Rotating sample acquisition in ultra-low-field MRI

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TARGET AUDIENCE Scientists interested in data acquisition methods—in particular—in ultra-low-field MRI

PURPOSE Ultra-low-field (ULF) MRI uses a magnetic field in the microtesla range for magnetization precession [1] and has advantages including acoustically silent acquisition, low projectile danger, safe operation, imaging compatibility with metal objects, and the possibility to allow open access**.** A technical challenge of ULF-MRI is the concomitant field effect [2-5], which causes image blurring and distortion if the image is reconstructed assuming that consider the magnetic fields generated by gradients are perfectly linear and with only the *z*-component. The associated blurring and distortion artifacts can be reduced by 1) carefully calibrating the local magnetization phase and the precession frequency, or 2) carefully describing the relationship between the received signal and magnetization dynamics in spatial-time domain [6] or in spatial-frequency domain [7]. ULF-MRI also suffers from low signal-to-noise ratio (SNR), which can be improved by a separate and strong polarization field [8] and using sensitive magnetic field detectors such as superconductive quantum interference devices (SQUIDs).

We propose the rotating sample acquisition (RSA) and applied it to ULF-MRI. Specifically, the sample rotates in the instrument and ULF-MRI signals are measured without any phase encoding. Because of using only frequency encoding, the concomitant field artifacts can be minimized. Using empirical data, we demonstrate that RSA is more efficient than conventional Fourier encoding when only a few SQUID sensors are used. The RSA method holds promise for a portable imaging system using only frequency encoding gradients and a small number of NMR signal detectors.

METHODS In RSA, ULF-MRI data are collected by repeating the process of first acquiring projection data generated by a frequency encoding gradient and then rotating the sample incrementally. Therefore, both the magnetic field and the coil sensitivity experienced by the sample change across measurements. In 2D imaging, the signal in one read-out when the imaging object rotated at $\boldsymbol{\theta_n}$ is:

$\boldsymbol{S(\omega_l,\theta_n)} = \int_{\mathbf{R}} I(\boldsymbol{x}\cos\boldsymbol{\theta_n} + \boldsymbol{y}\sin\boldsymbol{\theta_n}, x\sin\boldsymbol{\theta_n} - \boldsymbol{y}\cos\boldsymbol{\theta_n})\cdot C_{\boldsymbol{m}}(\boldsymbol{x},\boldsymbol{y})\cdot \mathbf{sinc}\left(\frac{\omega(x,y)-(l\Delta\omega+\omega_0)}{\Delta\omega}\right)\boldsymbol{dxdy}$, where $\boldsymbol{S(\omega_l,\theta_n)}$ is the acquired signal at frequency $\boldsymbol{\omega_l}$, $\boldsymbol{\omega(x,y)}$ is the magnetization precession frequency at (*x*,*y*), *I* denotes the magnetization distribution, C_m is the m^{th} coil sensitivity and $\boldsymbol{\Delta\omega}$ is the bandwidth per read-out pixel. The index $\boldsymbol{l}$ =1…*L* indicates the l^{th} frequency step. The index *n* = 1…*N* indicates the n^{th} rotation. Figure 1 shows the experimental setup and the phantom. In experiments, the magnetization was generated by a 20 mT pre-polarization field for 4 seconds. The pre-polarization field was then adiabatically turned off and a standard spin-echo pulse sequence was used to sample ULF-MRI data at 10 kHz for 250 ms (TE = 300 ms). The frequency-encoding gradient (50 μT/m) was turned on constantly. No phase encoding was used. The sample was rotated by 3.36^0 between measurements (*N* = 106) and measured with 9 averages. The coil sensitivity was measured using a homogeneous phantom and fitted by polynomials. Randomly selected 100 measurements were used to reconstruct images by solving the signal equation numerically with Tikhonov regularization.

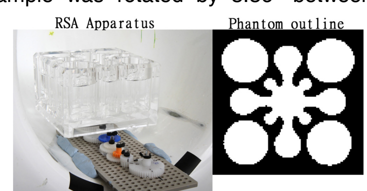

Figure 1

For comparison, we also acquired ULF-MRI data using conventional Fourier encoding with 10-fold averaging and 90 phase encoding steps of up to 75 μT/m and 80-ms duration to achieve about 2 x 2 mm^2 resolution. Fourier-encoded data was reconstructed by the 2D fast Fourier transform. For both the RSA and Fourier-encoded images, 900 read-outs were used. To investigate the performance of image reconstruction using accelerated acquisitions, we reconstructed images with 50% (50 angles with 9 averages in RSA; 45 equally-spaced *k*-space lines in Fourier encoding with 10 averages; ACC = 2) and 33% (34 angles with 9 averages in RAS; 30 equally-spaced *k*-space lines in Fourier encoding with 10 averages; ACC = 3) of data.

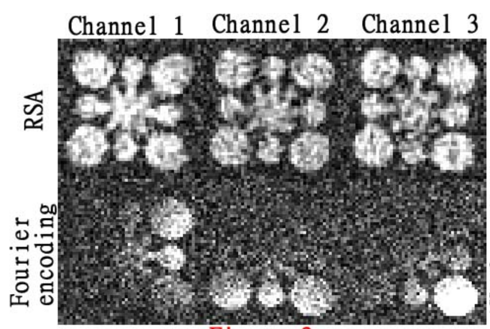

Figure 2

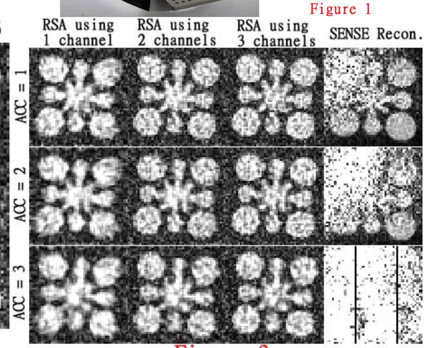

Figure 3

RESULTS Figure 2 shows three channels of the data reconstructed using RSA and Fourier encoding. It is evident that even with only one channel, RSA can provide sufficient data to generate a full-FOV image with some distortion. Figure 3 shows the reconstructed images using 1, 2, or 3 channels of RSA data (left, middle left, middle right columns) and the SENSE [9] reconstruction using three-channel Fourier-encoded data (right column). The SENSE reconstruction was found to have less noise around the sensor (close to the bottom and right margin of the image). However, SENSE reconstructions were noisy and failed to reconstruct images at 3-fold acceleration. Progressive improvement of SNR was found in reconstructions using RSA as the number of channel increased. Even at 3-fold acceleration, RSA generated reasonable image content, regardless of whether 1, 2, or 3 channels were used.

DISCUSSION We demonstrate that the RSA method is more efficient than Fourier encoding in ULF-MRI. Specifically, RSA used no phase encoding gradient, and generated reasonable images with only 33% of the fully Fourier encoded data from spatially clustered sensors. For a spatially homogeneous coil sensitivity and a linear read-out magnetic field, RSA is equivalent to radial trajectory. It should be noted that, in ULF MRI, a radial trajectory implementation is complicated by the fact that different projections have different concomitant and remanent fields. Consequently, RSA may be a more feasible approach to implementing projection reconstruction in ULF MRI. We expect that RSA can also work in other experimental setups, if 1) there is a static magnetic field with gradient to provide spatial information, 2) the sample or the instrument can rotate stably, and 3) the coil sensitivity is invariant (or known) when the sample/instrument rotates. Therefore RSA may be useful in low-field or portable MRI.

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