MRI endoscopy using intrinsically localized probes

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(Received 9 September 2008; revised 20 December 2008; accepted for publication 30 December 2008;
published 23 February 2009)

Magnetic resonance imaging (MRI) is traditionally performed with fixed externally applied gradient magnetic fields and is hence intrinsically locked to the laboratory frame of reference (FoR). Here a method for high-resolution MRI that employs active, catheter-based, tiny internal probes that utilize the spatial properties of the probe itself for localization is proposed and demonstrated at 3 T. Because these properties are intrinsic to the probe, they move with it, transforming MRI from the laboratory FoR to the FoR of the device itself, analogous to an endoscope. The “MRI endoscope” can utilize loop coils and loopless antennas with modified sensitivity, in combination with adiabatic excitation by the device itself, to restrict the MRI sensitivity to a disk-shaped plane a few mm thick. Excitation with the MRI endoscope limits the eddy currents induced in the sample to an excited volume whose size is orders of magnitude below that excited by a conventional body MRI coil. Heat testing shows maximum local temperature increases of $<1^\circ\text{C}$ during MRI, within regulatory guidelines. The method is demonstrated in a kiwifruit, in intact porcine and rabbit aortas, and in an atherosclerotic human iliac artery specimen, with in-plane resolution as small as $80\text{ }\mu\text{m}$ and 1.5–5 mm slice thickness. © 2009 American Association of Physicists in Medicine.
[DOI: [10.1118/1.3077125](https://doi.org/10.1118/1.3077125)]

Key words: interventional MRI, MRI coils, intravascular MRI, safety, atherosclerosis, endoscopy

I. INTRODUCTION

Conventional optical endoscopy permits routine identification and minimally invasive intervention for suspect lesions in the gastrointestinal tract, the bladder, and other body cavities, as well as the guidance of laparoscopic procedures.¹ It is responsible for identifying hundreds of thousands of new cancer cases in the world annually.² Endoscopy performed with alternative imaging modalities, such as optical coherence tomography (OCT)³ and intravascular ultrasound (IVUS),⁴ has extended access to the vascular system and vascular disease, which accounts for nearly one-third of all deaths globally.⁵ Of interest is the diagnosis of atherosclerotic plaques with high mobile lipid content and thin fibrous caps that are likely to rupture and cause thrombosis, heart attack, or stroke. Conventional x-ray catheterization procedures visualize the lumen only and therefore cannot provide an assessment of plaque vulnerability to rupture. On the other hand, optical penetration limits the depth that OCT can see into the vessel wall,⁶ and IVUS signals are diffracted by calcification that is commonly associated with the disease process. Magnetic resonance imaging (MRI) employing tiny intravascular detectors^{7,8} might offer performance advantages for plaque imaging compared to IVUS,⁹ and moreover, would provide access to the entire range of high-contrast, multifunctional, high-speed, and three-dimensional (3D) visualization capabilities afforded by MRI in the same exam. Yet, as a recent editorial notes,¹⁰ for MRI to be competitive with x-ray guided coronary procedures, better spatial resolution and efficiency or speed is essential.

One key difference between MRI and existing IVUS, OCT, and optical endoscopy is that the latter provide internal

high-resolution examination directly from the viewpoint of the probe. Intravascular MRI (IVMRI) cannot at present do this *directly* because localization depends entirely on the fixed external localizing gradient coils which are intrinsically locked to the laboratory frame of reference (FoR). Thus, with IVMRI, a probe is inserted into the body, and a sequence is applied either without slice selection or with excitation of a thick slab to obtain a projection image that includes the probe. The location of the probe is determined in the projection plane, for example, via its signal intensity, and then in an orthogonal plane by repeat MRI. Only then can high-resolution MRI be performed with the probe at the center of the MRI field of view (FOV).^{7,8} This is a less efficient use of scan time that could otherwise be used for signal averaging to increase the MRI signal-to-noise ratio (SNR) or to improve spatial resolution. Alternatively, the probe may be equipped with localizing radio frequency (RF) coils that generate intense local signals. The position of these coils is interrogated in 3D with a MRI pulse sequence that provides a projection in each of the x -, y -, and z -gradient directions,¹¹ for example. From the 3D location, image parameters that define a new scan plane on the probe are computed and fed to the scanner for MRI. The efficiency of this approach too, however, is suboptimal, requiring a delay while the probe is localized, the scan plane computed, and gradients prescribed, as opposed to acquiring signals that can maximize SNR and/or minimize scan time.

Here we introduce a novel method for performing “MRI endoscopy” which employs active internal probes that can potentially provide both real-time MRI from the probe’s viewpoint as well as high-resolution imaging of vessel walls.

FIG. 1. The concept of the MRI endoscope. A sensitive disk is intrinsically localized to a plane at the end of a guidewire, catheter, or other interventional device.

The method utilizes probes designed with MRI sensitivity that is substantially restricted to a disk-shaped volume on the probe itself (Fig. 1). Because the sensitivity is intrinsically locked to the probe, it moves with it, transforming the MRI from the laboratory FoR to the FoR of the device itself, creating a MRI endoscope or “MR-eye.” The localization properties of the modified probes are enhanced by using them for both MRI excitation with adiabatic pulses¹² and for MRI reception. The resulting one-dimensionally (1D) localized volume on the MRI endoscope could be cylindrically encoded in the other two dimensions using the phase shift between the antenna and the scanner’s main volume coil for the azimuthal direction, and radially, using the probe’s intrinsically radial RF sensitivity gradient, such as proposed elsewhere.¹³ However, we adopt here the simpler approach of using conventional MRI gradient localization in the other two dimensions. The image can be transformed to the center of the FOV based on a simple maximum-signal algorithm. Because by transmitting with the endoscope the eddy currents induced in the sample are limited to a tiny excited volume as compared to those excited by a conventional body MRI coil, local RF power deposition is minimal. We implement and demonstrate these techniques, for the first time, on biological specimens at 3 T, which offers better SNR and spatial resolution for internal MRI, as compared with 1.5 T.¹⁴

II. METHODS

Central to the concept of true MRI endoscopy is the development of a capability that intrinsically translates MRI from the FoR of the scanner to that of the device itself. Here we develop active small-diameter internal MRI probes whose MRI sensitivity is restricted to a narrow region of the probe by virtue of the local RF magnetic field profile and/or local static or gradient magnetic field mechanisms that are provided on the probe. The goal is to produce a “slice”-shaped sensitive region perpendicular to the long axis of a device that can be made the size of a guidewire or an intravascular catheter (Fig. 1). However, because the mechanisms for limiting sensitivity are all confined to the body of the probe and all involve magnetic field profiling of some sort, the thickness of the slice localized by these means may not be strictly constant but can broaden with radial distance from the wire, depending on the design, for example, as field lines flatten with distance from the device. Such broadening, while

FIG. 2. Schematic of 3 T prototypes of (a) a loop coil endoscope, and (b) a loopless antenna endoscope, connected to their matching circuits (left). The antenna cable length is $\lambda/4$ (depicted), or an odd multiple thereof. The switchable diode D is engaged during conventional MRI or during MRI where the probes are used as receivers only. Capacitor C tunes the loop coil to resonance in the medium. A cable trap balun B is connected to the cable shield. The loopless antenna whip is adjusted to about half its resonant length while maintaining a significant portion (80%) of the maximum SNR. Prototype loop (c) and loopless (d) MRI endoscopes built from flexible nitinol cable are pictured below.

a detriment to resolution, can partially offset the radial decline in MRI sensitivity by providing an increase in signal contributions from the larger volume.

II.A. Localized RF excitation and detection

The MRI sensitivity of active internal MRI devices depends on (i) the underlying MRI coil or antenna design; (ii) the presence of field-profiling elements; and (iii) the transmitter field, each of which can be used alone or in combination with other mechanisms to affect the desired localization. Here we evaluate RF probes based on tiny loop coils^{15,16} and loopless antenna⁸ geometries, both of which have previously been implemented in intravascular MRI receivers, modified, as illustrated in Fig. 2.

The MRI sensitivity (or transverse magnetic field, B_1) is computed using a full-wave electromagnetic method-of-moment (MoM) analysis (FEKO, EM Software and Systems, South Africa) for the solenoidal loop coil with axis perpendicular to the connecting lead and main field B_0 depicted in Fig. 2(a), and for the modified loopless antenna of Fig. 2(b), but without the balun. The probes are immersed in a medium with conductivity of $\sigma=0.65$ S/m, dielectric constant of $\epsilon_r=80$, and density of $\rho=1000$ kg/m³ which is comparable to that of skeletal muscle or 0.35% saline.^{14,17} The loop coil is modeled as a four-turn 3 mm diameter helix constructed from insulated 0.3 mm diameter copper wire resonated with a 40 pF capacitor, presenting zero reactance to a voltage source placed in series with the helix at 128 MHz (3 T). The computed loaded coil quality factor (Q) is $Q_L=23$. The loopless antenna is modeled as a 400 mm, 2.2 mm outer diameter coaxial cable with the inner 0.51 mm diameter conductor extended to form a whip. In this case, the voltage source is placed at the cable-whip junction. For endoscopy, the whip length is shortened from the 42 mm required to satisfy the conventional resonant length criterion¹⁴ to 23 mm to limit the device’s sensitivity along its long axis. As shown in Fig. 3, the sensitivity of both antennas is maximum at the con-

FIG. 3. [(a)–(c)] Transverse receiver sensitivity of a 3 mm diameter 3 T receiver coil in the coronal (a) and [(b) and (c)] axial planes plotted from radius $r \geq 3$ mm. [(d)–(f)] Sensitivity of a loopless antenna with 23 mm whip in the coronal (d) and [(e) and (f)] axial planes. Plots (c) and (f) are experimental data for the same devices in 0.35% saline for comparison with (b) and (e). The scale has been normalized to a 100 at the edge of the detector at $r=3$ mm.

ductor and is symmetric about the antenna's principle axis: that is, the y -axis of the loop detector and the z -axis of the loopless antenna. The maximum signal of the loopless antenna occurs at the cable-whip junction. The plots show that the sensitivity of the loop is confined to a smaller volume than the loopless antenna.

The SNR along the axis of the loop coil and at the junction of the loopless antenna is numerically computed by the MoM as¹⁸

$$\Psi = \frac{\sqrt{2}\omega M_0 |B^+|}{\sqrt{4k_B T R_{\text{load}}}},$$

where ω is the angular frequency, $M_0 = 9.33 \times 10^{-9}$ J/ml is the equilibrium nuclear magnetization per unit volume for water at 3 T, $|B^+|$ is the magnitude of the circularly polarized transverse magnetic field generated by the device with unit current, k_B is the Boltzmann's constant $= 1.38 \times 10^{-23}$ m² kg s⁻² K⁻¹, T is the temperature at the detector assumed to be 300 K, and R_{load} is the real part of the impedance seen at the voltage source. For the particular coil/antenna placements used in Fig. 3, $B^+ = B_y / \sqrt{2}$ on the axis of the loop coil and $B^+ = B_x / \sqrt{2}$ on the junction of the loopless antenna, where B_x and B_y are the transverse B_1 components. Figure 4 (broken lines) predicts that the computed SNR of the loop is nearly fivefold higher 3 mm from its center than at 3 mm from the junction of the loopless antenna. Differences in the radial dependence of the computed SNR of the two geometries result in the SNR of the shortened (23 mm) loopless antenna exceeding that of the loop at $r \geq 12.5$ mm,

which is comparable to $r \geq 11$ mm for the conventional loopless antenna with a 42 mm resonant whip. For $r < 12.5$ mm, the SNR of the loop outperforms the loopless antenna. The SNR cost of reducing the whip length of the

FIG. 4. SNR performance of the loop probe and the loopless antenna as a function of distance r from antennas placed in 0.35% saline measured on the axis of the loop [computed, dashed line and measured, solid line (a)]. Computed SNR of resonant loopless antenna with 42 mm whip (dotted line), 23 mm whip loopless antenna (dashed with circles), and measured SNR for the 23 mm loopless antenna with balun [solid line (b)]. The inset shows a portion zoomed for clarity.

FIG. 5. Cross-sectional MRI at the junction of the 23 mm (balunless) loopless antenna used for both excitation and detection with (a) a conventional 1.3 ms, $B_1=20\ \mu\text{T}$ pulse, and (b) a 4 ms, 15 kHz-sweep adiabatic 90° BIR-4 pulse.

loopless antenna from 42 to 23 mm is approximately 20%.

Using the antenna itself for excitation further confines sensitivity, but modulates the resultant images by signal voids that occur where the transverse excitation field strength, B_1 , corresponds to an integral multiple of 180° MRI pulses as shown in Fig. 5(a). In fact, such modulation can occur whenever tiny local coils are used for excitation due to their intrinsically highly nonuniform fields, from which we are attempting to derive benefit.

The undesirable side effect of localized excitation is overcome by use of either small flip-angle (FA) pulses ($\text{FA} \leq 180^\circ$ everywhere), or adiabatic excitation pulses such as adiabatic half passage (AHP) pulses, which provide 90° excitation, or BIR-4 pulses which provide excitation with $\text{FA} \leq 90^\circ$ that can be arbitrarily prescribed.^{19,20} Adiabatic pulses are characterized by excitation profiles that have a constant FA independent of B_1 above a threshold intensity, and a rapidly declining FA when B_1 falls below that threshold. They are therefore ideally suited to the present application. Figure 6 plots the FA of a 90° BIR-4 pulse as a function of B_1 , computed numerically from the Bloch equations excluding T_1 and T_2 relaxation effects. This BIR-4 pulse delivers an essentially constant FA for $B_1 \geq 10\ \mu\text{T}$.

When the loop and loopless probes of Fig. 2 are deployed for both reception and transmission with the adiabatic pulse in Fig. 6, the signal voids are eliminated and a relatively uniform disk-shaped volume is excited, as demonstrated by MRI at the whip junction of the loopless antenna in Fig. 5(b). The uniformly excited volume is at $r \leq r_t$, wherein the threshold radius r_t is determined by the contour of the threshold B_1 required for adiabatic excitation. The precipitous drop in FA as B_1 falls below the adiabatic threshold (Fig. 6) attenuates the excitation field in outlying regions, further con-

FIG. 6. Flip angle as a function of RF field amplitude B_1 , as computed numerically from the Bloch equations excluding relaxation effects for a 4 ms BIR-4 pulse with a 15 kHz frequency sweep. The BIR-4 pulses deliver an essentially constant flip angle for $B_1 > 10\ \mu\text{T}$.

fining sensitivity. The slice thickness w can be defined as the full width half maximum (fwhm) of the integrated signal magnitude along the z -axis at each point in the sensitive plane, with the caveat that this measure excludes the effects of any B_0 inhomogeneity. In practice, phase variations in the signal due to local B_0 inhomogeneity will reduce the signal integral along the z -axis, thereby ameliorating anticipated increases in w with r such that w may overestimate the slice thickness, particularly for large r .

The computed variation in w with r in Fig. 7 shows that w increases with r from 2 to 4 mm for the loop endoscopic probe, albeit over a volume in which sensitivity varies by 33-fold (down to 3% of the maximum sensitivity at the center). Thus, the net effect of the spatial selectivity afforded by

FIG. 7. The fwhm width w of the sensitive volume on the z -axis as a function of the radial distance r , 3 mm from the loop center (computed, dotted line; measured, solid line); 3 mm from the junction of the 42 mm whip loopless antenna (computed, dashed line with solid circles); 3 mm from the junction of the 23 mm whip loopless antenna (computed, empty circles); and 3 mm from the junction of the rigid cable 23 mm whip loopless antenna with balun (measured, dashed line). All probes are deployed for both detection and 90° adiabatic excitation. The plots are obtained by integrating the sensitivity along the z -axis and by determining the thickness of the disk that contributes 50% of the signal. The slice thickness w is annotated at the radii where the sensitivity has fallen to 3% of the maximum, which occurs at $r = 7\ \text{mm}$ for the loop design and 18 mm for the loopless design. The detectors are sketched at left.

FIG. 8. The 3D shape of intrinsically localized sensitive volumes produced by (a) the 3 mm loop endoscope and (b) the 23 mm whip loopless antenna with balun. The slice thickness of the 3D volume is given by the fwhm integrated signal magnitude (see text). The disk is extended in the xy -plane out to the 3% of maximum sensitivity contour. The disk surface is color-coded commensurate with the sensitivity decrease with radius r (maximum=white, minimum=black; see Fig. 4 for SNR values). All dimensions are in mm.

highly localized adiabatic transmission and reception by the loop is a discoidal sensitive volume, illustrated by the 3D plot in Fig. 8(a).

On the other hand, Fig. 7 also shows that the computed slice thickness for the resonant 42 mm whip loopless antenna only reduces from about 52 mm to $w=35$ to 38 mm with the shorter, 23 mm whip antenna. Thus, while the discoidal volume excited by the loopless antenna better mimics a slice with w remaining fairly constant with r , its response is much broader than for the loop even after the $\sim 35\%$ reduction achieved by shortening the whip.

II.B. Magnetic field profiling

The broader computed response afforded by the shortened loopless antenna (Fig. 7) suggests that the combined effects of localized excitation, adiabatic pulses, shortened whip length, and localized detection are insufficient to provide adequate spatial resolution along the device's axis for MRI endoscopy. Yet a loopless antenna comprised of a single imaging wire has an intrinsic advantage of being able to provide the minimum possible cross-sectional area for accessing tiny blood vessels in potential applications to arterial wall and plaque imaging. Can additional localization mechanisms be deployed to sharpen the resolution of the loopless antenna's sensitive volume?

The disposition of ferromagnetic and/or paramagnetic materials applied to the antenna shaft offers one potential means of restricting the sensitivity by means of a local static magnetic field. A gap is left in the coating or layer where MRI sensitivity is maximum at the "sensitive disk," while MRI signals adjacent to the coating are dephased and provide only attenuated contributions to the observed signal.

Another effective alternative employs RF magnetic field profiling in the form of a "sleeve" or "bazooka" balun (or "quarter-wave choke"), applied alone to the loopless antenna's shaft to modify its MRI sensitivity. As depicted in Fig. 2(b), the balun consists of a $\lambda/4$ conductive sleeve, where λ is the electromagnetic (EM) wavelength of the sleeve with the device placed in the medium. The sleeve is shorted at the

FIG. 9. Fractional SNR loss with loop (computed using MoM analysis, dashed line; measured, filled symbols) and the loopless antenna with balun (measured, diamonds) endoscopes rotated by θ relative to B_0 (inset).

proximal end to the antenna body²¹ to create a high impedance at the distal (whip) end. The balun can also be coated with ferromagnetic/paramagnetic particles to enhance localization. Surface currents on the antenna shield, which are the source of its extended sensitivity, are "choked" by the balun, reducing the sensitive length. A 3D plot representing the sensitive disk for this balun-type loopless antenna endoscope, including the effects of adiabatic excitation, is depicted in Fig. 8(b).

II.C. Off-axis performance

Thus far the field analysis has only considered MRI endoscopy with the antenna oriented parallel to B_0 which is appropriate for studying structures that are substantially parallel to the z -axis, including portions of the aorta and major blood vessels in a conventional MRI system employing a cylindrical magnet. When the probe is not parallel to the scanner bore, its transverse field may no longer be substantially perpendicular to B_0 . The computed fwhm of the excited slice in Fig. 7 does not vary significantly as a function of the azimuthal angle θ of the probe's long axis relative to B_0 for either endoscopes. While adiabatic RF transmission by the off-axis probe does excite a smaller FOV, this can be compensated for by increasing the input power. However, during reception, the SNR decreases as a function of θ . The loss can again be computed from the transverse field using the MoM. The sum of the signals across the entire FOV is taken as a measure of the received signal, and the noise is assumed to remain constant at all angles. The results for the loop endoscope are plotted in Fig. 9 and show a 15% drop with the probe at $\theta=30^\circ$ vs $\theta=0^\circ$ (decreasing monotonically for $\theta > 30^\circ$).

The MRI endoscopes described so far have intrinsic localization in just one (transaxial) dimension. Conventional MRI gradient localization can be used to localize in the other two dimensions. Gradient encoding purely in the transverse x - and y -directions is appropriate for MRI endoscopy when the device is oriented with long axis substantially parallel to the z -axis. However, when vessels deviate from the z -axis, "partial volume" effects can cause off-axis structures to

FIG. 10. Quantification of the partial volume effect which causes resolution loss when the MRI gradients are rotated by θ° relative to the detector (cable) axis. (a) A loop detector with intrinsic fwhm slice thickness w is placed against a phantom comprised of an $L=2.8$ mm thick signal-generating plane parallel to the z -axis sandwiched between two thick sheets of acrylic (hatched). The thickness of the signal plane l is measured with read-out gradient m . When $\theta=0^\circ$, $l=L$, but as θ increases, l increases due to the partial volume effect. (b) V , the relative increase in l , is a measure of the partial volume effect and loss of resolution (points, experimental; solid line, calculated).

bleed into each other, blurring the image to an extent that depends on the anatomy of the sample, the image resolution, and the orientation of the probe.

To quantify the blurring effect with MRI gradients rotated off-axis by θ° , we construct a phantom with a plane of signal-generating material of thickness L , comparable to the diameter of a rabbit aorta, or human coronary (2–4 mm), sandwiched between two thick acrylic sheets that do not generate MRI signal. An endoscope with intrinsic fwhm slice thickness w is placed next to the phantom, as shown in Fig. 10(a), and the apparent thickness l of the signal plane is measured with a read-out gradient m . As the gradient orientation is rotated away from B_0 , signal from the edge of the plane bleeds into the antenna's sensitive region reducing the image "sharpness" due to the partial volume effect, causing l to increase. The measured increase in l expressed as a percentage, $V=(l-L) \times 100\%/L$, can be taken as an index of the partial volume effect and the potential blurring it may cause. From the phantom geometry, the partial volume index can also be calculated as

$$V \approx \left\{ \frac{L \cos \theta + w \sin \theta}{L} - 1 \right\} \times 100\%.$$

Figure 10(b) shows that a calculated partial volume effect of up to $V=70\%$ can occur for probe-phantom orientations of up to $\theta=30^\circ$ relative to the imaging gradients, with a $L=2.8$ mm phantom. Its effect on resolution depends on the characteristics of the sample being imaged—the sensitivity of the endoscope affects the analysis only indirectly via its affect on the ability to visualize structures.

II.D. Local SAR

Internal MRI probes are conventionally deployed as receivers only, immersed in external whole-volume transmit fields which induce eddy currents in both the sample,²² and directly in the conductor. MRI experiments demonstrating that conducting guidewires placed inside a saline phantom can heat $>20^\circ\text{C}$ in 30 s but not heat without the phantom²³ suggest that the coupling of the eddy currents induced in the

larger sample volume to the wire is the major source of device heating rather than the direct induction of currents in the wire.²⁴ Nevertheless, temperature increases can vary considerably depending on experimental factors including the tissue RF electrical properties and geometry, the length and orientation of the conductor, and obviously, the RF power being applied.²² The power deposited in the body—the specific absorption rate (SAR, measured in W/kg)—is directly proportional to the incident power required for the coils, which, for probes with tiny excitation volumes, is a small fraction of that required by the body transmit coils. Consequently, the body average SAR for an MRI endoscope used for excitation is orders of magnitude below that generated by conventional MRI. However, local heating measured directly as a temperature rise, or the peak SAR determined in any 1 or 10 g average in accordance with regulatory guidelines^{25,26} may be a concern.

For endoscopy the input power is determined by the amplitude of the transverse B_1 field required to attain the adiabatic threshold. The radius r_t at which this threshold is reached sets the peak SAR. Peak SAR occurs close to the wire at $r < r_t$. While increasing the power provides a larger sensitive disk diameter, in practice this offers diminishing returns because the sensitivity falls as $\sim 1/r$ for the loopless antenna and even faster for loop endoscope, regardless (Fig. 4). It is therefore prudent to set r_t to coincide with the range of r over which useful SNR can be derived.

For the 3 T endoscopic loop in Fig. 3 with achievable $\text{SNR} > 10^6 \sqrt{\text{Hz/ml}}$ close to the device, setting r_t at 3% of the peak sensitivity at the coil center results in an effective disk radius of $r_t=7$ mm. This is achievable with the 4 ms, ± 15 kHz sweep adiabatic BIR-4 pulse in Fig. 6, with a B_1 amplitude $10 \mu\text{T}$ at r_t . The root-mean-square (rms) power required to transmit this pulse with the loop coil antenna, computed numerically using the MoM, is 2.25 W. The local SAR is given by²² $\{\sigma|\mathbf{E}|^2/2\rho\}$, where $\mathbf{E}$ is the electric field computed from the loop current, with the loop placed in an infinite sample with electrical properties equivalent to the 0.35% saline. The result for the loop normalized to 1 W input transmit power and numerically averaged over 10 g of the sample is shown in Fig. 11(a) in the yz -plane where it is maximum. The maximum local SAR is 0.82 W/kg for a pulse repetition period $\text{TR}=44$ ms, which is within regulatory guidelines.^{25,26}

The loopless antenna's transmission efficiency is lower than the resonant loop due to higher sample conduction losses and its larger excitation volume (Fig. 7).¹⁴ It therefore requires greater power to generate the same transverse B_1 field. For the short balunless loopless MRI antenna in Figs. 3(d)–3(f) operating at the same $r_t=7$ mm with the 4 ms, $10 \mu\text{T}$ BIR-4 pulse at 3 T, the power required is 6.25 W. The local 10 g average SAR numerically computed from the MoM for 1 W input power is cylindrically symmetric about the z -axis and is depicted in Fig. 11(b). It has a maximum value of 0.26 W/kg for $\text{TR}=44$ ms, again below regulatory guidelines.^{25,26}

FIG. 11. Computed 10 g average local SAR in W/kg for (a) a four-turn loop and (b) a balunless 23 mm loopless antenna in a biological sample with $\epsilon_r = 80$, $\sigma = 0.65$ S/m, and $\rho = 1000$ kg m⁻³ with the 90° adiabatic pulse in Fig. 6 applied at an input power of 1 W and a duty cycle of 1/11. The capacitor in the loop and the cable in the loopless antenna cause left-right asymmetry along the cable axis, although the loopless antenna's distribution is cylindrically symmetric about the z axis. S1–S4 denote the location of temperature sensors for heat testing. S1 is placed at the distal ends of each device, against the wire. S2 is 10 mm off the axis of the loop in (a), and at the whip-antenna junction in (b). S3 is at the probe-phantom junction. S4 is a control located remote from the probe.

III. EXPERIMENT

III.A. Devices

MRI endoscopy is reduced-to-practice with both the endoscopic loop and the loopless antenna design on a Philips 3T Achieva MRI scanner (Philips Medical Systems, Best, NL). The MRI endoscopes are connected to the scanner through a head-coil transmit/receive interface modified for single channel use. The existing preamplifier is replaced by two cascaded 24 dB gain, 0.5 dB noise figure (NF) nonmagnetic preamplifiers (ARR, Burlington, CT) to increase the total available receiver gain to 80 dB. The system NF is measured at 0.9 dB using

$$NF = -10 \log \left[1 - \frac{\sigma_c}{\sigma_h} \right] - 1.28,$$

where σ_h and σ_c are the noise variances measured with a 50 Ω load at room (300 K) and liquid nitrogen (77 K) temperatures.²⁷

The 3 mm loop coil endoscope is fabricated from four turns of 0.3 mm diameter insulated copper wire tuned to resonate at 128 MHz with a 51 pF ATC nonmagnetic 1.7 mm ceramic chip capacitor. The unloaded Q is $Q_U = 40$, with $Q_L = 16$ in the 0.35% saline. The coil is attached to a length of 2.75 mm diameter UT-85C semirigid copper coaxial cable, forming a MRI endoscope which is matched to 50 Ω at the scanner interface with the circuit shown in Fig. 2(a). The coaxial cable is an odd multiple of $\lambda/4$ (=40 cm) to provide coil decoupling when the diode that is included in the matching circuit is turned on. The diode allows the endoscope to be switched off for conventional MRI employing body-coil excitation, and/or for enabling operation of the endoscope's detector in a conventional receive-only mode. The entire MRI endoscope is insulated with PTFE heat-shrink tube to provide a smooth profile for internal use.

The loopless antenna, which until recently was limited to 1.5 T,¹⁴ is fabricated for 3 T use from a 40 cm length of the same UT-85C semirigid copper coaxial cable with the inner conductor extended 23 mm to form the whip. For MRI endoscopy with the loopless antenna, experiments are per-

formed with sensitivity along the antenna shaft restricted by means of (a) dysprosium oxide suspended in an adhesive emulsion (Plasti-Dip, Blaine, MN), (b) steel filings in the same suspension, (c) magnetic paint (Magically Magnetic Inc., Saxonburg, PA), (d) magnetic audio recording tape, and (e) magnetic ferrite tape, applied to the antenna shaft. For the balun-type loopless antenna endoscope, a dielectric layer of heat-shrink PTFE is provided over which copper tape is attached to form the sleeve balun depicted in Fig. 2(b), with no additional ferromagnetic or paramagnetic coatings. The balun affords a maximum impedance of about 400 Ω at the open end, with a 35 cm long sleeve. Loopless antenna endoscopes are connected to the scanner via the matching network in Fig. 2(b).

Because the rigid cable is unsuitable for IVMRI, a flexible loop and a balun-type loopless MRI endoscope are also built with a biocompatible superelastic nitinol coaxial cable lead for evaluation in small vessel imaging in animal models. These are pictured in Figs. 2(c) and 2(d). The nitinol cable is formed from 0.8 mm commercially available nitinol hypotube (Nitinol Devices and Components, Fremont, CA), such as is used for conventional interventional catheters, pulled over a 0.1 mm gold-plated nitinol core wire insulated with heat-shrink polymer.²⁸ The nitinol cable loop endoscope has five turns of insulated copper wire tuned with a 91 pF capacitor and is 2.3 mm diameter at its widest. With the balun insulated from the hypotube by heat-shrink polymer, the flexible nitinol version of the loopless antenna is 1.1 mm at its widest. The whip, formed by extending the gold-plated nitinol core, is again insulated.

III.B. MRI studies

MRI is performed using adiabatic BIR-4 excitation (duration=4 ms; sweep width ± 15 kHz; FA=80° or 20°) applied to each endoscope with a forward incident power of $W_f = 2.25$ W into the loop device, and 6.25 W for the loopless antenna. Except for off-axis experiments, probes are oriented parallel to the z -axis and MRI localization is provided in the x - and y -dimensions by conventional gradient-echo

FIG. 12. Localization of MRI signals on a 23 mm whip loopless antenna made from a rigid copper coaxial cable using a ferromagnetic coating. (a) Sagittal GRE image (TR=4 s; echo-time, TE=1 ms; FA 90°; FOV=250 mm; and acquisition time, $t=260$ s) acquired from a 120×4 mm² strip of audio cassette tape wrapped around the shaft, leaving a 7 cm gap. (b) Sagittal GRE image (TR/TE=300/3 ms; FA 90°; FOV=128 mm; and $t=76$ s) acquired with ferrite tape wrapped on the antenna's shaft and magnetic paint leaving a 2 cm gap. Note the signal artifacts close to the shaft. (c) Endoscopic image from a grapefruit (TR/TE=1000/3 ms; FA 40°; FOV=80 mm; and $t=160$ s) acquired with the probe in (b), showing how the signal artifacts on the shaft distort the image in the inner region of the sensitive disk.

(GRE) imaging, with no slice selection or z -axis encoding. Planar images are produced that are characterized by intense local signals, but which fall at arbitrary locations in the image FOV depending on the location of the device in the xy -plane. These may be easily detected using a maximum-signal algorithm and moved to the center of the FOV to provide an image stream from the viewpoint of the probe.

The results from the studies employing para- or ferromagnetically coated loopless antennas show that neither dysprosium oxide nor steel could effectively suppress MRI signals ($w > 70$ mm). As depicted in Fig. 12, magnetic tape does afford some localization capability [Fig. 12(a)]. However, when the gap is decreased in order to reduce w (using the magnetic paint), the dephased signals from the rapidly changing local field [Fig. 12(b)] cause artifacts in the integrated signal near the shaft [Fig. 12(c)], with the signal cancellation also reducing SNR. Consequently, these approaches are not pursued in the remaining experiments, owing to the better performance of the balun-type loopless antenna.

The spatial localization, SNR, and off-axis performance of the loop, and balun-type loopless rigid cable test devices are measured in a 3 g/l agarose gel phantom made from the 0.35% saline. When the slice thickness of the sensitive volume along the wire axis is measured by applying a MRI readout gradient along the probe's axis with no localization in the two orthogonal directions, a fwhm slice thickness of 1.5 mm for the loop and 5 mm for the loopless endoscope are obtained. However, for the loop, this measurement is biased in favor of small r by the high signal intensity there. At the radius at which the signal intensity has fallen to 3%, integration of the image intensity yields $w=4$ mm for the loop coil, in agreement with the computed result. The loop endoscope made with nitinol cable also has $w=4$ mm, while w

$=5$ to 6 mm was measured for both rigid and nitinol loopless endoscopes by the same criterion. The experimental data for w , as a function of r , are included in Fig. 7 and summarized in Table I.

SNR measurements are done using fully relaxed GRE images and separately acquired noise images with the gradients and RF transmitter turned off and the endoscopes operating in the conventional receive-only mode. The absolute measured SNR in $\sqrt{\text{Hz}}/\text{ml}$, corrected to eliminate the system noise, is given by¹⁸

$$\Psi = \frac{\psi \sqrt{BW}}{V \sqrt{N_x N_y}} \times 10^{(NF/20)}$$

for voxels of volume V in ml, where BW is the receiver bandwidth in Hz, N_x and N_y are the number of readout points

TABLE I. Performance comparison of prototype loop and balun loopless antenna endoscopes.

	Balun-type loopless endoscope	Loop endoscope
Maximum diameter	1.1 mm	2.3 mm
fwhm slice width	5 mm at $r \leq 18$ mm ^a	1.5 mm at $r=3$ mm 4 mm at $r=7$ mm ^a
SNR at $r=3$ mm ^b	$30 \times 10^4 \sqrt{\text{Hz}}/\text{ml}$	$250 \times 10^4 \sqrt{\text{Hz}}/\text{ml}$
SNR at $r=12.5$ mm ^b	$5.8 \times 10^4 \sqrt{\text{Hz}}/\text{ml}$	$5.7 \times 10^4 \sqrt{\text{Hz}}/\text{ml}$
SNR at $r=20$ mm ^b	$2.7 \times 10^4 \sqrt{\text{Hz}}/\text{ml}$	$1.0 \times 10^4 \sqrt{\text{Hz}}/\text{ml}$
Input power ^c	2.25 W	6.25 W

^aMeasured at the 3% sensitivity contour.

^bMeasured with the endoscope in receive-only mode with body-coil excitation.

^cPower required to achieve an adiabatic BIR4 threshold at 7 mm.

FIG. 13. Exemplary endoscopic images with transaxial resolution achieved entirely by axially constraining probe sensitivity. (a) Loopless antenna image of a kiwifruit (TR/TE=3 s/12 ms; FA 80°, FOV=50 mm; and $t=10$ min) with in-plane resolution of 250 μm . (b) Loop endoscopic image of an intact porcine aorta (TR/TE=500/20 ms; FA 20°; FOV=15 mm; and $t=75$ s) with in-plane resolution of 100 μm . (c) Loop endoscopic image of a human iliac artery specimen immersed in saline (TR/TE=500/21 ms; FA 20°; FOV=20 mm; and $t=127$ s) with 80 μm in-plane resolution. A $1/r$ inverse scaling has been applied to images (a)–(c). The large black spot in (c) is due to the presence of the probe only with a slight ghost artifact above it. (d) VVG-stained histology section corresponding to (c), showing a plaque (arrows). The shape was distorted during sectioning.

and phase encoding steps, respectively, ψ is the voxel signal divided by the rms noise, and NF is the measured system noise figure in dB. The experimental SNR data from the loop and balun-type loopless MRI endoscopes are included in Fig. 4 (solid lines). The SNR of the loop device is in reasonable agreement with the computed SNR (curve a) which did not account for losses in the tuning capacitor and solder joints.²⁹ The addition of the sleeve balun to the shortened loopless antenna reduces its SNR by almost 40% at all radii compared to the computed SNR of the balunless antenna. The loopless

endoscope's SNR exceeds that of the loop at $r \geq 12.5$ mm but is less than the loop for $r < 12.5$ mm (Table I).

A 250 μm resolution 3 T endoscopic image performed with the semirigid balun-type loopless endoscopic antenna in a kiwifruit [Fig. 13(a)] shows excellent resolution of the fruit's soft-tissue structure. An endoscopic image acquired with the loop endoscope in the aorta of an intact pig with an in-plane resolution of 100 μm is shown in Fig. 13(b). The ability of the loop MRI endoscope to identify plaque is illustrated by the image of a human iliac artery specimen in a

FIG. 14. A portion of the kiwifruit in Fig. 13(a) showing the blurring that occurs with the loopless antenna probe rotated 10° off-axis (bottom, TR/TE=3 s/12 ms; FA 55° ; FOV=55 mm; and $t=220$ s) as compared to $\theta=0^\circ$ (top).

saline bath, with in-plane resolution of just $80\ \mu\text{m}$ in Fig. 13(c). Although a dark signal void surrounded by a bright intensity artifact is evident at the location of the coil, the image shows clear resolution of the vessel wall, a plaque, and surrounding tissue, as identified in the corresponding Verhoeff–van Gieson (VVG) stained histological section [Fig. 13(d)]. Again, no external slice localization is applied here: Slice localization is affected entirely by means of the probe's localized sensitivity.

With the orientation of the two test (rigid cable) MRI endoscopes skewed relative to the z -axis, the experimental drop in SNR vs azimuthal angle is also included in Fig. 9. The measurements for the loop device agree with the calculated curves, which also overlap the measurements for the balun-type loopless antenna endoscope. Thus, the measured off-axis performance of both loop coil and loopless antenna-based devices appears comparable. The partial volume index V , measured in the resolution phantom, is included in Fig. 10 and is in agreement with the computed curve. The blurring of image resolution that occurs with the (balun-type) loopless antenna endoscope in a kiwifruit parallel to the z -axis with the gradient axes rotated by $\theta=10^\circ$ is evident in the images shown in Fig. 14.

III.C. Heat testing

Endoscopic probes are safety tested for RF heating by immersing them in the gel phantom made from 0.35% saline. A high-SAR MRI pulse sequence (4 ms pulse; TR=44 ms; duty cycle $d=1/11$; and FA= 90°) is applied continuously for 10 min, during which time temperature is monitored by Neoptix (Québec City, Canada, resolution= $0.1\ ^\circ\text{C}$) fiber optic temperature sensors S1–S4. These probes have a sampling volume of approximately $1\ \text{mm}^3$. The sensors sampled locations on the devices indicated in Fig. 11, including sites that were predicted to exhibit maximum SAR.

With the loop endoscope operating at $W_t=2.25\ \text{W}$ rms input power, as computed earlier, only sensor S1 on the coil at its distal end, recorded any heating, which was limited to $0.8\ ^\circ\text{C}$ [Fig. 15(a)]. This falls within regulatory guidelines of $<1\ ^\circ\text{C}$ for local temperature increases.^{25,26} Because the device is totally enclosed by the phantom, the total time-averaged power deposited in the body at thermal equilibrium

FIG. 15. Temperature recordings ($^\circ\text{C}$, vertical axis) from sensors S1–S4 for (a) loop endoscopic antenna at 2.25 W input power and (b) loopless endoscopic antenna at 6.25 W input power and 4 ms adiabatic pulses applied with a 1/11 duty cycle. Maximum temperature increase is $0.8\ ^\circ\text{C}$ over 10 min. (c) Temperature data from sensor S1 for loop and loopless endoscopes acquired from the heat-test phantom with the same pulse sequence but with input power reduced to 0.25 W rms: no detectable temperature change is apparent. The location of temperature sensors S1–S4 are marked in Fig. 11.

should conservatively be assumed to equal the input power of 2.5 W. This would be divided by the sample mass to obtain the body average SAR, and therefore the experimental average SAR is also well below the guideline for average SAR.^{25,26} With the balun-type loopless endoscope operating at $W_t=6.25\ \text{W}$ rms input power, the only sensor recording any temperature rise is S1, also located at the distal end of the whip. The maximum temperature rise is just $0.7\ ^\circ\text{C}$ [Fig. 15(b)], which again falls within the regulatory guidelines for temperature. The temperature rise at S2 is undetectable [Fig. 15(b)]. This compares to a theoretical SAR level 40% lower than that at S1 [Fig. 11(b)]. However the analysis did not include the effects of the high-impedance balun, which we could not model with the numerical MoM EM software.

Although the above experiments are performed at input powers of 2.25 and 6.25 W for the two designs, endoscopic MRI can also be successfully performed with much lower power levels. For example, reducing the forward power W_t to 0.25 W results in an adiabatic threshold that is not achievable at any usable radius from the device. The detrimental 180° modulation (Fig. 5) is avoided by using conventional pulses of FA $<180^\circ$ near the probe, albeit at the expense of a smaller FOV ($\sim 4\ \text{mm}$ radius). Nevertheless, useful images are still obtainable, as shown by a nitinol loop-detector image in an intact rabbit aorta in Fig. 16. At this reduced power level, no temperature rise is detectable with either the loop probe or the loopless antenna-based endoscopes in any sensor S1–S4 [Fig. 15(c)].

IV. DISCUSSION

We have introduced a novel method for performing MRI endoscopy that employs active internal probes that can per-

FIG. 16. Image of an intact rabbit aorta from the nitinol intravascular loop endoscope at 0.25 W rms input (TR/TE=500/20 ms; FOV=15 mm; t =75 s; and in-plane resolution=100 μ m).

mit both real-time MRI from the probe's viewpoint and high-resolution imaging of vessel walls. The method utilizes modified loop coils or loopless antenna probes designed with MRI sensitivity that is substantially restricted to a disk-shaped volume on the probe itself (Fig. 8). The localization properties of the modified probes are enhanced by using them for both MRI excitation with adiabatic pulses such as BIR-4 or AHP, as well as for MRI reception. The endoscopes are implemented at 3 T, with high-resolution MRI of a kiwifruit (250 μ m), an intact porcine aorta (100 μ m), a human iliac artery specimen (80 μ m), and a rabbit aorta (100 μ m), demonstrated [Figs. 13 and 16]. Heat testing of the endoscopes demonstrated a maximum increase of 0.8 $^{\circ}$ C for a 10 min scan, and computed 10 g average local SARs were <1 W/kg per watt of input power [Figs. 11 and 15], supporting the view that the method and devices can be made safe from RF heating during MRI.

The multiturn loop performed better in terms of the constrained slice thickness (as small as 1.5 mm) as well as SNR close to the coil [Figs. 4 and 7]. However, factors such as component size and a decreased sensitivity as the loop diameter is reduced, may constrain utility of the coil design for small body cavities and vessels. On the other hand, while the loopless antenna design may be better suited for smaller diameter blood vessels, restricting its sensitivity along the wire proved more difficult than with the loop. In addition, while the balun design performs well when a portion of the antenna ($\sim$ 6 cm) is immersed in the sample such as the kiwifruit, antenna loading is a confounding factor when different lengths of the antenna are immersed in the sample. In our unpracticed hands, we were less successful with the application of paramagnetic and ferromagnetic coatings to restrict the sensitivity of the loopless antenna (Fig. 12), and further work will be required to optimize this approach. Note that all of the signal-constraining approaches noted above can be applied to other types of internal coils, such as opposed solenoids,^{16,30} expandable coils³¹ and elongated loops⁷ which, however, have inherently more extensive sensitivity profiles.

Performing endoscopy when the probe is skewed relative to the B_0 -direction resulted in SNR losses of <15% (Fig. 9). However, the associated blurring that results from partial

volume effects, compromises the resolution of off-axis structures (Figs. 10 and 14). A practical solution to counter changes in resolution that occur with off-axis orientations is to occasionally update the gradient directions that define the image plane dynamically via "joystick control" by the operator. Alternatively, since the orientation of an endoscope designed primarily for use in the vasculature or other vessels will basically reflect the orientation of the vessel surrounding it, the probe orientation could be determined using a signal projection method,³² or by extrapolating the history of previous probe positions to provide updated corrections to the gradient encoding directions. Inasmuch as such corrections are performed only intermittently, they should not appreciably impact the efficiency of an MRI endoscopy procedure. Note also that for the loop design, the SNR loss from performing MRI endoscopy in vessels whose typical orientation differs from the z -axis (B_0) can be addressed fairly simply by remounting the coil in the appropriate orientation for B_1 .

Transmission using internal coils avoids the coupling effects and heating associated with the extensive eddy currents induced by the scanner's whole-volume external body and head MRI coils by dramatically reducing the excited volume. Indeed, small-vessel imaging was possible at input power levels of only 0.25 W that produced no detectable temperature rise at 3 T [Fig. 15(c)]. This compares with the 10–30 kW RF power levels typically used in MRI scanners, which can cause over 20 $^{\circ}$ C increases in conventional catheters in just 30 s of scanning at 1.5 T, for example.^{23,24} For this reason, receive-only internal MRI detectors such as the conventional loopless antenna must be "shut off" during excitation,¹⁴ and indeed, a diode is provided in Fig. 2 for operating the endoscope in a receive-only mode. Thus, diode switching of receive-only internal MRI detectors¹⁴ and MRI endoscopy described herein (Fig. 15) offer two different approaches for successfully limiting heating to below 1 $^{\circ}$ C. Although the numerical SAR analysis for the loopless antenna (Fig. 11) did not include the effect of the balun, its efficacy in choking off currents on the surface of the cable is evidenced both by its ability to limit the thickness w of the sensitive disk and by the attenuation of temperature rise seen experimentally at the junction [Fig. 15(b)].

Image encoding in the two dimensions perpendicular to the endoscope's axis is demonstrated here with a conventional Cartesian approach employing MRI gradients. We are currently investigating using projection imaging methods such as spiral imaging, with a sliding window reconstruction to facilitate real-time endoscopy. An elegant alternative perhaps is to use the intrinsically radial RF gradient in one of the dimensions analogous to rotating frame zeugmatography,³³ combined with the intrinsic phase difference between an external excite coil and the polar angle of the loopless antenna, to provide cylindrical encoding¹³ that is intrinsically locked to the endoscope in all three dimensions.

Thus, we are hopeful that the MRI endoscopic methods and devices described here will offer a new window to the vasculature and indeed other vessels such as the esophagus, stomach, the urinary tract, and colon. Such approaches may

afford an inside out yet transluminal view of the vessel wall and any associated pathology, with the ability to switch between the point of view at the head of the internal device directly and the scanner FoR with all of the functionality that MRI scanners currently enjoy. In this manner, the promise of MRI endoscopy is a probe's-eye high-resolution view of soft tissue and function to aid diagnosis and intervention.

ACKNOWLEDGMENTS

This work was supported by NIH R01 Grant Nos. EB007829 and HL090728. The authors thank Dr. Michael Schär for assistance with the 3T MRI scanner, Dr. Ronald Ouwerkerk for assistance in simulating the adiabatic pulses, Dr. Abdel-Monem el-Sharkawy and Dr. Ananda Kumar for assistance with heat testing, coil construction, and EM field simulations, Dr. William A. Edelstein for useful discussions about antenna geometries, Di Qian for assistance with animal experiments, and Dr. Barbara Crain and the Autopsy Service at The Johns Hopkins Hospital for supplying the pathology specimen in Fig. 13.

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