicted in Supporting Information Figure S6. In summary, 55% of correlation coefficients were between 0 ± 0.25, and the cumulative sum of the first 10 eigenvalues reached 80% of the total eigenvalues energy. This means that several coils can be replaced with a single coil (probably with an irregular shape) and 1 more powerful amplifier. The obtained results are based on the magnetic field generated in a spherical FOV with a diameter of 200 mm centered on the scanner isocenter.

Figure 4A shows the phase evolution of the NMR field probes during current alteration in channel 29 of the multi-coil. Channel 29 is mounted in the upper part of the cylinder above the frontal cortex of the volunteer and is overlapped with channels 3, 11, 24, and 25. The $\Delta B_0$ estimation from the closest field probe to the coil is plotted. No eddy-current contamination has been observed in the fitted $\Delta B_0$. Similar results have been achieved for other channels. Furthermore, the estimated gradient waveform reveals that there is a small interaction between gradients and the setup which can be compensated at an acceptable level when the amplifiers are switched on (Figure 4B).

Figure 5 displays the in vivo performance of the 32-channel optimized multi-coil in $B_0$ shimming. On average over all subjects, the whole-brain SD of off-resonance after performing global and dynamic slice-wise multi-coil shimming decreased from 71.9 to 48.4 Hz and 31.9 Hz, respectively. Table 2 shows SD of off-resonance in the whole brain for all volunteers. The average SD of global shimming in the measurement (48.4 Hz) for 5 volunteers is slightly higher than in the simulation (42.0 Hz); however, the average SD after second-order SH in the measurement (71.9 Hz) is also higher than the average SD of $B_0$ maps used in the simulations (68.6 Hz).

Figure 6 illustrates the impact of the shimming on correction of the geometric distortion in EPIs and the corresponding voxel shift map. An anatomical image acquired with GRE sequence is used as an undistorted reference image. Ventricles align better with anatomy, and a large portion of the distortion in the frontal cortex is recovered for both global and dynamic multi-coil shimming at the cost of stretching in anterior. On average over all volunteers, voxel shifts larger than 5 mm after global and dynamic slice-wise shimming decreased by 49% and 64%, respectively, for all utilized EPI protocols (discrepancy was below 2.5% for different resolutions). The EPIs for other resolutions and subjects are provided in Supporting Information Figures S7 and S8.

Figure 7 compares the banding artifacts in bSSFP images. Three slices covering the cerebellum, ear canals, and frontal cortex are chosen for the comparison. Global multi-coil shimming reduced the banding artifacts in different areas, and dynamic slice-wise multi-coil shimming was successful in a higher degree to eliminate a large portion of the banding artifacts. The bSSFP images for other subjects are provided in Supporting Information Figures S7 and S8.

The outcome of the $T_2^*$ calculation after second-order shimming as well as global and dynamic multi-coil shimming is depicted in Figure 8. Two slices that cover areas of the brain with the highest $B_0$ inhomogeneity from each subject are shown. Both dynamic slice-wise and global multi-coil shimming resulted in $T_2^*$ gain for the areas with

**Table 1** Different metrics and subregions used to evaluate improvement in shimming performance of the optimized multi-coil with respect to the symmetric design (with side length of 60 mm) in simulation

| Shim scope: | Global [2.0 A, unconstrained] % | Slice-wise [2.0 A, unconstrained] % |
|------------|----------------------------------|-------------------------------------|
| Criterion: | SD | RMS | SD | RMS |
| Original  | Whole brain | [14.7, 20.8] | [14.7, 20.8] | [8.6, 4.7] | [8.6, 4.7] |
| Frontal cortex | [11.5, 20.4] | [14.4, 24.0] | [10.5, 10.7] | [10.9, 11.0] |
| Near ear canals | [17.4, 23.3] | [19.4, 25.9] | [6.6, 1.9] | [6.6, 1.9] |
| Layered  | Whole brain | [12.4, 19.9] | [12.4, 19.9] | [6.3, 4.9] | [6.2, 4.9] |
| Frontal cortex | [9.3, 19.6] | [11.8, 23.1] | [8.0, 11.3] | [8.4, 11.5] |
| Near ear canals | [14.4, 22.7] | [16.0, 25.1] | [4.0, 0.7] | [3.9, 0.6] |

Global and dynamic shimming for both the constrained and unconstrained case are studied.
severe $B_0$ inhomogeneities. The $T_2^*$ ratio increased by 18.1% and 28.0% in the vicinity of the ear canals as well as 8.8% and 12.5% in PFC after global and slice-wise shimming, respectively.

The information of the coils layering and description of wiring pattern according to the public multi-coil information policy can be found in Supporting Information Tables S2 and S3, respectively.
The aim of this study is to design and construct a prototype of a multi-coil shim setup that is optimized for human brain shimming. The optimization procedure resulted in new coil arrangements and sizes, which allow boosting of shimming performance without increasing the number of coils. The optimization was limited to 3 DOFs per coil; axial and angular value in the cylindrical coordinate system and the coil size. Avoiding constraints in the coil geometry (i.e., permitting for irregular coil shapes) increases the number of DOFs; however, it may either result in a coil shape with difficulties to build or in overfitting of the model.

The designed multi-coil arrangement in this work may not be suitable for combined RF and $B_0$ shim array, as is proposed in earlier works\cite{16,17} attributed to possible degradation in sensitivity profile of the receive coils and more challenges in coils decoupling. Additionally, the optimized multi-coil may not be a proper choice for applications beyond $B_0$ shimming similar to what is suggested in Umesh Rudrapatna et al\cite{19} for imaging and in Scheffler et al\cite{20} for acceleration because of nonuniform coverage of the FOV.

**TABLE 2** The SD of $B_0$ inhomogeneity after different shimming strategies for individual volunteers and on average across all (experimental data)

| Subjects No. | Target volume (mL) | Shimming method (STD) |
|--------------|--------------------|-----------------------|
|              |                    | Second-order SH | Global multi-coil | Dynamic multi-coil |
| 1            | 1199               | 65.8          | 45.4 | 28.1 |
| 2            | 1502               | 75.7          | 51   | 32.4 |
| 3            | 1321               | 75            | 48   | 30.1 |
| 4            | 1046               | 61.5          | 42.4 | 33.4 |
| 5            | 1470               | 81.6          | 55.2 | 35.8 |
| Average      | 1307               | 71.9          | 48.4 | 31.9 |

**FIGURE 5** Comparison of the residual $B_0$ inhomogeneity after different shimming approaches. The whole brain is shimmed at the beginning with the scanner’s second-order built-in spherical harmonic shims. After having the second-order shim applied, global and dynamic multi-coil shimming is performed.
FIGURE 6 Evaluation of the geometric distortion for 3 representative slices of the brain at 1.0-mm isotropic resolution. Voxel shift maps are calculated based on $B_0$ maps and the EPI acquisition protocol. Better shimming results in less distortion, which is also apparent from the voxels shift maps. The amount of distortion in high-resolution accelerated 1.0 mm imaging is approximately 2 times less than in low-resolution 2.0-mm isotropic EPI without acceleration (depicted in Supporting Information Figure S7).

FIGURE 7 Effect of the improved $B_0$ homogeneity on banding artifacts in bSSFP images. Three representative slices from 3 areas of the brain with the highest $B_0$ inhomogeneity are selected. Global and dynamic multi-coil shimming apparently reduces banding artifacts.
Another approach suggests optimizing the wiring pattern with genetic algorithm, which can result in a complex design. In contrast, we used simple square coils and optimized the position and size rather than the shape. The optimization of multi-coil for position, geometry, and number of the segments for 2 slices of a human brain B0 map is also reported in Zivkovic et al. However, no in vivo measurement has been reported to date, and neither any performance analysis of the design on the whole brain nor any comparison with higher-order SH and other multi-coils are yet available.

The target of the optimization in this study was global shimming of the human brain, which is more general and applicable for whole-brain 3D sequences or multiband 2D sequences. However, the performance was evaluated for dynamic slice-wise shimming in both simulations and in vivo measurements. The position of the coils could be optimized to improve the performance of dynamic slice-wise shimming, but different results are expected for different slice orientations, which may limit the optimization output for more specific applications. Furthermore, the optimization of the shimming performance for dynamic slice-wise shimming may not result in a significant improvement, because the region of shimming is small and thus symmetric configuration of coils can even yield similar shimming performance.

Optimization of size and position of a limited set of external coils similar to this study is presented in Juchem et al for magnetic field homogenization of the human PFC at 4T. However, the optimization routine and the utilized algorithm were not explained clearly enough for proper comparison. The resultant improvement of SD reported in Juchem et al for PFC and whole brain with respect to SH shimming is 29% and 3%, respectively, which in this study improvement is 10.6/9.5/32.7% and 37.6/34.9/55.6% (based on RMS is 30.9/14.8/35.2% and 57.1/38.8/55.1%) with global and slice-wise shimming, respectively (in experimental measurements).

However, it has to be noted that the target of optimization for these 2 studies is different. The presented combined 32-channel RF-shim coil setup in Stockmann et al for magnetic field homogenization of the human PFC at 4T, which in this study is 40% and 65.3% in simulation (after layering) and 32.7% and 55.6% in measurement for constrained global and slice-wise shimming, respectively. In Juchem et al, a SD of 25 Hz was reported in the experimental study after dynamic slice-wise shimming with a 48-channel multi-coil at 7T (100 turns per coil and shimming target volume of 1269 mL), which in this study is 31.9 Hz at 9.4T (25–50 turns per coil and shimming target volume of 1307 mL). A previously reported 16-channel multi-coil setup utilized for shimming of human brain at 9.4T yielded 13.1% and 38.1% improvement with respect to the second-order SH after constrained global and slice-wise shimming, respectively. In Juchem et al, where the same multi-coil setup as described in Juchem et al was used, it was reported that the T2 ratio increased by approximately 15% and 30% in PFC and TL, respectively, after dynamic multi-coil shimming at 7T, while in the current study the gain is 12.5% and 28% in PFC and TL, respectively (both studies were performed with an isotropic resolution of 2.0 mm). Overall, the achieved results in this study are in agreement with the reports in recent publications. There is an improvement in some of the obtained results, and the rest do not contradict with the results from previous literature.
The utilized solver for optimization was based on the SQP algorithm. The natively implemented SQP algorithm in MATLAB allows for nonlinear constrained optimization, fast execution, less memory usage, and no explicit definition of the gradient of the cost function (the gradient is estimated by finite differences). SQP converged faster than interior point algorithm in MATLAB; however, it may not be the best choice among all published solvers to this end. Figure 2C compares the performance of the 2 nonlinear constrained algorithms, SQP and interior point, in different iterations. Given that SQP updates the Hessian matrix at each major iteration, it performs better in locating the local minima and moves faster to its final result. The runtime of both algorithms at the same number of iterations was similar; however, SQP converged into a lower loss. Using a quad-core 3.5 GHz Intel Xeon processor, computation time to obtain the optimal coil arrangement was approximately 48 hours. The objective function of the problem may not be convex, and therefore the obtained solution may not necessarily be a global minimum. To this aim, optimization with 9 random initial coils positioning was performed. Coil arrangements after optimization for 9 different starting points did not yield an identical result; however, none of them performed considerably better than the case when symmetric coil positioning was chosen as the starting point (Figure 2C). The obtained optimized coil arrangements (with SQP algorithm) starting from random initial positions improved the shim performance by 13.2 ± 2.8% and 10.9 ± 3.5% with respect to their initial positions and the symmetric design, respectively. Therefore, it can be concluded that the objective function is not a convex function, and there is no guarantee that the obtained result after optimization (with the initial symmetric coil positioning mentioned earlier) is the most optimal coil arrangement. Nevertheless, there was an improvement and shimming with this new coil arrangement performs better by 14.7% in comparison to the conventional symmetric design. The optimization allowed coil overlapping, which was addressed later by layering the overlapped coils. Two coils may be overlapped entirely during the optimization iterations, which results in a non-full-rank basis-set matrix. However, this is not a problem as long as constrained shimming was performed in every iteration. Unconstrained shimming can also be applied during the iterative optimization process; however, the final arrangement must be evaluated with constrained shimming.

The initial bench test of the setup revealed slight in-place mechanical vibration of some local coils. Although the coils were fixed tightly, small imperceptible local vibration of the coils in outer layers did not allow adjusting the proportional-integral-derivative in the amplifier. As a consequence, the transient time of the output current did not follow a similar pattern after several sharp changes in input and even was getting worse. To address this issue, the free space in the cylinder was filled with an epoxy which eliminated the shaking of the coils. To be on the safe side, a thin plastic tube was circulated in the cylinder before using epoxy to set up a water-cooling system in the case of overheating. The weight of the setup increased considerably after epoxy filling from 15 to 27.4 kg.

Although the human brain is almost symmetric in anatomy, the coil arrangement shown in Figure 1A is not symmetric with respect to the YZ (sagittal) plane. There are 3 possible reasons. First, the center of the brain in the train data may not agree with the isocenter of the scanner, which is the center of the coordinate system in the optimization. Second, as seen in the first row of Figure 5, the B₀ map of the brain is not perfectly symmetric respect to the center sagittal plane. Magnet imperfection, other contamination sources, and head orientation can cause asymmetric brain B₀ maps. Third, even when ignoring the 2 reasons above, the solver may converge to an asymmetric arrangement to shim a symmetric B₀ map; for example, 2 small coils in the right side can perform equally as well as 1 big coil in the left side. However, the resultant coil arrangement after optimization matches the overall B₀ pattern of the human brain. Several coils are located in the anterior to address the inhomogeneity in the PFC. Some of the coils are small to cancel very local inhomogeneities or compensate for the adverse effect of the large coils in vicinities. On the contrary, only a few large coils are positioned in the posterior. There are also many coils in the left, left-anterior, right, and right-anterior to cancel out inhomogeneities around ear canals and to also support better shimming of the PFC.

The simulation results depicted in Figure 3 reveal that constrained and unconstrained global shimming yielded a similar shim performance for the 32-channel symmetric design. However, unconstrained shimming improved the performance of the optimized design. As a consequence, the performance of the symmetric arrangement is intrinsically limited regardless of current constraints, while the performance of the optimized design is limited by the current bounds. Increasing the current constraints to ±6.0 A resulted in 19% better performance in global shimming for the optimized design after layering with respect to the symmetric arrangement (which was 12.4% with current constraints based on coil size as reported in Table 1). It has to be noted that the absolute sum of the currents in all channels did not exceed 77.0 A on average over 14 brain B₀ maps. Such a performance gain is simply possible by increasing the coil windings to 100 turns (similar to Juchem et al14) if supplying higher currents is not desired.

Interestingly, adding more coils increased the difference in shim performance between constrained and unconstrained global shimming as can be observed in Figure 3. Smaller coils have a smaller penetration depth and their shim fields might not contribute to compensate more distant inhomogeneities while in case of unconstrained shimming the shim fields are strong enough to penetrate deeper and contribute
also to compensate more remote inhomogeneity. Another reason might be that smaller coils can more effectively shim small targets (e.g., near ear canals or PFC) without significantly affecting the neighbor voxels and degrading the magnetic field produced by adjacent coils. Therefore, their impact is limited by the supplied current. However, unconstrained shimming is not a realistic solution because of current amplifier limitations and thermal issues. For the case of constrained dynamic slice-wise shimming (axial slices), a multi-coil with 24 channels can shim with the same performance level like a multi-coil with 65 channels (assuming the proposed geometry). Hence, there is no performance gain for multi-coil optimization (with the DOF as mentioned earlier) if dynamic slice-wise shimming is the goal.

Figures 5-8 show the performance of the constructed multi-coil for different applications. The off-resonance level in the PFC and ear canals proximity is remarkably reduced. Although the multi-coil is optimized for global shimming, the performance for dynamic shimming has been preserved as well (and slightly got better). With dynamic shimming, the residual inhomogeneity at 9.4T became very close to what is obtained at the conventional 3T scanner after second-order global shimming.

Considering Figure 7, it is important to note that the bSSFP frequency response is periodic with 1/TR. Supporting Information Figure S10A displays a simulation of the bSSFP profile based on the utilized acquisition protocol as well as literature white matter and gray matter T1 and T2 values at 9.4T. The periodic profile illustrates that even when the field homogeneity is improved, new banding artifacts can be generated for certain areas, which were in the pass-band before shimming and then shifted into the stop-band of the bSSFP profile after shimming. An example for this issue is shown in Supporting Information Figure S10B after global shimming. Therefore, the number, density, and distance of the bandings after shimming have to be considered rather than individual artifacts.

Despite a reduction of the geometric distortion in the center area of some slices covering brain ventricles, some stretching artifacts after shimming have to be considered rather than individual artifacts. The MATLAB source code of the implemented multi-coil optimization in this study is available under the MIT license from the project website: https://github.com/Aghaeifar/optimized_multi_coil.

5 | CONCLUSIONS

Multi-coil shim setups can be optimized for specific targets to increase shim performance without adding additional coils. The performance of the 32-channel optimized multi-coil which was proposed in this study is comparable to a 65-channel multi-coil with conventional symmetric design. In comparison to the symmetric arrangement with the same number of coils for which constrained and unconstrained global shimming yields similar results, the efficiency of the proposed optimized coil is limited by the supplied current.

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SUPPORTING INFORMATION

Additional supporting information may be found online in the Supporting Information section at the end of the article.

FIGURE S1 The designed supports allowed easy layering of the overlapping coils. The thickness of the supports was 7.5 mm

FIGURE S2 Comparison of the $B_1^+$ field in the presence and absence of the constructed multi-coil. The spatial distribution and strength remained similar. An actual flip angle imaging (AFI) sequence was used to measure the $B_1^+$ field. The phantom filled with equivalent tissue properties (c558.6, r50.64 S/m)

FIGURE S3 Comparison of the SNR and noise correlation matrix in the presence and absence of the constructed...
multi-coil. The difference is minor and may be attributed to discrepancies in the positioning of the phantom. The SNR was calculated based on the pseudo multiple replica approach.

**FIGURE S4** Comparison of the temporal SNR and noise correlation matrix in the presence and absence of the constructed multi-coil. The temporal SNR was determined in a spherical agar phantom from 100 measurements.

**FIGURE S5** The produced heating attributed to the resistance of the coils is a major concern. To this end, all channels were fed with 1.5 A for an hour and the temperature was monitored with temperature sensors that were attached to the setup. Furthermore, infrared images with a thermal image (E6 thermal imager; FLIR Systems, Wilsonville, OR) were acquired at the end of the measurement. The hottest point was in the outer surface of the multi-coil where several coils are overlapped in the anterior. However, this area is not in contact with the subject.

**FIGURE S6** Redundancy in the magnetic fields generated by the individual coils before and after optimization. (A) Matrix of correlation coefficients. Numbers close to zero represent less similarity in the magnetic field generated by 2 individual coils. (B) The cumulative sum of shim coils eigenvalues. Summation of the first 14 and 10 principal components reached to 80% of total eigenvalue energy for symmetric and optimized design, respectively.

**FIGURE S7** Evaluation of the geometric distortion and voxel shift map at resolution of 1.5-mm isotropic and 2.0-mm isotropic for the same subject depicted in Figure 6. EPIs with 2.0-mm isotropic resolution are acquired without and with an acceleration factor of 2.

**FIGURE S8** Evaluation of the geometric distortion and voxel shift map at resolution of 1.0-mm isotropic (GRAPPA factor = 3) and 2.0-mm isotropic (no acceleration) for subjects 2 to 5.

**FIGURE S9** Effect of the improved B0 homogeneity on banding artifacts in bSSFP images for subjects 2 to 4.

**FIGURE S10** (A) The periodic profile of bSSFP frequency response simulated based on the acquisition protocol used in this study (in our case, 1/TR corresponds to a periodicity of 66 Hz). (B) An example of generating new banding artifact after global shimming while the shimming is improved. Because of the periodic profile, some areas may be shifted from pass-band into stop-band after shimming.

**TABLE S1** Evaluation of improvement in shimming performance of optimized multi-coil with respect to the symmetric design in global shimming for individual subjects.

**TABLE S2** Information of coils layering to overcome the overlapping between the coils after optimization. The base for choosing a layer for a coil was to keep the most coils in layer1, then layer2 and so forth.

**TABLE S3** The wiring pattern of the designed 32-channel optimized multi-coil according to the public multi-coil information policy.

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