Gradient and frequency modulated excitation for a tailored spatial trajectory with two-dimensional time encoding for Fourier-free imaging

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INTRODUCTION: In recent years, time-encoded MRI has been implemented in several forms (1-2). However, all of these implementations involve at least one dimension of frequency or phase encoding with Fourier transform (FT) reconstruction, preventing them from being purely temporal-spatial. Described here is a new methodology for two-dimensional time encoding (2DTE) using only time encoding to produce an image by moving a resonance region along a trajectory in space. Accordingly, no FT is needed for reconstruction, offering the unique capability to treat each region independently, and thus creating the possibility to directly counter spatial imperfections, such as B0 and B1 inhomogeneities.

THEORY: For initial demonstration, a single spin echo implementation with an Archimedean spiral and constant radial velocity over 2π radians is presented. In this case, a chirped frequency sweep terminating at zero frequency offset is applied concurrently with gradients on two axes driven by sinusoids with 90° phase difference. The resonance point follows a spiral trajectory in space, from the maximum radial distance to the center as shown in Figure 1b. A RF 180° pulse is applied to reverse the phase evolution, and during this pulse a slice selective gradient may be applied. A hard pulse or sinc pulse is used for refocusing in order to avoid introducing additional spatially varying phase. During acquisition, the gradient functions are reversed which results in a sequential local refocusing of magnetization along the trajectory, from the center to the maximum radial distance. The sequence can be repeated, changing the angle each time to produce a series of spirals as shown in Figure 1b (dotted lines), and thereby covering the plane (i.e., the selected slice). The excitation is not in fact a point, but a forward-moving region of excitation traveling along a vector, exciting a plane in space perpendicular to that vector. In 2D, the wave is a line, with the tangential velocity driving the wave along the trajectory and the rotational velocity driving the rotation of the line about its center point. The instantaneous velocity of a point along the line can be described by $ds/dt \propto (r^2+L^2)^{1/2}$ where s is the position, t is time, r is the distance between the center of the spiral and the center point of the line of excitation, and L is the distance of a given point along the line from the center point of the line. Thus the center point has the lowest velocity, and velocity increases the further a point is from the center, quadratically if $|L| < r$ and linearly if $|L| > r$.

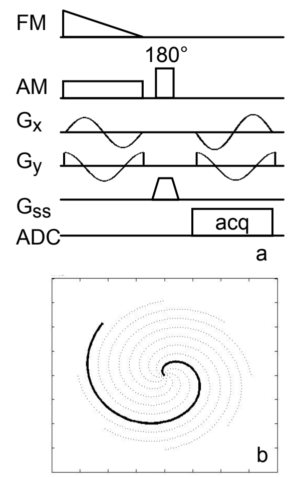

Figure 1 a) 2DTE pulse sequence b) spiral trajectories

METHODS: A 4T magnet with a 90 cm bore (Oxford) was used with a clinical gradient system (Siemens Sonata) and an imaging console (Varian) for all experiments. The phantom is a 50 ml tube containing water and Gd-DTPA positioned upright in a quadrature T/R surface coil, the scout image of which is shown in Figure 2d. The experiment was performed with b_W = 40 kHz, pulse length = acquisition window = 6.0 ms, TE = 21 ms, TR = 4 s, and FOV=20 cm, 240 points along the trajectory, and 128 rotations. To validate experimental results, simulations were performed using Bloch simulations on Matlab. The simulations were adjusted to obtain data in the same manner as the experiment. Figure 2a shows the simulated object.

Because the excitation profile is not sufficiently close to a delta function, reconstruction cannot be done by assigning signal along the trajectory according to the known spatial-temporal relationship, which has been used along one dimension with previous time-encoding sequences (1-2). Figure 2b,e shows the results using the simple assignment reconstruction method, demonstrating its inadequacy. The figure also shows strong correlation between the data from experiments and the data from simulations, suggesting that an inverse mapping problem could be employed for reconstruction by using simulations to generate a transfer matrix. The general form of the inverse mapping problem comes from earlier literature and has been employed successfully with other imaging modalities (3,4) and has been employed in various forms in a few MRI applications (5-8). Consider first the forward problem $\boldsymbol{g}=\boldsymbol{Hf}$, where $\boldsymbol{g}$ is a vector describing the observed data, $\boldsymbol{f}$ is a vector describing the source signal, and $\boldsymbol{H}$ is a transfer matrix. In implementing the inverse problem solution for 2DTE, the afore-mentioned Bloch simulations were used to approximate the forward problem and thereby obtain an estimate of $\boldsymbol{H}$. The pseudoinverse ($\boldsymbol{H}^+$) can be calculated using the least-squares solution of the inverse mapping with functions available in Matlab or the Regularization Tools package (9). The pseudoinverse can then be applied to the experimental data to generate an image which is the estimate of the source signal, ($\boldsymbol{f}_{est}= \boldsymbol{H}^+ \boldsymbol{g}$).

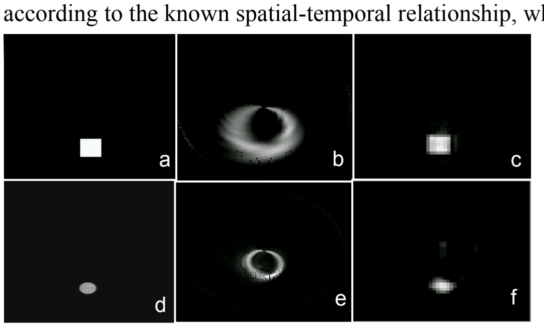

Figure 2 a) simulated object, b) simulation results reconstructed with an assignment method, c) simulation results reconstructed with an inverse problem method, d) scout image of the phantom used in the experiment, e) experimental results reconstructed with an assignment method, f) experimental results reconstructed with an inverse problem method

RESULTS: While not a highly accurate depiction of the object, Figure 2b,e shows strong consistency between the simulation and experiment. The improvements using the inverse problem solution are apparent comparing the assignment reconstruction to the inverse mapping problem reconstruction both in simulation (Figure 2b-c) and experiment (Figure 2e-f). Admittedly, some artifacts are present.

DISCUSSION: Experimental results demonstrate imaging by moving a resonance region in the spatial domain, acquiring in a time dependant manner, and reconstructing without the use of a Fourier transform. The challenge of addressing the artifacts remains, but numerous regularization and filtering options can be explored for improving the images and opening a myriad of possibilities for applications where spatial independence is important, such as for B0 and B1 inhomogeneity compensation.

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