A NEW RF COIL DESIGN FOR PROSTATE MRI

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Introduction: Despite increasing the number of receiver coils and upgrading to larger $\mathbf{B_0}$ fields, prostate MRI using the torso-pelvic receiver array coil still lacks sufficient sensitivity to visualize subtle transcapsular tumor spread or seminal vesicle involvement [1-2]. The accuracy of prostate MRI using only a torso-pelvic receiver array coil in defining the local tumor stage is only 57% in clinically confined tumors [3-4]. Hence, the current trend in 1.5T and 3.0T clinical imaging is to use an endorectal surface coil in combination with a torso-pelvic coil [5-8]. With this approach, the endorectal coil provides the necessary sensitivity to acquire high-spatial resolution images and spectroscopic data of the prostate while the torso-pelvic coil provides a large enough FOV to assess pelvic lymph nodes and pelvic bones for metastatic disease. However, use of an endorectal coil is an invasive technique and requires careful monitoring to avoid excess RF deposition. Therefore, we propose a non-invasive RF coil that provides both comparable SNR and a large FOV for prostate MRI. We modeled the new RF coil geometry and analyzed the $\mathbf{B_{1-}}$ field it generates using a full wave electromagnetic field simulation program. A prototype coil was then constructed and evaluated through 3T MR phantom studies.

Methods: For prostate imaging, we propose a diaper-shaped coil consisting of an array of five receiver-only channels. Four rectangular loop coils (width: 130mm, height: 80mm) and a butterfly shaped coil (width: 80mm, height: 150mm) were mounted on a half-cylindrical acrylic tube (radius: 135mm) as shown in figure 1. The diaper-shaped coil is to be positioned anteroposterior to the inferior pelvic abdomen and rectum of a patient. The design was first modeled in a full wave electromagnetic field simulation program (SEMCAD X Ver. 14.2.1 Schmid & Partner Engineering AG, Zürich, Switzerland). We simulated the $\mathbf{B_{1-}}$ field dissipated in a dielectric phantom (width = 300mm, height = 190mm, and length = 300mm) with the coil tuned at 3T (127.74MHz). The interior region of the phantom had relative permittivity (ε_r = 69.062), electric conductivity (σ = 1.5053 S/m), and loss tangent ($\tan\delta$ = 3.0851) values to simulate body fluid. Two spherical balls were also inserted in phantom. One ball simulated the bladder (diameter = 100mm, ε_r = 21.883, σ = 0.29788, and $\tan\delta$ = 1.9266) and the other ball simulated the prostate (diameter = 35mm, ε_r = 72.226, σ = 0.92588, and $\tan\delta$ = 1.8144). For coil construction, copper foil strips (thickness = 0.0341mm, width = 5 mm) were used to trace the circuit pattern on the coil frame. The coil was tuned for 3T MRI and matched to 50Ω. The passive and active combination detuning method was utilized to decouple the proposed receive-only RF coil from the RF transmitter. The outputs of RF coils were connected to an interface box developed by Philips Medical Systems. Low noise amplifiers (LNA) were integrated in this coil interface box. MR images were acquired using a 3T Philips Achieva system (Philips Medical Systems, Netherlands). For comparison, the phantom was also imaged using a Philips torso-pelvic RF coil and an endorectal coil (Medrad Inc, USA). The MR pulse sequence parameters were: sequence type = 2D T1 weighted spin echo, TR/TE = 300ms/10 ms, matrix = 512x512, FOV = 150 x 150 mm, slice thickness = 2.0 mm, NEX = 2.

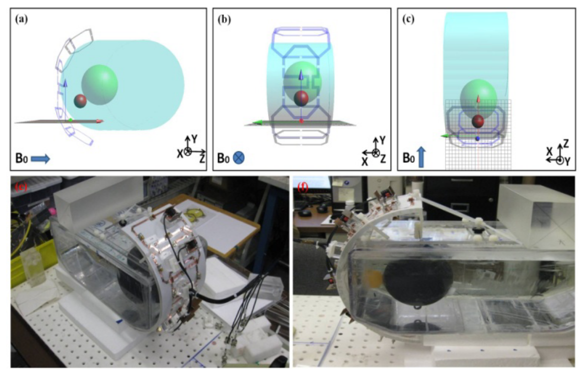

Figure 1. (a) Side view, (b) front view, (c) top view, and (d-f) experimental setup for the diaper-shaped RF array coiland phantom. The five channels consist of four loop coils and a butterfly loop.

Results: From simulation results (figure 2), the conventional torso-pelvic coil produced a relatively homogenous $\mathbf{B_{1-}}$ field over the phantom. In contrast, our proposed RF array coil generated a more non-uniform, but focused $\mathbf{B_{1-}}$ field intensity in the prostate region. As seen in the MR phantom images (figure 3), the torso-pelvic coil also produced a homogenous $\mathbf{B_{1-}}$ field across the human pelvic-lower abdomen region. The endorectal coil generated a very non-uniform, focused $\mathbf{B_{1-}}$ field distribution. Our proposed coils generated a non-uniform $\mathbf{B_{1-}}$ field across the whole phantom, but produced a homogenous $\mathbf{B_{1-}}$ field around prostate region. For the prostate ball phantom located 5cm from the side of the phantom wall, the proposed RF coil resulted in a significantly better SNR (549.55) than the torso-pelvic coil (140.45) but a lower SNR than the endorectal coil (719.62).

Discussion and Conclusion: We constructed a diaper-shaped array coil and demonstrated the feasibility of non-invasively imaging of prostate. The proposed coil produced the competitively high SNR and uniform $\mathbf{B_{1-}}$ field in region of the human prostate. For this design, the alignment of the RF coil generates a linearly polarized $\mathbf{B_{1-}}$ field. We expect a $\sqrt{2}$ increase in the SNR if more coil elements are added to form a circularly polarized coil. We also anticipate a better SNR by combining our proposed coil with the endorectal coil.

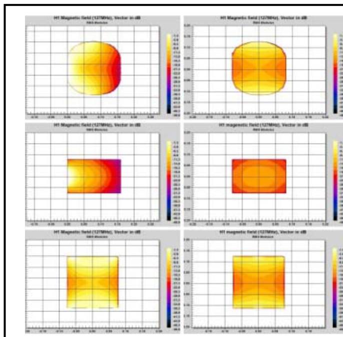

Figure 2. Simulated $\mathbf{B_{1-}}$ magnetic field distribution in the phantom using our proposed five-channel array coil (left column) and the second column is data produced from a six channels wrap around array coil.

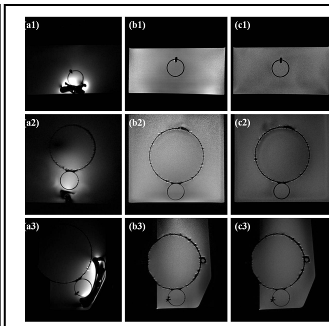

Figure 3. Axial (top row), coronal (middle row), and sagittal (bottom row) images acquired from the endorectal coil (left column), the diaper-shaped coil (middle column), and the conventional torso-pelvic coil (right column).

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