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RF Coil Considerations for Short-T₂ MRI

R. Adam Horch^{1,2}, Ken Wilkens², Daniel F. Gochberg^{2,3}, and Mark D. Does^{1,2,3,4}

¹ Dept. of Biomedical Engineering, Vanderbilt University

² Institute of Imaging Science, Vanderbilt University

³ Radiology and Radiological Sciences, Vanderbilt University

⁴ Electrical Engineering, Vanderbilt University

Abstract

With continuing hardware and pulse sequence advancements, modern MRI is gaining sensitivity to signals from short-T₂ ¹H species under practical experimental conditions. However, conventional MRI coils are typically not designed for this type of application they often contain proton-rich construction materials which may contribute confounding ¹H background signal during short-T₂ measurements. An example of this is shown herein. Separately, a loop-gap style coil was used to compare different coil construction materials and configurations with respect to observed ¹H background signal sizes in a small animal imaging system. Background signal sources were spatially identified and quantified in a number of different coil configurations. It was found that the type and placement of structural coil materials around the loop-gap resonator, as well as the coil's shielding configuration, are critical determinants of the coil's background signal size. Although this study employed a loop-gap resonator design, these findings are directly relevant to standard volume coils commonly used for MRI.

INTRODUCTION

Modern magnetic resonance imaging techniques, such as ultra-short Echo Time (uTE) imaging (1), Sweep Imaging with Fourier Transformation (SWIFT) (2), and Water- and fat-suppressed projection MR imaging (WASPI) (3), allow the use of MRI for studying short-T₂ signals, such as ¹H signal from cartilage and cortical bone. Such imaging is sensitive to numerous background ¹H signals that are commonly overlooked in conventional MRI due to their short T₂ characteristics. For example, the proton-rich engineering plastics (4,5), adhesives, and lubricating oils present in standard RF coils may present a significant background signal when imaging on the timescales necessary for detecting short-T₂ signals. Similarly, other fast-relaxing ¹H sources throughout the magnet bore may present problematic background signals, because their broad lineshapes may be excited by an RF pulse well off-resonance. Therefore, unwanted fast-relaxing background signals may originate from large areas both inside and outside the RF coil. With such a broad spatial distribution, components of the background signal may fold-over into the imaging field of view, confounding the underlying signal of interest. While gradient- and RF-based spatial selection techniques may avoid background signals in some cases, it is often impractical to maintain sensitivity to short-T₂s with these techniques due to gradient strength and RF power deposition limits. Thus, for MRI of short-T₂ signals, it is desirable to minimize or

shield all physical sources of background ^1H rather than relying on pulse sequence methods for filtering out the background NMR signal.

Herein, we identify and characterize background signal sources observable in a standard small bore imaging system and discuss approaches to lessen the contribution to short T_2 imaging. Given the widespread use of ^1H -bearing materials for in-bore MRI hardware, these results will aid investigators in devising effective means for improving data quality by reducing background signal across a broad range of coil designs and short- T_2 pulse sequences.

METHODS

All NMR studies were performed at 200 MHz using a 4.7T 31-cm horizontal bore Varian/Magnex magnet with a DirectDrive console (Varian Inc, Palo Alto, CA). The magnet bore was equipped with two concentrically nested gradient inserts: an outer 210 mm i.d. set, and an inner 120 mm i.d. set. All imaging measurements in this study used the outer gradient set, which allowed an imaging field of view large enough to include the inner gradient set. Constant time imaging (CTI), wherein every point in k-space is acquired at the same time, TE, after excitation (6), was used to acquire both 2D projections and full 3D images. Acquisition parameters included: 500 μs gradient settling time prior to 4 μs duration 20° hard excitation pulses, 15 μs or 30 μs TE (defined between midpoint of the excitation pulse and start of acquisition), a single-point acquisition at 1.25 MHz bandwidth, and 15 ms TR.

In order to demonstrate the background signal effects in a representative short- T_2 imaging scenario, 3D images of a 4.5 cm long segment of human cadaveric femur were acquired using a standard 63 mm i.d. birdcage-style coil (Varian Inc, Palo Alto, CA) with 30 μs TE. A 10 cm $\times$ 10 cm $\times$ 15.2 cm field-of-view (FOV) (large enough to include the entire RF coil, Fig 1) was encoded with 50 $\times$ 50 $\times$ 76 samples in k-space. Data were zero-padded by 2 $\times$ prior to reconstruction, resulting in a nominal 1 mm isotropic resolution. In order to mimic a typical imaging situation where the FOV is only large enough to encompass the sample, a second 3D image was reconstructed after the k-space data were sub-sampled by a factor of 2 $\times$.

To characterize potential short- T_2 signal contributions from different RF coil and/or sample holder materials, 2D and 3D images and non-localized free induction decays (FIDs) were collected using each of three variations of a 20 mm-diameter series-tuned loop-gap resonator (hereafter, Coils A, B, and C), built in-house (Fig 2). In all coils, high-purity copper ribbon was formed into the loop-gap shape; all electrical connections were made with a low-flux solder and cleaned with methanol; variable PTFE capacitors (Polyflon, Norwalk, CT) were used for tuning, matching, and balancing and B-type or C-type chip capacitors (American Technical Ceramics, Huntington Station, NY) were used for the main tank capacitance. Coil material selections, summarized in Table 1, were as follows: Coil A was built using polycarbonate pieces for the platform and coil support. Flexible polyethylene-based BNC cable was used as a transmission line, and the gap support was formed from copper-clad G-10/FR4 Garolite (McMaster-Carr, Atlanta, GA), an epoxy-bonded fiberglass. Polycarbonate and fiberglass are commonly used as structural materials for both small animal holders and volume coil forms, and, being relatively hard materials, might be assumed to have negligible background signal beyond a few microseconds. In Coil B, the polycarbonate and fiberglass materials were replaced with virgin PTFE (McMaster-Carr, Atlanta, GA) (i.e., “Teflon”, generally considered to be proton-free), and the common BNC cable was replaced with a PTFE-dielectric semi-rigid transmission line. In Coil C, the internal loop-gap support was removed and replaced with external support rings at the top and bottom of the coil. For all coils, a copper-clad mylar sheet with a 0.25 mm-thick copper

layer was rolled, copper layer on the inside, around the entire coil assembly to form a full-coverage RF shield.

Transverse relaxation rates (R_2^*) of materials were measured using a pair of 2D CTI projections with TEs of 15 and 30 μ s. The FOV and number of samples varied between variations of the RF coil, but in each case the nominal in-plane resolution after $2\times$ zero-padding was $2\text{ mm} \times 2\text{ mm}$. A coronal projection (parallel to long-axis of the loop gap, Fig 2) allowed unambiguous identification of different materials because there was no overlapping material in that direction. Three dimensional images were also acquired, with 15 μ s TE, 2 mm isotropic resolution and other parameters as defined above. For both the 2D and 3D images, the RF power was calibrated with a small water sample (0.1 mL, 6 mM CuSO_4), and then images were acquired from the coil in the absence of the water sample in order to best detect the background signals. To account for the varying background signal amplitude between coils, between 4 (2D, Coil A) and 1150 (3D Coil C) excitations (NEX) were averaged. Finally, for each coil, with and without the 0.1 mL water sample, FIDs were collected (8 μ s 90° hard pulse, 4 μ s receiver dead-time, 5 MHz bandwidth, 15 s TR, 16-64 NEX) and linearly extrapolated to $t = 0$ (midpoint of excitation pulse) to provide a quantitative measure of the total background signal in units of water volume. Neglecting variation in T_1 -weighting between materials, the measured R_2^* s and total signal amplitudes were used to convert the 3D images into units of apparent proton density. This was an apparent, not absolute, measure because the 3D image volume encompassed a wide variation of RF sensitivity and the effect of transverse relaxation during RF excitation was not considered.

RESULTS

Isosurface renderings of 3D images from the human femur segment in the 63 mm volume coil are shown in Figure 1a–c. In addition to the femur (green), signals from several coil elements such as plastic substrates of the birdcage rungs and BNC cables are clearly resolved (red) because of the large field of view. The total integrated signal from the bone and background were equivalent to 6.8 mL and 22.5 mL of water, respectively.

Rather than deconstruct the 63 mm volume coil, which was in routine use in the authors' lab, background signal characterization was conducted in three in-house loop-gap coils fabricated from known materials. The background signal amplitudes, expressed both as water equivalent volume and apparent proton density relative to water, as well as R_2^* values from various material in Coils A, B, and C are reported in Table 2. The total background signal in coil A was equivalent to $\approx 6.4\text{ mL H}_2\text{O}$, compared with the $\approx 4\text{ mL}$ uniform RF field volume. Each of the coil construction materials was clearly visible in CTI images (Fig. 3a), with the dominant source of signal arising from polycarbonate plastic inside the coil's loop. Coil B was built by replacing coil A's proton-bearing materials with low-proton PTFE, resulting in a much reduced background signal equivalent to $\approx 7.2\text{ }\mu\text{L H}_2\text{O}$, which originated predominantly from PTFE in the loop and gap (Fig. 3b). Coil C was built with all PTFE materials placed external to the loop and gap, further reducing the background signal to $\approx 2.1\text{ }\mu\text{L H}_2\text{O}$, which arose predominantly from variable PTFE and fixed chip capacitors (Fig. 3c). The localization of signals at the top and bottom of Coil C's loop and gap (Fig. 3c, sagittal view) likely corresponded to the PTFE external support frame. It should be noted that the C-type chip capacitors in Coil C contributed a larger apparent signal than the physically smaller B-type capacitors in Coil B.

RF shielding was also investigated as a means to reduce stray background signals originating from outside of the coil. Bench testing of the unshielded coil B with a network analyzer (relevant to any of the loop-gap coils studied) showed that a 90° RF pulse resulted

in $\approx 2^\circ$ RF at the inner gradient bore, which can potentially excite proton-bearing materials in the gradient insert. This effect is seen in practice (Fig. 4), where CTI projections from the PTFE-based coil without the RF shield in place reveal a spatially broad signal originating from the gradient insert. This signal amplitude was equivalent to approximately 50 μL of H_2O (Table 2), was only partially removed by a slotted RF shield (as is often used to avoid adding material area in which eddy currents can be generated), but was effectively entirely removed by the full coverage RF shield.

DISCUSSION

Short- T_2 imaging of a human femur with a conventional volume coil indicates that, even at $\text{TE} = 30 \mu\text{s}$, the coil background signal cannot be ignored. To demonstrate the potential effect of ignoring the background signal, compare Fig 1d and 1e. Fig 1d shows a single slice through the 3D volume, presenting a cross-sectional view of the femur and cropped to remove the background signals. This is representative of a bone image in the ideal case where no background signal exists. In contrast, Fig 1e shows the corresponding image that resulted when the acquisition FOV was reduced by down-sampling k-space. In this case the FOV was not large enough to encompass the entire RF coil and the signal from its proton-bearing material is aliased back into the image. Note that the nature of such aliasing depends on the k-space sampling scheme and will generally create incoherent artifacts for the non-cartesian sampling (7) common to many short- T_2 imaging methods.

Coil A, which was built from commonly used construction and sample holder materials to emulate the conventional volume coil, exhibits a prohibitively large background signal equivalent to 60% more water than can be contained in the coil's RF-homogenous volume. In Coil B, a simple substitution of the coil materials with low-proton PTFE yields a thousand-fold drop in the net background signal. The remainder of Coil B's signal, dominated by PTFE materials inside the loop and gap (Fig. 3b), is consistent with a previous study showing that PTFE contains a trace proton NMR signal (8). Coil C minimizes this PTFE signal by moving all PTFE supports to external locations around the resonator, resulting in a final background signal equivalent to $\approx 2.1 \mu\text{L}$ H_2O , which is approximately $1/3000^{\text{th}}$ the size of the coil's RF-homogenous volume.

To put the size and T_2^* values of these background signals into context, consider that uTE signals are derived from a combination of solid/macromolecular protons and the protons from water bound to these macromolecules. Quantitative magnetization transfer studies and compositional analysis in numerous tissues point to solid proton $T_2^* \approx 10 \mu\text{s}$ (9) and bound water T_2 or $T_2^* \approx 50\text{--}500 \mu\text{s}$ (10–12), with concentrations relative to bulk water $\approx 0.05\text{--}0.5$ (9,13). Thus, the fiberglass, gradient insert, and BNC cable signals pose the greatest concern for uTE images with relatively long echo times ($\text{TE} \geq 80 \mu\text{s}$), which tend to image bound water signals. When imaging with much shorter echo times (14,15), or with SWIFT ($\text{TE} \approx 0$) (2), any of the background signals are long enough lived to contribute to the image, but the polycarbonate, fiberglass, and to a lesser extent, the BNC cable are the most problematic due to their relatively large apparent proton densities. PTFE components and C-type chip capacitors may present significant signal when looking at solid signals from small samples. For example, the solid proton signal ($T_2^* \approx 10 \mu\text{s}$) from a 0.25 mL sample of human cortical bone has a signal equivalent to $\approx 40 \mu\text{L}$ H_2O (Table 2), as estimated from a previous study (16), which would be significantly corrupted by the apparent $\approx 6 \mu\text{L}$ signal from the PTFE loop support in Coil B.

In general, the relevant threshold for an acceptable background signal when studying short T_2 signals primarily depends on the sample size and RF coil geometry. It is expected that a linear increase in RF coil dimensions (diameter and length) results in a correspondingly

linear increase in the amount of coil construction materials and, therefore, a linear increase in the short- T_2 background signal. However, this linear increase in coil size corresponds to an approximate cubic increase in uniform RF volume. For example, in comparing the signals of the 63-mm coil and Coil A we see $\approx 3\times$ difference in linear dimensions, $\approx 3.5\times$ difference in background signal amplitudes, and $\approx 24\times$ difference in uniform RF volume. Hence, the background signal problem is greatest for small samples and coils. For moderate to large RF coils (> 6 cm diameter) and samples that are close to filling the uniform RF volume, it may be sufficient to avoid proton-rich materials within the RF coil and along the entire in-bore electrical signal pathway. However, for small samples and coils, it will also be necessary to minimize the PTFE near the RF coil volume, use high-grade low-proton capacitors, and robustly shield the coil from external NMR signals. Additionally, the proton content of any non-PTFE coil construction materials inside the shielded region should be carefully considered. For commonly-used materials not employed in this study, the reader is directed to previous studies that have identified short-lived signals from plastics such as polypropylene (4) and acrylic (17), as well as longer-lived ($T_2 \approx 2$ ms) signals from acrylonitrile-butadiene-styrene (ABS) and commercial coil casing plastics (18). Furthermore, foams and tapes used for sample placement and positioning cannot be neglected when assessing net background signal.

In addition to coil materials, RF shielding was found to be important for reducing background signal. The 120 mm i.d. gradient insert, like all common magnetic field gradient constructions, contains numerous proton-bearing materials such as fiberglass, epoxy, and water coolant. Although well outside the uniform region of the RF coil, these materials were found to contribute significant signal in the absence of the full coverage RF shield, increasing the background signal of Coil B nearly 8-fold. A full-coverage RF shield was necessary to block signals from the gradient insert, as any gaps in the shield conductor permitted extraneous NMR signals.

Finally, it is worth noting that in Coil C and using the full RF shield, a signal persisted throughout the center of coil's apparently vacant loop (Fig 3c). This signal amplitude was consistent with water vapor at 66% relative humidity (100% humid air at 20°C is ≈ 1.2 kg/ m^3 , holds $\approx 1.5\%$ w/w water (19), and thus contains $\approx 1.8 \times 10^{-5}$ g $\text{H}_2\text{O}/\text{mL}$), approximately the expected lab humidity, and was removed by purging the magnet bore with dry nitrogen gas. This implies that the humidity of laboratory air may set the ultimate lower-limit for background signals from freestanding, unenclosed coil designs.

CONCLUSIONS

In a series of RF coil constructions for short- T_2 MRI, it was found that material selection, placement, and shielding were important design parameters for mitigating total coil background signal. For minimal background signal, we recommend that the coil designer 1) use only low proton density materials such as PTFE, and in minimal possible quantity, in the immediate vicinity of the resonator and in-bore RF path, 2) likewise, utilize only low-proton RF electrical components such as PTFE-based BNC cables and small ceramic chip capacitors throughout the in-bore RF path, and 3) encase the resonator with an inward-facing full-coverage RF shield to attenuate stray NMR signals from outside the coil. Since the coil background signal generally becomes less significant as NMR sample size is increased, the first recommendation may be sufficient when using large coils and samples (> 6 cm diameter), but for study in small animals and tissue specimens, signals from BNC cables and the gradient insert may also be significant.

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Figure 1.

3D CTI of a human femur segment in a conventional volume imaging coil at 30 μ s TE. Isosurface renderings of coil components (red) surrounding the femur segment (green) are shown in sagittal (A), oblique (B), and axial (C) views. Axes in all images are expressed as distances from the coil isocenter. Background signal in the shape of the coil rung pattern is clearly seen, which presumably originates from the coil substrate plastics. BNC cables are also resolved. An axial slice from the 3D data is shown (D) in which the FOV has been cropped in image space, simulating a femur image over a conventional FOV in the absence of coil background signal. Coil background signal is introduced into the conventional FOV (E) by subsampling the 3D CTI k-space prior to image reconstruction, simulating a femur image analogous to D but in the presence of coil background signal. Via fold-over artifacts, the coil background signal severely degrades the femur region of interest.

Figure 2. Loop-gap resonator schematic. Key resonator elements and dimensions are shown in a perspective view. With its 20 mm diameter and 40 mm height, the resonator can hold samples of up to ≈ 4 mL in its homogenous region.

Figure 3.

Maximum intensity projections (MIPs) from 3D CTI of different loop-gap coil configurations. Coronal (left) and sagittal (right) MIPs are shown for the polycarbonate-based coil (A), PTFE-based coil (B), and copper-in-air coil (C). For comparison of all coils, color scales are expressed in units of spin density, normalized to the spin density of water ($\rho/\rho_{\text{H}_2\text{O}}$, see text for details). Axes in all images are expressed as distances from the coil isocenter. Images are shown in the same orientation as in Figure 2. Materials such as capacitors and BNC cable are labeled for reference. Note that the small region of signal to the right of the loop in C is capacitor-related signal wrapped around from the far left side of the FOV. Also, the spatial banding artifacts originating from the center of the FOVs in B and C were attributed to coil ringing and were excluded from signal quantification.

Figure 4.

Effects of RF coil shielding on background signal. 2D CTI projections of the PTFE-based coil are shown in the presence (A) and absence (B) of a full-coverage RF shield. Since these images were formed from 2D CTI projections, image intensity cannot be readily converted to spin density as in Figure 3 because of a lack of spatial depth information. As such, the images are shown on the same, normalized color scale. Nonetheless, it is clear that the RF shield eliminates a significant amount (see Table 2) of unwanted signal from outside the coil.

Table 1

Material selections in the three loop-gap coil variants.

	Coil A	Coil B	Coil C
Transmission Line	Polyethylene-dielectric BNC	PTFE-dielectric semirigid line	PTFE-dielectric semirigid line
Loop Support	Polycarbonate	PTFE	Air / external PTFE frame
Gap Support	Epoxy-bonded fiberglass	PTFE	Air / external PTFE frame
Coil Platform	Polycarbonate	PTFE	PTFE
Main Tank Capacitance	B-type chip caps	B-type chip caps	C-type chip caps

Table 2

Sources of background signal in different loop-gap configurations. Various coil materials observed with CTI are reported with their observed T_2^* , apparent ^1H spin densities relative to the spin density of water, and total ^1H signal sizes. Observed signal sizes and spin densities depend on the excitation flip angle and will vary according to the coil's spatial sensitivity. As such, spin density of the gradient insert material was not determined (see Fig. 4).

Proton Source	T_2^* (μs)	Apparent Spin Density/ H_2O	Signal Size ($\mu\text{L H}_2\text{O}$)
Polycarbonate Support ¹	7.5	0.85	6.1×10^3
Fiberglass Gap Support ¹	110	0.017	105
Gradient Insert ³	120		47
Polyethylene BNC Cable ¹	40	0.02	18
PTFE Dielectric/Support ²	20	3.4×10^{-4}	5.9
C-type Chip Capacitors ⁴	7.1	5.6×10^{-4}	1.0
PTFE Capacitors ⁴	15	4.0×10^{-4}	0.6
Laboratory Air ⁴	100	1.2×10^{-5}	0.1

¹ From shielded polycarbonate-based coil with total signal $\approx 6.4 \times 10^3 \mu\text{L H}_2\text{O}$

² From shielded PTFE-based coil with total signal $\approx 7.2 \mu\text{L H}_2\text{O}$

³ From unshielded PTFE-based coil with total signal $\approx 55 \mu\text{L H}_2\text{O}$

⁴ From shielded copper-in-air coil with total signal $\approx 2.1 \mu\text{L H}_2\text{O}$