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Toward 7T breast MRI clinical study: safety assessment using simulation of heterogeneous breast models in RF exposure

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Abstract

Purpose: To facilitate assessment of RF power deposition and temperature rise within the breast, we present a method to seamlessly join heterogeneous breast models with standard whole-body models and demonstrate simulations at 7 T.

Methods: Finite-difference time-domain electromagnetic and bioheat simulations are performed to analyze the specific absorption rate (SAR) and temperature rise distributions in 36 Breast Imaging Reporting and Data System (BI-RADS) categorized breast models fused to 2 female whole-body models while transmitting from a 7T breast volume coil. The breast models are uncompressed in the prone position and feature heterogeneous tissue contents; fusion with human models uses affine transformation and the level-set method.

Results: The fusion method produces a continuous transient from the chest region to the posterior portion of breast models while preserving the original volume and shape of breast models. Simulation results of both Ella and Hanako models indicate that the maximum local SAR, partial body SAR, and local tissue temperature rise are positively correlated with both breast density and the highest BI-RADS density classification. Additionally, maximum local tissue temperature rise is positively correlated with maximum 10-g SAR values.

Conclusion: Fibroglandular tissue content plays an important role in the distribution of SAR and temperature rise within breast tissue. The combined body-breast models preserve the integrity of breast models while concurrently exhibiting the loading of whole-body human models. The procedures presented in this simulation study facilitate safety assessments for breast MRI across the population at both clinical and ultrahigh field strengths.

Keywords

breast density; FDTD; high field; safety; specific absorption rate; temperature rise

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SUPPORTING INFORMATION

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1 | INTRODUCTION

Breast cancer is the most commonly diagnosed cancer and leads to the highest death rate among women in the United States. In 2016, roughly 246 660 new incidences of breast cancer were diagnosed, corresponding to 29% of all new cancer cases, while breast cancer caused 40 450 deaths, equaling 14% of all cancers deaths.¹ Compared with other imaging modalities, MRI offers high sensitivity for breast cancer detection^{2,3} and is commonly indicated for screening high-risk subjects (e.g., women with familial history, genetic predisposition, significant chest radiation history, or lobular cancer^{4–7}). Furthermore, owing to superior contrast between tumor and fibroglandular tissue, MRI has been reported to improve sensitivity within dense breasts^{8,9} (i.e., those clinically classified as heterogeneous or extreme fibroglandular tissue according to the American College of Radiology Breast Imaging Reporting and Data System [BI-RADS] Atlas).¹⁰ The increased SNR at 7 T enhances both temporal and spatial resolutions of dynamic contrast-enhanced breast MRI, thus improving sensitivity; the diagnostic specificity of breast MRI may be extended by using supplementary techniques including diffusion-weighted MRI and MRS, which also have improved performance at 7 T.^{11–13} However, RF power deposition—quantified as specific absorption rate (SAR)—must be well characterized, particularly at high fields. Both the U.S. Food and Drug Administration and the International Electrotechnical Commission (IEC) include SAR as a criterion for assessing significant risk in MRI devices.^{14,15} The IEC60601-2-33 standard specifies SAR limits and temperature limits at normal operating mode and first-level controlled operating modes, whereas second-level controlled operating mode requires SAR limits be specifically approved by an appropriate investigational review board. For a volume transmit coil at normal operating mode, the partial body SAR limit is 2 to 10 W/kg (determined by the ratio of exposed patient mass to patient mass), and the maximum local temperature limit is 39°C. For a local transmit coil at normal operating mode, the local SAR (10-g averaged SAR) limit is 10 W/kg for trunk and 20 W/kg for extremities, and temperature constraints are the same as for a volume transmit coil.

Experimental methods to determine SAR values in tissue rely on either measuring the temperature rise of a loaded phantom or by measuring the energy per pulse delivered by RF coils.¹⁶ At present, the real-time in vivo determination of regional SAR values in tissue is unfeasible. The finite-difference time-domain method (FDTD) simulation method for electromagnetics (EM) uses a simple numerical integration algorithm and facilitates SAR calculation in human body models of different resolution.^{17–20} Although the FDTD method is robust for whole-body simulation, caution must be undertaken if simulating cropped regions of the body (e.g., head only). Wolf et al. showed that adjoining body parts outside the coil affect the resulting EM fields inside the coil, and a cropped body part alone will exaggerate the resulting SAR values.²¹

Researchers have access to a limited number of noncommercial adult female whole-body models for SAR simulations; 4 prominent examples are detailed in Table 1. All 4 models are oriented in the standing or supine positions,^{22–25} where breasts are flattened with suboptimal visualization, thus requiring adaptation to accurately simulate breast MRI in the prone position. Prone-position breast imaging provides separation of deep breast structures from the heart, liver, and pectoral muscles, offering greater resolution of small lesions in deep

breasts²⁶ and avoiding respiratory and potential cardiac artifacts.^{27–29} Most recently, Kim et al. studied the SAR and $\mathbf{B}_1^+$ field in various types of breast models³⁰; however, there was no discussion on the method of fusion of the breast models to human voxel models, and the influence on SAR from the chest region of the body model was unknown. Van der Velden et al. combined 2-mm-resolution unilateral breast models generated from segmented MRI images of 5 study volunteers—a personalized modeling paradigm requiring a priori 3D image sets—with a whole-body model for FDTD simulations³¹; however, the overlap-fusion method is susceptible to producing embedded skin layers, and the study participants did not represent all BI-RADS breast composition categories.

To accurately assess and quantify the breast regional 10-g SAR, partial body SAR, and tissue temperature values, we propose EM and thermal simulations of patient-based breast models covering all BI-RADS breast composition categories—designated as a, b, c, and d—in a breast-specific volume transmit coil.³² In this work, the patient-based breast models are created from segmented MRI and CT images of uncompressed (i.e., free of compression plates) breasts of human subjects in the prone position. Our fusion technique then joins the breast models with a human model by applying (1) affine transformation to create a curved breast posterior portion to be positioned on the top of the chest of human models; and (2) the level-set method of mean curvature motion, a general method in the computer vision community to seamlessly fuse joints or reconstruct surfaces.^{33–35} Our simulations illustrate homogenous $|\mathbf{B}_1^+|$ field distributions in the coil effective volume; and the 10-g SAR, partial body SAR, and tissue temperature values are quantified at a mean $|\mathbf{B}_1^+|$ within the coil effective volume, based on the TR-averaged $|\mathbf{B}_1^+|$ used in practical breast MRI scans at 7 T.^{32,36–38}

2 | METHODS

2.1 | Models

We adapted 9 MRI image-derived breast models (0.5-mm isotropic resolution) made available by the University of Wisconsin Cross-Disciplinary Electromagnetics Laboratory³⁹ and 9 CT image-derived breast models (0.35-mm isotropic resolution) from the Carl E. Ravin Advanced Imaging Laboratories at Duke University.⁴⁰ All breast models were uncompressed and in the prone position; the set included models from all 4 BI-RADS classifications: a, b, c, and d. Dielectric properties were assigned from the IT'IS Foundation Database.⁴¹ For quantitative analysis, we labeled the breast models by the volumetric fibroglandular tissue (FGT) percentage (FGT%).

Tissue types in each breast model included skin, breast adipose, FGT, and 3 (intravoxel) FGT-adipose combinations of varying FGT makeup, by volume percentage: 25%, 50%, and 75%. The FGT-adipose tissue with higher proportions of FGT border the densest regions, which are typically in the center of the breast. We linearly distributed the dielectric properties at 298 MHz for the 3 FGT-adipose mixture definitions (i.e., 75%, 50%, and 25% FGT-adipose mixture) between that of 100% FGT ($\epsilon_r = 62.5$, $\sigma = 851$ mS/m) and adipose tissue ($\epsilon_r = 5.54$, $\sigma = 32.7$ mS/m). Our motivation to define these tissue types derives from Zastrow et al. who showed linearly decreasing dielectric properties between 0 and 5 GHz for 4 tissue types dubbed “glandular/fibroconnective-1,” “glandular/fibroconnective-2,”

“glandular/fibroconnective-3,” and “transitional.”⁴² Likewise, for the CT image–derived breast models, Erickson et al. defined adipose tissue, 4 heterogeneous proportions of FGT and adipose mixtures, and skin based on linearly increasing voxel intensities in breast CT data.⁴⁰

For this study, we used 2 adult female models with superior tissue counts and relatively high resolution: Ella and Hanako. The 0.5-mm isotropic Ella model is part of the Virtual Family 1.2 and represents a 26-year-old female of height 160 cm and mass 58 kg.²⁵ The 2-mm isotropic Hanako model is from Japan’s National Institute of Information and Communications Technology and represents an average 18- year-old to 30-year-old Japanese female of height 159 cm and mass 52.6 kg.²²

2.2 | Model fusion algorithm

2.2.1 | Morphological filtering—Closing and opening morphological operators⁴³ are applied to smooth the pectoral muscle wall. First, the pectoral muscle wall is converted to a binary image volume, with ones representing the pectoral muscle and bones, and zeros representing empty space. Next, we smooth the binary image volume by a 3D image-closing morphological operator; the kernel is a 5-voxel-diameter solid sphere for the Hanako model and a 15-voxel-diameter solid sphere for the Ella model. Subsequently, we fill empty voxels inside the pectoral muscle wall with a voxel value of 1. Finally, we smooth the binary image volume with a 3D image-opening morphological operator, using the same kernel sizes defined previously.

2.2.2 | Affine transformation—The posterior portion of the breast model is projected on top of the smoothed pectoral muscle wall surface using affine transformation. N voxels from location $[x, y, z]$ are transformed to $[x, y, z - d_i]$, where $i = 1, 2, 3 \dots N$, and d_i is the z projection values of the pectoral muscle wall surface. Our initial implementation of this method was previously described.^{44,45}

2.2.3 | Level-set method of mean curvature motion—The fusion of the posterior portion of the breast model to the subcutaneous adipose of the chest region is achieved through a mean-curvatures regularized level-set motion algorithm.³⁵ A minimal amount of subcutaneous adipose is added. Consider the common level set function

$$\frac{\partial \Phi}{\partial t} = -b\kappa|\nabla \Phi|, \quad (1)$$

where Φ is the time-dependent level-set function that determines a dynamic contour when $\Phi(x, y, z, t) = 0$; b is a constant that controls the speed of the active contour; $\nabla \Phi$ is the inward vector normal at the dynamic contour; and κ is the mean curvature. Equation (1) can be decomposed to

$$\frac{\partial \Phi}{\partial t} = -\frac{b}{2}(\kappa_1 + \kappa_2)|\nabla \Phi| \quad (2)$$

where κ_1 and κ_2 are the maximal and minimal curvatures (i.e., the principle curvatures), derived from the mean curvatures (κ_M) and Gaussian curvature (κ_G) equations⁴⁶ as

$$\kappa_1, \kappa_2 = \kappa_M \pm \sqrt{\kappa_M^2 - \kappa_G} \quad (3)$$

Where

$$\kappa_M = (\Phi_x^2 \Phi_{yy} - 2\Phi_x \Phi_y \Phi_{xy} + \Phi_y^2 \Phi_{xx} + \Phi_x^2 \Phi_{zz} - 2\Phi_x \Phi_z \Phi_{xz} + \Phi_z^2 \Phi_{xx} + \Phi_y^2 \Phi_{zz} - 2\Phi_y \Phi_z \Phi_{yz} + \Phi_z^2 \Phi_{yy}) / |\nabla \Phi|^3$$

(4)

and

$$\kappa_G = - \frac{\begin{vmatrix} \Phi_{xx} & \Phi_{xy} & \Phi_{xz} & \Phi_x \\ \Phi_{xy} & \Phi_{yy} & \Phi_{yz} & \Phi_y \\ \Phi_{xz} & \Phi_{yz} & \Phi_{zz} & \Phi_z \\ \Phi_x & \Phi_y & \Phi_z & 0 \end{vmatrix}}{|\nabla \Phi|^4} \quad (5)$$

To consider the outward motions for both concave and saddle surfaces, we take the positive components of the principle curvatures and denote them by κ_1^+ and κ_2^+ . Thus, Equation (2) becomes

$$\frac{\partial \Phi}{\partial t} = -b(\kappa_1^+ + \kappa_2^+)|\nabla \Phi|. \quad (6)$$

2.2.4 | Implementation—The model fusion software is implemented in MATLAB (MathWorks, Natick, MA), and the source code is publicly available (<https://github.com/rispoli-lab/Bilateral-Breast-Fusion->). The implementation of the fusion scheme for bilateral breasts consists of the following steps:

1. Unilateral breast models are scaled to the desired size with aspect ratio using a 3D nearest-neighbor interpolation. The scaling factor is approximately the ratio of breast model voxel length to the human model voxel length. In our simulation study, the breast models are resized to cover all breast tissue contained in the Ella and Hanako models. Unilateral breast models are mirrored by referring to the center sagittal plane of human models to generate bilateral breast models. Circular rings representing a breast RF coil are positioned outside of the scaled breast models to ensure the tissue-to-coil spacing is at least 1 cm, and the coil's

coordinates relative to the human model are saved for positioning the breast coil in the EM simulation software in step 4. Notably, the breast coil will be fixed at the same location with respect to the human model (e.g., Ella and Hanako).

2. In preparation for becoming the interface between the breast and body models, the pectoral muscle and bone volumes in the female body model are extracted and smoothed by morphological filtering. Affine transformation is applied on the breast model to create a curved posterior portion that fits on the surface of the smoothed pectoral muscle and bone.
3. A 3D region is selected, containing the joints of the posterior portion of the breast model and the chest of the human model. The level-set method is applied on the selected region to smooth the sharp joints and edges at the outer layer of the integrated model. Extra tissue added by the level-set method is labeled as subcutaneous adipose tissue. A skin layer of fixed depth is sequentially added on the top of the subcutaneous adipose tissue layer. The smoothed 3D region is then merged with the breast-chest model to create the modified breast model. The modified breast's coordinates relative to the human model are saved for positioning the breast model in the EM simulation software in step 4.
4. The modified breast model is saved, imported into EM simulation software, and positioned at the desired location based on location coordinates generated from step 3. Breast models are assigned an elevated meshing priority to overwrite breast tissues contained in Ella and Hanako. The breast RF coil is positioned based on the location saved in step 1.

2.3 | Electromagnetic and thermal simulations

We use a commercial full-wave simulation software package (XFDTD 7.7, Remcom, State College, PA) for electromagnetic and thermal simulation. For this study, the modified breast models are saved in the voxel file (.vox) to facilitate importing into XFDTD. Following successful voxel file import, the modified breast models are assigned a high mesh priority and consequently overwrite the breast regions of the body models; notably, the body model is never explicitly modified and saved, facilitating compliance with common licensing terms that prohibit truncating or segmenting the models. To mitigate potential discretization effects from combining multiple gridded objects, mesh cells were created with averaged dielectric properties.

A previously reported quadrature breast coil composed of orthogonal Helmholtz and saddle pairs³² is modeled by copper sheet of conductivity 5.96×10^7 S/m. The contour path method is used to mesh curved coil conductor geometries.⁴⁷ All coil elements are segmented by capacitors and fine-tuned using the ideal drive method to match input impedances from bench measurements.⁴⁸ The 2 coil pairs are in quadrature operation and driven by 298-MHz sinusoidal current feeds with appropriate 90° phase separation to achieve equal input power for both pairs.

The RF coil is positioned around the right breast, with a minimum spacing of 1 cm from adjacent tissue, at a fixed height of approximately 1 cm above the chest wall of Hanako or

Ella. Detailed Yee cells, set from 0.5 to 1 mm, surround the RF coil and are edge-aligned with geometric extents of the coil conductors; these detailed cells extend in 3 dimensions throughout the mesh and include parts of the human body. The Yee cell size for other regions is set to a maximum of 20 mm. The coil effective volume is outlined by the black dashed line in Figure 1; less than 95% of total RF power is absorbed in a homogeneous adipose phantom filling this volume. Thirteen 20-mm cells, approximating one-quarter wavelength in free space, are included for padding. The boundary condition for the far field is set to 7 perfectly matched layers. Tissue properties at 298 MHz for the Ella model and imported breast models are assigned according to the IT'IS foundation database,⁴¹ and properties of remaining tissue types that are unique to Hanako are assigned to source tissues as detailed by Nagaoka et al.²² The EM simulation convergence criterion for a sinusoidal current feed is set to -50 dB (or 0.316%), based on the dissipation of additional frequencies introduced during the ramp-up period of the sinusoidal wave. The 10-g SAR sensor is enabled in XFDTD for calculating the 10-g averaged SAR values within tissue using the methods outlined by IEEE standard C95.3-2002.⁴⁹ Following successful completion of the simulation, steady-state $\mathbf{B}_1$, raw SAR, and 10-g SAR are imported into MATLAB to calculate the average $|\mathbf{B}_1^+|$ (in μT) within the tissue in the coil-effective volume (i.e., $|\mathbf{B}_1^+|_{\text{mean}}$), and power dissipated to different tissue types. According to IEC60601-2-33, the partial body SAR is calculated as the total power absorbed in the coil effective volume divided by the total tissue mass in the coil effective volume. The SAR values are normalized by $|\mathbf{B}_1^+|_{\text{mean}}^2$ producing normalized SAR in $\text{W/kg}/\mu\text{T}^2$. Reviewing the 7T breast MRI literature, we computed TR-averaged $|\mathbf{B}_1^+|$ values of 0.13 μT for fat suppression with 8° tip angle and 4-ms TR,^{36,37} 0.6 μT with an RF-intensive imaging protocol with 20 μT applied with 3% RF duty cycle,³² and 3.6 μT for WALTZ-16 proton decoupling of 18 μT applied with 20% RF duty cycle.³⁸ Based on these practical TR-averaged $|\mathbf{B}_1^+|$ values, we scale the output of raw SAR, 10-g averaged SAR, and partial body SAR to a moderate $|\mathbf{B}_1^+|$ value of 1.5 μT .

Whole-body thermal simulation is performed in XFDTD using the same gridding from the EM simulation. The heat transfer equation solved adheres to the Pennes bioheat equation,⁵⁰ with an additional heating term due to the RF dissipated power in tissue.⁵¹ The ambient (air) temperature is set to 25°C, and the perfusion medium is blood at 37°C. Both initial and final temperatures are calculated from the steady-state solution of the heat transfer equation. Initial temperature is tissue absolute temperature without RF exposure, and final temperature is absolute temperature with addition of dissipated power in tissue due to RF exposure. The total dissipated power in tissue of the thermal simulation is set to a value corresponding to an MRI sequence with TR-averaged mean $|\mathbf{B}_1^+| = 1.5 \mu\text{T}$ within tissue inside the coil effective volume. At a similar SAR level, the steady-state solution of the Pennes bioheat equation corresponds to 30 minutes of tissue exposure to RF.⁵² Final temperature data are exported to MATLAB.

3 | RESULTS

Figure 2 shows the sagittal and axial cross sections, in the same scale, of the Ella and Hanako models fused with 0.5-mm resolution breasts. Affine transformation results in the curved posterior breast portion fitting properly on the surface of the pectoral muscle and

bones. The level-set method added additional subcutaneous adipose tissue, depicted in blue color, to smooth the surface at the outer layer of the breast to the chest joint. Figure 1D illustrates 3D surface renderings of Ella in prone position and fused to modified breast models while loaded with the unilateral RF coil for EM and thermal simulation. Figure 1A–C shows the $|\mathbf{B}_1^+|$ profile of a loaded breast coil in the 3 principal anatomical planes. A slight discontinuity of $|\mathbf{B}_1^+|$ appears along the boundary between air and skin tissue, owing to surface currents on the skin. The volume RF transmit coil $|\mathbf{B}_1^+|$ is distributed homogeneously (–20 to –29 dB) within the extended volume encompassed by the breast coil.

Figures 3 and 4 illustrate the sagittal slices of raw SAR and tissue temperature rise of 4 BI-RADS-category breasts. Sagittal slices of the raw and 10-g SAR and the local tissue temperature rise of all simulation cases are presented as Supporting Information Figures S1, S2, and S3, respectively; the corresponding axial slices are presented in Supporting Information Figures S4, S5, and S6. Figure 3 shows breasts fused with Ella, and Figure 4 shows breasts fused with Hanako. Figure 5 illustrates the sagittal slices of raw and 10-g SAR and the local tissue temperature rise of breast models of 0.5-mm and 2-mm voxel resolution fused with Hanako, with the same 0.5-mm breast model fused to Ella for comparison. In Figures 3, 4, and 5, skin and FGT have high raw- SAR levels compared with the remaining breast tissue owing to their high conductivity. The FGT of relatively high SAR levels branches out from regions near the nipple toward the chest, with SAR levels gradually attenuated near the pectoral muscle wall. In Figures 3 and 4, BI-RADS a, b, c, and d breasts show ascending values of FGT%, raw and 10-g SAR, and local tissue temperature rise. In Figure 5, the Hanako 2- mm-resolution breast skin with relatively high SAR levels is thicker compared with Ella and Hanako 0.5-mm breast skin. In Figures 4 and 5, the blank areas in the Hanako body represent internal air in the large intestine. The breast models in Figure 5 included chest wall tissues in the posterior portion that were unable to be segmented from the CT images; thus, these regions remained in the breast models and appear heightened in the raw SAR image in Figure 5.

Figure 6 shows the power deposition in tissue as a percentage of the same net power dissipated in Ella and Hanako. Muscle, skin, FGT, lung, heart, liver, and adipose tissue contain the most dissipated power in both models. In addition, connective tissue (cyan-labeled tissue in the Ella model in Figure 2) contained a high percentage of dissipated power in Ella, whereas blood (pink-labeled tissue in heart lumen of the Hanako model in Figure 2) contained a high percentage of dissipated power in Hanako. Breast fat has a mean value of 3% power dissipation, and FGT has a mean value of 9% in both 0.5-mm-resolution Ella and Hanako breasts. In the 2-mm-resolution Hanako breasts, the mean dissipated power in skin is significantly higher, whereas the mean dissipated power in FGT is slightly lower as compared with 0.5-mm resolution breasts.

Figure 7 shows the plot of the maximum 10-g averaged SAR as a function of FGT%, the plot of partial body SAR as a function of FGT%, and the plot of maximum local tissue temperature rise (i.e., the maximum temperature rise of tissue within the coil effective volume) as a function of the maximum 10-g averaged SAR for Ella and Hanako 0.5-mm-resolution breasts. For comparison of 10-g averaged SAR, partial body SAR, and

temperature rise values among BI-RADS breasts within each human model, BI-RADS a and b breasts have the lowest values; BI-RADS c have moderate values; and BI-RADS d have the highest values. Both Ella and Hanako breasts have similar slope of 10-g SAR and partial body SAR as a function of FGT%, and the maximum 10-g averaged and partial body SAR among all breasts is 1.5 and 0.7 W/kg, respectively. For both models, partial body and the maximum 10-g SAR value are positively correlated with FGT%, and the maximum local tissue temperature rise values are positively correlated with the maximum 10-g averaged SAR value. There is stronger linear regression between the maximum local tissue temperature rise and the maximum 10-g SAR value in the Hanako model as compared with the Ella model, as the R^2 values are 0.85 in the Hanako model and 0.49 in Ella model. This implies that the difference between Hanako and Ella anatomy may contribute to variance in bioheat thermal simulation results. Based on the bioheat simulation methodology in XFDTD, the simulated final tissue temperatures within human models are below 37.6°C, and the simulated tissue temperatures within breast tissue in the effective coil volume are below 37.2°C. Neither of these tissue temperatures increases as breast FGT% increases.

The transmit efficiency for each breast phantom simulation case is provided in Table 2. The transmission efficiency is calculated as mean $|\mathbf{B}_1^+|$ within breast tissue inside the coil effective volume divided by the square root of the net input power. Comparably, the transmit efficiency of the quadrature coil reported by van de Bank et al. was $0.9487 \mu\text{T}/\sqrt{W}$, given 30 μT within breast tissue and input power of 1kW.¹²

4 | DISCUSSION AND CONCLUSIONS

The simulated partial body SAR and local temperature obey the IEC60601-2-33 constraints for a volume transmit coil at normal operating mode. Furthermore, because local breast transmit coils are also used at 7 T,^{11,12,36} we provided 10-g averaged SAR results to facilitate comparison, and our simulation results also fulfill the 10-g averaged SAR constraints for local transmit coils. Breast transmit coils, as opposed to a whole-body transmit coil, will be necessary for potential future clinical adoption of 7T breast MRI. Although the presented SAR and temperature results corresponding to a time-averaged $|\mathbf{B}_1^+|$ of 1.5 μT are well below the recommended limits, it is important to note that SAR increases proportionally with the square of $|\mathbf{B}_1^+|$. Furthermore, although temperature will increase proportionally to SAR for short exposure durations, new temperature simulations would be required for different time-averaged $|\mathbf{B}_1^+|$. Because the practical safety hazard from RF energy deposition is manifested by the thermal dose, and not total power deposition, it is notable that dense breasts may be more vulnerable to local temperature increases. Adequate airflow during prone breast imaging can play a critical role in skin cooling.⁵³

Our results show FGT%, BI-RADS breast composition categories, and human model resolution are important factors affecting SAR and thermal distributions. As shown in Figure 7, models of BI-RADS a, b, c, and d breasts have FGT% in ascending order. Because FGT has the highest conductivity among breast tissues, both 10-g averaged SAR and partial body SAR values are positively correlated with breast FGT contents. This result is congruent with the transmit efficiency data presented in Table 2, with a general trend of higher FGT% requiring greater transmit power. In addition, FGT will experience temperature rise given its

higher metabolic rate and greater vascular perfusion compared to adipose tissue. As shown in Figure 5, 2-mm Hanako breast skin is thicker than 0.5-mm breast skin at a detailed meshing scheme (e.g., 0.5–1.0 mm mesh) in FDTD simulation, due to the larger voxel length of 2-mm-resolution breasts as compared with 0.5-mm-resolution breasts. Hence, power dissipation in skin of 2-mm Hanako breasts are higher than in skin of 0.5-mm Hanako breasts as illustrated in Figure 6. Additionally, as 2-mm-resolution breast models fused with Hanako are down-sampled 4-fold as compared with 0.5-mm Hanako breasts, detailed FGT structures in breasts are not reserved, which results in loss of FGT-adipose mixture of high FGT% (e.g., 75% FGT-25% adipose mixture). Therefore, Figure 6 illustrates higher power dissipation in 0.5-mm Hanako breasts as compared with 2-mm Hanako breasts. However, the power dissipation in the FGT and breast fat is similar in both Ella and Hanako with 0.5-mm breasts.

In practice, safety simulations must include the specific RF coil design being used in the imaging study, as coils may differ in transmit efficiency and produce unique $\mathbf{B}_1^+$ and E field patterns. Notably, we previously compared a quadrature breast coil of 2 orthogonal loops, akin to that used,^{11,12,36} with the Helmholtz and saddle pair used in this study; our results⁵⁴ illustrated similar trends with respect to breast and body models and thus we exclude simulation of other RF coils in this paper. Furthermore, the whole-body transmit coils used at 1.5 T and 3 T may be modeled with the fused phantoms presented herein; preliminary results using the FDTD EM solver in the Sim4Life software package (Zurich MedTech AG, Zurich, Switzerland) were recently presented.⁵⁵

The observed maximum tissue temperatures based on the XFDTD thermal simulation are below 37.6°C. In addition to XFDTD, the Pennes bioheat equation is also applied for the thermal simulation in Sim4Life version 3.4. In both cases, the blood temperature is constant during the thermal simulation, and thus the simulation results will not account for body core temperature. The generic bioheat transfer model takes into account spatial and temporal blood temperature variation due to RF heating. Comparisons between in vivo and in silico data show the temperature results based on generic bioheat transfer model have better fitting, whereas the temperature reported from the Pennes bioheat equation shows less fitting with in vivo data.⁵²

Because various shapes and anatomical structures of breasts exist across the population, it is difficult to define a standard breast template, which is a primary criterion used for image registrations. In contrast to image registration, the level-set method can preserve the original structures of the middle and anterior portions of breast models as well as whole-body human models. The level-set function discussed in this paper is regularized by the surface curvatures of both the human models and the breast models. Considering that recreating natural-appearing breast surface curvatures is considered the gold standard for breast reconstruction surgery,^{56–58} the fusion method presented here produces models that are analogous to the actual population.

Supplementary Material

Refer to Web version on PubMed Central for supplementary material.

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FIGURE 1. $|\mathbf{B}_1^+|$ spatial profile of a loaded breast coil in the sagittal (A), coronal (B), and axial (C) planes. The coil-encompassed volume accessible by the subject, with 1-cm spacing to adjacent coil boundary, is outlined by the black dashed line; this volume is defined as the coil effective volume and used for averaging $|\mathbf{B}_1^+|$. Breast and chest are outlined by the red long-dashed line. Local discontinuity of $|\mathbf{B}_1^+|$ appears at the skin-to-air boundary. (D) Right breast of prone position Ella model is positioned inside breast coil for unilateral RF excitation, with the coil boundaries at least 1 cm to skin

FIGURE 2.

Bilateral breasts (0.5-mm resolution) fused to whole-body models. Sagittal (A) and axial (B) view of bilateral breasts fused to Ella (0.5-mm resolution). (C,D), The same set of bilateral breasts are fused to Hanako (2-mm resolution). The modified breast models are assigned a higher meshing priority and consequently overwrite the chest region of the whole-body models. Tissue added by the level-set method is designated as level-set subcutaneous adipose tissue (SAT) and colored blue

FIGURE 3.

Sagittal plots of raw specific absorption rate (SAR) and temperature rise with 4 Breast Imaging Reporting and Data System (BI-RADS) categorized breasts fused to Ella. Raw SAR and temperature rise when tissue temperature reaches the steady state are quantified at mean $|\mathbf{B}_1^+| = 1.5 \mu\text{T}$ within breast tissue at 7 T. Each row represents a simulation case of 1 BI-RADS category, and the corresponding fibroglandular tissue (FGT) percentage is reported. All plots are taken from the sagittal plane across the coil center in a FOV of 22×19 cm. Raw SAR 0 dB = 1.3 W/kg

FIGURE 4.

Sagittal plots of raw SAR and temperature rise with 4 BI-RADS-categorized breasts fused to Hanako. Parameters are equivalent to those from Figure 3. Internal air in the large intestine of Hanako results in the blank internal regions of raw SAR

FIGURE 5.

Sagittal plots of raw SAR, 10-g averaged SAR, and temperature rise with the same set of bilateral breasts, but of different voxel resolution, fused to Ella and Hanako. The SAR and temperature rise when tissue temperature reaches the steady state are quantified at mean $|\mathbf{B}_1^+| = 1.5 \mu\text{T}$ within breast tissue at 7 T. Both the top and bottom rows include breast models with 0.5-mm resolution, whereas the middle row includes the same breast models at 2-mm resolution. This breast was categorized as BI-RADS d, and the FGT% is 47.6 for Ella and 50.4 for Hanako. All plots are taken from the sagittal plane across the coil center in the same FOV of $22 \times 19 \text{ cm}$ (SAR 0 dB = 1.3 W/kg). Internal air in the large intestine of Hanako results in the blank internal regions of raw SAR

FIGURE 6.

Mean power deposition percentage in tissues with the same net power dissipated under RF exposure in Ella (top) and Hanako (bottom). The same sample size of 0.5-mm and 2-mm resolution breasts are plotted in the bottom bar graph. Error bars demonstrate data distributed within 1 SD

FIGURE 7.

Plots illustrating correlations between FGT%, SAR, and temperature rise for Ella and Hanako breasts. Top plot shows the maximum 10-g averaged SAR values versus FGT%. Middle plot shows the partial body SAR versus FGT%. Bottom plot shows maximum local tissue temperature rise (at steady state) versus maximum 10-g averaged SAR values. All breast phantoms are in 0.5-mm resolution. Linear regression is used to fit Ella and Hanako breast data separately; the blue regression line is for Ella, and the black line is for Hanako. The R^2 values for 10-g averaged SAR as a function of FGT%, partial-body SAR as a function of FGT%, and temperature rise as a function of 10-g averaged SAR are 0.61, 0.70, and 0.49 for Ella, and 0.63, 0.86, and 0.85 for Hanako, respectively. All breast models are color-labeled according to their BI-RADS category

TABLE 1

Attributes of Selected Adult Female Whole-Body Voxel Models

Name	Resolution (mm)	Height (cm)	Tissue types	Year released
Ella	0.5 ^a	160	76	2013
Hanako	2	159	51	2004
Naomi	2	163	41	2005
High-fidelity female body mesh	5	165	36	2005

^aThe Virtual Population version 1.2 was the last release of a voxelized Ella model at 0.5-mm resolution; the finest voxelized resolution in subsequent releases is 1 mm. The Ella model is also available in CAD format with the purchase of commercial software (Sim4Life, Zurich MedTech AG, Zurich, Switzerland).

TABLE 2

Transmit Efficiency for Each Breast Phantom Simulation Case

Ella	Hanako			
	Fibroglandular tissue (%)	Transmit efficiency ($\mu\text{T}/\sqrt{\text{W}}$)	Fibroglandular tissue (%)	Transmit efficiency ($\mu\text{T}/\sqrt{\text{W}}$)
7.1	1.24	8	1.22	1.19
7.1	1.16	7.9	1.19	1.21
17	1.15	17.9	1.28	1.29
19.1	1.23	20	1.15	1.24
19.2	1.26	19.7	1.23	1.25
23.4	1.28	27.5	1.11	1.23
31	1.13	31.7	1.2	1.15
31.6	1.19	34.3	1.15	1.14
36.3	1.26	36.6	1.2	1.17
39.2	1.17	42.6	1.17	1.14
40.3	1.03	40.7	1.15	1.14
41.9	1.15	43.8	1.15	1.14
42.4	1.09	42.9	1.15	1.14
43.3	1.08	43.4	1.15	1.14
47.6	1.15	50.4	1.15	1.14
54.2	1.06	56.2	1.15	1.14
62.5	1.16	62.5	1.15	1.14
66.2	1.08	66.6	1.15	1.14