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15 **A 31-channel integrated “AC/DC” B₀ shim and RF receive array coil for**
16 **improved 7 Tesla MRI**
17
18

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ABSTRACT:

PURPOSE: To test an integrated “AC/DC” array approach at 7 Tesla, where B_0 inhomogeneity poses an obstacle for functional imaging, diffusion-weighted MRI, MR spectroscopy, and other applications.

METHODS: A close-fitting 7T 31-channel brain array was constructed and tested using combined Rx and ΔB_0 shim channels driven by a set of rapidly-switchable current amplifiers. The coil was compared to a shape-matched 31-channel reference receive-only array for RF safety, signal-to-noise ratio (SNR), and inter-element noise correlation. We characterize the coil array’s ability to provide global and dynamic (slice-optimized) shimming using ΔB_0 field maps and echo planar imaging (EPI) acquisitions.

RESULTS: The SNR and average noise correlation were equivalent to the 31ch reference array. Global and slice-optimized shimming provide 11% and 40% improvements respectively compared to baseline 2nd-order spherical harmonic shimming. Birdcage transmit coil efficiency was similar for the reference and AC/DC array setups.

CONCLUSION: Adding ΔB_0 shim capability to a 31ch 7T receive array can significantly boost 7T brain B_0 homogeneity without sacrificing the array’s RF performance, potentially improving ultra-high field neuroimaging applications that are vulnerable to off-resonance effects.

INTRODUCTION

Ultra-high field (UHF) brain imaging at 7 Tesla provides improved signal-to-noise ratio (SNR) and contrast-to-noise ratio (CNR) compared to conventional field strengths of 1.5 and 3 Tesla MRI. The SNR gain enables high resolution imaging useful for depicting small brain structures, while the improved CNR derives from increased contrast such as T_2^* for blood oxygenation level-dependent functional MRI (BOLD fMRI) and improved T_1 contrast for MR angiography. However, it has been difficult to achieve the full potential of UHF

1 MRI in practice due to both static and time-varying perturbations of the static
2 magnetic field (ΔB_0) introduced by the human body. These field inhomogeneities
3 lead to a variety of image artifacts (1). *Static* ΔB_0 inhomogeneities arise primarily
4 at the air-tissue interfaces (near the sinus, ear canal, and oral cavities). They
5 cause a variety of problems including: (i) disruption of the RF pulse excitation in
6 limited bandwidth pulses (such as shaped excitation pulses) as well as frequency
7 selective pulses such as fat saturation pulses; (ii) spectroscopic line broadening;
8 and (iii) signal voids, geometric distortion and blurring along the phase-encode
9 direction in echo planar imaging (EPI) data. In standard gradient-echo BOLD
10 fMRI, this leads to a loss in BOLD sensitivity (2), region-dependent resolution
11 loss, difficulty in registering distorted EPI to a reference anatomical image (3),
12 and in some cases singularities in the distortion that cause unrecoverable losses
13 of information (3). The B_0 inhomogeneity is also problematic for measurements of
14 fine-scale microscopic and mesoscopic anatomy based on T_2^* since the
15 overlaying macroscopic ΔB_0 patterns can mask subtle microstructural details.
16 *Time-varying* ΔB_0 shifts arise from motion and motion-related physiological
17 processes such as respiration. These lead to ghosting and ringing artifacts in
18 multi-shot structural images as well as phase instability (time-varying distortion)
19 in single-shot EPI time series. Unfortunately, the 1st- and 2nd-order spherical
20 harmonic (SH) based shim coils available on most 7T scanners offer limited
21 capacity for nulling the high-spatial order ΔB_0 profiles. Moreover, commercial 2nd-
22 order SH shim coils are typically not set up for rapid switching to compensate
23 time-varying ΔB_0 .

24
25 Both post-processing (4) and acquisition acceleration methods (5) have been
26 used to reduce ΔB_0 induced geometric distortion in single-shot EPI at the
27 expense of sacrificed spatial resolution and SNR. While now standard practice,
28 significant geometric distortion remains, especially as measured in units of pixel
29 displacement in high-resolution single-shot EPI (6). In contrast, improved B_0
30 shimming addresses geometric distortion at its source. Higher-order SH coils
31 built into the gradient coil or operated as shim insert coils have the longest track
32 record for B_0 shimming at 7T. Global 1st-to-4th order SH shimming of the whole

1 brain at 7T has been used to improve B_0 homogeneity by 25% (σ^{WB} : standard
2 deviation of ΔB_0 over the whole brain) and reduce geometric distortion in EPI
3 (typically using up to 10A per SH channel) (7). SH shim currents can also be
4 dynamically switched to optimally shim each slice in 2D acquisitions, however the
5 size of these structures requires attention to eddy current effects through the use
6 of pre-emphasis (8,9), probe monitoring (10), and/or constraints on current
7 switching (11).

8
9 A second approach uses close-fitting multi-coil (MC) shim arrays with small
10 loop coils patterned around the anatomical region-of-interest to generate non-
11 orthogonal B_0 shim fields at high spatial orders (12),(13). These small, low-
12 inductance coils can be driven by low-voltage, low-cost amplifiers (14). Because
13 they are mounted nearer to the head, away from conductive structures in the
14 magnet, they can be switched rapidly without inducing eddy currents (15)(16)
15 and experience less induced voltage from gradient coil slewing. It has been
16 shown that B_0 -shim capability of small loops can share the structure (loop) of the
17 Rx-coil element itself. In this approach, inductive chokes provide a DC bridge
18 across the RF tuning capacitors (17–19). This approach has been employed for
19 Rx/Tx loops as *Integrated Parallel Receive, Excitation, and Shimming (iPRES)*
20 and for Rx-only arrays as the “AC/DC” coil array. Either enables a single array of
21 loops on a close-fitting helmet to be used for both B_0 shimming and RF purposes.
22 Despite the single-turn nature of most RF loop array, their close proximity to the
23 body allows sufficient shim performance with modest current amplitudes (1–2A).
24 While originally proposed as a tool for static or slice-by-slice optimized shimming
25 with a given setting applied for the whole TR period, MC arrays have recently
26 been used as a tool for dynamic *local field control* by exploiting the ability to
27 rapidly switch to different B_0 states within the TR period. For example, a B_0
28 pattern can be created during RF excitation which facilitates zoomed imaging
29 (20,21), improves lipid suppression for spectroscopy (22)(23), or reduces flip-
30 angle inhomogeneity (24) and then returns to a “uniform B_0 ” setting during
31 readout. MC arrays with greater numbers of turns have also been explored for

1 image encoding as a supplement or replacement for conventional linear gradient
2 coils (25,26)(27).

3
4 While beneficial at 3T (28),(29) and for primate imaging at 7T (30), the
5 AC/DC or iPRES approach has not been shown for human imaging at 7T, where
6 the need for shimming is greater. Additionally, technical barriers increase at
7 ultrahigh field due to higher current needs, increased number of distributed RF
8 capacitors, and exacerbated RF tuning and coupling challenges. In this work, we
9 designed a 7T AC/DC coil array – with 31 RF receive channels and 31 B_0 shim
10 channels – that addresses these technical barriers, constructed the array, and
11 assessed its RF and dynamic B_0 -shimming performance. We validated the
12 shimming capabilities through B_0 field mapping and EPI-based brain imaging on
13 healthy volunteers.

14
15 We previously explored the feasibility of adding B_0 shim capability to
16 individual 7T receive elements (33); the AC/DC modifications for shimming
17 produced similar SNR maps before/after the modification. Preliminary results
18 from the full AC/DC array were reported in a recent review paper (31) and
19 abstract (32).

20 21 **METHODS**

22
23 The complete hardware setup for B_0 shimming experiments is outlined in
24 Figure 1. The equipment components can be divided into three categories by
25 their position either in the magnet room, the console room, or the equipment
26 room.

27 28 **Magnet Room subsystems**

29 30 **Coil element design**

31 To inform the coil element design, we first assessed RF transmit and receive
32 performance losses associated with the extra circuitry for the DC current path in

1 the AC/DC coil. The loaded-to-unloaded quality factor (Q) ratio of a single RF
2 test loop was measured before and after adding the shim components to the loop.
3 The coil impedance match, PIN diode detuning, and preamp decoupling
4 characteristics were also measured.

5
6 We first compare coil performance metrics to an industry-standard, vendor-
7 provided 32-ch array with birdcage transmit (Nova Medical, Wilmington, MA).
8 We then compared coil performance metrics to an existing, in-house-built 31-
9 channel (31-ch) RF Rx-only array with the same helmet geometry but lacking the
10 shimming capacities. The basic RF circuit design was the same for the Rx-only
11 and AC/DC arrays except for the number of distributed tuning capacitors. Fewer
12 distributed capacitors were used on the AC/DC coil elements (4 instead of 6) to
13 reduce the number of bridging chokes needed. Wire gauge choices were made
14 after simulating the tradeoff between B_0 shim performance and maximum
15 current-per-channel (see “ B_0 shimming simulations”).

16
17 Details of the Rx and Tx designs are provided as Supporting Information. In
18 summary, both in-house arrays used 31 RF receive elements rather than 32 to
19 allow one of the system receive channels to be available for volume coil
20 reception (useful for testing). Of the Rx elements, 25 were set up as AC/DC
21 elements. The two RF loops above the eyes, which share a conductor where the
22 two loops meet, were set up as a single DC shim loop. The five most inferior and
23 posterior neck loops were not set up for shimming based on (i) the difficulty of
24 running wires all the way through the coil to these loops, and (ii) preliminary B_0
25 shimming simulations suggesting that they provide only minor gains for B_0 -shim
26 performance. To bring the number of shim channels up to 31, six multi-turn
27 shim-only loops were added over the face area to target prefrontal cortex, an
28 idea previously explored experimentally by Juchem et al. (34) and Zhou et al.
29 (35), as well as in simulations (1). These 4-6cm dia. “shim-only” coils consisted
30 of 3–4 turns of 22 AWG wire with 3 chokes in series to ensure that any self-

1 resonances were shifted at least 50 MHz above or below the Larmor-frequency
2 (297.2 MHz).

3
4 B_0 shim capability was added to RF loops by using 325nH toroidal (self-
5 shielded) inductive chokes to bridge the loop's tuning capacitors (Figure S1).
6 Additional chokes placed at intervals of roughly $\lambda/10$ apart were used to block RF
7 on the twisted-pair DC feed wires.

9 **Helmet design and layout**

10 Figure 2 shows the CAD model for the close-fitting helmet alongside
11 photos of the two halves of the RF array before and after the addition of shim
12 chokes and feed wires. Figure 2c shows the receive array mounted inside the
13 birdcage transmit coil. The birdcage and top half of the receive array slide back
14 to make it easier for subjects to position their head inside the housing.

16 **Shim amplifiers**

17 Figure 2d shows the open-source 8-channel amplifier boards used to
18 provide digitally-programmable-shim currents that can be switched rapidly (as
19 fast as 30 μ s) (36),(37). Schematics and layout files along with Matlab circuit
20 simulation software are available in the support material and online
21 (<http://rflab.martinos.org>). The shim amplifiers are placed at the back of the
22 magnet room and are connected to the coil with a cable along the scanner's
23 energy chain (see *Supporting Information for further details*).

25 **Control components**

26 The amplifiers are controlled via fiber optic lines entering the scanner room
27 through a waveguide. A custom graphical user interface written in MATLAB
28 (Mathworks, Natick, MA, USA) is used to process *in vivo* brain ΔB_0 field maps
29 and calculate optimal shim currents using the shim coil basis set of 31 ΔB_0 maps.
30 A laptop computer sends the current amplitudes via USB cable to a Teensy 3.5

1 microcontroller (PJRC, Sherwood, OR, U.S.) which in turn relays commands to
2 the 8-channel LTC2656 digital-to-analog-converter on each amplifier board.

4 **Human subjects**

6 Eight healthy volunteers (4 female, 4 male) provided informed consent
7 and were enrolled for system performance evaluations. All experiments were
8 approved by the Institutional Review Board of Massachusetts General Hospital.
9 Four of the volunteers were used for RF performance assessment and the
10 remaining four were used for B_0 shimming experiments.

12 **RF performance assessment**

15 All experiments used a Siemens Magnetom whole-body 7 Tesla scanner
16 with an SC72 gradient coil ($G_{\max} = 70$ mT/m, slew rate = 200 T/m/s) equipped
17 with 2nd-order static shim coils. Standardized phantom tests were performed to
18 identify any RF focusing or component heating from interactions between the Tx
19 and Rx coil or between the gradient and RF components (38). RF performance
20 evaluations were run on the vendor array, the Rx-only 31-ch reference array, and
21 the AC/DC array, including transmit B_1^+ mapping, receive SNR, noise correlation,
22 and geometry factor (g-factor) of each Rx array. B_1^+ maps were acquired using
23 the Actual Flip-angle Imaging (AFI) method (39) to assess the impact of shim
24 chokes and wiring on birdcage coil transmit efficiency (3D gradient-echo,
25 TR=5.8/28 ms, TE=2.73 ms, 3 mm isotropic, 64×64×56 matrix, nominal FA=60°,
26 readout BW=260 Hz/pixel, TA=2:07). The maps were then normalized by the
27 transmit voltage to obtain B_1^+ efficiency in units of nT/V. The receive SNR
28 measurement followed the method of Kellman *et al.* using a noise-covariance-
29 weighted optimal coil combination (equation 5 in (40)). A proton-density-
30 weighted gradient-echo image was acquired using a nominal flip-angle of 90° to
31 limit the impact of B_1^+ inhomogeneity on signal intensity over the FOV (axial,
32 TR=10s, TE=3.82ms, matrix=[256×176], FOV=[256×176] mm, 2mm slice,

readout BW=340 Hz/pixel, TA=14:42). Following the approach in Ref. (41), the impact of B_1^+ variability was further reduced by dividing the SNR maps by the sine of the flip-angle from the AFI maps. The noise correlation matrix was computed using raw complex-valued noise data acquired on each receive channel with the transmit voltage set to zero.

■ A 3D multi-echo magnetization-prepared rapidly-acquired gradient-echo (ME-MPRAGE) acquisition was used to qualitatively compare assess the impact of B_1^+ inhomogeneity and efficiency on image quality for AC/DC and Rx-only 31-ch arrays (42) (3D gradient-echo, TR=2550ms, TI=1100ms, TE=[1.62, 3.44, 5.26, 7.08] ms, 0.88×0.88×1 mm, GRAPPA $R=3$, matrix=[226×226×176], FA=9°,TA=3:37). Anatomical imaging was also assessed in high-resolution structural gradient-echo images (axial, TR=1000ms, TE=21ms, 300μm or 400μm in-plane, matrix = [512×512], FOV=208×208 mm, 10 slices with 600% gap, 1-mm slice thickness, a phase stabilization navigator was used to track B_0 drifts, TA=4:34). Full-detail protocol summaries are included as Supporting Information.

B_0 shimming simulations

To assess the benefits provided by the 6 face-shim elements, we simulated B_0 shim performance of three array geometries: a 30-ch design where all RF coil elements were used as AC/DC elements (with eye-loops tied together); the hybrid 31-ch design with 25 AC/DC elements and 6 face-shim elements which we planned to build for this study; and a 36ch design with 30 AC/DC elements and 6 face-shim elements. Wire patterns for all three geometries are included in PUMCIN format as Supporting Information (43). Bio-Savart simulations (44) were used to compute the static magnetic field generated by each loop. Using the field basis set for each array we attempted to shim the

1 measured field distributions from all 8 volunteers. The ΔB_0 field maps for these 8
2 individuals were acquired using a vendor-provided gradient-echo two-echo field
3 mapping sequence with relatively thin slices (1.3mm) to limit through-plane
4 dephasing. The scan parameters included TE=[2.0 3.02] ms, TR=465ms, FA=40°,
5 matrix=112×112, FOV=212×212mm (2mm in-plane resolution), readout
6 BW=1207 Hz/pixel, 100 slices, 1.3mm slice thickness, TA=1:47. The magnitude
7 image obtained from the field mapping sequence was masked using the FSL-
8 Brain Extraction Tool (BET) (45) using a threshold of 0.2 and residual phase
9 wrap was removed using FSL-PRELUDE (46). The “whole-brain” shimming
10 simulations used the entire brain mask over the full cerebrum and cerebellum
11 (typically spanning 14–15cm in the superior-inferior direction). Stray voxels were
12 removed by applying the MATLAB ‘imerode’ function and by thresholding to
13 remove voxels with intensities less than 0.3 times the average of the magnitude
14 image. The maps were smoothed using the Matlab ‘wiener2’ pixel-wise adaptive
15 low-pass filter with a 3×3 kernel.

16
17 MATLAB’s ‘quadprog’ solver was used to calculate optimal shim currents
18 within the mask using a least-squares penalty on the residual ΔB_0 subject to
19 current constraints for each loop and for the entire array. The optimization
20 problem is framed as a quadratic program with a Tikhonov-regularization term
21 added to prevent noise in ΔB_0 field maps from causing large changes in the
22 optimal computed solution. Regularization also reduces the size of current steps
23 between adjacent slices. The first step in the simulations was to globally shim
24 using 1st- and 2nd-order SH terms to ensure that the starting maps were
25 corrected to this order using the same algorithm, optimization region, and metrics
26 applied to the higher-order shim terms. The final shim was characterized by the
27 standard deviation over the brain region mask (σ_{WB}), the standard deviation
28 within each slice (σ_{SL}), as well as the 80%, 90%, and 95% whole-brain ΔB_0
29 residuals ($\Delta B_{80\%}$, $\Delta B_{90\%}$, and $\Delta B_{95\%}$). Multi-coil B_0 shimming was examined (a)
30 globally over the whole brain and (b) dynamically on a slice-by-slice basis. Slice-

1 optimal shims were computed using three adjacent slices to limit through-slice
2 field variation of the applied shim fields.

3
4 We compared conventional static shimming using up to 2nd-order spherical
5 harmonics (“2SH”) for both global shimming and slice-optimized dynamic MC
6 shimming. For all MC shim simulations and experiments, we included the 1st- and
7 2nd-order SH shims applied as global (whole brain) shims prior to adding the MC
8 shim terms. For dynamic shimming simulations, we allow the first order SH
9 components (gradient offset currents) to be dynamically updated on a slice-by-
10 slice basis since this is, in principle, achievable on most scanners through offset
11 currents to the gradients. The benefit of doing this in conjunction with MC
12 shimming has been previously demonstrated (13). Additionally, we simulated
13 shimming with SH basis fields up to 5th order. The SH shim results were
14 checked for consistency by computing the solution both with an unconstrained
15 pseudoinverse and a constrained ‘quadprog’ solver with very high current limits.

16
17 Simulations were also used to evaluate the shim performance of 31-ch
18 AC/DC coil as a function of current limits varying from 1A to 15A per element and
19 total current constraints from 10A to 400A; the results were used to guide the
20 choice of wire gauge for the DC shim components (see “Coil element design”
21 above). To provide some basis for comparison, we also simulated the SH basis
22 up to 5th as well as a reference 48-ch 7T MC shim array from the literature
23 consisting of 100-turn loops patterned on a cylinder around the head (47).

24
25 Electric field (E-field) simulations in a realistic head model (Sim4Life, Zürich,
26 Switzerland) (48,49) were computed for a 1 KHz sinusoidal current to assess
27 whether rapid switching of the shim fields could potentially induce peripheral
28 nerve stimulation (PNS). Specifically, we examined whether the maximum
29 current applied at 1 KHz generates E-fields in excess of 5 V/m (50) or dB/dt fields
30 in excess of the IEC guideline of 20 T/s (51).

31 32 **B₀ shimming experiments**

1
2 The shimming basis functions needed to compute the *in vivo* shim
3 currents were obtained once in a phantom measurement. The phantom was
4 designed in CAD software to closely match the helmet to maximally extend the
5 field measurement region. The 3D printed phantom (Figure 2b) was filled with
6 silicone oil (Super Lube, P/N-56101, Bohemia, NY) rather than water to provide a
7 more homogeneous B_1^+ profile. The B_z -field component generated by each shim
8 element was obtained using the two-echo B_0 field mapping sequence after
9 applying +750mA and -750mA to each coil loop. The difference between these
10 two field maps was computed to remove the background field. The maps were
11 then normalized to provide ΔB_0 in Hz/A. We also applied a similar method to
12 create basis maps for the scanner's 1st-and-2nd-order SH shims.

13
14 For B_0 shimming experiments, we adjusted the 1st- and 2nd-order SH global
15 shims using two methods: (i) the vendor-provided online field-mapping and shim
16 optimization tool and (ii) our own offline mapping and optimization tools which
17 use a more detailed masking procedure.

18
19 After setting the SH shims, we acquired a field map of the remaining ΔB_0
20 field and attempted to remove these variations with the shim array. Shim array
21 currents were computed using the optimization described in the shim simulation
22 section above. After applying the MC shims, a final field map was acquired to
23 assess performance. All calculations were run on a laptop equipped with 16GB
24 RAM and a 4-core 2.7GHz i7-processor (Apple, Cupertino, CA, USA).

25
26 Validation using EPI acquisitions. Single-shot gradient-echo 1mm EPI scans
27 were acquired with phase-encoding (PE) set to the anterior-posterior (AP) as well
28 as posterior-anterior (PA) directions to observe the geometric distortion arising
29 from residual ΔB_0 . (1mm axial slices, 300% slice gap,
30 TR/TE/flip=2560ms/25ms/80°, matrix=174×174, FOV=192×192mm, GRAPPA
31 R=4, readout BW=1596 Hz/pixel, echo spacing=0.83ms yielding an effective
32 echo spacing of 207 μ s after considering the R=4 acceleration. Trigger output

1 pulses were added to both the field mapping and EPI sequences (including
2 during GRAPPA autocalibration pre-scans) to signal the microcontroller to update
3 the shim currents.

4
5 The EPI slices before and after shimming were compared to a contrast-
6 matched T_2^* -weighted gradient-echo volume, providing a distortion-free (in PE
7 direction) reference. Temporal SNR maps (52) were computed from EPI time-
8 series data for the baseline, global MC, and slice-optimal MC shims using 60
9 consecutively-acquired volumes that were motion-corrected using FSL.

10
11 Acoustic noise levels during current switching from 0 to 1.5 amps on every
12 channel were recorded inside the coil helmet using a Bruël-and-Kjaer sound level
13 meter (Naerum, Denmark) and compared to sound levels generated by the
14 scanner gradient coil running localizer and EPI sequences.

17 **Results**

20 RF performance

21
22 Figure 3 shows RF bench performance metrics measured on a single RF
23 test coil element before and after the addition of DC shim chokes and twisted-
24 pair feed wires. The coil element input-impedance changes only slightly when
25 the DC shim components are added. Supporting Information Figure S2 shows
26 that properly tuned chokes approximate an open circuit and thus cause a
27 negligible shift in the RF resonant frequency. The shim hardware caused a small
28 shift of roughly 3 MHz, which was easily corrected using a trimmer capacitor.
29 Figures 3c–d show double-probe $|S_{21}|$ measurements for assessing preamp
30 decoupling and PIN-diode detuning. Although the $|S_{21}|$ curves shift slightly, a
31 large peak-to-trough distance is preserved in both cases (more than 35dB),

1 showing that the DC shim components do not adversely impact PIN-diode
2 detuning or preamp decoupling.

3
4 The change in coil-element quality factor (Q) before and after adding DC
5 shim components was as follows. Without shim components the Rx-only element
6 had $Q_{\text{unloaded}}=207$ and $Q_{\text{loaded}}=22$ giving an unloaded to loaded Q ratio of $R_{\text{ul/l}} =$
7 9.4. With the components needed for the AC/DC array: $Q_{\text{unloaded}}=183$, $Q_{\text{loaded}}=22$,
8 $R_{\text{ul/l}} = 8.3$.

9
10 The RF transmit-receive coil system passed all RF and gradient eddy-current
11 heating safety tests in the 7T scanner with less than 4°C temperature rise on the
12 coil surface. B_1^+ maps in Figure 4 quantitatively compare the transmit efficiency
13 of the birdcage coils used with the vendor array, “Reference” (non-shimming)
14 array, and the AC/DC array. For the reference and AC/DC arrays, the mean and
15 standard deviation of B_1^+ values over the brain volume agree to within 10%. A
16 few differences are noted, such as greater left-right asymmetry in the temporal
17 lobes for the AC/DC array setup. The vendor birdcage shows modestly higher
18 efficiency, which may be partially attributable to its smaller size (e.g., rung
19 diameter of 29.5 cm vs. 33.8 cm for the in-house birdcages). Figure 4b shows
20 co-registered T_1 -weighted MPRAGE images acquired using the in-house coils.
21 Some loss of contrast occurs due to B_1^+ dropout, particularly in the inferior
22 temporal lobes and cerebellum, but these flip-angle patterns are typical for
23 single-channel 7T birdcage transmit coils and overall image quality is comparable
24 for the two arrays.

25
26 Figure 5 shows SNR maps, noise correlation coefficient matrices, and
27 inverse g-factor maps. The SNR maps show roughly equivalent performance for
28 the three arrays. The average off-diagonal noise correlation coefficient is also
29 comparable for the two in-house arrays (13.3% and 13.7%) and inverse g-factor
30 maps show similar retained SNR for accelerated imaging. Figure S3 shows 300
31 μm and 400 μm in-plane T_2^* -weighted gradient-echo images acquired with the

1 AC/DC array demonstrating that the coil has adequate sensitivity and stability to
2 perform high-resolution 7T structural imaging.

3 4 5 B₀ shim simulations

6
7
8 Figure 6 summarizes simulated B₀ shim performance over eight ΔB_0 brain
9 field maps (current was limited to 3.5A/ch or 40A total). Further shim
10 performance metrics including residuals are summarized in Supporting
11 Information Table S1. Figure 6a shows the three simulated array geometries.
12 Global shimming (Figure 6b) with the 31ch “hybrid” AC/DC array including face
13 elements improves σ^{WB} by 21% compared to 2SH shimming, versus a 9%
14 improvement for a 30ch AC/DC-only array. For slice-optimized shimming (Figure
15 6c), the hybrid 31ch array achieved a 50% improvement over the 2SH baseline
16 compared to 44% for the 30ch array. As shown by the 36ch hybrid design
17 results, adding shim capability to the remaining 5 RF-only coils omitted from the
18 31ch design brings only modest additional gains.

19
20
21 Figure 7 plots simulated shim performance (σ^{WB}) on 8 brain B₀ maps for the
22 31ch hybrid design (which we chose to construct) as a function of maximum shim
23 current. The “L-curves” show steep performance gains up until the knee of the
24 curve is reached, beyond which the performance slowly converges to the
25 theoretical limit (pseudoinverse solution). We chose the wire gauges to allow
26 3.5A per channel and 40A total for the fabricated array (without excessive
27 heating), lying near the knee of the L-curves for both global and slice-optimized
28 shimming. At this operating point, the gains in σ^{WB} compared to 2SH shimming
29 are 17% and 49% for global and slice-optimized cases (similar to the 4SH basis),
30 respectively, compared to 29% and 65% in the unconstrained cases (similar to
31 5SH basis). Figures 7c and 7e show the simulated channel-wise current used for
32 the 8 field maps for global and slice-optimized shimming. Above the 3.5A/40A

operating point, the remaining performance gains come largely from improved shimming of the frontal lobes, with an outsized role played by the face-elements. For slice-optimized shimming, by contrast, increasing the current limits results in a more even spread of the added current across the array.

B₀ shimming experiments

Table 1 lists σ^{WB} and residuals for both the predicted and acquired ΔB_0 field maps with global 2SH, global MC, and slice-optimized MC shims applied. Across all four subjects, σ^{WB} improved by 11% and 40% for global and slice-optimized MC shimming, respectively (or 14% and 44% for the 95% residual). The table also summarizes EPI voxel-shift metrics (computed for each voxel using the ΔB_0 maps and effective echo-spacing) as well as the fraction of voxels that improve or worsen. Figure 8 provides a more detailed look at shim performance for each case, showing representative slices for each volunteer, plots of the st. dev. within each slice (σ^{SL}), and ΔB_0 histograms over the whole brain. MC shimming narrows the ΔB_0 histogram and reduces the number of outlier voxels that correspond to the largest geometric distortions in EPI slices.

Case	ΔB_0 Residuals (Hz)				EPI voxel shift (#)		Percentage of voxels with less vs. more voxel shift relative to 2SH shim	
	St. dev. (σ^{WB})	80%	90%	95%	Mean	St. dev.	Less (%)	More (%)
2 nd order shim baseline	53.8	85.7	126.4	208.5	1.1	1.8	N/A	N/A
Global MC (predicted)	47.9 (46.2)	62.0 (58.0)	107.0 (114.8)	187.3 (181.6)	0.8	1.6	69.8%	30.2%
Slice-optimal MC (predicted)	32.2 (30.8)	47.0 (39.9)	78.8 (72.3)	120.4 (111.5)	0.6	1.1	67.4%	32.5%

Table 1. Metrics of B_0 shim performance and EPI voxel shifts for experimental data shown in Figs. 8, 9, and S3-S5. The mean, st. dev., and residual metrics are computed over the 4 subjects scanned in the B_0 shimming experiments.

For 2SH shimming, σ^{WB} improves from 61.2Hz to 53.8 Hz (12%) when the off-line shim optimization and brain mask are used instead of the vendor-provided online shim adjustment.

Figure 9 shows representative EPI slices and ΔB_0 field maps for a dynamic shimming experiment on one volunteer (results for the other 3 volunteers are shown in Figures S5, S7, and S8). MC shimming reduces geometric distortion by an average of 41.3% over the brain, bringing many features back into closer alignment with the T_2^* -weighted anatomic image. The change in distortion is more apparent in Supporting Information Video S2, which toggles between EPI slices acquired with the two phase-encode directions (see also Videos S1, S3 and S4 for the other 3 subjects).

The EPI slices do not show ghosting, spiking, or other effects that typically result from coil instability. Temporal SNR maps derived from EPI time series data are similar for the static 2SH and dynamic MC shim cases, showing that shim current settling times and mechanical torque transients during shim updating are too small and/or too short-lived to measurably impact coil stability. A fuller presentation of the tSNR maps is shown in Figure S9.

Figure S10 shows measured ΔB_0 maps in the three cardinal planes for all 31 shim channels. The maps have been made available as Supporting Information along with the eight co-registered *in vivo* ΔB_0 field maps.

In dynamic shimming EPI acquisitions, reduced signal intensity was observed in some brain regions when slice-optimized shim currents were updated *prior* to the fat saturation pulse at the beginning of each TR. Figure S11

1 shows that this effect is caused by applied B_0 offsets outside the acquired slice
2 that frequency-shift water into the bandwidth of the fat saturation pulse, resulting
3 in incomplete T_1 recovery in those areas. The problem was solved by adding an
4 additional trigger to disable the MC shim fields during fat saturation.

5
6 Maximum sound pressure levels measured inside the coil helmet during
7 current switching were 90 dBA. This is two orders of magnitude lower than the
8 110 dBA recorded from the scanner gradient coil during a localizer scan.
9 Volunteers reported hearing a sound on the coil helmet during dynamic shim
10 updating (qualitatively described as “tapping” or “buzzing”), but no subjects
11 reported discomfort.

12
13 Figure S12 shows maximal-intensity projections of the E-field distribution in
14 the face area for a four-turn shim-only coil driven by 1A of 1KHz sinusoidal
15 current. The E-field in this case reaches a peak amplitude of approximately 0.015
16 V/m inside the body. Since this amplitude scales with rise time and the current,
17 for switching from -3.5A to $+3.5\text{A}$ in $250\mu\text{s}$, the E field is expected to be 0.45 V/m,
18 which is still safely below the 5 V/m PNS threshold seen for head-gradient
19 systems (50). The maximum B-field change is 0.6mT and dB/dt is 2.4T/s. This is
20 below the IEC guidance of 20T/s (51).

31 DISCUSSION

32

1 We extend the integrated “AC/DC” brain array approach to 7T, building upon
2 our previous realization at 3T (28). RF performance tests show broadly similar
3 SNR, average noise correlation coefficient, g-factor maps, coil stability, and
4 transmit efficiency for the AC/DC and a size-matched reference array. By
5 providing a large number of channels for both RF receive and B_0 shimming, the
6 AC/DC coil gives access to multiplicative reductions in geometric distortion: a
7 factor provided by parallel imaging acceleration and a further factor of nearly 2
8 provided by dynamic slice-optimized B_0 shimming. This gives the user the option
9 of either (i) using conventional acceleration-factors and reducing the image
10 distortion via the improved shim, or alternatively (ii) reducing the acceleration-
11 factor, thereby increasing SNR and image quality, and achieving conventional
12 levels of distortion.

13
14 A key challenge of the AC/DC approach is making efficient use of the space
15 inside the birdcage transmit coil while minimizing coupling between the transmit
16 fields and the DC shim components. Spatial nonuniformity of 7T birdcage coil
17 B_1^+ profiles resulting from dielectric wavelength effects is a well-known issue,
18 which has been shown in simulations of ideal birdcage coils (53). However, the
19 increased right-left asymmetry in the AC/DC case suggests that the chokes used
20 in the DC wiring have not entirely eliminated coupling between the birdcage coil
21 and the DC components. A possible remedy may be to route shim feed wire
22 paths more carefully so as to avoid areas of high electric-field and/or add more
23 chokes to the wires. Further investigation of this topic is needed.

24
25 The DC shim components cause an 11% drop in unloaded quality factor in
26 single-element bench testing. The cause of this Q loss is not well understood by
27 us, but factors could include (a) the AC resistance of the chokes and/or (b)
28 “copper shading” of the loop fields due to RF eddy-currents induced in the
29 chokes.

1 A discussion of practical issues related to the dynamic shimming
2 experiments has been included as Supporting information.

3
4 Slice-optimized shimming reduced σ^{WB} to 32.2 Hz across four volunteers at
5 7T. For comparison, the 2SH shim routine of the manufacturer at 3T typically
6 achieves σ^{WB} values in the low-to-mid-20s of Hz. Thus, the shim array does not
7 quite bring the 7T homogeneity down to what users are accustomed to at 3T. As
8 shown in Fig. 8c, the gains provided by slice-optimized shimming vary
9 considerably from slice to slice. The superior half of the brain shows large
10 improvements, with low residual ΔB_0 after AC/DC shimming (σ^{WB} =5–15 Hz). By
11 contrast, the inferior slices show less consistent gains, since focal B_0 “hotspots”
12 in the frontal and temporal lobes are impossible to fully null with the available
13 AC/DC coil basis set, which cannot generate field profiles of sufficiently high
14 spatial order.

15
16 Global MC shimming provided only 11% improvement over 2SH shims,
17 suggesting that there is room for improvement in future iterations of the coil
18 design. However, we note that the global shim performance of any MC shim
19 array will converge toward an asymptote as more and more shim sources are
20 added. For B_0 offset fields caused by sources outside the ROI, LaPlace’s
21 equation implies that the field can be shimmed to any desired level of
22 homogeneity if a sufficiently large number of shim coil basis functions are used
23 (e.g., shimming an empty magnet bore). But when there are tissue susceptibility
24 interfaces located inside the body, LaPlace’s equation is no longer valid (54),(55).
25 The resulting fields can be represented as a combination of regular and irregular
26 solid harmonic functions (56,57), but only the regular solid harmonic field terms
27 can be fully compensated using shim sources placed outside the body. Thus,
28 using successively greater numbers of terms of shim terms, even for an
29 orthogonal basis set such as the spherical harmonics, does not provide smooth
30 convergence toward a completely uniform shim.

1 Simulations in Figure 7 show that for the 3.5A/40A current limits chosen, the
2 AC/DC coil provides global shim performance comparable to the 4SH shim basis
3 and slice-optimized shim performance in between 3SH and 4SH. The same
4 simulations show that with higher current limits, the AC/DC performance could in
5 principle be further improved, reaching or surpassing the 5SH basis. With
6 unconstrained current, the AC/DC array performance is comparable to the
7 reference 48-ch MC array, but with the 3.5A/40A constraint, the 48-ch array
8 achieves a further 13-15% reduction in σ^{WB} relative to the AC/DC array. While
9 increasing the current limit above 3.5A could be challenging in practice,
10 performance of the face loops could potentially be boosted by using target-field
11 approaches for designing the wire patterns (58,59) or by increasing the number
12 of wire-turns.

13
14 Other groups have investigated joint optimization of the linear and MC shims
15 in slice-optimized shimming (13). MC shimming for simultaneous multi-slice
16 (SMS) acquisitions has also been explored in simulations (31) and EPI
17 acquisitions at 3T (60). Finally, MC shim arrays have also been used for
18 emerging local field control applications like volume-tailored lipid suppression
19 (22) and zoomed-imaging (20,21).

20
21
22 An alternative MC shimming approach for UHF MRI is to place dedicated
23 multi-turn shim coils just outside the transmit coil shield, as recently
24 demonstrated at 9.4T (13,61). Compared to AC/DC arrays, this approach
25 provides (i) reduced interaction of DC shim components with the transmit coil(s)
26 and (ii) less complexity for the receive-array design, and (iii) greater flexibility to
27 use target field-based shim coil designs optimized for the human head anatomy
28 (56,59,62). However, shim coil placement outside the shield could be
29 geometrically constrained, for example if a head-gradient insert coil is used. Also,
30 the higher inductance may be an obstacle for some emerging local field control
31 applications (63),(64) where waveforms with KHz-range frequencies (1–100 KHz)
32 are played out on MC arrays to provide supplementary spatial encoding. Larger

1 inductances will require higher amplifiers voltages to slew at these frequencies.
2 An additional complication is the potential for coupling to the gradient coils, which
3 increases the amount of amplifier voltage compliance required to reject gradient
4 slew-induced disturbances. Finally, since these coils typically use >20 turns of
5 wire, they experience stronger mechanical torques, which can create problematic
6 noise and vibrations inside the coil if they are not addressed with suitable
7 mechanical constraints. Nevertheless, in existing realizations, many of these
8 concerns have already been addressed (13,61).

10 **CONCLUSION**

12 The feasibility of an integrated 32ch AC/DC B_0 shim array was explored for
13 7T brain imaging. RF performance was found to have comparable SNR, noise
14 correlation, g-factor, and Tx efficiency to a size-matched 32ch reference array.
15 Dynamic, slice-optimized B_0 shimming experiments showed that shim currents
16 can be rapidly switched between TRs without introducing artifacts or
17 compromising temporal SNR. Geometric distortion in 7T EPI slices was
18 significantly reduced, especially in superior slices.

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Captions for Supporting Information Videos

Supporting Information Figure S1. Details of circuit used for each integrated AC/DC coil element. (a) Coil circuit schematic showing the shares wire loop in gray. RF components include tuning capacitors and PIN diode detuning are shown in black. DC components including feed wire (twisted pair), high impedance inductive chokes (L_{SC}), and blocking capacitors (C_B) are shown in red. (b) Photo of one coil element (along with parts of neighboring coils) showing the location of toroidal chokes. (c) Photo of a toroidal choke along with core dimensions. Fifteen turns of 22 AWG wire are wound on a 3D-printed acrylic core, providing a DC inductance of ~ 325 nH and self-resonance frequency in the 310-320 MHz range. Care is taken when adding choke to minimize shifts in the RF coil resonant frequency and to avoid parasitic resonances caused by choke interactions with one another or with other components.

Supporting Information Figure S2. Example of how the RF receive coil resonant frequency (f_0) changes when a toroidal choke with a given self-resonant frequency (f_{srf}) is added across one of the tuning capacitors in the loop. (A) When S_{11} is measured on a network analyzer for a choke at the end of a calibrated cable, the trace passes from the inductive to capacitive region of the Smith chart at f_{srf} . The level of interaction between the choke and the coil can be inferred from the shift in f_0 , as illustrated by the three cases shown. (B) Case 1: When $f_{srf} < f_0$, the choke presents an inductive impedance which reduces the effective capacitance of the parallel LC branch, causing an increase in f_0 . Case 2: When $f_{srf} > f_0$, the choke presents a capacitive impedance which adds in parallel with the tuning capacitor, causing a decrease in f_0 . Case 3: When $f_{srf} \approx f_0$, the

impedance of the choke is close to an open circuit, causing little or no shift in f_0 (the desired condition). (C) Table summarizing experimental measurements of a given choke's f_{sr} , and the resulting loop f_0 , as the choke's windings were slightly adjusted to tune its impedance so as to present inductive, capacitive, and open-circuit impedances to illustrate the three example cases.

Supporting Information Table S1. B_0 shim performance metrics for the three simulated B_0 shim array loop geometries compared in Figure 6. The standard deviation and 3 residual metrics are listed for the example brain field map shown in Figure 6 as well as the average values over the set of all 8 brain field maps used in the simulations.

Supplemental Figure S3. High-resolution gradient echo T_2^* -weighted structural images demonstrating the RF receive sensitivity of the AC/DC array. Parameters: Top row: 400um in-plane, FOV=208×208mm, 512×512 matrix, 1mm slice, TE=21ms, TR=1s, FA=50 deg, 21 slices, NA=1, phase stabilization, TA=4:34. Bottom row: 300um in-plane, FOV=192x192, 640x640 matrix, 1mm slice, TE=21ms, TR=833ms, FA=43°, 10 slices, NA=1, phase stabilization, TA=4:40.

Supporting Information Figure S4. Axial, sagittal, and coronal ΔB_0 field map slices from one representative subject are shown for baseline 2SH, global MC, and dynamic MC shim cases.

Supporting Information Figure S5. Images from Subject #1 showing the same ΔB_0 field maps, EPI slices, and performance metrics as those shown in Figure 9. Yellow arrows show example areas with reduced distortion: orbitofrontal cortex (OF) and cerebellum (CB). The animation in Video S1 provides a clearer depiction of the changes in geometric distortion.

Supporting Information Figure S6. An expanded version of Figure 9 showing ΔB_0 field maps, EPI slices, and performance metrics for Subject #2 (more slices

are shown here as compared to Figure 9). Yellow arrows show example areas with reduced distortion: cerebellum (CB), anterior temporal lobes (AT), and orbitofrontal cortex (OF).

Supporting Information Figure S7. Images from Subject #3 showing the same ΔB_0 field maps, EPI slices, and performance metrics as those shown for Subject #2 in Figure 9. Yellow arrows show an example area of reduced distortion in the anterior temporal lobes (AT). The animation in Video S3 provides a clearer depiction of the changes in geometric distortion.

Supporting Information Figure S8. Images from Subject #4 showing the same ΔB_0 field maps, EPI slices, and performance metrics as those shown for Subject #2 in Figure 9. Yellow arrows show an example area of reduced distortion in the anterior temporal lobes (AT). The animation in Video S4 provides a clearer depiction of the changes in geometric distortion.

Supporting Information Figure S9. Temporal SNR (tSNR) maps computed from EPI time series volumes. The AC/DC coil tSNR is consistent across baseline, global, and dynamic slice-optimal shim acquisitions, indicating that shim updating and settling times do not adversely impact tSNR. The AC/DC and RF-only coils show comparable tSNR. For protocol details, see manuscript text.

Supporting Information Figure S10. ΔB_0 field maps acquired in the three cardinal planes on a silicone oil phantom for all 31 coils in the AC/DC array. The field maps show the variety of independent spatial profiles provided by the different shim channels. Field maps are normalized to 1 ampere of current. Directions are marked for Anterior, Posterior, Superior, Inferior, Right, and Left.

Supporting Information Figure S11. Acquired ΔB_0 field maps and EPI slices showing how slice-optimized MC field offsets can shift water voxels into the fat saturation band, leading to loss of contrast from incomplete T_1 recovery. The

1 shim field offset applied during acquisition of the slice outlined in purple shifted
2 part of the frontal lobe into the saturation band (transition outlined in yellow). The
3 loss of contrast is visible in the orange axial slice acquired during the same TR,
4 as well as the sagittal view showing the effected region of the brain ("naïve
5 trigger"). The problem is corrected by adding triggers to switch off the MC shim
6 fields before the fat saturation pulse. The fields are then switched back on just
7 before the water excitation pulse.

8
9 **Supporting Information Figure S12.** Maximal intensity projections (MIP) for the
10 electric field (E-field) distribution inside the head generated by a 4-turn face shim
11 element (white outline) driven by 1A of 1 KHz sinusoidal current. The maximum
12 E-field in this case reaches a peak amplitude of approximately 0.015 V/m.
13 Scaling these fields linearly to model the rise time ($250\mu\text{s}$) and maximum current
14 swing possible in dynamic shimming experiments (-3A to 3A), the E-field would
15 be 0.45 V/m, which is still safely below the conservative 5 V/m PNS safety
16 threshold.

17
18 **Supporting Information Video S1.** Animation showing changes in geometric
19 distortion of EPI slices acquired on Subject 1 when dynamic slice-optimized B_0
20 shimming is used. The images correspond to those shown in Figure S4. Left
21 column: T_2 -weighted anatomic image with no distortion. Middle column:
22 Animation toggling between A-P and P-A phase-encoded EPI slices with 2SH
23 static shims applied. Right column: Animation toggling between A-P and P-A
24 phase-encoded EPI slices with slice-optimized shims applied.

25
26 **Supporting Information Video S2.** Animation showing changes in geometric
27 distortion of EPI slices acquired on Subject 2 when dynamic slice-optimized B_0
28 shimming is used. The images correspond to those shown in Figure 9. Left
29 column: T_2 -weighted anatomic image with no distortion. Middle column:
30 Animation toggling between A-P and P-A phase-encoded EPI slices with 2SH

static shims applied. Right column: Animation toggling between A-P and P-A phase-encoded EPI slices with slice-optimized shims applied.

Supporting Information Video S3. Animation showing changes in geometric distortion of EPI slices acquired on Subject 3 when dynamic slice-optimized B_0 shimming is used. The images correspond to those shown in Figure S5. Left column: T_2 -weighted anatomic image with no distortion. Middle column: Animation toggling between A-P and P-A phase-encoded EPI slices with 2SH static shims applied. Right column: Animation toggling between A-P and P-A phase-encoded EPI slices with slice-optimized shims applied.

Supporting Information Video S4. Animation showing changes in geometric distortion of EPI slices acquired on Subject 4 when dynamic slice-optimized B_0 shimming is used. The images correspond to those shown in Figure S6. Left column: T_2^* -weighted anatomic image with no distortion. Middle column: Animation toggling between A-P and P-A phase-encoded EPI slices with 2SH static shims applied. Right column: Animation toggling between A-P and P-A phase-encoded EPI slices with slice-optimized shims applied.

Other supporting materials

- Silicone oil B_0 field mapping phantom
- CAD model for AC/DC array helmet housing
- Imaging protocols run on 7T Magnetom scanner
- Simulated multi-coil shim array geometries in PUMCIN format
- 8 *in vivo* brain B_0 field map datasets in Matlab format

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mrm_29022_f1.tiff

mrm_29022_f2.tiff

(b) S_{11} Impedance match (loaded)

(c) S_{21} PIN diode detuning (not loaded)

(d) S_{21} preamp decoupling (not loaded)

mrm_29022_f3.tiff

mrm_29022_f4.tiff

mrm_29022_f5.tiff

Simulations of face loop B₀ shim efficacy

mrm_29022_f6.tiff

mrm_29022_f7.tiff

mrm_29022_f8.tiff

mrm_29022_f9.tiff