Three-channel flexible phased array using circular coils with annex structure for decoupling

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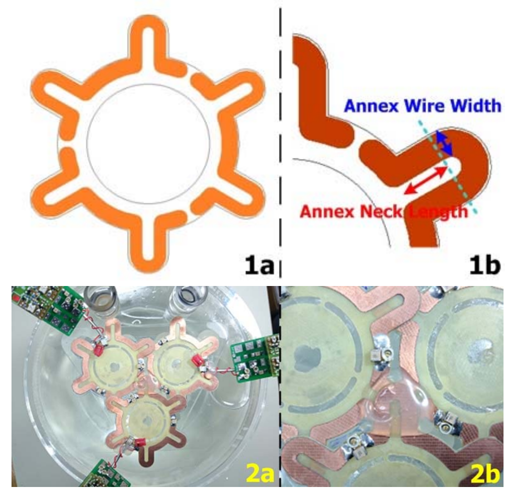
TARGET AUDIENCE: RF engineers interested in coil array development.

PURPOSE: MRI coil array can achieve large field-of-view (FOV) coverage and high signal-to-noise ratio (SNR) simultaneously. Sufficient decoupling between coils is necessary to ensure these advantages. Coil decoupling can be achieved by either overlapping coil elements[1] or adding gaps between coil elements[2-4]. However, either approach only allows a fixed coil arrangement for the optimal decoupling between coil elements. To obtain high quality MR images, the MRI coil should also have a high filling factor[5]. Because the same anatomical structure can vary significantly across subject, a coil array need to be flexible in order to achieve a high filling factor in different experiment setup[6,7]. Taken together, decoupling can be challenging in flexible coil array design.

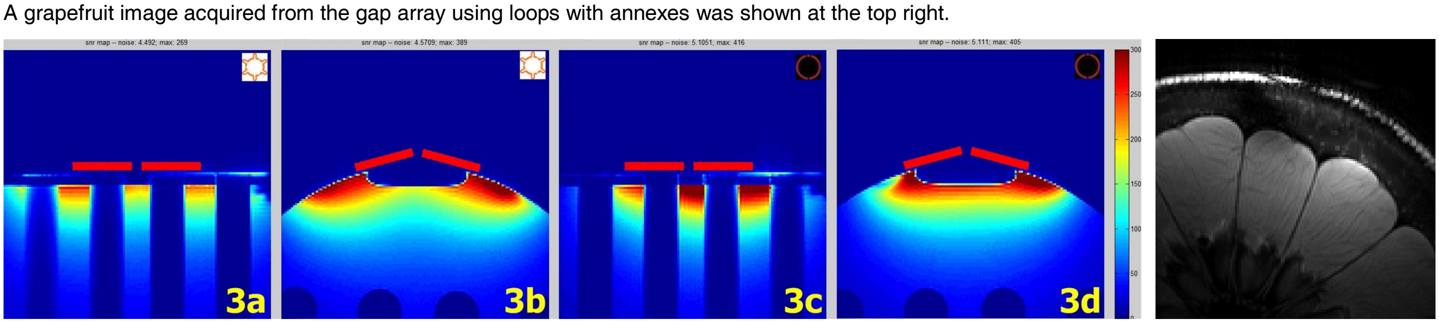
Here we propose a coil array design using circular coil loops with branch-like structures (annex) to achieve decoupling. We hypothesize that such an annex structure is particularly useful in decoupling coils in a flexible array, because there is a clear gap between coils allowing geometrical deformation in bending[4]. Our design differs from the previously study[4] in 1) constructing the coil using discrete electronic components on PCB rather than using the monolithic transmission line resonators, 2) morphing the circular loop itself instead of additional small annex, and 3) using capacitive coupling to receive NMR signal. In a bended position, the maximum SNR of our array using circular loop with annexes was roughly √2 times of that when array was on a flat top.



METHODS: Three circular coil loops (40 mm diameter and 4 mm width) with annexes were constructed using FR4 PCB (0.4 mm thickness, 18 μm single copper layer, NYPCB corp.). Each coil had six annexes evenly distributed on the circumference of the coil with 8 mm annex neck length (Fig.1). Each coil was tuned to 123.25 MHz with three evenly distributed capacitors (Voltronics, Salisbury MD, USA). Coils were decoupled by overlapping annexes (Fig.2) and low-noise pre-amplifiers[1]. Each coil was mounted on a circular disc (acrylic, thickness = 1mm) to ensure the mechanical rigidity and to force the deformation at annexes only when the array was bended. For comparison, a 3-channel coil array using circular loops (no annex) was constructed. Coils in this control array were decoupled by overlapping. Experiments were performed on a 3T MRI scanner (MAGNETOM Skyra, Siemens, Erlangen, Germany) using a gradient echo sequence (TR = 400 ms, TE = 10ms, flip angle = 25°, slice thickness = 5mm, FOV = 150mm, image matrix = 256x256) and a cylinder (17 cm diameter) saline phantom. The coil array was either bended and placed at the side of the cylinder or placed flap at the top of the cylinder. We also imaged a grapefruit using a turbo spin echo sequence (TR = 3000 ms, TE = 13 ms, slice thickness = 2.5 mm, FOV = 150 mm, image matrix = 256x256).

RESULTS AND DISCUSSION: Fig.3a shows the saline phantom image when our proposed 3-channel array using circular loops with annex was positioned on the flat top of the phantom. The maximum SNR was 269. Fig.3b shows image when the 3-channel array using circular loops with annex was bended to fit to the curved side of the phantom. In a bended position, the maximum SNR of this array was roughly √2 times of that when array was on a flat top. Fig.3c shows the image acquired from the 3-channel array using circular (no annex) placed at the flat top of the phantom. The maximum SNR was 416. Fig.3d shows the image acquired 3-channel array using circular (no annex), which was bended to fit to the curved side of the phantom.

Since the bending an array inevitably deformed the loop coil without annexes, the resonance frequency was no longer the proton Larmor frequency and the coupling between coils can be significant after such geometry distortion. Consequently the image quality will be degraded, consistent to our observation (Figs. 3c and 3d). The 3-channell array using circular loops with annex, on the other hand, showed of √2 SNR improvement when the array was bended. A grapefruit image acquired from the gap array using loops with annexes was shown at the top right.



CONCLUSION: We empirically demonstrated that a 3-channel flexible coil array consisting of circular loops with annex structure can be used to acquire localized MRI with high quality even when the array was bended. While we only demonstrated an array of three coils, we expect that more coils of the same geometry can be added to the array to increase the FOV coverage while maintaining the decoupling between coils with annexes.

REFERENCES:

1. Romer PB., *et al*, Magn Reson Med 1990;16:192-225.
2. Charlton T., *et al*, Proc ISMRM 2011;#3875.
3. Keil B., *et al*, J. Magn Reson 2013;229:75-89.
4. Kriegl R., *et al*, Magn Reson Med 2014,doi:10.1002/mrm.25260.
5. de Zwart JA., *et al*, Magn Reson Med 2002;47:1218–27.
6. Keil B., *et al*, Magn Reson Med 2011;66:1777-1787.
7. Duan Y., *et al*, Proc ISMRM 2013;#4383.