Constellation Coil

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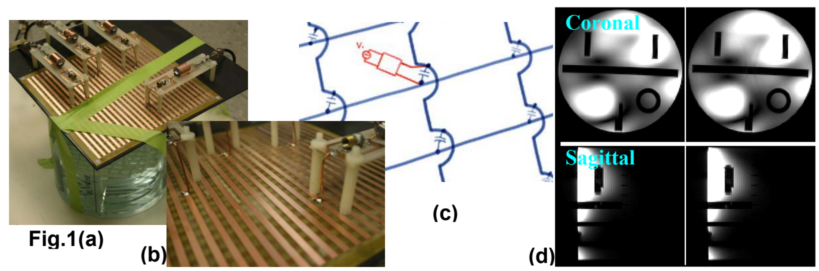
Introduction: A multi-port array coil provides individual RF field profiles that, under the static control of B1 shimming coefficients or dynamic control of parallel RF pulses, are weighted and superimposed to create the RF field. The coil therefore plays a central role in the induction of a temporally/spatially varying B1 field for creating an excitation profile, and meanwhile, a concomitant E field that leads to RF loss and SAR. Fundamentally a coil structure that supports flexible current distribution control is essential for effective management of both the excitation profile and RF power, and is hence a key factor in parallel Tx performances (1,2). In this work we developed a "constellation coil", which prioritizes field optimization-based Tx/Rx improvement with a continuum structure, and accommodates scalability supporting highly parallel Tx/Rx. This differs from the conventional approach where an array coil is composed of decoupled and tuned element coils such that each element provides a discrete current path. We present below the new approach and preliminary 7T MRI results obtained with parallel Tx and Rx constellation coils.

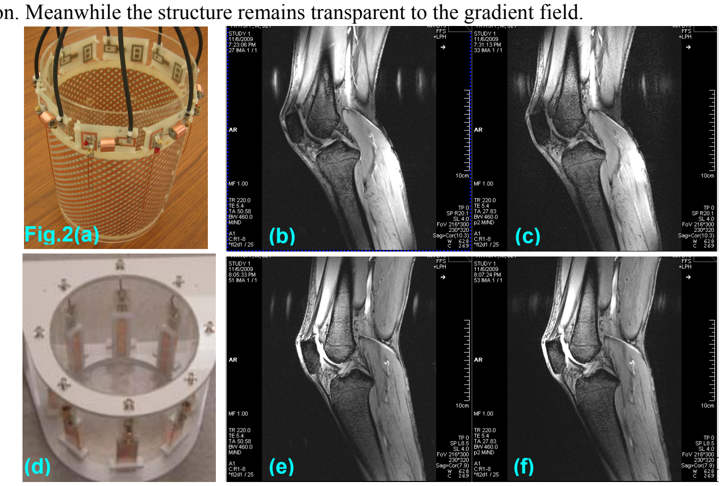
Methods and Results: With the goal of inducing diverse and strong B1 profiles at low RF loss, we introduce the idea of a constellation coil, a new type of parallel Tx / Rx array coil that supports, with a continuum structure, flexible current distribution control and RF field optimization. Compared to a conventional parallel Tx / Rx array coil, a constellation coil's structural components tend to couple to each other capacitively, inductively, or through direct connection. While decoupling recedes in importance, several characteristics are emphasized to form a foundation for high performance parallel Tx / Rx operations, as explained below with examples.

Fig.1a shows a 4-port grid pattern-based planar constellation coil that was used in 4-channel parallel Tx and Rx MRI of a phantom. **Fig.1b** is a close-up showing the grid structure that was fabricated by etching 4mm x 200mm conductors on double-sided FR-4 circuit board with 4mm gap between conductors and the top layer conductive traces rotated 90° with respect those on the bottom layer. **Fig.1c** illustrates capacitive coupling between the conductors on the two sides and the introduction of a port with a matching network (shown in red). Phantom imaging results were obtained on an 8-channel parallel transmit and receive 7T scanner with a GRE sequence without parallel Rx acceleration (**Fig.1d**, left column) and with 2x acceleration (**Fig.1d**, right column). Four spatially arbitrary ports were introduced to the planar array. The ports were not tuned in the traditional sense, but simply matched to 50Ω through single stage LC networks as required for minimum preamplifier noise figure during Rx and maximum power delivery during Tx. The capacitive links in the structure are important as it is desirable to make the capacitive reactance relatively small at the Larmor frequency but large at audio frequencies, in order to allow RF current flow while disallowing gradient-induced currents. This capacitance can be estimated as $C = \varepsilon A/d$, where ε is the substrate dielectric constant, A is the area of conductor overlap, and d the substrate thickness. In the present example $C = 4.34\varepsilon_0 \cdot 16\text{mm}^2/0.078\text{mm} = 7.9\text{pF}$ – the dense grid structure approaches a continuous conducting structure for RF currents, making the structure behave like a RF shield. This continuum provides the support for accommodating sophisticated RF current patterns required to support field optimization. Meanwhile the structure remains transparent to the gradient field.



Creating a continuum 3D structure is desirable. In a second example, we wrapped a flexible PCB with a similar grid pattern (4.5mm x 282mm traces with 4.5mm gap, rotated 45° with respect to the main cylinder axis giving 200mm z-length) around a ∅20cm cylindrical former (**Fig.2a**). We introduced 8 ports separated azimuthally by 45° at the mid section of the coil and matched each port to 50Ω through LC circuits.

Volunteer imaging results were obtained on the 7T scanner with a GRE sequence without parallel Rx acceleration (**Fig.2b**) and with 2x acceleration (**Fig.2c**). Additional parameters include: 150V Tx voltage, TR=220ms, TE=5.4ms, number of signal averages=1, 230x320 matrix, FOV=21.6x30.0cm^2. For comparison, imaging results from the same volunteer were obtained with same transmit and sequence parameters, but with a conventional 8-element stripline coil approximately driven in CP mode (**Fig.2d-f**). In a quantitative study using a loading phantom, data indicated that the constellation coil could produce larger average |B1+| at several tested axial slice locations given same total output power from the 8 RF power amplifiers. Additional measurements are needed to provide a comprehensive assessment, including parallel Tx / Rx performance. Further investigations will also include optimizing port locations, assessing performance on a 3T scanner, and deploying many more ports.



Discussions: These examples illustrate the rational behind the new coil approach. First, a conventional array coil's structure tends to be too discrete to support sophisticated current patterns needed for approaching ultimate performance allowed by electrodynamics, and coupling / mode structure put additional constraints on current flow. Structural components of a constellation coil are well coupled, approximating a mesh-type generic RF shield to maximize support for complex / broadband RF current patterns without concerns for decoupling or mode structure. Desired current patterns are realized by driving / receiving from the constellation coil structure through distributed multiple ports (the constellation), as analyzable with Kirchhoff's current law. In practice current patterns are in effect generated with parallel Tx RF pulses / parallel Rx image reconstruction which are predetermined based on calibrated $B1^+$ / $B1^-$ sensitivity maps and power- / noise-correlation matrices. Explicit knowledge of current patterns or special RF pulse design / image reconstruction is not required. Second, RF loss may be better managed in the constellation coil case. Direct E field coupling to the sample, a significant loss mechanism in high field MR, is expected to be reduced with the amount and density of distributed capacitance across the coil structure. The densely populated conductors collaborate in conducting currents and tend to reduce the resistance seen by the RF currents. Promoting geometrically confined currents as is the case with a conventional resonate structure may not be efficient. While the B1 field increases, so does the E field and RF loss. Third, for a conventional coil to work robustly, sophisticated design, construction, tuning and testing are required. Substantial challenges include tuning of individual element coils in the presence of complex interactions amongst them, integration of a large number of parts (capacitors, inductors and blocking circuits), and the management of failure modes. The fact that one structure may not perform adequately for Tx, Rx, various parts of the body, multiple frequencies and various parallel MR acceleration configurations, further leads to the need for a large arsenal of coils that are costly to operate and maintain in clinical practice.

The new coil approach reflects a belief that field optimization-based performance enhancement is a sensible way to go. The approach targets SNR and SAR performance enhancement based on electrodynamics principles, focusing on optimizing current distribution that is the foundation of the performance-dictating B1 and E fields. Its simple construction and versatility may also positively impact the fabrication and use of RF coils in MR.

References: 1. Y. Zhu, *14th ISMRM*, p 599, 2006. 2. R.Lattanzi, et al., *16th ISMRM*, p 614, 2008.