

An MRI endoscope?

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Introduction: Existing internal clinical imaging devices such as endoscopes and IVUS, provide internal high-resolution examination directly from the viewpoint of the inserted probe. However, the efficiency of high-SNR internal detectors that are ideally suited to MRI Endoscopy, is compromised by conventional MRI being intrinsically locked to the laboratory frame-of-reference (FoR). This is because the location of the probe in the body must first be determined, and the scan plane coordinates computed and fed to the system's external gradient coils, before local MRI from the probe's viewpoint can be done. The resulting latency period costs SNR per unit time. We propose a novel solution using active internal probes, which utilizes the spatial properties of the probe itself, for MRI. Because these properties are intrinsic to the probe, they move with it, transforming the MRI FoR from the laboratory to the device itself, creating a true MRI endoscope— an “MR-eye”.

Methods and Results: Central to the idea of shifting the FoR to the device, is slice localization intrinsic to the probe, and employing the probe for excitation. A loopless catheter antenna, which to date has been limited to 1.5T use and usually only for detection [1], was fabricated from UT-85C copper coaxial cable, by tuning it for 3T (128 MHz). A sleeve balun tuned to $\lambda/4$ was added to reduce the sensitive region of the probe to ~5mm Full-Width-Half-Max as shown by MRI along the probe's axis (Fig 1). Transmission of RF by this modified probe excites only a small region confined to a disk shaped area. Reception of signal using the same probe ensures an image arising from this excited slice (Fig 1).

The excitation field of the probe is very nonuniform across the sensitive disk. To compensate for this, phase-cycled adiabatic BIR-4 RF pulses [2,3], are applied to deliver a constant flip-angle above a threshold B_1 , chosen so as to achieve the desired sensitive disk diameter (Fig 2). Thus, knowledge of the probe's location in the scanner is unnecessary as the probe inherently acquires an image of what it 'sees'. As the probe moves, the imaged slice moves with it. No time is wasted in tracking the probe.

Excitation by the detector requires lower power (<1W), generating reduced eddy currents and Specific Absorption Rate (SAR), compared to whole-body excitation, as confirmed by numerical EM-field analysis: the SAR <0.6 W/kg over any 10g (Fig 3).

The MRI endoscope was implemented on a Philips 3T *Achieva* scanner using a custom single-channel transmit-receive interface. Localization in the other 2 dimensions was provided by conventional gradient-echo imaging (without slice-selection). This generates an image at an arbitrary location in the field-of-view which is easily moved to the center based on image intensity. Note that the image plane is distorted to some degree when the plane is not orthogonal to the probe axis. However, this is not considered detrimental because the image plane or “eye” can simply be rotated by a joystick-control.

The modified probe and MRI sequence was first validated in a phantom to determine the localization properties, and then in a grapefruit to evaluate its ability to differentiate heterogeneous tissue structure as shown in Fig 4.

Conclusion: RF transmission and reception by a modified internal probe can inherently localize the MRI signal to the probe-head, removing the need for probe tracking within the scanner's FoR. Such a probe can be used for real-time internal MRI Endoscopy with much lower power deposition than occurs with the probe when using body excitation.

References: [1] Ocali O, Atalar E. *Magn Reson Med* 1997; 37:112–118. [2] Garwood M, Ke Y. *JMR* 1991; 94:511-525. [3] Bottomley PA, Ouwkerk R. *JMR* 1993; 103, 242-244. *Supported by NIH grant R01 15396*

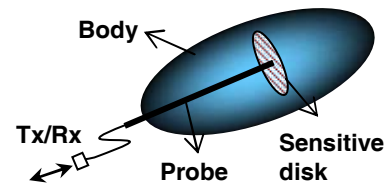
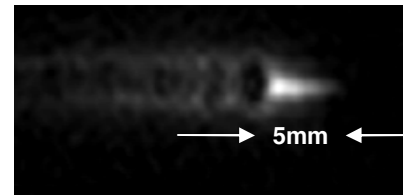


Fig 1: A sleeve-balun reduces sensitive region to 5mm (top). Transmission and reception by the probe inherently localizes signal.

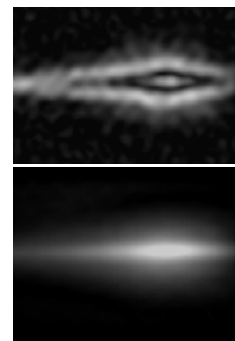


Fig 2: Sensitivity of a bare loopless antenna with square pulse (top) and BIR-4 pulse (bottom). Non-uniformity is corrected by adiabatic pulse.

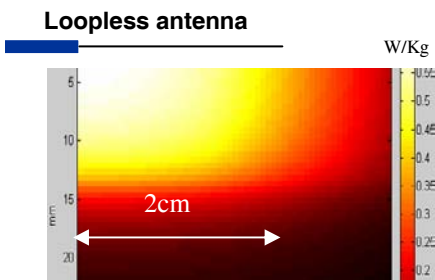


Fig 3: Computed 10g avg. SAR distribution of probe with 0.4W, 4ms adiabatic BIR-4 pulse (TR 10ms).

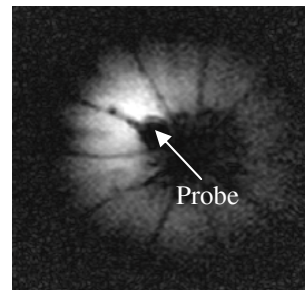


Fig 4: Gradient echo MRI from a grapefruit. Slice localization arises inherently from probe sensitivity (12cm FOV; 128x128).