A mechanically rotating RF transceive system and method for applications in Magnetic Resonance

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Introduction: A new concept relating to a fast mechanically rotating radio-frequency (RRF) transceive coil and method for application in MRI is presented. The introduction of controlled RF system rotation as additional degree of freedom during imaging facilitates the use of (at least) one coil sensitivity pattern to effectively emulate a conventional RF coil array over time. This approach aims to substantially diminish the number of coil elements and channels, lumped circuit components and tedious mutual decoupling while preserving the benefits of very large RF coil arrays. This feasibility study presents and discusses first experimental results of the RRF system in *in-vitro* imaging and spatially selective excitation.

Methods: This RRF approach is based on the notion that the Larmor frequency of the excited nuclear spin is at least six to seven orders of magnitude larger than the angular frequency of the rotating RF coil. Therefore, in the time reference frame of the precessing spin, the RRF coil and its RF field sensitivity distribution may be perceived as physically stationary. As the magnetization signal is decaying due to T_1 and T_2^* relaxation phenomena, the high-speed RRF coil should be perceived as multiple stationary RF coils analogous to a multi-element transceive phased coil array.

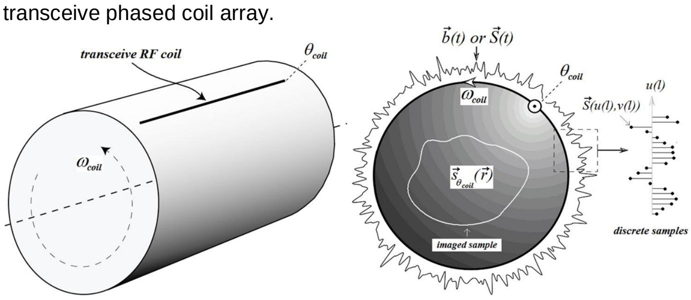

Fig. 1. Illustration of a RF coil that rotates about a samples at the angular frequency ω_{coil} during which the coil 'illuminates' the sample with a RF field profile $\mathbf{s}_{\omega t}$.

Consider a transceive RF coil in Fig.1 situated on a hollow cylinder that rotates around its axis at the angular frequency $\omega_{coil} = \theta_{coil} / t$. At any point in time of one coil revolution, the rotating RF coil creates a spatially unique field profile $\mathbf{s}_\theta(\mathbf{r}) \equiv \mathbf{s}_{\omega t}(\mathbf{r})$ as a function of θ_{coil}, where $\mathbf{s}_\theta(\mathbf{r})$ is equivalent to a field profile produced by a fixed coil positioned at θ_{coil} in a conventional RF coil array. In contrast to the standard expressions for net magnetization excitation $\mathbf{M}_{exc}(\mathbf{r})$ and MR signal $\mathbf{S}(t)$ detection, according to Eqs.(1) and (2), the rotating RF coil adds an additional degree of freedom for the manipulation of the spin, signified by the subscript ωt in the weighting (filter) function $\mathbf{s}_{\omega t}(\mathbf{r})$:

Transmission:
$$\mathbf{M}_{exc}(\mathbf{r}) = i\gamma\int_{T_b} \mathbf{b}_1(t)\mathbf{s}_{\omega t}(\mathbf{r})\mathbf{M}_0(\mathbf{r})e^{-2\pi i\mathbf{k}(t)\cdot\mathbf{r}}dt \qquad \text{Eq.(1)}$$

Reception:
$$\mathbf{S}(t) = \mathbf{S}(\mathbf{k}(t)) = \int_R \mathbf{s}_{\omega t}(\mathbf{r})\mathbf{M}_{exc}(\mathbf{r},t)e^{-2\pi i\mathbf{k}(t)\cdot\mathbf{r}}d\mathbf{r} \qquad \text{Eq.(2)}$$

When Eqs.(1) and (2) are written in discrete forms to comply with the digital sampling theory, in one period of revolution, the rotating RF coil emulates $N=2\pi f_s/\omega_{coil}$ virtual stationary coils, where f_s is the digital sampling frequency of the RF coil signal. To test the practical viability of the proposed method, a prototype 2T shielded RRF system for small animal imaging was designed and constructed, as shown in Fig. 2. All experiments were conducted in a 2T whole-body MRI system (Centre for Magnetic Resonance, University of Queensland, Australia).

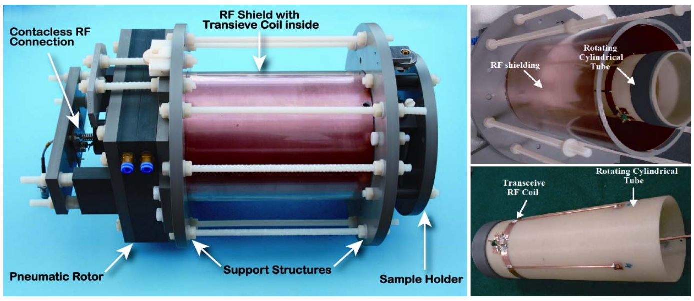

Fig. 2. The rotating RF coil platform measuring about 335mm in diameter and 500mm in length. To avoid magnetic field and eddy current interactions when operated within the 2 Tesla MRI system, most of the system was assembled from plastic materials.

A custom-designed, Tesla turbine (*1*) was used to rotate a Ǿ75mm RF coil. A second, fixed concentric cylinder of Ǿ65mm then acted as the sample holder. An RF shield reduced power losses and a frictionless inductively-coupled RF link connected the rotating transceive coil to the MRI system. The rotational speed of the coil was governed by air pressure drive to the Tesla turbine and measured with an infrared photo-interrupter. This scheme constituted an open-loop system and the observed maximum rotational speed was about 1500rpm.

Results and discussion: Fig.3 shows images and simulated results of a homogeneous saline-based cylindrical phantom when the coil was stationary and rotating at 1104rpm.

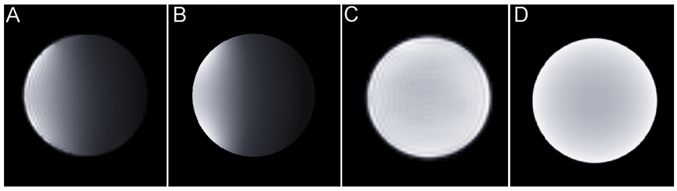

Fig. 3. Axial MR images of the homogeneous cylindrical solution sample (σ = 0.2 S/m and ε_r = 76) using the following FLASH sequence parameters: time of repetition (TR) of 70ms, slice thickness of 5mm, FOV of 25x25cm, matrix size of 256x512, flip angle (FA) of 45^o and a truncated Gaussian pulse of 3ms in duration; Experimental **(A)** and simulation result **(B)** with RRF transceive coil firstly positioned to the left side of the phantom prior to spinning it; Only part of the sample was imaged due to the limited FOV of the RF coil **(C)** Complete, full FOV image of the same sample obtained by rotating the RF transceive coil at a constant speed of 1104rpm; **(D)** Simulated image result of a equivalent stationary 2717 coil element array.

Considering the FLASH sequence parameters, f_s = 50kHz, 1104rpm and $N=2\pi f_s/\omega_{coil}$, we were able to estimate that the single rotating RF coil has emulated a conventional *N*-element array consisting of 2717 coils per revolution. Building an equivalent stationary array for verification purposes would be impractical. Using a hybrid Method-of-Moments/Finite-Element scheme (*2*) instead, we have simulated a 2717 element stationary RF coil array, where the results from each coil were combined to produce the overall image as shown in Fig.3D. The experimental and simulated results in Fig.3 are consistent.

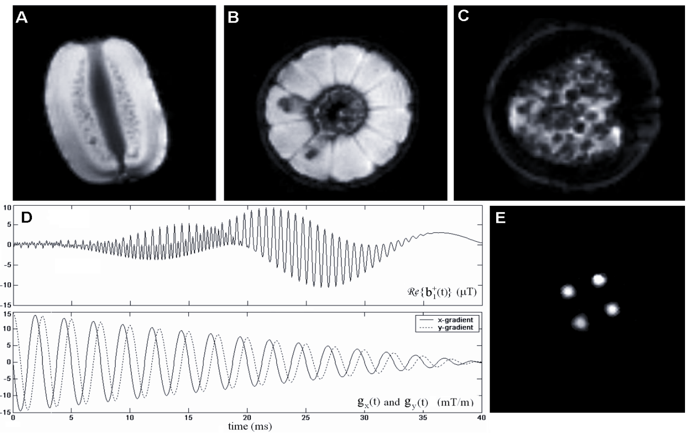

Fig. 4. Images of heterogeneous samples obtained by rotating RF transceive coil at 1200rpm: **(A)** kiwi fruit, **(B)** mandarin and **(C)** passion-fruit. The FLASH sequence parameters were as follows: TR=65.8ms, slice thickness of 3mm, FOV=17x17cm, matrix size of 256x512, FA=33^o and 6ms sinc pulse. **(D)** Tailored RF and gradient waveforms for spatially selective excitation in a sample of Fig.3. **(E)** The excited four Gauss shapes. MSME sequence parameters were as follows: TR=80.2ms, time of echo TE=38.5ms, slice thickness of 5mm, FOV=22x22cm, matrix size of 256x512, FA=60^o.

The second experiment imaged heterogeneous biological samples. The acquired images, as shown in Fig.4A-C, exemplify high uniformity and well-resolved anatomical structures of the samples at ~1200 rpm. In the third experiment, multi-dimensional RF pulses and gradient field waveforms (see Fig.4D) were synthesised based on Eq.(1) and the MSME imaging pulse sequence to selectively excite four small Gauss profiles. Thereafter, the RRF system was operated at ~1200 rpm, and the pulses were transmitted during a time period of 40ms. Fig.4E shows the result of this spatially selective excitation.

Conclusion: The results suggest that a single rotating transceive RF coil can be used to acquire MR images. The benefits of such a scheme are that coil coupling interactions and the requirement for multiple signal pathways cease to be a practical impediment. We are currently developing accelerated imaging techniques using the RRF approach.

Acknowledgements: Financial support from the Australian Research Council is gratefully acknowledged.

References:

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