Published in final edited form as:

J Magn Reson Imaging. 2012 October ; 36(4): 865–872. doi:10.1002/jmri.23719.

Field shaping arrays: a means to address shading in high field breast MRI

Ileana Hancu, PhD¹, Seung-Kyun Lee, PhD¹, W. Thomas Dixon, PhD¹, Laura Sacolick, PhD², Ricardo Becerra, PhD³, Zhenghui Zhang, PhD³, Graeme McKinnon, PhD³, and Vijayanand Alagappan, PhD⁴

¹GE Global Research Center, Niskayuna, NY, USA

²GE Global Research Center, Munich, Germany

³GE Healthcare, Waukesha, WI, USA

⁴GE Healthcare, Aurora, OH, USA

Abstract

Purpose—To develop a simple correction approach to mitigate shading in 3T breast MRI.

Materials and Methods—A slightly modified breast receive (Rx) array, which we termed field shaping array (FSA), was shown to mitigate breast shading at 3T. In this FSA, one Rx element was selectively unblocked and tuned off the Larmor frequency during the transmit (Tx) phase. The current flowing in this element during Tx created a secondary transmit field; the vector addition of this field and the one created directly by the body coil resulted in a more uniform excitation profile over the entire breast area. The receive Rx element was returned to its intended tuning during the Rx phase, ensuring unperturbed signal reception.

Results—Using the FSA, improved Tx field uniformity better fat suppression, increased image homogeneity and reduced power deposition was seen in all volunteers studied.

Conclusion—A simple modification of a standard breast Rx array, converting it to a field shaping array, was shown to mitigate breast shading in all volunteers studied.

Keywords

breast; MRI; shading; high field; RF coil; coupling

Introduction

3T MRI is becoming more and more common, and is the preferred choice for imaging pathologies of the brain or musculoskeletal system. Although high field offers inherently higher image signal to noise ratio (SNR), it also comes with its own set of challenges. One of such challenges, the inhomogeneous transmit field (B_1^+), can impact the quality of fat suppression and cause image shading. This shading, particularly visible in liver, breast, and non-contrast angiography exams, documented by users of equipment from all manufacturers (1–6), limits the enthusiasm of clinicians for performing such exams at 3T.

Typical image post-processing approaches, used in the past for image shading correction (7), do not work in areas of the field of view (FOV) characterized by almost complete signal

loss. Moreover, they cannot recover lost contrast or SNR due to an improper local flip angle. Going forward, it is hoped that multi-transmit channel MR systems, in conjunction with RF shimming or more generic pulse design strategies, will be able to correct the RF field inhomogeneity at high fields (8–12). The higher specific absorption rate (SAR) associated with the use of multi-transmit channel technology, however, may limit the degree of improvement achievable in the clinic. The large number of single channel MR systems currently in use and the significant lifetime of such systems also limit the rate of transition of the new multi-transmit channel systems to mainstream use. A simpler, easier to implement approach, capable of reducing shading in problem exams in the hundreds of existing 3T MRI scanners, without fundamental hardware changes, is desired. Such an approach would allow a quicker transition of breast, liver or angiography studies to 3T, where the higher image SNR afforded by the higher field strength may translate into better diagnosis of pathological conditions.

One such approach, in which the typical circular polarization is replaced with fixed elliptical polarization, has been proposed to address shading at high field (13). Given the diversity of anatomies studied in full body MRI scanners, and the particularities of the shading patterns in each anatomy, it is expected for such approach to alleviate the shading problem for most anatomies, yet work less than perfectly for each individual anatomy. Local solutions, correcting shading in one particular anatomy of interest, without impacting other anatomies, may also be envisioned. One such example was proposed (14), and showed to improve image homogeneity in abdominal imaging.

In this work, a local solution, capable of addressing breast shading was developed. First, a study was performed to understand the inter-subject variability of breast shading. Consistent with previous literature reports (2,3,5), a bimodal distribution of the excitation field, characterized by a left breast/right breast B_1^+ ratio of around 1.43, was identified in all volunteers. The low variability (<10% (5)) of the ratio of the transmit (Tx) field in the right and left breast measured in multiple volunteers suggests that a single solution may alleviate the problem in all subjects. Although image shading is due to the inhomogeneous excitation created by the body coil, it is one of the elements of the breast receive (Rx) coil (usually existent in the needed location), which was chosen to solve this problem. Normal receive coil operation requires all array elements to be blocked during Tx, to minimize Tx field perturbation. This requirement assumes, however, that the body coil is capable of generating the desired excitation profile. If this requirement is not met (which is often the case in high field imaging), allowing current to flow in selected Rx element(s) during the Tx phase (through inductive coupling of these elements to the body coil), while actively controlling the impedance of the coil at the operating frequency (128MHz in our case), allows a controlled perturbation of the excitation field. Such a perturbation can result a more uniform Tx field.

In a manner somewhat similar to the one described in two previous reports for correcting the excitation field in 7T brain imaging (15,16), a modification of a standard breast Rx array, transforming it to a field shaping array (FSA), was demonstrated here. The modification is based on selective unblocking and tuning of a single right breast Rx element off Larmor frequency during Tx, while returning the Rx element to its intended tuning during the Rx phase. The use of the FSA at 3T was shown to mitigate breast shading, improve fat suppression and reduce average power in all volunteers studied.

Materials and Methods

Phantom experiments

All imaging experiments described in this work were performed using a 3T GE HDx system (Waukesha, WI), equipped with a standard RF body coil. An 8-channel breast coil manufactured by GE (GE part number 2415544) was used for all experiments done in the standard configuration. A torso phantom, with an elliptical cross-section (of diameters of 26 and 41 cm), an empty cylindrical inner tube (20 cm diameter, simulating the lungs), and two 10 cm diameter spheres attached to it (simulating the breasts), was built to understand the origin of the shading artifact. The phantom was filled with Wesson canola oil, or 1g/l CuSO_4 dissolved in de-ionized (DI) water, or 1g/l CuSO_4 and 2.8g/l NaCl dissolved in DI water.

A recently developed technique, based on the Bloch-Siegert shift (17), was used to map the excitation profile *in vitro*. Five slices of B_1^+ maps, spaced 2 cm apart, were acquired with a 5mm through plane and 3mm in plane resolution, respectively.

In vivo experiments

Following a protocol approved by our Institutional Review Board, ten female volunteers (of 1.52 to 1.78m height, and of 54 to 86kg weight) were recruited for this study. The first four volunteers were imaged with the standard breast coil to evaluate cross-cohort variability of B_1^+ maps *in vivo*. The same protocol described above for phantom experiments was used to map the excitation profile in all volunteers. Following the conversion of a standard breast coil to a FSA, the next four volunteers were scanned in the standard configuration, as well as with the FSA. The protocol included a B_1^+ mapping acquisition, an axial VIBRANT acquisition (RF-spoiled gradient echo (SPGR)), with 10 degree flip angle and at 1mm isotropic resolution). This SPGR acquisition is a standard sequence used in the clinic for dynamic contrast enhanced (DCE) imaging. The last two volunteers were scanned using a T_2 -weighted, fast spin echo sequence (TE/TR=104/4000ms), at 5mm through-plane and 1mm in-plane resolution, in both the standard configuration and with the FSA, to assess the improvement in the quality of images acquired with a high flip angle sequence due to the FSA.

Coil design

In a conventional Rx only coil array, the Rx elements are actively blocked during Tx with the help of a parallel resonant tank circuit tuned to the Larmor frequency. This tank circuit introduces a high impedance in series with the coils (few k Ω) and makes the Rx coils transparent to the Tx field. If the parallel resonant tank circuit is not tuned to the exact Larmor frequency, however, the Tx field induces some currents into the Rx array coil during the transmit phase. The local field generated by the current flowing in the modified coil adds constructively or destructively with the Tx field from the body transmit coil, depending on whether the imaginary part of the blocking circuit is capacitive or inductive.

In the 8-channel breast coil, the Rx element encompassing the right breast area, situated approximately parallel to the chest wall, was modified. The approximate location of this element on a top view of the coil is shown in Figure 1a. The simplified schematics of the modified circuit are depicted in Figure 1b. The loop is roughly square shaped, with each side measuring 22cm. For standard operation, the Rx coil is tuned to 127.8 MHz with the help of tuning capacitors (C_t). For simplicity, only the equivalent capacitance C_t is shown in Figure 1b, replacing the highly distributed tuning capacitance characteristic of a high field Rx coil. The coil is matched to 50 ohms in the human loaded condition with the capacitor C_m . This matching capacitor is resonated with an inductor L_m and the input impedance of the low

input impedance preamplifier to enable preamplifier decoupling. The first part of our modifications consisted in incorporating a detuning trap circuit on one of the tuning capacitors C_{detune} , along with a pin diode D_1 (MA4P4002B-402, Macom) in parallel. The detuning circuit was tuned lower than the Larmor frequency (108.6MHz), in order to achieve the desired amount of induced current into the modified coil. At 127.8MHz the impedance of the off-tuned tank circuit, which determines the amount of current induced and the resulting field from it, was $17-j172 \Omega$. This impedance was determined experimentally, by changing the inductance L_{detune} until we obtained the inverse right to left B_1^+ ratio in a silicone oil phantom (GE part number 2417235) to the one seen *in vivo* (i.e. $B_1^+ (\text{right})/B_1^+ (\text{left})^{\text{silicone phantom}} = B_1^+ (\text{left})/B_1^+ (\text{right})^{\text{in vivo}}$). Alternatively, the inductance of the main loop needed to restore field homogeneity could be calculated using the formalism described in (15). Once this inductance is known, the inductance of the detuning circuit could be determined using the theory of inductively coupled circuits described in (18). For simplicity, however, we have chosen to perform this determination experimentally. The second part of our modification consisted in adding a diode D_2 parallel to the matching capacitor C_m , so that the high current flowing in the coil would not destroy the preamplifier.

During the Tx phase both the diodes D_1 and D_2 are forward biased with a help of a 120mA current supplied by the scanner. Upon activation of the detuning circuit an impedance of $17-j172\Omega$ appears in series with the coil. Forward biasing the diode D_2 bypasses the matching capacitor C_m , thereby protecting the preamplifier from the high currents flowing in the coil. During the Rx phase, the diodes are reverse biased with the help of $-7V$ from the scanner. This deactivates the detuning circuit and enables the preamp decoupling as seen in a traditional Rx coils. With this modification the breast coil acts like a normal Rx array coil in the receive phase; one of the elements of the FSA acts like a local Tx coil during the transmit phase, correcting the Tx field inhomogeneity.

A fast acting fuse F (Littelfuse 459 series PICO) was incorporated in series with the Rx coil, to protect the subject in the case of an accidental current spike. The value of the fuse was determined after calculating the induced current flow in the coil in the worst-case condition (the coil resistance $R_C=0$ and load resistance $R_L=0$) both analytically and with full-wave EM simulations. It was verified on the scanner that all sequences intended to be run *in vivo* could be run without the fuse blowing; a high duty cycle sequence (which would not be allowed in a patient by the power monitor) was also shown to blow the fuse, validating the fuse's operating principle.

SAR simulations

To qualitatively replicate the shading, and give an estimate of local SAR, simulations were conducted after the modification of the receive element (as described above), and prior to human scans. The finite element analysis software Comsol Multiphysics 4.2 (Stockholm, Sweden) was used to compute the excitation field (B_1^+) and SAR maps in both the standard scanner configuration, and with the modified receive coil. Figure 2 depicts the geometry used for the simulations, which includes the Tx body coil and a female torso phantom, of $\epsilon_r = 50$ relative electrical permittivity and $\sigma = 0.5S/m$ electrical conductivity. Note that the simulation geometry is only vaguely representative of breast/torso physiology. The shading will likely depend on the size of the torso/breast/lung combination, rather than on solely the size of the breasts; this dependency was, however, not systematically explored in this work. A square loop, with 22 cm long, 5 mm wide copper sides (depicted in black, and highlighted by the arrow in Figure 2) was also added in the simulations for the non-standard configuration at the approximate location seen *in vivo*, and represents the un-blocked Rx element tuned off Larmor frequency. The impedance of the detuning circuit used in the simulations at the excitation frequency was the one measured experimentally at the end of

the coil modification process, after ensuring a ratio of B_1^+ (right) to B_1^+ (left) in the silicone phantom inverse to the ratio seen *in vivo* (see above). The resonant RF body coil model was driven in quadrature with the same drive voltage for the two simulation configurations (with and without the off-tuned Rx coil). The current flowing in the unblocked Rx loop as a function of Tx power was also determined through the simulations, and used to compute the rating of the fuse added in the main circuit of the Rx coil (as described above), in order to ensure patient protection. For a nominal B_1 of $15\mu\text{T}$ in the center of the magnet, a maximum current of 2.5A was assessed to circulate through the loop; a 250mA fuse was therefore installed in the circuit of the modified Rx coil to prevent accidental current spikes. Note that the fuse burn time is a function of current through the fuse; a fuse rated for 250mA needs, in effect, 1s of current at 250mA, or 2ms of current at $\sim 2.5\text{A}$, in order to burn.

Results

The first series of experiments was performed *in vitro*, to understand the B_1^+ distribution in a torso phantom filled with different fluids. The geometry of the experimental setup is shown in Figure 3a. The largest circle depicts the edge of the scanner warm bore (of 60cm diameter); the dotted circle shows the scanner specified homogeneity region, of 40 cm diameter. The phantom is supported by the breast coil, which raises the “sternum” by 6cm, placing it at magnet isocenter. Such positioning is done in breast exams to locate the areas of concern (breasts, axillae and sternum) in the region of best B_0 homogeneity. B_1^+ maps acquired in oil, water and salted water with a FOV of 48cm are presented in Figures 3b, 3c and 3d, respectively. For display purposes, only the voxels with more than 5% of the maximum signal intensity are shown; all three images are scaled such as an average field of 0.073G is obtained in each slice. The uniform excitation profile seen in the oil phantom transforms into a “focused” field pattern in the water phantom, with high B_1^+ signal intensity in the center of the phantom, and low (and equal) signal over the “breasts”. It is only the addition of salt to the torso phantom that creates a right to left excitation field imbalance, similar to the one reported *in vivo* (2,3). The transmit field is the solution of Maxwell’s equations, depicting propagation of high frequency waves in lossy objects; their wavelength in water, of approximately 25cm, is comparable to the size of the objects. Both a high dielectric constant and non-zero conductivity will dictate the shape of the resulting field. While a complete elucidation of the source and parameter dependence of breast shading is beyond the goal of this study, Figure 3 indicates that high dielectric constant tends to result in central brightening, while the introduction of conductivity results in the observation of right/left asymmetry.

The second series of experiments was performed *in vivo* to assess the degree of transmit inhomogeneity in a population of normal volunteers. Figure 4 presents a slice of B_1^+ maps in the initial four volunteers. As can be easily noticed from this figure, there is a high degree of similarity in the transmit maps between the various subjects scanned. The average ratio of left to right B_1^+ field is 1.43, varying between 1.37 and 1.53 for the four volunteers scanned. No correlations between the volunteers’ anatomical measures and the amount of shading were observed.

Figure 5 presents the finite element simulations, which match the data acquired *in vivo* well. Figure 5a shows the simulated B_1^+ map in the female torso phantom, which is very similar to the transmit field maps acquired *in vivo* (Figure 4). Figure 5b presents the SAR maps over the same slice displayed in Figure 5a. By comparison, Figures 5c and 5d present simulated B_1^+ and SAR maps respectively, in the same geometries as the previous simulations, but with the receiver coil unblocked and up-tuned as described in Methods. Note the increased uniformity of the transmit field and the lack of local SAR increase associated with modified coil. Additional simulations indicate that this approach is also relatively insensitive to the

impedance of detuning circuit; a 10% change in the impedance of this circuit at the Larmor frequency resulted in only a 1.5% change in the right to left average B_1^+ ratio.

Improved uniformity of the Tx field also comes with decreased average power deposition. The simulations displayed in Figure 5c indicate an improved uniformity transmit field, with the maximum local SAR being approximately equal in the two cases (standard configuration, and FSA). Experimentally, power calibration aims at obtaining a given average B_1^+ in a slice (which corresponds, e.g., to an average 90 degree flip for the spins in that slice). Such calibration is typically performed for body exams using the body coil for transmit and receive, in order not to unintentionally bias the calibration towards a given part of the anatomy. Correspondingly, in our simulations, the drive voltages were initially adjusted to obtain a given average B_1^+ in a slice (Figure 5a). In the case of the FSA, the body coil was driven with the same voltages; due to the current induced in the FSA, a higher average B_1^+ is noted in Figure 5c as compared with Figure 5a. Since power calibration would require the same average B_1^+ for the standard configuration and the FSA case, a reduction in the drive voltages (or transmit gain, in practice) would be needed to obtain the same average flip angle (or average B_1^+) in the case of the FSA compared to the standard configuration. This fact was confirmed by our experiments. For the 4 volunteers we have studied in both the standard configuration and the FSA case, an average reduction in the power needed to achieve a given flip angle of 0.6 dB was measured.

Figure 6 presents data from two of the four volunteers scanned in the second series of experiments; B_1^+ maps and VIBRANT images are displayed for both volunteers in both configurations. Note the improved homogeneity of the transmit field, the improved fat saturation in the left axilla (where spins were over-flipped in the standard configuration), and the right to left signal homogenization in the VIBRANT images. While these next 4 volunteers had the same left to right signal B_1^+ ratio as the initial volunteers (of 1.43) when scanned in the standard system configuration, this average ratio decreased to 1.08 with the FSA.

Figure 7 presents fast spin echo data acquired from the last two volunteers studied, in both the standard configuration and with the FSA. Note the significantly improved fat suppression when the FSA was used as well as the equalization of the signal in the glandular tissue of the right and left breasts.

Discussion

An investigation was conducted to determine the extent and variability of the Tx field homogeneity at 3T over the breast area. A large similarity in the pattern and distribution of the B_1^+ field was seen among the volunteers studied; all subjects displayed relatively uniform field over each breast, and a ratio of the left to right Tx field of 1.43 (Figure 4). Power calibration aims at obtaining a given average B_1^+ in an axial slice, chosen to maximize signal or contrast; because of the inherently bimodal B_1^+ distribution in the two breasts, some parts of the slice (including the left breast) will exhibit higher, and some parts of the slice (including the right breast) lower than average B_1^+ . Both breasts will end up with lower signal or contrast than ideal, although the intensity of the voxels in the two breasts may appear similar (e.g. for a gradient echo sequence, measured signal is proportional to $\sin(\text{flip angle})$, and $\sin(80^\circ)=\sin(100^\circ)$). The data acquired in our scanner indicate that left breasts typically had a flip angle 1.43 times that in the right. Choosing the transmitter power so the left breast receives more than the desired flip while the right receives less results in one of the breasts receiving an average flip that is off the desired flip by at least a factor of $\sqrt{1.43}=1.2$. The impact of the 20% error in the flip angle on the SNR or contrast to noise ratios of an image depends on the ensuing pulse sequence. Locally

higher errors are also possible, as local flip angles can deviate even more than the average breast flip from the intended flip angle. Marked signal loss, which is corrected by the modified coil, is visible in the VIBRANT image of Figure 6, particularly in the axilla. Improving the uniformity of the transmit field could reduce the errors in quantitative perfusion measurements, and increase the diagnostic value of the measured parameters. Increased signal to noise ratio in diffusion weighted images, e.g., may also aid differentiation between benign and cancerous tissue, improving breast cancer detection specificity.

The inter-volunteer consistency of the right to left Tx field imbalance seen *in vivo* enabled us to develop a mitigation approach, in which one of the eight elements of the Rx coil was unblocked and tuned off Larmor frequency during Tx. This loop coupled inductively to the body coil; the current circulating through this local coil created its own B_1^+ field, which added constructively to the field created by the body coil, and reduced Tx inhomogeneity. Returning the Rx element to its intended frequency ensured un-perturbed reception sensitivity. This approach is expected to entirely address the shading problem for the subjects whose initial left/right B_1^+ ratio was 1.43. It is also expected to significantly reduce shading for volunteers who started with initial left/right B_1^+ ratios different from 1.43. While we studied a somewhat diverse range of volunteers, we have clearly not covered the entire range to be seen in the general population. Should the range of B_1^+ imbalance remain between 1.3 and 1.6 in the general population, right/left B_1^+ ratios of between 0.93 and 1.12 will be expected while using this modified coil. Even if the range of shading will increase in a larger study of a diverse population, the use of the FSA should offer significant improvement in the shading artifact, with little or no drawback.

The particular implementation of creating additional B_1^+ field through inductive coupling of one Rx element to the body coil presented here in detail was, in fact, a 2nd generation solution tested by us. In the first approach, a simple, passive loop, tuned higher than 128MHz was employed over the right breast to couple to the body coil in Tx, and “add the missing field”, in a manner similar to the one previously described for 3T abdominal imaging (14). While this first approach worked as well as the one presented here in detail for correcting the Tx field, it slightly degraded the reception profile of the four right breast coils, which shifted their resonance frequency as a consequence of coupling to the passive coil. Selectively unblocking and tuning off Larmor frequency of one of the Rx array elements (already existent in the right location), while returning that element to its intended frequency during Rx, provided a means to improve the uniformity of the Tx field and maintain the intended Rx profile of all Rx elements. More than one means for achieving this goal exists; a simpler one may be to remove sufficient capacitance from the main circuit, driving the resonance frequency at the desired location. The widely distributed capacitance of typical high field product coils prevented the use of this solution in our case.

Note that this modification of the Rx coil should significantly improve the Tx field in all patients imaged with a particular type of RF body coil, in any 3T scanner. Opposite polarity of the B_0 field (possible in clinical settings housing more than one magnet in close proximity) will require the modification of the Rx element on the opposite side of the coil. Last, but not least, for a centered symmetric phantom, and a quadrature driven coil, the B_1^+ field will have a point symmetry. Since it is typical for the “sternum” to be placed at magnet isocenter in breast exams, even the point symmetry is destroyed. In such case the degree of shading can depend on the mode spacing and loading sensitivity of the coil. Consequently, different kinds of RF body coils (coils with different lengths, high pass vs. low pass birdcage vs. TEM coils, accommodating the patient in the center or slightly off-center, etc) will produce different degrees of shading. Note, for example, that previous reports (presumably using different kinds of body RF coils) mentioned a left to right Tx field ratio of 2 (3) or

1.67 (2), while we measure 1.43 in our *in vivo* experiments. A very simple change of the inductance of the blocking circuit (Figure 1b) will control the amount of current induced in the main loop, therefore the amount of shading correction performed by the un-blocked coil.

In conclusion, we have demonstrated a very simple approach for improving the uniformity of the 3T excitation field over the breast. The results obtained appear comparable to the ones obtained with a parallel transmit scanner (19,20), requiring only minor modifications of already existing hardware. Improved image homogeneity and reduced power deposition were shown with this approach.

Acknowledgments

Grant support: 5R01EB005307, 1R01CA154433

References

1. Andrews, T.; Ghostine, J.; Gonyea, J.; Ebert, G.; Braff, S.; Filippi, C. Reduction in Dielectric Shading in Liver on Clinical 3T Parallel Transmission MR System. Proceedings of the 18-th Annual Meeting of ISMRM; Stockholm, Sweden. 2010; p. (abstract 2590)
2. Azlan CA, Di Giovanni P, Ahearn TS, Semple SI, Gilbert FJ, Redpath TW. B1 transmission-field inhomogeneity and enhancement ratio errors in dynamic contrast-enhanced MRI (DCE-MRI) of the breast at 3T. J Magn Reson Imaging. 2010; 31(1):234–239. [PubMed: 20027594]
3. Kuhl CK, Kooijman H, Gieseke J, Schild HH. Effect of B1 inhomogeneity on breast MR imaging at 3.0 T. Radiology. 2007; 244(3):929–930. [PubMed: 17709843]
4. Miyazaki, M.; Yamashita, Y.; Wheaton, A.; Dannels, W.; Anderson, R.; Zha, L.; Sugiura, S. Non-contrast MRA at 3T. Proceedings of the 19th Annual Meeting of ISMRM; Montreal. 2011; p. (abstract 3349)
5. Sung, K.; Daniel, B.; Hargreaves, B. Transmit B1 Field Inhomogeneity and T1 Estimation Errors in Breast DCE MRI at 3T. Proceedings of the 19-th Annual Meeting of ISMRM; Montreal, Canada. 2011; p. (abstract 3086)
6. Zheng, B.; Cheong, D.; Au, C.; Ng, E.; Lee, S.; Ng, T. Quantitative DCE-MRI in breast with direct measurement of AIF using Tofts and ATH models: a simulation study. Proceedings of the 19-th Annual Meeting of ISMRM; Montreal, Canada. 2011; p. (abstract 1042)
7. Behrenbruch CP, Petroudi S, Bond S, Declerck JD, Leong FJ, Brady JM. Image filtering techniques for medical image post-processing: an overview. Br J Radiol. 2004; 77(Spec No 2):S126–132. [PubMed: 15677354]
8. Grissom WA, Xu D, Kerr AB, Fessler JA, Noll DC. Fast large-tip-angle multidimensional and parallel RF pulse design in MRI. IEEE Trans Med Imaging. 2009; 28(10):1548–1559. [PubMed: 19447704]
9. Katscher U, Bornert P. Parallel RF transmission in MRI. NMR Biomed. 2006; 19(3):393–400. [PubMed: 16705630]
10. Kukuk GM, Gieseke J, Weber S, Hadizadeh DR, Nelles M, Traber F, Schild HH, Willinek WA. Focal liver lesions at 3.0 T: lesion detectability and image quality with T2-weighted imaging by using conventional and dual-source parallel radiofrequency transmission. Radiology. 259(2):421–428. [PubMed: 21330565]
11. Nelles M, Konig RS, Gieseke J, Guerand-van Battum MM, Kukuk GM, Schild HH, Willinek WA. Dual-source parallel RF transmission for clinical MR imaging of the spine at 3.0 T: intraindividual comparison with conventional single-source transmission. Radiology. 257(3):743–753. [PubMed: 20858848]
12. Willinek WA, Gieseke J, Kukuk GM, Nelles M, Konig R, Morakkabati-Spitz N, Traber F, Thomas D, Kuhl CK, Schild HH. Dual-source parallel radiofrequency excitation body MR imaging compared with standard MR imaging at 3.0 T: initial clinical experience. Radiology. 256(3):966–975. [PubMed: 20720078]

13. McKinnon, G.; Weyers, D. General Electric, assignee. System and method of elliptically driving an MRI coil. US patent 7135864. 2009.
14. Schmitt, M.; Feiweier, T.; Voellmecke, E.; Lazar, R.; Krueger, G.; Reykowski, A. B1 homogenization in abdominal imaging at 3T by means of coupling coils. Proceedings of the 13th Annual Meeting of ISMRM; Miami. 2005; p. (abstract 331)
15. Wang S, Murphy-Boesch J, Merkle H, Koretsky AP, Duyn JH. B1 homogenization in MRI by multilayer coupled coils. IEEE Trans Med Imaging. 2009; 28(4):551–554. [PubMed: 19336276]
16. Wiggins, G.; Zakszewski, E.; Wiggins, C.; Wald, L. B1 transmit field manipulation using controlled decoupling of array coil elements. Proceedings of the 15th Annual Meeting of ISMRM; Berlin. 2007; p. (abstract 1054)
17. Sacolick LI, Wiesinger F, Hancu I, Vogel MW. B1 mapping by Bloch-Siegert shift. Magn Reson Med. 2010; 63(5):1315–1322. [PubMed: 20432302]
18. Terman, F. Circuit theory Radio engineers' handbook. New York and London: McGraw Hill; 1943. p. 135-251.
19. Katscher, U. Basic and tailored RF shimming in a multi-transmit whole body MR system. Prog Electromagn Res Symposium; Cambridge, USA. 2008. p. 701-704.
20. Kuhl, C.; Kukuk, G.; Mekes-Rijckaert, Y.; Schild, H. Parallel RF transmission for breast MRI at 3.0 Tesla: preliminary results. Proceedings of the 17th Annual Meeting of ISMRM; Hawai'i. 2009. p. (abstract 4133)

Figure 1.

a) Top view of the breast coil, with the approximate location of the modified element depicted in gray and highlighted by the 4 arrows **b)** Circuit of the receive (Rx) element situated at the entrance of the right breast. The diode D_1 is used to detune the Rx element during the transmit (Tx) phase; the diode D_2 protects the preamplifier during Tx, and the fuse F protects the patient in the case of an accidental current spike.

Figure 2.

Geometry of the SAR simulations, including the Tx body coil, a female torso phantom (of $\epsilon_r = 50$ relative electrical permittivity and $\sigma = 0.5 S/m$ electrical conductivity), and a 22cm side square loop (depicted in black, and highlighted by the arrow).

Figure 3.

a) Geometry of experimental setup. The largest circle depicts the edge of the 60cm scanner warm bore; the dotted circle shows scanner specified homogeneity region (of 40cm diameter). The phantom is supported by the breast coil, which raises the “sternum” by 6 cm, placing it at magnet isocenter. Sample B₁⁺ maps (displayed in Gauss) of the torso phantom filled with **b)** Wesson canola oil **c)** 1g/l CuSO₄ dissolved in de-ionized (DI) water and **d)** 1g/l CuSO₄ and 2.8g/l NaCl dissolved in DI water.

Figure 4.

Typical B_1^+ maps (displayed in Gauss) acquired with a Bloch-Siegert sequence from 4 volunteers. The standard system configuration was used for these acquisitions, and the center slice is displayed from each volunteer.

Figure 5.

Comsol simulations presenting **a)** B_1^+ maps and **b)** SAR maps in the standard scanner configuration (transmitting with the body coil, receiving with the clinical 8 channel GE breast coil). **c)** B_1^+ maps and **d)** SAR maps obtained while receiving with the FSA. B_1^+ maps are displayed in Gauss and SAR maps in W/kg, respectively.

Figure 6.

Data from two volunteers: B_1^+ maps (displayed in Gauss) in the standard system configuration (Figures 6a and 6c) and with the FSA (Figures 6b and 6d). VIBRANT images in the standard configuration (Figures 6e and 6g) and with the FSA (Figures 6f and 6h). White arrows highlight the improved fat suppression in the left axilla.

Figure 7. Fast spin echo images acquired from two volunteers (different rows), acquired in the standard system configuration (Figures 7a and 7c) and with the FSA (Figures 7b and 7d).