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A 32-channel receive array coil for bilateral breast imaging and spectroscopy at 7T

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Abstract

Purpose: This work describes the construction and evaluation of a bilateral 32-channel receive array for breast imaging at 7T.

Methods: The receive array consisted of 32 receive coils, placed on two 3D-printed hemispherical formers. Each side of the receive array consisted of 16 receive loops, each loop having a corresponding detachable board with match/tune capacitors, active detuning circuitry, and a balun. Coil performance was evaluated on homogeneous canola oil phantoms using a Philips Achieva 7T system. Array coil performance was compared with a bilateral forced current excitation volume coil in transmit/receive mode and with a previously reported 16-channel unilateral coil with a similar design.

Results: The 32-channel array had an increase in average SNR throughout both phantoms by a factor of five as compared with the volume coil, with SNR increases up to 10 times along the periphery and three times in the center. Noise measurements showed low interelement noise correlation (average: 5.4%; maximum: 16.8%). Geometry factor maps were acquired for various acceleration factors and showed mean geometry factors <1.2, for combined acceleration factors of up to six.

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SUPPORTING INFORMATION

Additional Supporting Information may be found online in the Supporting Information section.

Conclusions: The improvements achieved demonstrate the clear potential for use in dynamic contrast-enhanced or diffusion-weighted MR studies, while maintaining diagnostically relevant spatial and temporal resolutions.

Keywords

breast imaging at 7T; forced current excitation; receive array

1 | INTRODUCTION

MRI of the breast is commonly used as a supplemental tool to mammography throughout various stages of disease management because of its ability to detect breast cancers with high sensitivity.^{1–6} Specifically, techniques such as DCE MRI and DWI provide qualitative and quantitative information about morphology and kinetics of breast lesions, which can enable the distinction between malignant and benign tumors with improved diagnostic accuracy.^{7–10} Several studies have reported the benefits of high-field (7T) imaging in DCE and DWI studies,^{11–14} where the use of array coils is particularly beneficial to increase the achievable temporal resolution.^{15–18}

The challenges of high-field imaging with respect to homogeneous RF excitation and specific absorption rate are well-documented.^{19,20} Our group has previously reported on the forced current excitation (FCE) technique as a way to transmit effectively at 7T, achieving homogeneous excitation in the presence of the asymmetric load presented by the patient in a prone position, while remaining within specific absorption rate regulations.²¹ The insensitivity of the FCE approach to loading conditions was additionally demonstrated with the insertion of a 16-channel receive breast array designed for unilateral breast imaging.²² The FCE approach is also beneficial for bilateral configurations because of its ability to provide homogeneous excitation throughout the region of interest without the need for multiple transmit channels, its ability to mitigate effects of the mutual coupling between the left and right volume excitation coils, and the potential for a switchable, unilateral single breast excitation implementation.^{23,24}

Here we describe the practical aspects of the design, construction, and evaluation of a bilateral 32-channel receive array for imaging at 7T, together with a bilateral FCE transmit coil modified to accommodate the array. Because of the the lack of commercially available high-channel count ¹H breast coils for 7T, the performance of the array coil was evaluated by comparison with the bilateral FCE volume coil in transmit/receive (T/R) mode, as well as to the performance previously reported on the 16-channel unilateral version of this coil.²² The coil was evaluated on homogeneous canola oil phantoms to assess performance with respect to temporal and spatial resolution.

2 | METHODS

2.1 | Transmit volume coil

A modified version of a previously reported FCE bilateral breast coil was designed and constructed to allow space for the integration of the 32-channel receive array.²³ The coil was

composed of separate quadrature volume coils for each breast, each consisting of a saddle pair and a Helmholtz coil. The left and right quadrature coils were connected to produce B_1^+ fields with the same polarity, consistent with the previously reported version.²⁴ Concentric 3.5-mm-wide coplanar shields surrounded each saddle and Helmholtz loop to improve B_1^+ transmit efficiency at high fields.²⁵ All four saddle elements were connected in parallel at a common voltage point located at the center of the bilateral volume coil along the x -axis via $\lambda/4$ twin-axial transmission lines (RG108; Pasternack Enterprises, Irvine, CA) to ensure equal currents through all elements, irrespective of loading.²¹ The use of twin-axial cable avoided the unbalanced-to-balanced connection associated with using coaxial cable, eliminating the need for the integrated baluns and/or RF shields previously needed, and creating space for the receive coil array and its corresponding hardware and cabling.^{21,22,24} A short 7-cm twin-axial cable was used to connect the common voltage point to a match-and-tune board, adjacent to the left quadrature volume coil, containing variable matching and tuning capacitors (SGNMNC1206E; Sprague-Goodman Electronics, Westbury, NY) and a fixed matching capacitor (100C Series-7.2kV; American Technical Ceramics, Huntington Station, NY) to provide a semibalanced match. The remaining parts of the coil (including the integrated balun, bias tee, and biasing signal) were kept consistent with previous designs that simplified the coil active-detuning network.^{21,22,24} The volume coil dimensions and simplified version of the FCE configuration are shown in Figure 1A.

2.2 | Receive coil array

Size and placement of each element on the bilateral 32-channel receive array was based on the design of the previously reported unilateral 16-channel receive array, which used the “soccer-ball” configuration presented by Wiggins et al.^{22,26} The coil former for each breast was a hemispherical 3D-printed shell with a volume of 900 mL, designed to accommodate approximately 80% of the general female population.²⁷ Extruded cuts on the outer surface of the shell were designed to hold each element and passive detuning trap in place. The layout of one side of the receive array, looking down from the apex, is shown in Figure 1B.

The array elements were printed on flexible (0.51 mm) copper-clad FR-4 and conformed to the 3D-printed shell, as described in By et al.²² From this point forward, component values, part numbers, and functions are indicated in Supporting Information Table S1. Each receive element had six equally spaced breaks: four breaks containing fixed capacitors, one break containing a passive detuning network, and another break containing a male SMA connector connected to a detachable board ($2.5 \times 5.5 \text{ cm}^2$) that contained the match/tune/decoupling network for each element. The passive detuning network was located opposite the detachable board and consisted of two fast-switching crossed diodes, a fixed inductor, and a variable capacitor, as shown in Figure 1C.

The detachable boards for each element allowed the array to be easily inserted into and removed from the close-fitting volume coil and allowed streamlined initial tuning and troubleshooting of each element. The match/tune/decoupling networks are shown in Figure 1C, based on the design presented by Reykowski et al.,²⁸ which lends itself well to coils tuned to high-operating frequencies. Each network consisted of three variable capacitors, a fixed inductor, and a PIN diode used for active detuning. Custom-made can baluns after the

network suppressed common-mode currents. Flexible coaxial cable (7805-010; Belden Inc., St. Louis, MO) was soldered onto each detachable board, with a BNC connector (2-331350-4; TE Connectivity, Schaffhausen, Switzerland) on the opposite end for connection to the remote Philips (Amsterdam, the Netherlands) 32-channel LNA interface box. The cable length required to reach the interface box and achieve the appropriate decoupling requirement was 113 cm. This cable length, including the balun and cabling within the interface box, corresponded to a total length from the coils to the preamplifiers of four half-wavelengths. Cables from the detachable boards were bundled with a cable sleeve and two floating cable traps²⁹ were added on each bundle.

2.3 | Bench measurements

Receive elements were initially matched and tuned individually using an S_{11} measurement with additional coils mounted, but open-circuited at their detachable boards. S_{12} measurements using a double-loop probe were taken to measure the active detuning of each receive element. Active detuning was measured by taking the difference in the S_{12} measurements (at 298 MHz) with the receive element terminated in $50\ \Omega$, and again with the active detuning PIN diode forward-biased by a -5-V , 100-mA supply. Finally, the preamplifier decoupling of each coil was measured in the same manner as the active detuning except, instead of forward-biasing the PIN diode, a mock preamplifier box (representing the input impedance provided by the preamplifiers in the Philips interface box) was used.³⁰ Detachable boards for the remaining elements were then attached, and measurements for each element were repeated after placing the array into the transmit coil, while actively detuning all other elements.

2.4 | Imaging

All phantom data were acquired on a Philips Achieva 7T system. Two homogeneous canola oil phantoms encased in watertight 3D-printed hemispherical shells were used to mimic the lipid properties of the breast.³¹ B_1^+ maps were first generated using a 3D T_1 -weighted fast field echo pulse sequence with the actively detuned array in place. Images to demonstrate the SNR gains achievable with the array were obtained using the standard 3D T_1 -weighted turbo field echo pulse sequence for anatomical imaging with the following parameters: TR: 5 ms, TE: 1.77 ms, FOV: $152 \times 131 \times 352\text{ mm}^3$ [AP $\times$ FH $\times$ RL], resolution: 1-mm^3 isotropic, scan duration: 2.5 minutes. Given the composition of the phantoms used, no fat suppression was applied. Images were acquired using either the volume coil in T/R mode, with the receive array kept in place, or using the 32-channel array to receive and the volume coil in transmit-only mode. Noise-only data sets were also acquired for both configurations using identical imaging parameters, but with the transmit pulse and gradient pulses disabled.

The potential benefit provided by the SNR gains was demonstrated via geometry factor (g-factor) mapping and the acquisition of high spatial and temporal resolution imaging within the bounds of maintaining a clinically relevant scan time (~ 90 seconds) and image quality comparable to the volume coil. The g-factor maps were acquired using the same imaging parameters as detailed above with SENSE (sensitivity encoding) acceleration factors of $1\times$, $2\times$, or $3\times$ in the FH and/or $1\times$, $2\times$, or $3\times$ in the left/right directions.

2.5 | Data analysis

Image processing and analysis were performed using inhouse Matlab (MathWorks, Natick, MA) scripts on images reconstructed from the Philips system. The B_1^+ inhomogeneity was measured (with the detuned array in place) by taking the SD and dividing by the mean of the same region throughout the phantoms on both sides of the bilateral coil.^{15,18} SNR comparisons between the array and the volume coil were performed on an axial slice through the center of the phantoms by taking the average and maximum signal values throughout both phantoms and dividing by the SD of the noise in the noise-only image then scaling by a factor of 0.655.³² SNR maps were generated for each image by dividing the signal intensity by the SD of a selected region in the noise-only image on a pixel-by-pixel basis. Finally, to show the relative SNR improvements of the array, an SNR map comparison was made by dividing the SNR map of the array by that of the volume coil. As with the B_1^+ map, SNR and SNR-comparison maps were masked manually during postprocessing in Matlab, and profiles along a centerline in the phantom were taken for quantification.

To assess coupling between receive elements, a 32×32 noise-correlation matrix was generated by computing correlation coefficients between all combinations of images reconstructed from the noise-only data sets from single-array elements. Average, minimum, and maximum noise correlation was evaluated on receive channels 1 to 16 and 17 to 32 separately because the two sides had negligible coupling with each other. The maximum and average g-factors were calculated within the masked region for all g-factor maps.

3 | RESULTS

Side and front views of the array inside the transmit coil are shown in Figure 2A,B, respectively.

3.1 | Bench measurements

S_{11} measurements showed an average of -24 dB (min: -15 dB; max: -40 dB) of matching and tuning for all coils. S_{12} probe measurements showed an average of 23 dB (min: 17 dB; max: 26 dB) active detuning, and an average of 12 dB (min: 10 dB; max: 15 dB) preamplifier decoupling. The average coil Q_{unloaded} was 71 for the small elements and 47 for the large elements, a ratio comparable to that reported in the original design.²⁶ Given the composition of the phantoms used, changes in Q from loading were negligible. Overall, the measurements indicated adequate matching and tuning, active detuning, and preamp decoupling, as evidenced by the ability to straightforwardly match and tune each element and the volume coil with no indication of element-to-element coupling or coupling between the array and the volume coil.

3.2 | B_1^+ measurements and SNR measurements

The coefficient of variation (SD/mean) showed a B_1^+ variation of 17% throughout both phantoms in the middle axial slice. Supporting Information Figure S1 shows the B_1^+ map and horizontal and vertical profiles through both sides of the bilateral coil, with similar profiles

on both sides of the bilateral coil and more signal being produced on the left (in horizontal profile) and top (in vertical profiles) sides of each phantom, similar to previously reported work.^{15,18,21}

SNR measurements show an average SNR of 16 for the volume coil, and an average SNR of 79 for the array coil. SNR maps for both cases are shown in Figure 3A,B. The SNR map for the volume coil is relatively homogenous throughout both phantoms, whereas the SNR map for the receive array shows a larger increase in SNR along the periphery and drops off toward the center of the phantom, as expected. The comparative SNR map (array SNR map/volume SNR map) shown in Figure 3C indicates the increase in SNR from the array throughout the entire phantom, even at the center—specifically, an increase of approximately three times in the center and 10 times along the periphery, as shown in Figure 3D.

3.3 | Noise-correlation measurements

The noise-correlation matrices shown in Figure 4 depict a mean noise correlation of 5.7% and maximum of 16.5% for coils 1 to 16, and a mean of 5.0% and a maximum of 16.8% for coils 17 to 32. Noise correlation is minimal between coils on the left breast side (receive elements 1 to 16) and coils on the right breast side (receive elements 17 to 32), 3.1%.

3.4 | Geometry factor maps

The g-factor maps were acquired for combined acceleration factors up to nine and demonstrated the feasibility of accelerated imaging without excessively sacrificing SNR (average g-factor <1.2) for acceleration factors up to six.³³ A middle axial slice of the SENSE g-factor maps for multiple acceleration combinations is shown in Figure 5. Among these acceleration schemes, the average g-factor rose above 1.2 only when accelerating by factors of three in the FH direction. Even in these cases, the average g-factor reached a maximum of 1.23 for nine times acceleration. The shortest scan duration within a mean g-factor of 1.2 was 45.1 seconds, which is well within a clinically relevant 90-second scan duration.

4 | DISCUSSION

Overall, the modified FCE volume coil showed B_1^+ field variation (coefficient of variation = 17%), comparable to other previously reported breast coils tailored for transmission at 7T,^{15,18} and the 32-channel coil array showed significant improvements in SNR and receive element isolation.

SNR comparisons between the array and volume coil had a mean improvement of five times throughout both phantoms compared to the volume coil, with improvements up to 10 times along the periphery and three times in the center. These SNR improvements are higher than our previously reported 16-channel unilateral version, which had a mean improvement of 3.3 times throughout the phantom and 2.1 times in the center.²² This difference is likely caused by the loading and shielding effects from the presence of the (detuned) array during volume coil imaging. Unlike this study, in the 16-channel unilateral case the array was removed

when acquiring volume coil data because the volume coil could be matched/tuned within the range of the variable capacitors. Therefore, the absence of loading and shielding from the array attributed to a slightly higher volume coil SNR, or a less favorable SNR improvement relative to the bilateral configuration.

The noise-correlation data indicated minimal coupling between the two sides of the bilateral array. These measurements also were comparable to the previously reported unilateral version, which had mean noise correlation of 6.6%, (min: 3.6%; max: 17.7%).²² However, this does not correlate with the preamplifier decoupling bench data, which showed low decoupling values (average 12 dB) likely caused by the cable losses (0.45 dB) introduced by placing the preamplifier four half-wavelengths away from the preamplifier decoupling network. Instead, the low noise-correlation data were correlated with the unloaded Q values of the receive elements: Lower Q values yielded a lower noise correlation. These lower Q values could be attributed to (1) the number of breaks on the elements, (2) the presence of a passive detuning trap, and (3) losses introduced from the cable to the preamplifier box. However, all of these design criteria were necessary to (1) utilize reasonable segmenting capacitor values (to minimize the effects of parasitic capacitance from patient loading), (2) minimize coupling between the receive elements and transmit coil during transmit, and (3) connect the elements to the preamplifier interface box while allowing space for the patient to rest her head and arms.

5 | CONCLUSIONS

This work described the design, construction, and evaluation of a bilateral 32-channel receive array and FCE volume coil for breast imaging at 7T. The use of the modified FCE design for integration of the receive array and active detuning network, detachable boards for each receive element, preamplifier decoupling network for high operating frequencies, and an addition of a passive-detuning trap simplified the construction and integration of receive and transmit elements. Evaluation of the FCE volume coil and 32-channel coil array showed homogeneous excitation, significant improvements in SNR, and receive element isolation. Additionally, g-factor mapping showed the feasibility of accelerating up to a factor of six times while staying within acceptable g-factor values. Future work could include replacing the receive elements located partially along the z-axis with butterfly surface coils to help receive a more transverse field, phantom studies on a heterogeneous phantom with more clinical significance, and the use of the array in clinical/translational studies.

Supplementary Material

Refer to Web version on PubMed Central for supplementary material.

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FIGURE 1.

Bilateral forced current excitation transmit volume coil, receive coil array, and schematic of single receive element. A, The volume coil is composed of two quadrature volume coils, left and right, each consisting of one saddle pair and a Helmholtz coil. The saddle elements (width, 15.3 cm; length, 8.1 cm) were mounted on a cylindrical 3D-printed former (i.d. 14.7 cm; aperture angle 120°). The Helmholtz elements (i.d. 16.0 cm) were placed 8-cm apart. All saddle elements, as well as all Helmholtz elements, were joined at a common voltage point (CVP) with twin axial cables $\lambda/4$ -distance away from the coil's feed-point (example configuration for the saddle coil shown in blue). A short cable (~7 cm) was then used to connect the match/tune (M/T) to the CVP. For simplicity, neither the CVP and M/T circuitry for the Helmholtz elements, nor the coplanar shielding around each element of the coils is shown. B, One side of the bilateral receive array (16 of 32 channels) is shown. The diameter of the array former is 13.3 cm. The array consists of 10 large (7-cm diameter) and six small (5-cm diameter) elements. C, Circuit schematic of a single receive element showing four segmented capacitors, a passive detuning network (in the green dashed box), and a matching network with active detuning integrated (in blue dashed box). Component names, descriptions, and values are shown in Supporting Information Table S1. C_p , C_t , C_x , C_m , variable capacitors; D_d , a PIN diode; D_p , fast-switching crossed diodes; L_d , a fixed inductor; LNA, low-noise amplifier; L_p , a fixed inductor

FIGURE 2.

32-channel receive array in volume coil. A, The side view of the bilateral coil shows the 32-channel receive array within the volume coil, with the inset showing an example of the detachable match/tune/decouple boards. B, The top view of the coil shows the 3D-printed top cover, cushioned edges, and rounded corners, which were all incorporated in order to maximize patient comfort. Additionally, an acrylic platform was placed over the coaxial cables to prevent the raised arms of the prone patient from coming in contact with them

FIGURE 3.

SNR maps and profiles through the maps using the volume coil in transmit/receive mode and the 32-channel receive array. A, SNR map of the volume coil in transmit/receive mode showed an average SNR of 16 throughout the phantom and an average SNR of 14 in the center. B, SNR map of 32-channel receive array showed an average SNR of 79 throughout the phantom and an average SNR of 42 in the center. C, SNR-comparison map (32-channel receive array SNR map/volume coil SNR map) showed an increase in SNR throughout the entire phantom with larger increases toward the periphery. D, Profiles along the center of the phantom (shown by dashed red/blue lines) show an increase in SNR of about three times at the center and an increase of up to 10 times along the periphery

FIGURE 4.

Noise correlation matrix for 32-channel receive array with the middle diagonal removed. The noise correlation matrix shows a mean noise correlation of 5.7% and maximum noise correlation of 16.5% for coils 1 to 16, and a mean noise correlation of 5.0% and a maximum noise correlation of 16.8% for coils 17 to 32. Additionally, minimal noise correlation (3.1%) is shown between the left and right sides of the bilateral coil

FIGURE 5.

Geometry factor maps for the 32-channel array for axial slices. Acceleration factors of 1×, 2×, and 3× were applied in the foot-to-head direction and left-to-right direction. Average and maximum geometry factors are reported under each image, as well as the total scan duration