Multidimensional Spatial Encoding by Parallel Excitation

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Introduction: In classical MRI and spectroscopic imaging spatial information is encoded into NMR signals by means of magnetic gradients which are applied after exciting the spins by means of RF pulses so that delayed signal reception leads to signal loss due to relaxation. Additional spatial information can be added by using spatially selective RF pulses, e.g. thus confining the source volume of the received NMR signals. Further, for simultaneous selective excitation of separate volumes 1-dimensional encoding schemes for phase encoding these volumes during the excitation have been used to resolve the received signal's source volume [1,2]. This study describes a method of MRI and spectroscopic imaging which allows multidimensional spatial encoding during the excitation period. It is called SPEEDI (Spatial Encoding by Excitation with multidimensional RF Pulses) and uses multidimensional RF pulses [3] not only for spatial selection and modulation of the transverse magnetization magnitude but also for defining spatial distributions of the transverse magnetization phase. In order to alleviate the disadvantage of long RF pulses parallel excitation techniques [4,5,6] have been used to perform a first experimental realization of SPEEDI.

Theory: The basic principle of SPEEDI is to realize spatial encoding by generating specific relative phase distributions of the transverse magnetization $M(\mathbf{x})$ during the excitation period of an MRI experiment for a series of phase encoding steps. If this relative phase distribution is preserved – in principle - in a subsequently acquired FID or spin echo, images with different contrasts or spatial distributions of spectral information can be reconstructed. When playing out multidimensional RF pulses using a transmit array the generated $M(\mathbf{x})$ can be expressed as a linear, sensitivity-weighted combination of virtual single-element patterns $M_n(\mathbf{x})$:

$$M(\mathbf{x}) = \sum_n S_n(\mathbf{x})\, M_n(\mathbf{x}), \quad \text{where} \quad M_n(\mathbf{x}) = i\gamma M_0(\mathbf{x}) \int_0^T B_{1,n}(t)\, e^{i\mathbf{x}\cdot\mathbf{k}(t)}\, dt \qquad (1)$$

(S_n is the sensitivity of the n-th array-element in which the RF-waveform $B_{1,n}$ is played; $\mathbf{k}(t)$ is the (reduced) k-space trajectory traversed during RF-transmission). To set up a spatial encoding scheme a different phase distribution of $M(\mathbf{x})$ is defined for each phase encoding step, while the amplitude distribution of $M(\mathbf{x})$ is kept constant and can be chosen according to requirements re spatial selectivity and modulation, e.g. a uniform distribution within a small field of view (FOV). For each phase encoding step a specific set of RF-waveforms $B_{1,n}$ has to be calculated. The approach proposed in [7] applies a direct temporal and spatial discretization of the equations (1) that leads to the following linear system:

$$\mathbf{m} = \mathbf{A}\mathbf{b}, \quad \text{where} \quad A_{u,(n,v)} = i\gamma M_0(\mathbf{x}) S_n(\mathbf{x}_u) B_{1,n}(t_v) e^{i\mathbf{x}_u\cdot\mathbf{k}(t_v)} \Delta t \qquad (2)$$

where u and v are the spatial and temporal indices respectively. For a given magnetization vector $\mathbf{m}$ the waveforms $\mathbf{b}$ can be calculated by solving the linear system (2).
Due to the spatial discretization the targeted transverse magnetization has to be defined for a grid of excitation cells exactly covering the field of excitation (FOX) which in turn has to include the object to be studied. Within this FOX, according to the region to be imaged, a FOV is defined which can have any (2D or 3D) shape. Inside this FOV $|M(\mathbf{x})|$ is set for the excitation cells according to a given modulation, e.g. a uniform distribution, while zero-amplitudes are assigned to all other excitation cells. Phase encoding according to the given spatial encoding scheme affects only the excitation cells within the FOV. In general, this encoding scheme will provide a set of codes each of which is assigned in a unique manner to a certain region within the FOV. Each code is given by a tuple of phase values each of which corresponds to the (relative) phase to be set for $M(\mathbf{x})$ within the assigned region in the respective phase encoding step. This prescription defines how to set the $M(\mathbf{x})$ phase values of all excitation cells within the FOV for all phase encoding steps. Combining several excitation cells to one so-called coding cell and assigning the same code to each member of this coding cell allows reducing the number of required phase encoding steps.
The following advantages and potentials of this approach should be emphasized: (1) In that the entire spatial encoding can be completed within the excitation period SPEEDI is suitable for imaging nuclear spins with very short relaxation times, if accordingly short RF pulses can be used. (2) SPEEDI is insensitive to imperfections of the gradient pulses since all phase encoding steps are performed with the same k-space trajectory (isomagnetic phase encoding) and the variation of the phase distributions it achieved exclusively by changing the RF pulses played out. (3) Due to the fact that excitation and spatial encoding can be restricted to subvolumes SPEEDI is appropriate for imaging small substructure within bigger objects (microscopy). (4) As arbitrary phase distributions can be realized SPEEDI offers many options re the choice of the spatial encoding scheme, completely unknown in conventional MRI. In order to assure high imaging fidelity the mapping of codes to coding cells has to comply with certain conditions, e.g. re code interpolation effects. (5) The coding cell concept permits establishing a non-uniform spatial resolution or to map distinct locations within the object on the same image position.

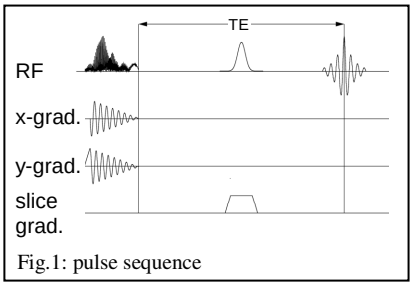

Fig.1: pulse sequence

Materials, Methods and Results: Our first experimental verification of the SPEEDI concept was performed on a 4-TX-channel 9.4 T, 20 cm bore BioSpec system (Bruker BioSpin MRI, Ettlingen, Germany) using a 4-element TX/RX volume array. As k-space trajectory a constant density, constant angular velocity 16-turn spiral was chosen. Test object was a cylindrical bottle with a T_1-doped saline water solution and 2 immersed plastic bars. As proof of principle we acquired a 2-dimensional (2D) image of a small FOV by realizing 2D spatial selection, amplitude modulation and phase encoding during the excitation with a 2-fold accelerated 2D RF pulse and subsequently forming a slice-selective echo (Fig.1). Fig.2 shows a pilot image taken with a conventional MRI method. It was used to set the FOX (5.6 cm x 5.6 cm) and FOV (1.4 cm x 1.4 cm), as indicated, for the SPEEDI experiment. Within this FOX a rectangular grid of 64x64 square excitation cells was defined while within the FOV 8x8 coding cells were fixed by combining 2x2 adjacent excitation cells. As encoding scheme a 2D Fourier scheme with 64 phase encoding steps was chosen and for each phase encoding step the RF pulses (6.6 ms length) were calculated as described above. For each phase encoding step a slice selective spin echo sequence was executed according to Fig.1 (TE=8ms). Only a single complex data point in the centre of the echo was recorded per excitation. A 2D FFT reconstruction resulted in an image as shown in Fig.3 and, as superposition on the pilot image, in Fig.4. Fig.5 shows the phase distribution realized in one of the phase encoding steps inside the FOX as imaged with a classical spin echo sequence. In Fig.6 the phase-difference distribution of 2 subsequent phase encoding steps is depicted.

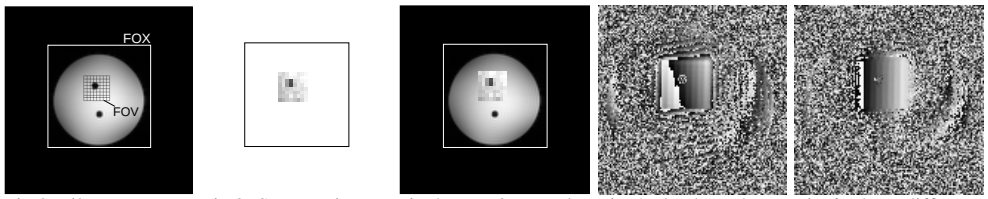

Fig.2: pilot scan Fig.3: SPEEDI image Fig.4: superimposed Fig.5: absolute phase Fig.6: phase difference

Discussion and Conclusions: The results of our proof-of-principle experiments demonstrate that it is feasible to achieve multidimensional spatial encoding for a restricted FOV within an extended object by means of Parallel Excitation. The resulting images nicely correlate with conventional imaging, but have limited spatial resolution. Therefore, the next efforts will focus on increasing the spatial resolution, e. g. by using more transmit elements. Further, the above mentioned potentials of SPEEDI are to be verified systematically.

References: [1] W. Dreher, D. Leibfritz, MRM 1994; 31:596-600; [2] P. Börnert, MAGMA 2003; 16:86-92; [3] J. Pauly et al., JMR 1989; 81:43-56; [4] U. Katscher et al., MRM 2003; 49:144-150; [5] Y. Zhu, MRM 2004; 51:775-764; [6] P. Ullmann et al., MRM 2005; 54:994-1001; [7] W. Grissom et al., MRM 2006; 56:620-629