MRI-Comptaible Ultrasound for Image Guided Therapy

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Introduction Use of multiple imaging modalities benefits image guided therapies (IGT) if the strengths of each modality can be fully appreciated. The temporal and spatial resolution of x-ray combined with soft tissue multi-planar characteristics of MRI has been shown to impact shunt placement in the liver [1] and cardiac ablation [2]. Ultrasound has the advantages of high temporal resolution and absence of ionizing radiation. We present a method for acquiring MR images in a known orientation with respect to the US image and demonstrate concurrent real-time imaging in a phantom with both modalities. US can be used outside the MRI bore to overcome limits of bore size for the placement of therapeutic probes or biopsy needles. Once the probe or needle has been tentatively advanced to clear the bore, MR imaging can be used to confirm placement or monitor therapy.

Materials and methods A portable ultrasound system (model T3000, Terason Ultrasound, Burlington, MA) with a 128 element linear array transducer was connected to an interface box within the magnet room and the box was connected to a laptop computer outside the room via a waveguide through a 25m fiber optic FireWire cable. To mitigate the affects of magnetic attraction the interface box steel enclosure was replaced with an aluminum enclosure. Aluminum foil was applied to the entire length of US probe cable and interface box to limit RF interference in MRI. The US probe window was covered with a single layer of aluminum foil to limit RF interference in both modalities, while still allowing US signal to penetrate. Compatibility tests were performed on 1.5T and 3T whole body MRI systems (Signa Excite HD, GE Healthcare, and Milwaukee, WI). A location and orientation for the interface box was established such that the fringe field did not deleteriously affect the ferrite core inductors in the signal conditioning circuit but close enough for the US probe to reach the iso-center. MR images of a spherical phantom in a standard head coil were acquired with US probe inside the coil. To evaluate the degree of interference of the US on MRI, MR images were acquired with the RF transmit power turned off (FSE-XL, TR/TE=600/6.16, BW=62.5kHz, ETL=16, FOV=480, NEX=1) while the US operated at various transmission frequencies (5MHz, 6MHx, 7.5MHz, 9MHz, 10MHz). The test was repeated with the RF transmit power turned on to establish the effects of MRI on US. The US probe and cable were held at the lip of the z-gradient against the bore to test the worst-case effects of heating with a gradient echo sequence. To show feasibility for probe placement, a fiducial target in a PVC breast phantom was localized outside the bore using US. A needle holder was used to position an MRI-compatible cryo-therapy probe (Galil Medical, Yokneum, Israel) along the desired trajectory. Four fiducial markers (Beekley MR-Spots, Bristol, CT) were attached to the US probe head for passive tracking in MRI and to establish the oblique MR scan plane through the trajectory path, orthogonal to the US plane. The cryo-therapy probe was advanced to the target fiducial during real-time 3T MRI (4 fps) and US (16fps). No calibration was performed between MRI and US.

Results With proper RF shielding of the cables and US probe, interference from US operation was much reduced. Fig. 1a shows the baseline MRI image noise before the US system was introduced. Fig. 1b shows the MRI image noise during US caputirng with US system shielded and US probe window unshielded. Fig. 1c shows the image noise dramatically reduced during US capturing with an additional thin aluminum foil shielding the US window. Image noise measurements are shown in Table 1. No heating of the US probe was detectable to touch. The affects of the MRI RF transmit pulses to US was negligible when the probe head window was covered with a single layer of foil. Fig.2 shows simultaneous US/MR scans of breast phantom with a fiducial target displaying on the same volume with cryo-probe inserted and passive markers attached on the US probe head. Image rendering was done in 3D-Slicer (in-house software). Accuracy of MR scan plane prescription is limited by the ability to localize the passive fiducials.

Measurement Conditions	Baseline	Capturing mode at isocenter w/ US probe shielded, 1 layer foil at window
Background noise ROI mean (sd)	4.79 (2.56)	5.06 (2.66)

Table 1 Image noise comparison

Discussion Preliminary results suggest that US and MRI can operate together but some interference still existed. The broadband noise is likely originated from digital switching in the US interface box associated with the Firewire signal. Broadband noise was reduced with proper shielding. Periodic zipper noise is likely due to the short high-voltage pulses apply to the piezoelectric crystal. Alternating the US transmit frequency changed the patterns of zippers. Covering the probe window with one layer of aluminum foil reduced the zippers but also caused some signal degradation in US. Blanking US transmission during MR readout and vice versa is an immediate approach to circumvent the problem. Simultaneous US/MR acquisitions were demonstrated for image guided therapeutic probe placement on a breast phantom. Future directions include active tracking of the US probe and biopsy needle, remote manipulation of the US probe and needle holder, proper calibration and real time fusion of US and MRI.

Reference

1. Kee ST et al., J Vasc Interc Radiol 2005 Feb 16:227-34
2. Rode KS et al., IEEE Trans Med Imag 2005 24(11):1428-40

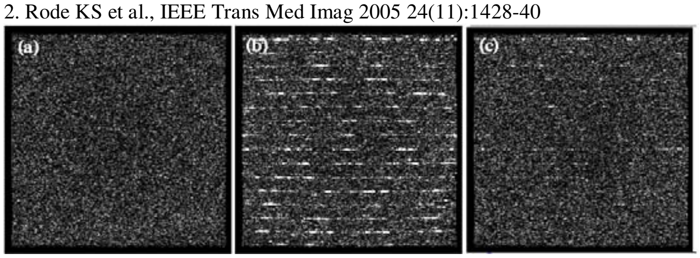

Fig. 1 MR image noise induced by US (a) Baseline, (b) During US capturing with US system shielded and probe window unshielded, and (c) During US capturing with shielded US system and probe window

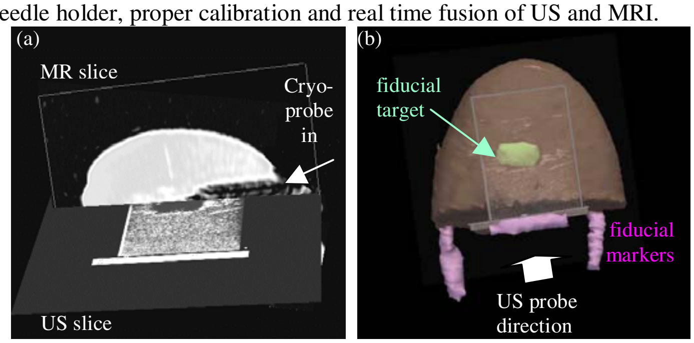

Fig. 2 Simultaneously acquired US and MR images of a breast phantom showing the targeting of fiducial with a cryo-probe (a) 2D US image coincided with a MR slice. (b) 2D US image fused with a rendered MR volume