Published in final edited form as:

J Magn Reson Imaging. 2010 September ; 32(3): 684–691. doi:10.1002/jmri.22270.

Pyrolytic Graphite Foam: A Passive Magnetic Susceptibility Matching Material

Gary C. Lee, MS¹, Patrick W. Goodwill, MS¹, Kevin Phuong², Ben A. Inglis, PhD³, Greig C. Scott, PhD⁴, Brian A. Hargreaves, PhD^{4,5}, Elizabeth Li, BS⁴, Alex C. Chen, BS², Rachana N. Shah, BS², and Steven M. Conolly, PhD^{1,2}

¹Berkeley/UCSF Bioengineering Joint Graduate Group

²Department of Bioengineering, University of California, Berkeley

³Helen Wills Neuroscience Institute, University of California, Berkeley

⁴Department of Electrical Engineering and Computer Science, Stanford University

⁵Department of Radiology, Stanford University

Abstract

Purpose—To evaluate a novel soft, lightweight cushion that can match the magnetic susceptibility of human tissue. The magnetic susceptibility difference between air and tissue produces field inhomogeneities in the B_0 field, which leads to susceptibility artifacts in MR studies.

Materials and Methods—Pyrolytic graphite (PG) microparticles are uniformly embedded into a foam cushion to reduce or eliminate field inhomogeneities at accessible air and tissue interfaces. 3T MR images and field maps of an air/water/PG foam phantom were acquired. Q measurements on a 4T tuned head coil and pulse sequence heating tests at 3T were also performed.

Results—The PG foam improved susceptibility matching, reduced the field perturbations in phantoms, does not heat, and is non-conductive.

Conclusion—The susceptibility matched PG foam is lightweight, safe for patient use, adds no noise or MRI artifacts, is compatible with RF coil arrays, and improves B_0 homogeneity, which enables more robust MR studies.

Keywords

Magnetic susceptibility; B_0 field homogeneity; shim; pyrolytic graphite

Introduction

In MR studies, the uniformity of the B_0 main field is critical for image signal and contrast. In particular, the 9 ppm magnetic susceptibility difference between air and tissue produces magnetic field perturbations near the skin and near the lungs. These are known to cause significant MRI artifacts including unreliable fat suppression, intravoxel dephasing, blurring, and voxel shifts. Certain pulse techniques, such as echo planar imaging (EPI), balanced steady state free precession (bSSFP), blood oxygen level dependent (BOLD) contrast, gradient recalled echo imaging (GRE) and fat suppression methods suffer

particularly from B_0 field inhomogeneities (1,2). Furthermore, the absolute size of the perturbations also increases as the strength of the B_0 field increases, making artifact-free imaging at 3.0 T and above a challenge. Susceptibility artifacts make it more difficult to conduct robust MR studies for any applications that require exceptionally uniform B_0 fields, such as fat suppression, BOLD fMRI (3), extremity MRI, and spectroscopy (4).

Susceptibility artifacts can be reduced by directly compensating for the B_0 inhomogeneities with either active or passive methods. Actively shimming the region of interest (ROI) with gradient shims and low order spherical harmonic shim coils is common in MRI studies. However, active and dynamic shim methods cannot correct for high spatial frequency perturbations (5–7), since the shim coils must be large to admit the patient. Large coils cannot generate high spatial frequency field patterns deep within a patient (2). However, these high spatial frequency perturbations are invariably present near abrupt skin-tissue interfaces, such as those found near the breast, shoulders, chin, and extremities. Other types of shims include diamagnetic crystal pyrolytic graphite (PG) shims (8–10) that have been placed in the subject's mouth to shim the field within the brain. Here the small, solid PG shims are shaped to partially compensate for field inhomogeneity induced by air within the sinuses. A similar, active shim method using localized external current carrying coils has also been developed to compensate for inhomogeneities in the human prefrontal cortex (11,12). These sets of well-defined shim coils were designed and tailored specifically for the local field distortions caused by the multi-air cavity system of the sinuses, and were located outside the patient.

A popular method to improve field homogeneity near the skin is to cover the skin with a tissue susceptibility matched material to move the inhomogeneities outside the field of view or ROI. This is a simple, passive, and direct technique for reducing B_0 inhomogeneities at air and tissue boundaries. Previous versions of these techniques include bags of fluid, such as perfluorocarbon (13), barium sulfate doped water (14), Kaopectate (15), or other MRI invisible agents. However, it can be difficult to shape a fluid to conform to the body, and these fluids are typically incompatible with an embedded RF coil. In addition, these materials have specific densities close to one, and hence can be quite heavy when of adequate size.

Previous finite element simulations by Bhagwendian on simple geometries such as spheres and cylinders suggest that increasing the size of the object can shift the high frequency perturbations at the interface outside the original ROI (16). This suggests that surrounding a target region with a layer of susceptibility matched material would result in a more homogenous field near the tissue/material interface and is depicted in Fig. 1. The necessary size of this matching layer, which is approximately a radius for cylinders and spheres, suggests a lower density material is preferable. For example, surrounding a patient head with a 3 in. shell of tissue susceptibility matched material in a typical head coil can improve homogeneity near the skin interface.

In this article we present a new passive method to reduce field perturbations and to achieve improved B_0 homogeneity near the skin. Here, instead of a fluid, we employ a lightweight, composite closed cell foam. The key innovation was to dope the closed-cell foam with a randomized, uniform dispersion of pyrolytic graphite (PG) microparticles to match the magnetic susceptibility of human tissue. Hence, the lightweight PG foam composite may become a convenient alternative to the fluid matching agents discussed above for moving the high spatial frequency perturbations outside the tissue of interest. Unlike fluid susceptibility matching agents, the PG foam is simple to shape, lightweight, and can be embedded directly within an RF coil. Because foam padding is ubiquitous in the MR suite,

PG foam can be easily adapted for use for a wide variety of applications, such as spectroscopy, patient studies, especially for extremity MRI, and small animal imaging.

Here we show experimentally that the PG foam composite matches the magnetic susceptibility of water. We calculate a tolerance of at least 10% in targeted PG volume fraction for improved susceptibility matching. We also show that the foam is three orders of magnitude less conductive than human tissue, so it is safe to use in an MRI scanner and it will add virtually no noise to the scan. We also demonstrate that PG foam is lightweight and produces no MRI signal.

Theory

Pyrolytic graphite is a polycrystalline form of carbon that has a hexagonal crystal structure, much like normal graphite (17). It crystallizes in a planar order resulting in stacks of graphene sheets, which produces a single cleavage plane. Due to these highly ordered layers, pyrolytic graphite has well-known anisotropic properties (18). For example, it is on the order of 10^4 times more electrically conductive parallel to its crystal plane than perpendicular to the plane (19). Pyrolytic graphite also has anisotropic diamagnetic susceptibility that is 70 times greater perpendicular to its crystal plane ($\chi_{\perp} = -595$ ppm) than parallel to the plane ($\chi_{\parallel} = -8.5$ ppm) (2).

Composite Bulk Magnetic Susceptibility

We take advantage of the anisotropic diamagnetic susceptibility of pyrolytic graphite to create an isotropic, two-phase composite material consisting of diamagnetic pyrolytic graphite particles embedded in dielectric matrices useful for tissue magnetic susceptibility matching. We have calculated that we can control the isotropic susceptibility of a material by embedding a small volume fraction of randomly oriented powdered pyrolytic graphite particles into a closed-cell polyurethane foam.

We first calculate the effect of randomly dispersing PG particles into the foam. The average susceptibility of a dispersion of small magnetic particles in a less magnetic matrix can be estimated in terms of the volume fraction f of the magnetic particles (2):

$$\chi_{avg} = \frac{f\chi}{1 + \alpha\chi}, \quad [1]$$

where χ is the overall magnetic susceptibility of the dispersed particles and α is a shape dependent demagnetizing factor between 0 and 1 (2). Because χ is on the order of part per million, the denominator of Eq. 1 approaches unity and the total equation reduces to

$$\chi_{avg} = f\chi. \quad [2]$$

Note that this implies that the shapes of the crystal (sphere versus rod or rhombus) have negligible effect on the bulk magnetic susceptibility of the composite, which greatly simplifies the design of the composite.

The magnetic susceptibility of the PG foam has contributions from a volume fraction f of the uniformly dispersed and randomly oriented PG, which has a magnetic susceptibility, χ_{Total} equivalent to the average of its directional components χ_1 , χ_2 , and χ_3 (20,21),

$$\chi_{Total} = \frac{(\chi_1 + \chi_2 + \chi_3)}{3} = \frac{(\chi_{\perp} + 2\chi_{\parallel})}{3}, \quad [3]$$

and from the remaining volume fraction $(1-f)$ of the foam, which has the equivalent susceptibility of air ($\chi_{air} = 0.36$ ppm) and is equal to

$$\chi_{PG\ foam} = \chi_{Total} f + \chi_{Air} (1-f). \quad [4]$$

Thus, the equation for the magnetic susceptibility of the pyrolytic graphite foam (χ_{PGfoam}) is

$$\chi_{PG\ foam} = -204.4 f + 0.36 [ppm], \quad [5]$$

where f is the volume fraction of the pyrolytic graphite foam. Eq.5 predicts a pyrolytic graphite volume fraction $f \sim 0.045$ to match human tissue susceptibility of ~ -9 ppm (2). Note that 10% error in the ideal volume fraction translates to susceptibility values between -8.5 to -9.5 ppm, or within 1 ppm of human tissue susceptibility. This tolerance is generous, which is encouraging for the robustness of the susceptibility matching under variable compression of the PG foam. To prevent significant compression, we rely on closed-cell foams.

Composite Electrical Conductivity

As with all materials to be used with patients in MRI, we are concerned not only with its magnetic susceptibility and effects on the main field, but also its MRI compatibility. In particular, we must avoid using a conductive material, which could burn the patient. Conductive materials support eddy currents which cause heating, which could ultimately lead to patient discomfort or even injury. Also we would not want the PG foam to add a significant amount of noise to the MRI scan. MRI noise scales linearly with the conductivity of the sample and quadratically with the radius (22). Because the foam will be packed around the patient, it is important for the conductivity of the pyrolytic graphite foam to be at least two orders of magnitude below that of human tissue, which is approximately $0.5\text{--}1.0$ S/m (23). Hence the electrical conductivity should ideally be below 5 mS/m so that the noise contribution from the PG foam will be negligible. This was a significant concern given the very high conductivity (in plane) of the individual crystals ($\sigma_{PG} = 1.9 \times 10^6$ S/m) (19). Fortunately, at the 4.5% volume concentration used here for human tissue matching, the PG crystals are largely insulated from each other, rendering the bulk conductivity to be nearly identical to the conductivity of the foam. This is explained below using the “effective medium theories (EMTs)”, which are commonly used to model and predict the electromagnetic properties of two phase conductor/insulator composites.

EMT includes the Maxwell and the symmetric media Bruggeman formulas to predict the conductivity σ_{em} of effective media. These media have a highly conducting component with conductivity σ_h and an insulating component with conductivity σ_l . The simplest effective media equations for dilute solutions of spherical inclusions ($f_h \ll 0.1$), the Maxwell equations (24,25), predict

$$\sigma_{em} = \frac{\sigma_l}{(1-3f_h)}, \quad [6]$$

when $\sigma_h \rightarrow \infty$ and where f_h is the volume fraction of the conducting component. Applied to the PG foam ($f_h \sim 0.045$) Eq. 6 suggests that the conductivity of the PG foam will approach that of the insulating foam.

The EMTs also suggest that the insulating components coat and insulate the conducting inclusions and prevent the formation of conduction paths within the composite material. As the concentration of inclusions increase, they begin to touch and form conduction paths, increasing the conductivity. Because the limiting conditions for the Maxwell equations are not met ($\sigma_h \rightarrow \infty$), we introduce the symmetric Bruggeman equation to evaluate the conductivity of our PG foams for general conditions.

The Bruggeman equation predicts a relationship between σ_{em} and the known quantities,

$$f_l \frac{\sigma_l - \sigma_{em}}{\sigma_l + 2\sigma_{em}} + f_h \frac{\sigma_h - \sigma_{em}}{\sigma_h + 2\sigma_{em}} = 0, \quad [7]$$

for spherical inclusions of high conductivity in a low conductivity matrix (25). Eq. 7 reduces to the Maxwell equations under the previous conditions in the dilute limit, as expected. Under our conditions for pyrolytic graphite foam with volume fraction $f = 0.045$, $\sigma_{PG} = 1.9 \times 10^6$ S/m, and $\sigma_{foam} = 10^{-4}$ S/m (26), solving for σ_{em} in Eq. 7 yields a solution $\sigma_{em} = 1.1 \times 10^{-4}$ S/m. In addition, increasing σ_{foam} up to 1 S/m to simulate far more conductive foams yields solutions $\sigma_{em} \sim 1.1 \sigma_{foam}$. Thus, the solutions of σ_{em} are on the same order of magnitude as σ_{foam} . Consequently, we expect that the composite pyrolytic graphite foam is approximately 3 orders of magnitude lower in conductivity than human tissue.

Eq. 7 holds for solutions below the vicinity of the percolation threshold, the point where the crystals begin to touch and to form conduction paths. However, calculations indicate that the crystals will not form significant conduction paths until the volume fraction of PG crystals is approximately 16% (24), which is approximately 3–4 times higher than the volume fraction required for human tissue susceptibility matching with PG foam. More complicated EMTs, such as the General Bruggeman formula, include parameters to account for percolation (24,25,27), but are not necessary for our desired concentration of particles. This theory would be important, however, if long rods, e.g. carbon nanotubes, were used instead of spherical particles. We believe that roughly spherical crystals are the best option since they minimize the chances of particles forming conduction loops.

Materials and Methods

Pyrolytic Graphite Construction

Pyrolytic graphite (PG) foams were created by dispersing high purity PG powder (Asbury Carbons, Asbury NJ) into a closed-cell, two-component polyurethane polymer foam (Silpak, Inc., Pomona, CA). PG particles were ~44 microns in diameter. PG powder was uniformly dispersed into one component of the foam prior to mixing both foam components. The foam was then poured in plastic containers of various sizes or custom built plastic molds and allowed to rise and cure. We targeted foams with volume fraction $f = 0.045$ to match human tissue susceptibility of ~9 ppm, based on our theoretical calculations (see Theory). Foams of $f = 0$ to 0.05 (0–5%) in 0.005 (0.5%) increments were also made for testing purposes. PG volume fraction was controlled by dispersing a calculated mass of PG based on density calculations relative to final targeted foam volume.

Phantom MRI Field Maps

Test phantoms were built to observe and quantify field gradients near susceptibility interfaces. A proof of concept phantom was constructed with three (1.5 in. diameter) plastic cylinders (United States Plastics Corp., Lima, OH) oriented orthogonal to the main B_0 field and immersed in a sealed, larger cylinder filled with $MnCl_2$ doped water. The water was doped to simulate tissue relaxation times (28). The smaller cylinders were filled with doped water, air, and PG foam, respectively, to create the relevant susceptibility boundaries. Tissue susceptibility matched PG foam ($f = 0.045$) was used in the phantom. Similar plastic cylinders containing other volume fractions of PG foam were also created for testing purposes and individually immersed in doped water.

For the proof of concept phantom, coronal high spatial resolution, 2D multislice gradient echo images were acquired for magnitude and off-resonance field maps on a 3T whole body Siemens Trio (Siemens Medical Solutions, Malvern, PA) scanner equipped with a standard 12 channel head coil. The field of view of the 256×256 magnitude images was 22.4×22.4 cm² in-plane (0.875×0.875 mm² spatial resolution) with 4 mm slice thickness. TR/TE of the gradient echo images was 261.9/5 ms. A series of two similar 3D gradient echo scans with the same FOV were acquired to compute off-resonance field maps. TR of both scans was 488 ms. TE was 4.92 and 7.38 ms for the two scans, respectively. Off resonance field maps were computed using the phase accrual data between the two sets of gradient echo images by dividing phase maps by ΔTE (29) using custom in-house code. This field map reconstruction did not include a phase unwrapping algorithm in order to produce a qualitative view of the extent of the phase wrapping caused by the field inhomogeneities.

A second series of field maps were acquired on a 3T GE Signa scanner (GE Healthcare, United Kingdom) using a custom GE IDEAL (Iterative Decomposition of water/fat using Echo Asymmetry and Least-squares estimation) pulse sequence on the various volume fractions of PG foam. The FOV of the 256×256 field maps was again 22.4×22.4 cm² in-plane (0.875×0.875 mm² spatial resolution) with 4 mm slice thickness. The TR of the IDEAL sequence was set at 6.3 ms, and utilized various TE (base TE = 2 ms). Field maps were auto-computed using the IDEAL sequence data and a built-in iterative reconstruction algorithm. These field maps included a phase unwrapping algorithm in order to produce quantitative view of the field inhomogeneities.

Pyrolytic Graphite Foam Safety and RF Compatibility Testing

Q measurements of the PG foams, with f from 0.5 to 5%, along with regular foam, PG powder, and human heads were measured with an HP 4195A 1–500 MHz spectrum analyzer (Agilent Technologies, Palo Alto, CA) and a tuned 4T MR head coil using a transmission method described previously (30) with $N = 6$ samples each to compare conductivities. Foam samples were approximately 4.5 L in volume to simulate a tissue volume larger than the average human head.

PG foam heating was investigated by placing foams in a 3T Siemens Trio (Siemens Medical Solutions, Malvern, PA) scanner equipped with a standard 12 channel head coil during two SAR intensive pulse sequences, including fast spin echo ($\alpha = 180^\circ$) and bSSFP ($\alpha = 70^\circ$), for 10 minutes. Sequence parameters were chosen to reach the SAR limits of the system for each sequence. Changes in foam temperature were measured with a Luxtron m3300 optical probe (Lumisense Technologies, Santa Clara, CA).

Results

Pyrolytic Graphite Foam and MRI Phantom Images

Figure 2a shows a large PG foam sample (10 in. diameter, 3.5 in. height) of $f = 3\%$. The foam has a final density of 0.2 [g/mL]. Figures 2b and 2c show other foam shapes of similar density with $f = 4.5\%$ created for testing use or for padding purposes. Figure 3 shows the magnitude image (a) and off-resonance field map (b) of the proof of concept phantom. In the magnitude image, there is no signal from the PG foam and air cylinders as expected in the magnitude image. There is also significant geometric distortion in the air cylinder portion of the phantom. In the off-resonance field map, the air/water interface causes the classic dipole field pattern outside the air-filled cylinder. The field maps also show that the PG foam demonstrates susceptibility matching to the doped water. Figure 4 shows individual field maps of (a) water, (b) air, (c) regular foam, (d) $f = 1.5\%$ PG foam, (e) $f = 4\%$ PG foam, and (f) $f = 4.5\%$ PG foam immersed in doped water. Figures 4b and 4c indicate that the regular foam shows a dipole pattern similar in magnitude to the pattern resulting from air indicating equivalent magnetic susceptibility. The poorly matched PG foam ($f = 1.5\%$) in Figure 4d shows a reduced dipole pattern, but still results in substantial field perturbations. The water matched PG foam ($f = 4.5\%$) in Figure 4f demonstrates 5x–10x reduction in off-resonance effects depending on location as compared to the air, suggesting field homogeneity within ± 1 ppm outside the PG foam cylinder as compared to ± 5 ppm near the air cylinder. In addition, the predicted $\sim 10\%$ deviation from the targeted $f = 4.5\%$ volume fraction is apparent in the significantly reduced dipole pattern found in Figure 4e.

Pyrolytic Graphite Foam Conductance

Q values were obtained of the head sized PG foams, PG powder, human heads and air (unloaded) in the 4T tuned MR head coil. A sample 4.5 L PG foam loaded in the coil is shown in Fig. 2b. Figure 5 shows that the regular foam and the PG foams have virtually the same Q and only a negligible effect on the coil Q value compared to the unloaded coil, indicating that the PG foams are non-conductive and add virtually no additional noise to the receiver coil. This is in contrast to the significant Q drop when the coil was loaded with conductive PG powder (not in a foam) or when loaded as normal with human heads ($N = 6$), which were significantly smaller in Q than the foams.

Pyrolytic Graphite Foam Heating

As a second corroborative test of conductance, the direct heating tests of the foam are shown in Figure 6. There were no discernible effects of heating ($|\Delta \text{temp}| < 0.2$ C) over the courses of the heating tests. Figure 2b shows a sample setup of the PG foam in the coil for heating tests.

Discussion

Superconducting magnets can be passively shimmed to have the static B_0 field with 1–2 ppm uniformity over typical regions of interest (22). The remaining static field inhomogeneities are usually comprised of low order spatial variations that can be reduced through the use of similarly low order spherical harmonic active shim coils (31). Thus the main sources of static inhomogeneities in the B_0 field arise from objects with varying magnetic susceptibilities in the body or sample. Consequently, air-tissue and tissue-tissue boundaries in the body are unavoidable and typically uncorrectable sources of field gradients within the main field. These perturbations occur across all susceptibility boundaries (some with diminished magnitude due to gross geometry) and contain higher order spatial variations in the field that cannot be corrected by the lower order active shim coils (6).

Previously, other passive techniques have been introduced to reduce or eliminate the effects of these susceptibility boundaries. These methods include the use of pads of external diamagnetic fluids, such as water or perfluorocarbon (13–15). Water or perfluorocarbon pads can be used to surround the relevant tissue of interest because their magnetic susceptibilities are naturally close to those of human tissues. In effect, this moves the field gradients from the mismatched susceptibility boundaries out of the field of view or away from the region of interest. Perfluorocarbon has no intrinsic water signal and water can be doped with paramagnetic ions to have no MRI signal so as not to saturate the image. The disadvantage of this technique is the incompatibility with embedded RF coils, bulkiness, and lack of patient comfort. Of course, PG foam can only address field perturbations due to external air cavities; it cannot improve the magnetic field disturbance due to sinus cavities or lungs.

Others have used the field gradients arising from oral and aural shims of the strongly diamagnetic pyrolytic crystals in an attempt to directly compensate for inhomogeneities caused by air-tissue interfaces in the sinus and aural cavities (8–10). However, it is difficult to compensate for the dipole-like field effects from the air cavities using the similarly dipole-like effects of the graphite shims at a distance away from the original source. This technique does have the advantage of improving the field in areas where the air cavity cannot be directly accessed or filled, but similar to an active shim, it cannot completely correct the higher order spatial variations. The active current-carrying coil methods have been shown to improve field homogeneity in the human prefrontal cortex (11,12), but has not been tested or modified for use in other body geometries and applications.

The uniformity of the main field is especially important in applications such as fat suppression. In MRI fat appears with a bright signal and can cover up relevant features in the region of interest. Fat suppression techniques in MRI are often particularly relevant around areas with natural susceptibility boundaries, such as the breast, head, back, and other extremities. Traditional fat suppression techniques include spectrally selective RF fat saturation pulses, short T_1 inversion recovery (STIR) imaging, 2 point Dixon, and 3 point Dixon/IDEAL techniques.

Spectrally selective fat saturation methods typically use the 3.5 ppm chemical shift between fat and water to exclusively excite and saturate the fat signal prior to MRI (32), or to excite only the water (spectral spatial). Because the chemical shift between fat and water is small, a uniform static field on the order of 1 ppm is required to correctly perform chemical shift selective RF pulses for fat suppression. Thus, these methods typically fail near areas with strong susceptibility mismatches (33), with either unreliable fat suppression or erroneous suppression of water (tissue) signal.

STIR methods use an inversion pulse to suppress fat by its short T_1 . While fat suppression can be uniform with STIR using a non-frequency selective pulse, any species with T_1 similar to fat, including contrast-enhanced tissue, is also suppressed (34). STIR typically has poor SNR performance (35) due to saturation from the inversion pulse. Frequency selective STIR pulse sequences do not suffer this SNR degradation, but again these methods require pristine 1 ppm field homogeneity.

A class of pulse sequence, often called 3-point Dixon methods, include IDEAL (Iterative Decomposition of water/fat using Echo Asymmetry and Least-squares estimation) methods (36,37), which generally use repetitive phase-shifted acquisitions (scans) and special reconstruction algorithms to account for field inhomogeneities to resolve chemical shift from field inhomogeneities. However, drawbacks to these methods include increased time necessary for acquisition and reconstruction (38). Specialized reconstruction techniques and

software are also often used. Hence, it would be advantageous to improve field homogeneity so that these 3-point Dixon methods are not required

The use of the tissue susceptibility matched PG foam is especially relevant to applications such as breast MRI, for example, in which the air/tissue susceptibility boundary is accessible to the foam and requires an extremely uniform static field for proper fat suppression using the faster spectrally selective techniques. PG foam cushions that would correct for a more uniform field to enable more robust frequency selective fat suppression could be pre-molded in various sizes for patient fit and comfort. Also, in breast MRI, the patient is typically imaged with a surface coil, in which SNR decreases as the distance away from the surface coil increases. Because PG foam is non-conductive, it is also compatible with embedded RF coils allowing maximal SNR from a given coil, something that could be unsafe with perfluorocarbon or doped water matching materials. In addition, in clinical practice, breast MRIs are often performed in conjunction with biopsy procedures. With the use of PG foam, the clinician can directly perform MR guided biopsies through the PG foam and around the surface coil without moving the patient. Also, anatomical features and positions can vary for each patient between scans depending on the placement of the breast. PG foam can be molded to the correct shape for a patient allowing for reliable and repeatable positioning prior to each scan.

PG foam also has applications in the general MR suite. In a typical MRI scan, the patient rests on a foam cushion on the exam bed and foam pieces may be used to secure the patient from moving during the scan. PG foam is lightweight and relatively inexpensive, so it can naturally replace other foams currently used in the MR suite, including the patient table bed. In addition, foam ear plugs are ubiquitous for patient safety inside the magnet bore and might be replaced with PG foam ear plugs, which may reduce susceptibility artifacts near the temporal lobe due to the air in the aural cavities.

Translating the PG foam from phantom applications to *in vivo* imaging raises two issues concerning compressibility and air gaps. First, using the foam as padding can result in compression for certain applications, such as under the head or back or inside the ear cavity. The change in volume will cause a change in density and PG volume fraction, and thus a susceptibility change. Our theory and phantom results show that a 10% error in volume fraction produces susceptibility matching to ~1 ppm, which would suggest that at least 10% compression would be tolerable for human imaging. The density and softness of the underlying foams used for the manufacturing of PG foams can be independently varied, so PG foams of varying compressibility and feel can be tailored towards specific applications, depending on whether pressure padding is required or the level of compression expected. Second, unlike the initial phantom experiments, the foam padding may not conform itself perfectly to the imaged body part, even if the foam is pre-molded, and thus air gaps are possible, which may be a source of dipole-like B_0 inhomogeneities. Solutions for simple ellipsoid patterns, such as spheres or cylinders have been published previously (2) and more complicated patterns can be predicted using a Fourier based method to calculate field perturbations from a heterogeneous magnetic susceptibility distribution (31,39,40). Because PG foam can be compressible and is relatively conforming, air gaps of less than a few millimeters are feasible, and our initial estimates of simple gap geometries suggest that the field homogeneities will be with a tolerable ~1 ppm.

In conclusion, we have described a tissue susceptibility matching pyrolytic graphite foam that will improve the homogeneity of the B_0 static magnetic field within the patient. PG foam reduces local inhomogeneities due to field gradients by moving the susceptibility boundary away from the field of view or region of interest and it produces no MRI signal. It is inexpensive, lightweight, non-conductive, does not heat, and safe to use with patients. It is

also compatible with embedded RF coils. Future work will include developing foams for use in specific applications, such as in breast MRI, cervical spine, other extremity imaging applications, and spectroscopy.

Acknowledgments

Grant Support

CITRIS Seed Grant #42, NIH R01 EB009055

We would like to thank Pamela Tiet and Carlos Ruiz (Berkeley Imaging Science Laboratory) for their assistance in foam production. We would also like to thank the Berkeley Brain Imaging Center and the Radiological Science Laboratory at Stanford for the generous use of facilities and technical support. In addition to CITRIS, we would also like to acknowledge the Berkeley Bioengineering department, the Siebel Scholars Foundation, and the CIRM Tools and Technology program for support.

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Fig. 1.

Concept illustration of magnetic field homogeneity in the breast. Field perturbations are present near the air/tissue interface without PG foam (**top**). By surrounding the skin with tissue susceptibility matched PG foam (**bottom**), the perturbations are shifted further away from the skin and the magnetic field is more uniform inside the breast. This pristine field homogeneity would allow for more efficient fat suppression, which is critical for the early detection of breast tumors.

Fig. 2.

Examples of tissue susceptibility matched PG foam. a) A large PG foam sample (10 in. diameter). b) A 4.5 L sample (8.5 in diameter) PG foam in a Varian 4T head coil. This setup was used for conductivity and heating tests. c) An example of 0.75 in. thick sheets of PG foam that can be used for padding or wrapping patients in an MRI exam.

Fig. 3.

Example of water susceptibility matching using PG foam. a) A 3T coronal image of a PG foam, air, and water phantom acquired with a gradient echo sequence, TR/TE = 200/5 ms, slice thickness = 4 mm, FOV = $22.4 \times 22.4 \text{ cm}^2$, matrix = 256×256 . Significant distortion is seen near the air cylinder. b) B_0 field map of the phantom. The classic $\pm 5\text{ppm}$ dipole effect is seen outside of the air cylinder but is significantly reduced outside the PG foam indicating improved susceptibility matching.

Fig. 4.

B_0 field maps of individual water, air and PG foam phantoms. 3T coronal field maps were acquired with a custom GE IDEAL pulse sequence with a built in iterative reconstruction algorithm, TR= 6.3 with variable TE, slice thickness = 4 mm, FOV = 22.4×22.4 cm², matrix = 256×256. a) Perfect susceptibility matching is shown with water. b)-c) Air and regular foam each shows the classic ± 5 ppm dipole effect indicating equivalent magnetic susceptibility. d) Mismatched $f = 1.5\%$ PG foam shows reduced perturbations, but still a significant susceptibility effect. e) $f = 4.0\%$ PG foam demonstrates improved susceptibility matching with the predicted $\sim 10\%$ tolerance in targeted volume fraction for water matching ($f = 4.5\%$). f) $f = 4.5\%$ PG foam demonstrates good susceptibility matching to water because it reduces off-resonance effects down to ± 1 ppm, relative to the ± 5 ppm off-resonance effects from air at the water interface.

Fig. 5.

Q Measurements on a Varian 4T head coil. Q measurements show the PG foams to be nonconductive, similar to air or regular foams, and thus add no additional noise to the receiver coil. Q for conductive loads of PG powder, and human head are also shown for comparison. Load volumes were approximately 4.5 L.

Fig. 6. Heating tests at 3T. 10 minute temperature changes using (**top**) fast spin echo ($\alpha = 180^\circ$), and (**bottom**) balanced SSFP ($\alpha = 70^\circ$) pulse sequences indicates no heating in the foams. $|\Delta\text{temp}| < 0.2^\circ\text{C}$.