Published in final edited form as:

Phys Med Biol. 2011 November 7; 56(21): 6809–6822. doi:10.1088/0031-9155/56/21/004.

Feasibility study of a unilateral RF array coil for MR-scintimammography

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Abstract

Despite its high sensitivity, the variable specificity of magnetic resonance imaging (MRI) in breast cancer diagnosis can lead to unnecessary biopsies and over-treatment. Scintimammography (SMM) could potentially supplement MRI for improving the diagnostic specificity. The synergistic combination of MRI and SMM (MRSMM) could result in both high sensitivity from MRI and high specificity from SMM. Development of such a dual-modality system requires the integration of a radio frequency (RF) coil and radiation detector in a strong magnetic field without significant mutual interference. In this study, we developed and tested a unilateral breast array coil specialized for MRSMM imaging. The electromagnetic (EM) field, specific absorption ratio (SAR), and RF coil parameters with cadmium-zinc-telluride (CZT) detectors encapsulated in specialized RF and gamma-ray shielding mounted within the RF coil were investigated through simulation and experimental measurements. Simultaneous MR and SMM images of a breast phantom were also acquired using the integrated MRSMM system. This work, we feel, represents an important step toward the fabrication for a working MRSMM system.

1. Introduction

Although mammography is very sensitive in detecting early breast cancer, it does not work well in women who have radiographically dense breasts, breast implants, or scar tissue. Alternatively, MRI has been proven to be the most sensitive imaging modality in delineating tumor extent and detecting multifocal or multicentric diseases in the breast (Lehman *et al* 2007). Many studies indicate that preoperative MRI is useful in local staging for surgical planning, especially for patients with lobular breast cancer (Boetes *et al* 1995, Davis *et al* 1996, Kriege *et al* 2004, Trecate *et al* 2006). However, the variable specificity of MRI can lead to unnecessary biopsies and over-treatment. Scintimammography (SMM) can potentially supplement MRI by improving the diagnostic specificity in breast cancer imaging.

The combination of MRI and SMM has great clinical potential for improving specificity through SMM while maintaining the high sensitivity offered by MRI. However, interference between the MRI and SMM components presents a significant challenge to the design of a combined system. This interference can lead to major artifacts and image degradation in both modalities. The primary concerns are electromagnetic interference and the effect of the

static field (B_0 field) of the MRI scanner on the SMM detectors. Traditional gamma-ray detectors based on scintillators coupled to photomultiplier tubes do not function within high magnetic fields. Instead, MR-compatible cadmium-zinc-telluride (CZT) semiconductor-based radiation detectors can be utilized (Azman *et al* 2007, Hamamura *et al* 2010a, Wagenaar *et al* 2006). Such detectors must be enclosed in RF shielding to prevent electromagnetic interference between the MRI and SMM systems. Furthermore, sufficient gamma-ray shielding is required to prevent the detection of uncollimated radiation. Both of these shielding layers have the potential to adversely affect MR imaging, particularly when they are placed within the RF coil.

In this study, we evaluated the feasibility of using a specialized RF coil and CZT radiation detectors for combined MR and SMM imaging (henceforth called MRSMM). First, a unilateral breast array coil was designed and constructed. A full-wave electromagnetic field simulation was performed to determine the change in the electromagnetic (EM) field and specific absorption ratio (SAR) distribution when RF shielding blocks for the CZT detectors were loaded into the RF coil. Experimental measurements of various RF coil and MR image parameters were then acquired when combinations of RF and gamma-ray shielding for the CZT detectors were loaded into the RF coil. Finally, operation of the integrated MRSMM system was demonstrated through the simultaneous MR and SMM imaging of a breast phantom.

2. Materials and Methods

2.1 RF coil and CZT detector configuration

The basic geometry of the proposed RF array coil consists of two parallel circular loops each 15 cm in diameter and separated by 6.5 cm. Each loop was etched on FR4 laminate board using copper strips of 0.0341 mm thickness and mounted on an acrylic plate with the material within the loops removed to allow for insertion of the breast. The separation between the plates allows for insertion of CZT detector modules through the side of the RF coil, as shown in figure 1. Each coil loop also contained fully integrated components that were tuned to 127 MHz (the larmor frequency at 3 T) and matched to an impedance of 50. Mutual coupling between the coils could not be eliminated by geometrical overlapping or connecting to low noise amplifiers (LNA) alone. Thus, to remove mutual coupling, we utilized both the inductive decoupling method, in which air inductors connected to each coil loop are crossed over, and integration of LNAs with low input impedance (below 2 Ω) (Nabeshima *et al* 1996). Since the proposed RF array coil operates only as a receiver, the combination detuning method using active and passive detuning circuits added to each loop was utilized to decouple the array coil during transmission of high-power RF energy.

For SMM imaging, each CZT-based radiation detector (Gamma Medica, Inc., Northridge, USA), consisting of $50.8 \times 50.8 \times 5$ mm of CZT crystal coupled to 1024 (32×32) detector elements, was mounted in a $17 \times 6.5 \times 6$ mm box. The walls of this box were made of a lead composite for gamma-ray radiation shielding. The MR compatibility of this lead composite when placed outside of an RF coil was demonstrated by Ha *et al* (2010). For RF shielding, a layer of copper mesh was wrapped around each detector box. This mesh was segmented not to connect electrically on the edges to suppress the flow of eddy currents caused by gradient switching in the MRI, thus preventing the formation of significant MR image artifacts such as image distortion (Alecci *et al* 2002). For testing the RF shielding without the presence of the gamma-ray shielding, a foam box the same size as the detector box was also wrapped with copper mesh.

For breast imaging, two opposing CZT detector modules with parallel-hole collimators were integrated with breast compression paddles and positioned between the acrylic plates of the

RF coil, as illustrated in figure 1a. The control knob shown in figure 1c allows for accurate adjustment of the compression under observation by the patient. Although breast compression is not required for MR imaging, it is of benefit to SMM since this modality's spatial resolution decreases as the distance from the imaged object to the collimator increases. With opposing detectors, the farthest location within the breast to either collimator is at the centerline. Compression can reduce this distance, thus improving the spatial resolution.

The proposed RF coil is positioned within the MRI bore such that it generates the radio frequency field ($\mathbf{B}_1$ field) parallel to the (vertical) y-axis. The MRSMM assembly including the RF coil can be adjusted along the y-direction so that the volume of interest within the breast can be centered within the RF coil and positioned within the field-of-view (FOV) of the CZT detectors. Furthermore, the assembly can rotate about the y-axis around the breast as illustrated in figure 2. This feature is important in SMM imaging for distinguishing multiple lesions. Thus, in the case of two lesions, one orientation (0°) provides maximum separation in the resulting images while others (90°) make it difficult to visually distinguish the lesions. An immediate benefit of combined MRSMM imaging is that the MR images can be used to determine the optimum orientation of the CZT detectors.

2.2 Effects of RF and gamma-ray shielding

The proposed RF coil for MRSMM imaging should provide high signal-to-noise (SNR) and field homogeneity in the presence of the RF shielding-encapsulating CZT detectors. However, these modules, particularly the RF shielding, may distort the EM field generated by the RF coil. Investigation of this distortion was first reported by Ha *et al* (2011). In that study, the system configuration allowed for the RF shielding to be placed up to 6 cm away from the RF coil. Therefore, the $\mathbf{B}_1$ field and parameters such as the quality (Q) factor, resonance frequency, and impedance of the RF coil were not significantly affected. In the current study, the RF shielding-encapsulating CZT detectors were positioned directly between the end plates of the RF coil. Due to the close proximity of these components, the effect of the RF shielding on the RF coil was investigated further.

To investigate this issue, the RF coil and RF shielding were modeled using a simulation program (SEMCAD X Ver. 14.2.1, Schmid & Partner Engineering AG, Zürich, Switzerland) to assess the EM field change, resonance frequency variation, and SAR distribution when the detector boxes were positioned through the side opening of the RF coil. SEMCAD X currently offers advanced computational solutions by employing the Finite-Difference Time-Domain method (FDTD) proposed by Yee (1966), which is a direct solution of Maxwell's curl equations in the time domain. While the actual RF shielding consists of copper mesh covering the detector boxes, it was modeled as pure copper conductive blocks with electric conductivity $\sigma = 5.96 \times 10^7$ S/m in the simulation. The SAR map was calculated as:

$$\text{SAR} = \frac{\sigma}{2\rho} |\mathbf{E}|^2, \quad (1)$$

where ρ is the mass density of the object and $\mathbf{E}$ the induced electric field vector calculated using the finite difference time domain method. To simulate a compressed breast, a $11.5 \times 11.6 \times 4.5$ cm brick-shaped dielectric phantom, modeling breast glandular tissue, was simulated with the following parameters: relative permittivity (ϵ_r) = 66.841, electric conductivity (σ) = 0.80383 S/m, and loss tangent ($\tan \delta$) = 1.702. These dielectric properties are estimated at 127MHz (Gabriel S *et al* 1996).

Using the actual RF array coil, we evaluated the changes in the Q-factor, resonance frequency, and impedance when the CZT detectors were positioned within the system.

These parameters were measured with the CZT detector boxes placed at various distances from the isocenter of the RF coil. The boxes were tested without RF and gamma-ray shielding, with RF shielding only, with gamma-ray shielding only, and with both RF and gamma-ray shielding. All parameters were measured without loading the RF coil.

A brick-shaped phantom with dimensions identical with the one used in the simulation ($11.5 \times 11.6 \times 4.5$ cm) was constructed for MRI tests. It was filled with solution of 4.0g/L $\text{CuSO}_4 \cdot 5\text{H}_2\text{O}$, 10g/L NaCl, and distilled water. The phantom was positioned at the isocenter of the RF coil and the detector boxes were positioned 10 cm away from this phantom. This setup was placed within a 3 T Philips Achieva system (Philips Medical Systems, Netherlands). Coronal MR images of the middle of the phantom were acquired using a gradient-echo sequence with the parameters: FOV = 120×120 mm, matrix size = 256×256 , echo time (TE) = 5 ms, repetition time (TR) = 30 ms, flip angle = 60° , slice thickness = 5mm, and number of signal averages (NSA) = 4. MRI data were acquired for detector boxes without RF and gamma-ray shielding, with RF shielding only, with gamma-ray shielding only, and with both RF and gamma-ray shielding. For each configuration, MR images were acquired without and with 2nd and 3rd order shimming to assess correction of any B_0 field distortions caused by the shielding.

From the acquired MRI, we calculated the SNR as [(signal average-noise average)/(noise standard deviation)]. To measure the phantom signal, a rectangular region-of-interest (ROI) was drawn over the central 80% of the phantom image. To calculate the noise, the standard deviation was measured from the background.

2.3 MRSMM phantom experiment

To demonstrate operation of the integrated MRSMM system, simultaneous MR and SMM imaging of a breast phantom was performed. A hollow, oval-shaped acrylic cylinder with a length of 80 mm, width of 68 mm, and height of 120 mm was used to simulate a compressed breast. Two vials each 10 mm in diameter and separated by 15 mm were used to simulate breast lesions. The phantom was placed at the center of the RF coil within the 3 T MRI system. The two vials simulating lesions were placed at the center of the phantom. Three different orientations for the vials were tested (0° , 45° , 90°), simulating three different angles of breast compression (figures 2).

Three different data sets were acquired, each testing the three different orientations for the vials. For all data sets, the vials were each filled with a solution of 4 mM CuSO_4 and 60 $\mu\text{Ci/mL}$ of $^{99\text{m}}\text{Tc}$ sestamibi (Cardiolite; Cardinal Health, Dublin, USA), a radiopharmaceutical used in nuclear medicine imaging. For the first data set, the hollow breast phantom “background” was left empty. In the second data set, the background was filled with a 10 mM CuSO_4 solution. In the third data set, the background was filled with a solution containing both 10 mM CuSO_4 and 1 $\mu\text{Ci/mL}$ of $^{99\text{m}}\text{Tc}$ sestamibi. For each vial orientation, a coronal MR image of the middle of the phantom was acquired using a 2D fast-spin-echo sequence with the following parameters: TR = 1.5 s, TE = 80 ms, FOV = 150×150 mm, matrix = 512×512 , slice thickness = 5 mm, and NSA = 4. Concurrent to the MRI acquisition, nuclear radiation counts were recorded from the CZT detectors for 10 minutes. The spectrum for each detector element was then windowed about the 140 keV photopeak ($\pm 5\%$). The resulting SMM images were adjusted for radioactive decay by scaling to the initial activity of the first image and corrected for inherent nonuniform detector sensitivity using the image of a flood-field phantom.

For the 0° orientation, a ROI was drawn about one of the vials in each of the SMM images. The number of radiation counts within the ROI was then computed. From the MR image, the distance from each voxel within the vials to the detector through the background was

measured. For each voxel in the vials, the attenuation of the radiation counts was then calculated as $e^{-\mu d}$, where μ is the attenuation coefficient of water at 140-keV and d is the MRI-measured distance. This information was then used to correct for attenuation by the water background in the second and third data sets. In the SMM images of the third data set, a second ROI containing only the background along with geometrical measurements from the MR images were used to determine the background activity concentration. This information was used to remove the background contribution to the counts measured within the first ROI prior to attenuation correction. After this background correction, attenuation correction was then performed to yield the corrected activity of only the vial.

3. Results

In our simulation studies, the RF array coil tuned exactly to the target frequency (127 MHz) without the presence of the RF-shielded blocks generated a homogeneous $\mathbf{B}_1$ field in the region between the coil loops as shown in figures 3a and 4a. With the addition of the conducting blocks, the resonance frequency of RF coil shifted to 138.2 MHz. Nevertheless, at this shifted frequency, the blocks did not produce any $\mathbf{B}_1$ field distortion within the phantom region (figures 3c and 4d). In contrast, the root mean square (RMS) $\mathbf{E}_1$ electric field sharply increased near the surface of the conducting blocks as shown in figures 3d and 4e. After re-tuning the RF coil with the conductive blocks back to the target frequency, the $\mathbf{B}_1$ field within the phantom remained uniform while its intensity slightly increased compared to the field generated by the RF coil without the conducting blocks. The RMS $\mathbf{E}_1$ electric field intensity on the surface of the phantom remained higher than the other areas due to electric charges concentrated on the surface of conductive copper blocks. As a result, a high SAR was localized on the surface of the phantom (figure 4f).

The Q-factors and impedances for each of the two circular loops of the RF array coil were first experimentally measured without the presence of any shielding. The Q-factor and impedance of the top loop was 118.5 and 251.6 Ω , while the measurements for the bottom loop were 116.8 and 247.0 Ω , respectively. When the RF-shielded boxes (without gamma-ray shielding) were placed 10 mm away from the phantom (26 mm from the isocenter of the RF coil), the Q-factor and impedance of both coil loops decreased 27.1% and 46.3%, respectively. The resonance frequency also shifted up about 6.5 MHz from the target frequency, which was a smaller change than the one observed in the simulation. The reason is because the sensitivity of the actual RF coil is reduced by the resistance of lumped elements such as capacitors and inductor which were not modeled in the simulation. In addition, mutual coupling between neighbor coils was not considered in the simulation study. Unlike the simulation results, the resonance frequency spectrum split into two peaks when the RF-shielded boxes were positioned closer than 10 mm from the phantom, as shown in figure 5a. The split of the resonance frequency lead to $\mathbf{B}_1$ inhomogeneity within the ROI of RF coil as well as low SNR. It also leads to the use of unnecessary RF transmit power because of the off-target frequency. However, when the RF coil was re-tuned to the target frequency, the frequency spectrum returned to a single peak, while the reduced Q-factor remained. When the gamma-ray shielded boxes were moved into the space between the RF coil loops, the frequency remained constant, but the Q-factor dropped. This finding is consistent with the previous study by Ha *et al* (2010). When both the gamma-ray and RF shielding were positioned within the RF coil, the resonance frequencies shifted up about 7 MHz while the Q-dropped even lower. When the RF coil was re-tuned to the target frequency, the Q-factor remained low, but the splitting in the frequency spectrum was again resolved. A summary of these measurements is given in table 1.

MR images of the brick phantom are shown in figure 6. Coronal images of the middle of phantom were selected. Image intensity non-uniformity appeared on the top and bottom

portion of MR images due to the nonlinear response of the surface coils. This non-uniformity is a typical sensitivity problem that arises with the use of surface coils (Axel *et al* 1987). A similar magnitude image for all shielding configurations indicates that the $\mathbf{B}_1$ field within the phantom region was not significantly affected by the presence of the RF or gamma-ray shielding, as predicted by the simulation studies. With linear (1st) order shimming, the phase images were also similar, except for the case with only the gamma-ray shielding. However, higher (2nd and 3rd) order shimming resulted in similar and more uniform phase images for all configurations. The SNRs calculated from the magnitude images acquired with high order shimming are listed in table 1. These values parallel the changes in the Q-factor for the different shielding configurations.

MR and SMM images of the compressed breast phantom are shown in figure 7. The number of radioactive counts measured within an ROI encompassing one of the vials is listed in table 2. As anticipated, the two vials are distinguishable in the SMM images for the 0° and 45° orientations, but indistinguishable for the 90° orientation, thus demonstrating the importance of using the proper compression angle. The image intensity and measured activity of the vials decrease as water is added to the background due to attenuation of the gamma-rays. For the 45° orientation with water background, the two vials have slightly different intensities due to the different attenuating distances through the water from each vial to the detector. The measured radioactivity within the ROI increases when radioactivity is added to the background since both the vial and the background contribute to the counts measured within the ROI. The variance in the activity across the background ROI was less than 5% of the average value, indicating a homogeneous distribution of the radiotracer. Attenuation correction of the measured counts from the water background configuration was able to restore the original (no background) counts to within 2%. Subtraction of the background activity within the ROI followed by attenuation correction was able to restore the counts from the active background configuration to within 1%.

4. Discussion

The results of our simulation and experimental measurements demonstrated the effects of placing CZT detectors with RF and gamma-ray shielding into an RF array coil for MRSMM imaging. With RF shielding, the operating MRI system did not adversely affect the measured radiation counts within the $\pm 5\%$ energy window. This finding is consistent with previous combined imaging studies (Hamamura *et al* 2010a, 2010b). The RF shielding did introduce a shift in the resonance frequency and an increase in the $\mathbf{E}_1$ electric field and SAR at the periphery of the test phantom. Although the RF shielding with segmented mesh produced no significant eddy currents in this study, further measurement when using fast gradient switching sequences such as echo planar imaging (EPI) should be investigated. The gamma-ray shielding introduced inhomogeneity in the $\mathbf{B}_0$ field. Both shielding layers did not distort the distribution of the $\mathbf{B}_1$ field in the region between the coil loops, but did result in a decrease in the Q-factor of the RF coil and the SNR of the MR images.

To avoid these adverse effects, MR and SMM imaging could be performed sequentially, where the RF coil is operated without the presence of the (shielded) CZT detectors. However, if simultaneous imaging is performed, various techniques can be utilized to reduce these adverse effects. To avoid manually re-tuning the RF coil whenever the CZT detectors are moved within the RF coil, an auto-tuning method using varicap diodes can be utilized (Hwang *et al* 1998 and Muftuler *et al* 2002). To increase the SNR of MR images despite geometric constraints, RF coil elements can be added on the side openings of the current frame. The added coils generate a $\mathbf{B}_1$ field along the x-axis and have quadrature polarization with the existing coil configuration without mutual coupling. To reduce SAR deposition, lower flip angles, shorter echo train lengths, increased inter-echo spacing, and/or longer TR

values can be utilized. Similarly, rapid acquisition with relaxation enhancement (RARE) and variable-rate selective excitation (VERSE) methods have been developed to improve MR image quality without exceeding SAR limits (Conolly *et al* 1969 and Hennig *et al* 1986). Our RF array coil may also be adapted for use with parallel imaging techniques, which can decrease RF exposure by reducing the number of phase-encoding steps (Pruessmann *et al* 1999). As previously demonstrated, distortion of the B_0 field by the gamma-ray shielded can be corrected through high order shimming. Finally, the drop in image SNR can be compensated by additional signal averaging at the expense of imaging time.

Despite a reduced SNR, our breast phantom experiment demonstrated that accurate MR images can still be obtained with simultaneous MRSMM imaging. Furthermore, the MR images were used to measure the geometry of the phantom to facilitate attenuation and background correction of the radioactive counts within the ROI of the SMM images. In more complex objects, such as the human body, segmentation of the MR images into different tissue types with their corresponding attenuation coefficients and uptake values could be utilized for such corrections. This could enable accurate quantification of radiotracer uptake by suspicious lesions, which may be of clinical value.

In addition to attenuation and background corrections, combined MR and SMM imaging has several other advantages. Co-registered anatomical MR images can provide a reference for the SMM images, allowing for improved localization. The higher spatial resolution MR images can be utilized to improve the spatial resolution of gamma-ray images through post-processing techniques such as the maximization of mutual information (Hamamura *et al* 2010b). Simultaneous imaging would enable the use of both MRI contrast agents and radiotracers to investigate multiple biological processes at the same time (Hamamura *et al* 2011). An additional advantage of simultaneous acquisition is the reduction in imaging time by eliminating the need for two separate scans. This could be beneficial in a clinical setting, where scheduling, patient convenience, and cost management are significant factors. A comprehensive rationale for combining MRI with gamma-ray imaging is given by Wagenaar *et al* (2006).

5. Conclusion

In this study, we evaluated the performance of an RF array coil integrated with CZT detector modules containing RF and gamma-ray shielding for unilateral breast imaging. The performance degradation of the RF coil due to the presence of CZT detector modules containing RF and gamma-ray shielding can be managed by techniques such as increasing the number of RF coil elements, auto-tuning, and pulse sequence development. Simultaneous acquisition of MR and SMM images of a compressed breast phantom demonstrates the feasibility of using our MRSMM system *in vivo* for human breast imaging. Ideally, the high sensitivity of MRI combined with better specificity from SMM will lead to improved breast cancer detection and diagnosis. The design and development of a high-quality RF coil for MRSMM is an important step in achieving this goal.

Acknowledgments

This research was supported in part by CIRM grant RT1-01120 and NIH NIBIB grant R44EB006712. It was also supported in part by the World Class University program through the National Research Foundation of Korea funded by the Ministry of Education, Science and Technology, Republic of Korea [grant no. R31-20004-(ON)].

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Figure 1.

(a) Illustration of the MRSMM assembly. Each CZT detector is enclosed in RF and gamma-ray shielding. (b) Modeling of the RF coil and RF shielded blocks in the simulation program. The blocks are each located 10mm from a brick-shaped phantom. (c) Experimental setup for testing RF and gamma-ray shielding in the RF coil. (d) Gamma-ray shielding box (left) and additional RF shielding (right).

Figure 2.

Illustration of breast compression at various angles. The two lesions (red circles) will be distinguishable in the SMM images acquired at 0° compression, but indistinguishable at 90° compression.

Figure 3.

The simulated RMS magnitude B_1 field distribution (left column) and RMS magnitude E_1 electric field distribution (right column) generated by the RF array coil tuned to 127 MHz without the conductive copper blocks (top row), the coil shifted to 138 MHz with the introduction of the conductive blocks (middle row), and coil with the conductive blocks re-tuned to 127 MHz. A shift in the electric field intensity around discrete components such as capacitors accounts for the asymmetry in (b). The B_1 field maps (unit=dB) were normalized to one input power for 2.37×10^{-10} A/m² source power in A and the electric field maps (unit=dB) were normalized to one input power for 2.59×10^{-8} V/m² source power in V.

Figure 4.

The simulated B_1 field distribution (top row), E_1 electric field distribution (middle row) and SAR distribution (bottom row) generated by the RF array coil tuned to 127 MHz without the conductive copper blocks (left column) and with the copper blocks (right column). The B_1 maps (unit=dB) were normalized to one input power for 2.37×10^{-10} A/m² source power in A, the electric field maps (unit=dB) were normalized to one input power for 2.59×10^{-8} V/m² source power in V and the SAR maps (unit=dB) were normalized to one input power 5.48×10^{-20} mW/g. source power in mW.

Figure 5.

The frequency spectra (impedance in Ω versus frequency in Hz) of the RF array coil with the RF shielding (top row), gamma-ray shielding (middle row) and both shielding (bottom row). It was measured following location at 125 mm (black solid line), 85 mm (black dotted line), 52 mm (black dashed line), 42 mm (gray dashed line), 36 mm (gray dotted line), and 26 mm (gray solid line) from the isocenter of RF coil. The red line was RF spectrum re-tuned at 26mm away location from the isocenter of RF coil.

Figure 6.

MR images of the brick phantom with no shielding (top row), RF shielding (2nd row), gamma-ray shielding (3rd row) and both shielding (bottom row) placed 10 mm from the phantom within the RF coil. All magnitude and phase images were each scaled to identical contrast levels.

Figure 7.

MR and SMM images from the top detector (left images of each column) and bottom detector (right images of each column) of the breast phantom with three different simulated compression angles for three different background configurations. Each SMM image was normalized to the maximum value of all the images. The yellow box denotes the ROI used to measure the radioactive counts.

Table 1

Average values of the parameters of the two circular loops of the RF array coil. The frequency and Q-factors were measured without loading the coil. The SNR was calculated from the MR images of the brick phantom.

	Frequency (MHz)	Q-factor	Q-factor after re-tuning	SNR
No shielding	127.0	117.5	---	301.1
RF shielding	132.5	86.1	87.9	231.3
Gamma-ray shielding	127.0	82.0	---	219.8
RF and gamma-ray shielding	134.0	68.6	61.3	185.6

Table 2

Radioactive counts within the ROI of figure 7 for the different background configurations. MRI-based attenuation correction (AC) and background correction (BC) were used to restore the original (no background) counts from the vial.

	no background	water background	active background
raw counts	59056	36175	65110
with AC	---	60242	---
with BC	---	---	35845
with AC and BC	---	---	59692