7T External Prostate Array with Single Channel Transmit: Simulation and Experiment

Bei Zhang[1], Andrew B Rosenkrantz[2], Daniel K. Sodickson[1], Samir S Taneja[2], Cornel Stefanescu[1], and Graham C Wiggins[1]

[1]The Bernard and Irene Schwartz Center for Biomedical Imaging, NYU Medical Center, New York, NY, United States, [2]Department of Radiology, NYU Langone Medical Center, New York, NY, United States

Introduction: Prostate imaging could benefit from the increased MR sensitivity at 7 Tesla, but such deep torso imaging is challenging due to extreme RF inhomogeneity. Several authors have demonstrated 8 channel transmit-receive (TxRx) arrays for prostate imaging, but these have relied on parallel transmit and RF shimming, which bring additional complexity to the equipment required and the acquisition of data. [1,2,3]. Given that the prostate presents only a small region of interest (ROI), we propose that it is possible to achieve excitation in the prostate through the use of only two transmitter elements, one anterior and one posterior, with fixed power split and phase relationship. Positioned appropriately these should interfere constructively in the prostate. Two different transmitter element designs were evaluated through simulation, traditional loops and radiative dipole antennas [1]. The strong interaction between the human tissue and the electromagnetic field requires full wave EM simulation to predict the coil behavior. We have used the simulations to help design and build a two-element transmit-receive plus six-channel receive-only array for prostate imaging at 7 Tesla and tested it in human imaging.

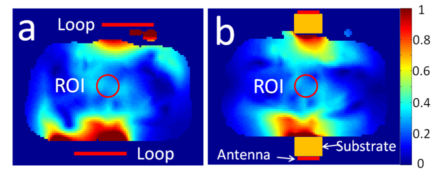

Fig. 1: (a) Combined B1+ for two offset loop system. (b) Combined B1+ for two radiative antennas

Methods: A system with two surface coil loops and a system with two radiative antennas were simulated with the FDTD method (Microwave Studio, CST, Germany). The simulated loop diameter is 16cm, corresponding to the approximate depth of the ROI in the HUGO body model. As shown in Figure 1(a), one loop is placed anterior and the other is placed posterior, with both loops shifted 5.5cm away from the center in opposite directions to compensate for B1+ twisting [4,5], to create maximum B1+ in the region of interest. The two radiative antennas were modeled according the descriptions in Ref [1], with a copper strip 0.7cm wide and 11cm long split in the center and placed on a block of high dielectric substrate (relative epsilon = 37) 6.7cm x 4.2cm x 14.3 cm in size. These are placed in the center, as shown in Figure 1(b), since the radiative antenna shows little B1+ twisting [1].

Based on the simulation results, the array was constructed using loop elements, with a row of 4 overlapped elements anterior and another row of 4 posterior (Figure 2). The loop diameter was reduced to 12cm based on phantom experiments. In each row of four, one coil was used both to transmit and receive, while the remaining coils were receive-only. A quarter lambda lattice balun was constructed at the output of each loop to balance current flowing in the coil. Each receive-only element was connected to a preamplifier (Siemens Healthcare, Erlangen, Germany) and preamp decoupling was implemented with a phase shifter to transform the input impedance of the preamp appropriately. Bridging the lattice balun with a diode provided active detuning, and additional detuning is provided by a passive detuning circuit located on the opposite side of the loop. For the two transmit-receive elements, equal power is provided to each through a Wilkinson power divider, with cable lengths chosen to provide 180 degree phase difference between the two coils. A T/R switch is placed in each transmit path routing the coils either to the RF transmitter or to two preamps. To allow conformation of the anterior array to different subjects it was constructed of a single piece of etched Pyralux flexible circuit board material (Dupont, Wilmington DE). Volunteer measurements are conducted in accordance with our institute's IRB. Flip angle maps were obtained with a turboflash sequence with preparation pulses (TR/TE/BW=5000ms/2.2ms/490Hz/pixel, TA=0:35, FOV=350mm*350mm) [4]. After careful calibration of the flip angle in the prostate, SNR maps were calculated from GRE measurements both with and without RF excitation (TR/TE/BW=200ms/4.1ms/300.0Hz/pixel, TA=0:53, FOV=220mm*220mm, 256 Matrix). SNR was also measured at 3T in a clinical patient with a 16 element body array (Siemens Healthcare, Erlangen, Germany). T2 TSE images of the prostate were acquired with resolution 0.6mm*0.6mm*3mm, TR/TE/BW=10,300ms/81ms/255Hz/pixel, TA=4:08 min.

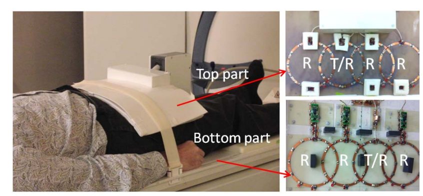

Fig. 2: 8-element prostate coil (T/R: Transmit and Receive; R: Receive only)

Results: Simulation of the two coil systems showed that with proper offsetting of the loop transmit elements, a B1+ field of slightly greater magnitude and uniformity could be achieved in the target ROI than with the radiative antennas (Fig. 3). Close examination of the B1 vectors through the RF cycle showed that the two loops create a region of circular polarization in the prostate, enhancing the B1+ efficiency. The radiative antennas, while they beam their energy directly towards the ROI, create only linear polarization with this two coil arrangement. Although more radiative antennas could be used to wrap around the body and create circular polarized B1+ in the center, this would involve a more complex splitting and control of the RF transmit signal.

Experimental measurements with a body sized tissue equivalent phantom allowed further optimization of the 2 loop transmit structure. Highest transmit efficiency was achieved with the coils offset 4.25cm from the center, less than was found in simulation, and with smaller 12 cm loops. The unloaded to loaded Q ratios were 10.3 or better for all coils. Temperature tests were conducted with meat to determine safe SAR limits. In volunteer measurements a 90 degree flip angle could be achieved in the prostate with a 550v 1ms hard pulse, 40% above the scanner's usual reference calibration range. Flip angle maps (Figure 3) show good correspondence to the simulated B1+ fields, with constructive interference creating a reasonably uniform excitation across the ROI. SNR comparison to 3T shows a gain of 3.3. T2 weighted TSE images show good depiction of prostate anatomy and clinically relevant details (Figure 5).

Conclusions: Full wave simulations allowed us to explore various design options before constructing a 7T coil for prostate imaging. This allowed us to construct a coil which does not rely on parallel transmit or RF shimming, considerably simplifying the equipment requirements and making 7T prostate imaging possible on standard single transmitter 7T systems. The SNR gain compared to 3T should provide comparable results to 3T endorectal coils, avoiding the need for this invasive procedure.

Acknowledgement: The authors are grateful to Department of Defense (PC100090) and The Joseph and Diane Steinberg Charitable Trust for their support.

[1] Raaijmakers A, et al, ProcISMRM2010, p48 [2] Metzger GJ, et al, MRM 64:1625-1639, 2010 [3] Scheenen TW, et al, ProcISMRM2011, p592 [4] Wiggins G, et al, ProcISMRM 2009 p394 [5] Duan Q, et al. ProcISMRM 2010 p50 [6] Klose U, Med. Phys. 19 (4), 1992

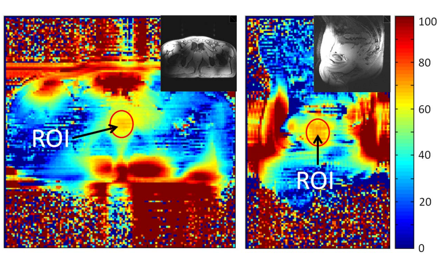

Fig.3 Flip angle maps in vivo. Left image is axial slice and right image is sagittal

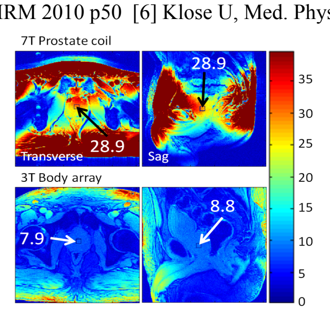

Fig.4 7T and 3T SNR (RSS)

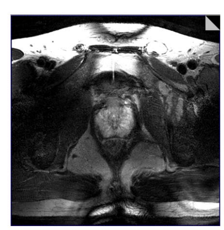

Fig. 5 T2 TSE .6 x .6 x 3mm