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MRI using Radio-Frequency Magnetic Field Phase Gradients

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Abstract

Conventionally, MR images are formed by applying gradients to the main magnetic field (B_0). However, the B_0 gradient equipment is expensive, power-hungry, complex and noisy, and can induce eddy currents in nearby conducting structures – including the patient. Here we describe a new silent, B_0 gradient-free MR imaging principle ‘TRASE’ (TRAnsmmit Array Spatial Encoding) based on phase gradients of the radio-frequency (RF) field. RF phase gradients offer a new method of k-space traversal. Echo trains using at least two different RF phase gradients allow spin phase to accumulate, causing k-space traversal. Two such RF fields provide 1D imaging and three are sufficient for 2D imaging. Since TRASE is a k-space method, analogues of many conventional pulse sequences are possible. Experimental results demonstrate 1D, 2D RF MRI and slice selection using a single channel transmit/receive 0.2 Tesla permanent magnet human MR system. The experimentally demonstrated spatial resolution is much higher than that provided by RF receive coil array sensitivity encoding alone, but lower than generally achievable with B_0 gradients. Potential applications are those in which one or more of the features of: simplified equipment, lower costs, silent MR imaging or the different physics of the image formation process are particularly advantageous.

Keywords: MRI, Radio-frequency, RF phase gradient, low field, TRASE

Introduction

MR imaging relies on the introduction of spatial dependence into the NMR experiment. The conventional form this takes is of audio-frequency modulated magnetic field gradients in the main magnetic field (B_0), (1-3). This requires the installation of a set of coils around the patient, driven with high current waveforms to produce three linear B_0 gradients (G_x, G_y, G_z), which add to the main magnetic field and so modify the resonant frequency depending on both spatial position (x, y, z) and (t) time like so: $f(x, y, z, t) = -(\gamma / 2\pi)(B_0 + xG_x(t) + yG_y(t) + zG_z(t))$, where $G_x = \partial B_z / \partial x$, etc. This imaging approach has proven highly successful, but incurs substantial costs and system complexity. Typical B_0 gradient system hardware requirements include waveform generation electronics, high power amplifiers, filters, cables, and heavy water-cooled self-shielded gradient coils. The B_0 gradient coils themselves occupy valuable space within the magnet bore. Auxiliary to this primary gradient system equipment are various support systems required on site including a 3-phase power supply and air and water cooling systems for the power electronics and gradient coils. As a result, a large fraction of the purchase, installation and operation costs incurred by MR systems are due to the B_0 gradient system.

The operation of a B_0 gradient system leads to some undesirable consequences. The pulsing of the B_0 gradient coil current within the main magnetic field gives rise to loud acoustic noise and vibration which can be disturbing for patients and may require safety mitigation measures (4). The production of switched B_0 gradients results in induction of eddy currents in any nearby conductors (including the patient), and may result in image artifacts. B_0 gradient switching rates are limited by regulatory agencies to avoid painful or potentially harmful nerve stimulation.

We present here a new MR imaging approach, which offers an alternative to B_0 gradient-based imaging. The new method ‘TRASE’ (TRAnsmit Array Spatial Encoding) makes use of a new type of gradient - the RF phase gradient. K-space traversal is achieved

by playing out echo train pulse sequences with each RF refocusing pulse generated using one from a small set of phase gradient RF transmit fields.

Rotating Frame Imaging

In the very early days of MRI, an alternative (and silent) imaging strategy based on gradients in the transmit radiofrequency (RF) field was proposed by Hoult (5). Specifically, gradients in the amplitude of the RF B_1 field were used to encode spatial information into the NMR signal via a spatially dependent flip angle ('Rotating Frame Zeugmatography'). The technique was successfully applied to localized spectroscopy, and was especially useful for 1D surface coil experiments (6,7), but did not have an impact on clinical imaging. The principle limitations were slow scanning speed, as only one data point is collected per excitation, and very high peak RF power requirements as large flip angle pulses ($>90^\circ$) are needed.

A more recent high speed rotary echo version of rotating frame imaging (8,9) has addressed the imaging speed and RF power issues by using a train of rotary echoes produced by very short RF pulses acting in effect as a readout gradient. The technique has been demonstrated for micro-imaging of small samples but has not been shown to scale up for in vivo imaging. A review by Canet (10) suggests that RF imaging techniques are under-used, suggesting one reason being artifacts arising from non-compensated off-resonance effects. Although rotating frame RF methods are able to selectively excite a single slice (11), an important limitation is that the entire sample (not just the selected region) is perturbed from equilibrium, so time interleaved multi-slice imaging - a key clinical MR technique allowing rapid and flexible access to the third spatial dimension - is not possible.

Parallel Imaging

In contrast to the very early transmit RF imaging approaches, much more recently imaging using arrays of receive fields has been highly successful clinically in providing low resolution spatial information to accelerate MR acquisitions when used in conjunction with B_0 gradient encoding (12,13).

Use of highly parallel arrays for very high speed acquisition without B_0 gradients in a volume at low resolution (14), or in a surface imaging configuration has also been recently demonstrated (15). Parallel transmission using a transmit array ('Transmit-SENSE') has also been used to speed up B_0 gradient based slice or volume excitations (16,17). The method proposed here, Transmit Array Spatial Encoding (TRASE) (18-20), differs from these parallel imaging methods in that it provides high spatial resolution volume imaging using RF transmit array fields without any B_0 gradients. The spatial resolution offered by TRASE is much higher than that available from RF receiver coil array sensitivity encoding methods, which are limited to providing spatial information on the order of the coil array element size. TRASE however operates on a phase accumulation principle which offers spatial resolution much finer than the coil dimensions, although not as high as the best resolution achievable with B_0 gradients.

TRASE requires neither a multi-channel transmitter nor the generation of a large number of different transmit fields. It differs fundamentally from rotating frame imaging as the spatial dependence of the phase, rather than the amplitude, of the RF field is exploited. We will show that the introduction of this third type of magnetic field gradient retains many of the powerful capabilities of conventional MRI – including simple k-space traversal and multi-slicing. Table 1 compares the three types of gradient.

Theory

Frequency Encoding

We first recap the conventional imaging approach. For a 1D imaging experiment consisting of a 90° excitation ('flip') followed by a constant B_0 gradient applied for a time T , the detectable magnetization is given by:

$$M_{xy} = M_0 e^{i\phi(\mathbf{r})} = M_0 e^{i\gamma T G \cdot \mathbf{r}} = M_0 e^{i2\pi \mathbf{k} \cdot \mathbf{r}} \quad (1)$$

where $\mathbf{k}$ is the spatial frequency coordinate (cycles/m). The NMR signals corresponding to different values of $\mathbf{k}$ contain the information needed to build up an image,

with large values of $\mathbf{k}$ corresponding to high resolution information. Since $\mathbf{k}$ -coordinates are controlled by B_0 gradients: $\mathbf{k}(T) = \frac{\gamma}{2\pi} \int_0^T \mathbf{G}(t) dt$, to collect all the data needed for a full image the gradients are modulated to scan the k_X and k_Y coordinates across a 2D plane, ('k-space traversal'(21)), and the image subsequently formed by frequency analysis (Fourier transform) of this data.

The RF Phase Gradient

Our new approach is based on special RF coils which produce linear phase gradients in the RF field (B_1), as given by:

$$\mathbf{B}_1 = |\mathbf{B}_1| e^{i\phi_1(\mathbf{r})}, \quad \phi_1(\mathbf{r}) = 2\pi \mathbf{k}_1 \cdot \mathbf{r} = \mathbf{G}_1 \cdot \mathbf{r} \quad (2)$$

assigning zero phase at $\mathbf{r}=0$. The vector $\mathbf{k}_1$ describes the direction and degree of the twist inherent in the phase of the RF field. So $\mathbf{k}_1$ characterizes the RF field and corresponds to a point in 3D $\mathbf{k}$ -space: $\mathbf{k}_1 = (k_{1X}, k_{1Y}, k_{1Z})$, which we refer to as the phase gradient "k-space origin" (Fig.1a). $\mathbf{G}_1$ is the RF phase gradient in radians/m. For a conventional RF coil with uniform phase: $\mathbf{k}_1 = \mathbf{G}_1 = 0$.

An RF phase gradient can be produced by either a dedicated coil, or by the superposition of fields from simultaneously driven elements of a coil array. One design procedure is to take a conventional RF coil design and twist it (22). At any one instant during the pulse sequence, only a single phase gradient is required. The RF field amplitude is not used for spatial encoding and is preferably uniform.

1D Experiments

Given an initial longitudinal magnetization, M_0 , a 90° RF excitation pulse applied with an RF phase gradient coil results in transverse magnetization of:

$$M_{xy} = M_0 e^{i\phi(\mathbf{r})} = M_0 e^{i2\pi \mathbf{k}_1 \cdot \mathbf{r}} \quad (3)$$

This is simply a statement that excitation with a B_1 phase gradient results in a spatially-dependent spin phase. The result is mathematically equivalent to encoding with a B_0 gradient, (see Eqn.1), however, the spatial encoding is immediately contained in the phase of the NMR signal using a single RF pulse (of any shape or type: square, sinc, adiabatic, etc.), see Fig.1b. The spin system is excited directly to the k-space coordinate $\mathbf{k}_1$. This k-space encoding can be observed by an ordinary (uniform phase) receiver coil. Note however that if the same phase gradient coil is used for both transmission and reception then although encoding occurs, it remains unobserved because the transmit and receive phases cancel. In other words, a phase gradient coil receives signal with a k-space shift of $-\mathbf{k}_1$.

Unfortunately, this approach of using a single excitation to achieve high resolution imaging is ultimately flawed, as it demands the generation of B_1 fields with impractically large phase gradients, which are unachievable for a volume coil configuration. This geometrical difficulty in achieving a very high degree of spin-dephasing in a single step is analogous to the difficulty encountered by rotating frame imaging. In that case the requirement for fields of very high amplitude to achieve high resolution, lead to RF power deposition problems, and was one reason these RF methods never made it to the clinic. However all is not lost. We will show next that high resolution is achievable by exploiting the properties of the NMR spin-echo to amplify the effect of the RF phase gradient strength by orders of magnitude.

Spin Echo

When applied with an RF phase gradient field, a 180° RF pulse rotates initial transverse magnetization M_{xy}^- about the B_1 field direction $\phi(\mathbf{r})$ to M_{xy}^+ thus:

$$M_{xy}^+ = M_{xy}^{-*} e^{i2\phi(\mathbf{r})} = M_{xy}^{-*} e^{i2\pi[2\mathbf{k}_1 \cdot \mathbf{r}]}, \quad (4)$$

where M^* denotes the complex conjugate of M . So, at a given location $\mathbf{r}$ in the sample, the spin phase after the RF pulse is determined (by Eqn.4) by the initial spin phase

of M_{xy}^- and the local RF pulse phase, $\phi(\mathbf{r})$. For the sample as a whole, the exponential phase term in Eqn.4 represents a spatially-dependent phase increment, i.e. a shift in k-space. So refocusing with an RF phase gradient results in firstly a reversal in sign ($\mathbf{k} \rightarrow -\mathbf{k}$) due to the complex conjugation of M_{xy} (the familiar action of an ordinary 180° pulse), but secondly a jump along the encoding direction of $+2\mathbf{k}_1$, so we can write (see Fig.1c):

$$\mathbf{k}_+ = -\mathbf{k}_- + 2\mathbf{k}_1. \quad (5)$$

The refocusing action is thus equivalent to a reflection about the coil k-space origin $\mathbf{k}_1$ as: $(\mathbf{k}_+ - \mathbf{k}_1) = -(\mathbf{k}_- - \mathbf{k}_1)$. This is consistent with the familiar behaviour of a uniform phase coil, for which $\mathbf{k}_1 = 0$. However from Eqn.5, if the same RF phase gradient is used for both excitation and refocusing, i.e. starting in state $\mathbf{k}_- = \mathbf{k}_1$, then no further encoding occurs as the k-space location after refocusing is unchanged ($\mathbf{k}_+ = \mathbf{k}_-$). No additional spatial encoding or further k-space traversal has been achieved. We also note that any scheme involving simply changing the phase of the transmitted RF pulses applies equally to all spatial positions, i.e. any resulting phase shifts would be global, and hence no encoding could come from this. So do we still have a problem with achieving high resolution imaging? No. The solution is to introduce a second phase gradient into the spin-echo experiment.

Echo Train

We will now proceed to show that the properties of RF phase gradient refocusing pulses (Eqns.4 & 5) can be used to construct an echo train, in which all echoes are sampled, with each subsequent echo having a different k-space encoding. Recall that during a B_0 readout gradient, spin phase evolves steadily in time (but with discrete sampling), and with a rate proportional to the distance from the gradient coil origin. For a TRASE echo train the situation is very similar, but the phase evolution does not occur continuously during the echo, but instead discretely at each refocusing pulse.

To see how this works in detail, consider a transmit array coil capable of producing either of two RF phase gradients, A and B. The two gradient fields are selected alternately to produce refocusing pulses: $90_A - 180_A - M_1 - 180_B - M_2 - 180_A - M_3 - 180_B - M_4 \dots$, where M_i denotes measured magnetization M_{XY} for the i^{th} echo. Equation 4 gives the effect of a single pulse so: $M_2 = M_1 \cdot \exp(2\phi_B i)$, and similarly: $M_3 = M_2 \cdot \exp(2\phi_A i)$. The combined effect of the BA pair is therefore just the phase factor: $\exp(2[\phi_A - \phi_B]i)$. Therefore for an echo train constructed from BA pairs, the spins will receive a phase increment of $2[\phi_A - \phi_B]$ from each pair of pulses. Substituting in the phase gradients: $\phi_A = G_{IA} \cdot x$ and $\phi_B = G_{IB} \cdot x$ the increment becomes $2[G_{IA} - G_{IB}]x$ for each pair. So moving down this AB echo train the spin phase is proportional to both time increment and spatial position. This is identical to the situation with a B_0 readout gradient, and so it is evident that the AB RF echo train can become the basis of a k-space imaging technique.

This process can be re-described in terms of k-space, writing the k-space origins of the two fields as $\mathbf{k}_{1A}$ and $\mathbf{k}_{1B}$ and using Eqn.5. For the above pulse sequence, the k-space coordinates of the echoes M_i , as observed by receive field A are: $(0, \Delta\mathbf{k}_{BA}, -\Delta\mathbf{k}_{BA}, 2\Delta\mathbf{k}_{BA}, -2\Delta\mathbf{k}_{BA}, \dots)$, where $\Delta\mathbf{k}_{BA} = 2(\mathbf{k}_{1B} - \mathbf{k}_{1A})$, see Fig.2. This makes use of the fact that k-space coordinates observed by a receiver ($\mathbf{k}_{OBS}$) is the actual k-space state ($\mathbf{k}_{ACTUAL}$) minus the k-space origin of the RF phase gradient receiver coil ($\mathbf{k}_{1RXCOIL}$), so: $\mathbf{k}_{OBS} = \mathbf{k}_{ACTUAL} - \mathbf{k}_{1RXCOIL}$. For an ordinary uniform phase coil $\mathbf{k}_{1RXCOIL} = 0$, and so $\mathbf{k}_{OBS} = \mathbf{k}_{ACTUAL}$. For this sequence, after $N+1$ refocusing pulses (N even), k-space is covered centrically from $-(N/2)\Delta\mathbf{k}_{BA}$ to $+(N/2)\Delta\mathbf{k}_{BA}$. The data reordering needed prior to FFT is illustrated in Fig.2b.

So in summary, the key to high resolution TRASE MRI is to repeatedly refocus using alternating RF phase gradients, allowing the signal phase or k-space trajectory to accumulate to higher k-space states. The effect of a pair of refocusing pulses (A, B) on any initial k-space vector is a translation of $2(\mathbf{k}_{1B} - \mathbf{k}_{1A})$, thus any pair of fields defines an encoding axis for k-space traversal (Figs.1d, 2b).

FOV

The field-of-view which can be encoded without aliasing, for any single-shot AB echo train, using a single receive field, and only the directly acquired data points, is given by the reciprocal of the k-space point spacing, i.e. $\text{FOV}_{\text{shot}} = 1/[2(\mathbf{k}_{1B} - \mathbf{k}_{1A})]$, and is equivalent to the distance over which $\phi_A(\mathbf{r}) - \phi_B(\mathbf{r})$ is equal to π . The sequence: $90_A - 180_A - M_1 - 180_B - M_2 - 180_A - M_3 - 180_B - M_4 - \dots$, detected by field A, yields points on both sides of k-space symmetrically arranged about $\mathbf{k}=0$, (assuming $\mathbf{k}_A = -\mathbf{k}_B$). A useful variant, using a 3rd coil U (uniform phase), is $90_U - 180_U - M_1 - 180_A - M_2 - 180_B - M_3 - 180_A - M_4 \dots$, which results in: $(0, +2\mathbf{k}_A, -4\mathbf{k}_A, +6\mathbf{k}_A, \dots)$, as detected by coil U. By complex conjugation and reordering the denser set of points $(\dots, -6\mathbf{k}_A, -4\mathbf{k}_A, -2\mathbf{k}_A, 0, +2\mathbf{k}_A, +4\mathbf{k}_A, +6\mathbf{k}_A, \dots)$ can be generated, resulting in a doubled FOV.

Resolution

For any k-space technique (TRASE included) pixel size is proportional to the reciprocal of the sampled k-space extent. Therefore for a 1D AB TRASE echo train, the pixel size, Δx , is inversely proportional to the product of the number of steps, ETL, (echo train length) and step size, so $\Delta x = \text{FOV}_{\text{shot}}/\text{ETL} = 1/[2 \text{ ETL } (\mathbf{k}_{1B} - \mathbf{k}_{1A})]$. Additionally, any mechanism leading to loss of signal down the echo train will decrease resolution.

2D Imaging

This new imaging principle can be extended to 2D imaging. Since any pair of fields with different k-space origins define an encoding axis, adding a 3rd non-collinear RF phase gradient ($\mathbf{k}_C$) provides three non-parallel encoding axes within a plane, i.e. $2(\mathbf{k}_B - \mathbf{k}_A)$, $2(\mathbf{k}_C - \mathbf{k}_A)$, $2(\mathbf{k}_C - \mathbf{k}_B)$, and so allow traversal throughout a 2D k-space plane. One procedure for acquisition of a $M \times N$ data matrix is to use $M/2$ trains with refocusing pulses $C - (AC)_i - (AB)_{N/2}$, and $M/2$ trains of $C - (AC)_i - (BA)_{N/2}$, where $i=0 \dots M/2-1$, and finally a single train of $(AB)_{N/2}$. The notation $(AC)_i$ indicates $180_A - 180_C$ repeated i times. Figure 3 shows an example set of 2D k-space trajectories using this scheme. In extension from the 2D experiments, four fields (all non-collinear and non-coplanar) are sufficient for 3D imaging.

Segmentation

There are a whole range of possible approaches to improve and refine TRASE. An alternative method of increasing the k-space sampling density is to segment the acquisition of one k-space line into two shots (without changing train length). Consider the two trains: $90_A - [180_A - 180_B]_{N/2}$ and $90_B - [180_B - 180_A]_{N/2}$. Data sampling occurs after each refocusing pulse and always uses the A coil. The two trains result respectively in acquired k-space points of: (0, +1, -1, +2, -2, ...) and (+0.5, -0.5, -1.5, +1.5, ...) in units of $\Delta \mathbf{k}_{BA}$. The k-space separation is identical in each dataset, but the points are shifted, so combination of the two doubles the FOV.

Time-domain multiplexing (Acceleration)

In the SMASH technique (12) local sampling density is increased without a time penalty (acceleration) by acquisition of data in parallel with multiple coils. In analogy, in TRASE acceleration can be performed with a single channel receiver by series multiplexed data reception with different RF phase gradients. This is achieved by switching PIN diode control currents in the coil array, so that each receive field is activated in turn during each echo. This is technically feasible as very high speed switching is not necessary because during each echo the system is parked at a single k-space location and only a small number of field switches are required (equal to $N_F - 1$, where N_F is the number of receive phase gradient fields). This serial activation of different fields to sample the same spin state is a form of time-domain multiplexing (TDM) (23). Switching between N_F different phase gradient fields on a single receive channel, during a single echo, yields N_F distinct k-space data points, each with an additional k-space shift of $-\mathbf{k}_1$ (Fig.4). From these N_F acquired k-space points an additional N_F signals can be generated using k-space conjugate symmetry (24), although the generated points are not necessarily unique. For an AB echo train the maximum acceleration available is $2N_F - 2$ for $N_F > 1$. TDM leads to the collection of more k-space data points in the same time, so the available SNR is shared over more of k-space. This is closely analogous to the use of the blipped B_0 gradient in EPI - a decision is made to shift the receiver's point-to-view to a different part of k-space. Sampling the echo off-center potentially makes the sampling more susceptible to $T2^*$ or off-resonance effects, study of these effects and TDM variants is beyond our current scope.

We have described three receive acceleration options for TRASE: half Fourier (complex conjugation), multi-shot segmentation and time-domain multiplexing. There is also a 4th option, because TRASE is also compatible with conventional receive arrays and parallel MRI (12,13). Image folding occurs when FOV_{shot} is less than the object size. Since FOV_{shot} is inversely proportional to the strength of the RF phase gradients, these methods of increasing k-space density and FOV will become increasingly important as stronger RF phase gradients are employed. Consideration of coil designs and achievable phase gradient strengths are beyond the current scope.

Slice Selection

RF phase gradients can also be used to define imaging planes by slice selection (25). Slice selection can be viewed as the traversal of excitation k-space with the modulated deposition of RF energy along that trajectory (26). In TRASE this is implemented by interleaving small flip angle excitation pulses between refocusing pulses in an $(AB)_N$ train, followed by a $(BA)_{N/2}$ train to re-center the k-space excitation pattern (Fig.5). The Fourier transform of the envelope of the excitation pulses (typically sinc) is a close approximation to the spatial excitation profile. This is seen most simply in Fig.5 by just considering the small flip angle pulses (u_1 - u_5 , square symbols in Fig.5b) whose flip angles are modulated by a sinc function. Figure 5b shows that the action of the refocusing pulses is to transform this temporal modulation (horizontal) into a centered k-axis modulation (vertical).

In close analogy to the receive TDM method, selective excitation can also be accelerated to increase the density in excitation k-space by using a series of small flip angle pulses on different coils (without intervening refocusing) to produce excitations at shifted k-space locations. This is shown in Fig.5 where the sinc envelope is jointly defined by small flip angle pulses, using both a uniform RF field (u_n , squares) and the RF phase gradient fields (a_n , b_n , circles). Increased k-space density increases separation between aliased slices.

Slice profile shape, width, position and flip angle are all fully controllable (25). Slice position is controllable by adding RF phases (ϕ_A , ϕ_B) to A and B pulses respectively. The

slice centre may be defined as $\mathbf{r}_0 = (\phi_A - \phi_B) / 2\pi(\mathbf{k}_{1B} - \mathbf{k}_{1A})$. In receive mode the same procedure produces an FOV shift. This makes use of the fact that for a linear RF phase gradient the effect of a transmitter phase shift is equivalent to a spatial shift of the coil along the gradient direction (ignoring any coil end effects).

Interleaved multi-slicing is also possible as spins outside the selected slice are not dephased, but returned to the equilibrium $+z$ axis. Oblique slices can be defined by use of three or more coils to produce an oblique k-space trajectory. Since this slice selection process does not require modulated RF pulse waveforms, the usual linear RF transmit chain (including Class A or A/B power amplification) could be replaced by a lower cost and more efficient non-linear design.

Methods

Imaging data were obtained using a 0.2 Tesla, 55cm gap, 49ppm (over 30cm DSV), 4-post permanent magnet MRI system, controlled by our NRC TMX MRI console (27). For all experiments only a single transmit channel and a single receive channel were used. All echo trains used the following pulse phase pattern: $180_A^0 - 180_B^0 - 180_A^{180} - 180_B^{180}$ as this sequence of phases was found to have better immunity to pulse imperfections than a sequence with all pulses the same phase. B_0 gradient amplifiers were powered-down for all TRASE experiments except for the measurement of slice profiles.

A single train 1D TRASE pulse sequence (see Fig.2) was implemented, ETL= 64, 800 μ s square refocusing pulses, echo spacing of 5ms, for a total acquisition time of 320ms. The two fields (± 1.25 deg/mm) were produced using a pair of cylindrical spiral-birdcage coils (18,19,22,28) as shown in Fig.6a. Because each spiral birdcage coil inherently produces a unique linear RF phase gradient, the experiment is carried out by simply transmitting with one coil at a time, with the other actively shut off using a PIN diode switch. A two compartment phantom (Fig.6e) was used to measure a 1D projection.

To produce k-space traversal in 2D, array elements with inherent RF phase gradients were again used, with $\mathbf{k}_A$ and $\mathbf{k}_B$ fields produced by a spiral birdcage pair ($\pm 1.25 \text{ deg/mm}$) and the orthogonal $\mathbf{k}_C$ field by a 4-loop crossed Helmholtz-Maxwell coil (1.2 deg/mm), Fig.7b. A 2D TRASE image was acquired with $800\mu\text{s}$ square refocusing pulses and 5ms echo spacing. The receiver was set to sample signal at 10 kHz, with 3 points per echo averaged to produce each k-space data point. From an SNR point-of-view averaging 3 points is equivalent to sampling at $10\text{kHz}/3 = 3.3\text{kHz}$, and improves SNR by $\sqrt{3}$. An advantage of over-sampling is that the acquisition window can be more precisely defined in time, which helps in rejecting transmitter or switching glitches. The TRASE image was acquired with trajectories of the form as shown in Fig.7a. For comparison, a standard 2D image was obtained using B_0 gradients and a 2D spin echo pulse sequence image ($\text{TE}=18\text{ms}$, $\text{BW}=10\text{kHz}$; single RF coil). Both images were acquired as projections without slice selection, $\text{TR}=1\text{sec}$, 4 averages, 128×128 matrix, acquisition time 512s, and $800\mu\text{s}$ square excitation pulse.

Acceleration Methods

To test the time domain multiplexing receive acceleration method, a 1D experiment was performed with the same parameters as described above, but with two receiver windows defined for each echo. Switching between the A & B coil windows occurred in the center of the echo, with a $200\mu\text{s}$ gap between windows, 8 points were collected per window, Fig.8.

Slice Methods

To test the TRASE slice selection method (Fig.5), profiling experiments were performed with sinc envelope excitation train lengths of $\text{NP}=32, 40, 50$, corresponding respectively to slice widths of 16mm, 12.5mm, 10mm, see Fig.9. Each excitation train was followed by $\text{NP}/2$ refocusing pulses in reversed order (BA). A single compartment phantom (Fig.7c) was used. All refocusing pulses were implemented as $1.6\text{ms } 90^X\text{-}180^Y\text{-}90^X$ composite pulses. Excitation pulses were $800\mu\text{s}$ in length and transmitted on the B coil only. Slice profiles were measured by following the TRASE slice selection process with a

B_0 gradient echo readout. The multi-slice method was evaluated in simulation using single-shot interleaved multi-slice profiles, (slice excitations separated by 160ms, $T_1 = 800$ ms, pulse length = 1.6ms, $TE=3$ ms, $T_2 = 200$ ms) with a 3-lobe sinc waveform of 50 excitation pulses using a uniform field, and refocusing with ± 1 deg/mm RF phase gradient fields. Simulations were performed both with and without transmit acceleration.

Results

1D TRASE results are shown in Fig.6c, and the corresponding frequency encoded profile in Fig.6d. The profiles correspond well, with the TRASE results exhibiting higher SNR and lower resolution.

The 2D TRASE image result is shown in Fig.7e. The 2D imaging results are remarkable for the high resolution achieved without the use of B_0 gradients. The images from the B_0 gradient based MRI method (Fig.7d) and the TRASE technique are very similar (both showing the phantom conical base and plunger) and so validating the TRASE approach. An air bubble trapped on the right side of the plunger is more clearly seen in the RF image. The TRASE image shows higher SNR, but lower resolution due to loss of signal down the echo train from relaxation and refocusing errors. Minimum echo spacing was limited by an interlock on the particular RF power amplifier used, leading to increased T_2 losses. Line artifacts visible in both images incidentally demonstrate the different encoding directions used.

Results for the acquisition with 2-windows receive acceleration are shown in Fig.8. As expected, the combination of datasets from both coils results in a doubling of the field-of-view.

Experimentally measured single slice profiles (Fig.9a,b) show good quality profiles and demonstrate that slice width and slice position are both controllable. A multi-slice simulation is shown in Fig.9c, where the first slice (top profile) shows maximum signal, while reduced signal from subsequent slices is mainly T_2 -weighting due to relaxation

during earlier composite inversion pulses, and is noticeable due to the relatively high ratio of pulse length to T_{2p} used. Simulated results for an accelerated slice selection procedure show increased separation between aliased slices, as expected (Fig.9d).

Discussion

RF Power

SAR measurements made according to the NEMA procedure (29) yield 2mJ deposited energy into the load for a 180° 500 μ s square pulse and a 8.2MHz linear solenoidal head coil. This is low enough that even when played out continuously these pulses would not exceed SAR limits. For a fixed flip angle and pulse shape, deposited pulse power is approximately proportional to $BW_{\text{PULSE}} B_0^2$. Using this relationship to extrapolate the 0.2T results to 1.0T, the IEC-60601-2-33 head SAR limit of 3.2W/kg would permit the use of 500 μ s square refocusing pulses at an average rate of 300/sec, which is sufficient for many TRASE imaging experiments. Factors affecting the required average pulse rate include the resolution, efficiency of the k-space trajectory (3D is preferred over 2D), and the use of acceleration methods. Unlike the rotating frame method (5), the maximum flip angle required by TRASE is only 180° , therefore there is no special requirement for high peak RF power.

Scan Duration

For commonly encountered experimental conditions, a higher k-space traversal rate will be achievable with B_0 gradients than with TRASE. Therefore for high speed MRI applications (e.g. cardiac imaging), B_0 gradient methods will almost always outperform TRASE. However not all MRI applications require a high k-space traversal rate. For instance, for low SNR situations (e.g. low field MRI or low pixel volume) the total acquisition time is determined more by sampling efficiency and the need to perform averaging to improve SNR, than by k-space traversal rate.

Typically for TRASE one k-space line per echo train will be acquired, and one or a few (if multi-slicing is used) echo trains per TR period. Since magnetization recovery by

T1 relaxation is effectively inhibited by the repeated inversions occurring during echo trains, pulse-free recovery periods (also compatible with limiting the SAR), are required during multi-shot sequences. Scan time is given by: $TR \cdot N_{2D} \cdot N_{3D} \cdot N_{AVE}$ where N_{2D} and N_{3D} are respectively the 2D and 3D acquisition dimensions, N_{AVE} is the number of averages, and $T_R = N_{SLICE}(T_{EXCITE} + T_{TRAIN}) + T_{FREE}$. T1 recovery happens during the pulse-free periods of total duration T_{FREE} . The choice of acquisition mode (multi-slice/3D/multi-slab) will depend upon the usual factors including: scan time, number of slices, SNR and contrast. In summary, TRASE scan durations will be in the same range as SE sequences on early clinical MRI systems at one to several minutes. The use of half-Fourier reconstruction, any of the acceleration methods, and development of efficient k-space trajectories will all be helpful in reducing scan duration.

Echo Train Duration

The duration of a single train, T_{TRAIN} , is an important factor in scan duration and is given by:

$$T_{TRAIN} = (T_{PULSE} + T_{ACQ} + 2 T_{SW}) \cdot ETL \quad (6)$$

where T_{PULSE} is the refocusing pulse length, T_{ACQ} the acquisition window length, and T_{SW} a switching time. Many factors directly or indirectly affect T_{PULSE} . Firstly, since there is no requirement for a specific frequency (i.e. slice) profile, hard pulses can be used. Coil factors influencing T_{PULSE} include: size, geometry, efficiency and available RF power. The need for composite pulses (e.g. $90^X 180^Y 90^X$) depends upon the B_0 and B_1 inhomogeneities present and on the sensitivity of the particular sequence to refocusing errors. For head-size coils at 0.2T, pulse lengths of a few hundred microseconds are feasible. Unlike B_0 gradient coils, RF coils are physically lightweight and easy-to-handle, and custom arrays can be designed for each anatomical situation, which is beneficial because smaller coils are more efficient, allowing shorter T_{PULSE} .

The second factor affecting T_{TRAIN} is the acquisition window length, T_{ACQ} , during which echo sampling occurs. Under conditions of good homogeneity (i.e. long $T2^*$), the

sampling SNR is proportional to $\sqrt{T_{ACQ}}$. It is therefore beneficial to maximize T_{ACQ} , although benefits diminish as T_2^* dephasing leads to signal loss away from the echo center. In general, TRASE is insensitive to off-resonance effects (T_2^* decay) due to continual refocusing by hard pulses. T_{ACQ} is also constrained by total acceptable T_{TRAIN} , which is mainly determined by T_2 . For ballpark estimates we assume $T_{ACQ} = T_{PULSE}$.

The 3rd parameter affecting T_{TRAIN} is ETL, which is proportional to pixel resolution. In summary, typical train lengths may be anywhere between a few tens of ms to a few 100's of ms. For example, the representative parameters: ETL=100, $T_{ACQ} = T_{PULSE} = 0.4\text{ms}$, $T_{SW}=0.05\text{ms}$ give $T_{TRAIN} = 90\text{ms}$.

B₀ Homogeneity, Bandwidth & SNR

The use of readout and slice-selection B_0 frequency gradients requires a very high initial level of B_0 field homogeneity so that the spectral line broadening produced (by 2 orders of magnitude) provides faithful spatial encoding. In contrast, RF imaging operates with an unbroadened sample, so the refocusing and acquisition bandwidth requirements are determined by the natural sample spectral bandwidth, i.e. B_0 field inhomogeneities and chemical shift. As for any MRI sequence, the trade-off for a higher SNR is a lower data rate. From the sampling point-of-view TRASE is just another MR Fourier imaging method – so fundamentally the SNR of RF encoding is neither better nor worse than frequency encoding - it depends on the specifics of the sequences.

Since in TRASE the 1st dimension (frequency) is not used for spatial encoding it becomes a 'free' dimension, available for other uses. Time and/or frequency domain processing of the acquired echoes to eliminate or correct for off-resonance or other imperfections becomes possible. The relaxed B_0 homogeneity requirement for TRASE suggests that a favourable configuration for low cost MRI may be a lower field magnet with homogeneity 1 to 2 orders of magnitude lower than conventionally required, with the proviso that the refocusing RF pulse bandwidth at least matches the sample bandwidth.

Gyromagnetic Ratio

Spatial encoding with an RF phase gradient relies on neither the manipulation of the magnetization precession rate (conventional imaging), nor of the nutation rate (rotating frame imaging); therefore spatial resolution is not related to the gyromagnetic ratio, γ . This feature may be beneficial for low γ nuclei imaging experiments, which can be especially demanding of B_0 gradient equipment.

B_1 Homogeneity

The B_1 transmit fields should be as uniform as possible to promote accurate refocusing. Using square pulses for readout and composite for slice selection a B_1 amplitude homogeneity of approximately $\pm 10\%$ was found to be sufficient for good performance. Transmit field inhomogeneity can also be straightforwardly mitigated by limiting the imaging FOV to the central volume of the RF phase gradient coil by employing small (and sensitive) receive coils.

Slice Selection

TRASE slice selection has many of the features of B_0 gradient slice selection but also has some limitations. Firstly, TRASE slice selection is a multi-pulse technique, so the selection process will tend to be of longer duration, with the length being inversely proportional to slice width. Secondly there is the potential for pulse imperfections and T_1 effects to cause out-of-slice interference or saturation effects. These conditions favour the selection of a smaller number of slices and/or wider slices. This would be compatible with 2D imaging of targeting regions using a limited number of slices, and also with multi-slab 3D volume data acquisition.

Pulse Train Phases

Conventional MRI uses B_0 spoiler gradients for elimination of unwanted coherences, which can lead to image artifacts; however this approach is not available in RF imaging sequences. There is however an alternative based on the refocusing pulse phases. These are free parameters because k-space traversal occurs regardless of the pulse phases ϕ_i applied to the RF power amplifier. The B_1 phase seen at position r for the i^{th} pulse is: $\phi(i, r) = \phi_i +$

G_1 .r. So each point in the sample sees a different series of phases (i.e. a different pulse sequence), and this provides a lever to control the evolution of both wanted and unwanted coherences. A simple example ($180_A^0 - 180_B^0 - 180_A^{180} - 180_B^{180}$) was mentioned above. A full examination of these techniques is beyond the scope of this paper.

Contrast

Various image contrasts are available with TRASE including T_1 , T_2 and T_{2p} . T_1 contrast is available as TRASE is usually a multi-shot technique. During the refocusing pulses the magnetization decays with time constant T_{2p} , where $1/T_{2p} = 1/2 [1/T_1 + 1/T_2]$ (30). TRASE sequences are centric-out with little inherent T_2 loss. T_2 and T_2^* weighting can however be introduced during excitation or by incorporating preparation modules (31).

Conclusions

This paper introduces a new, silent MR encoding principle based on echo trains produced by RF phase gradient fields. 2D MRI and slice selection have been demonstrated, and 3D MRI is expected to be feasible with the construction of a suitable coil array. The technology needed for TRASE is considerably simpler than for conventional MRI. The minimum equipment required consists of a single RF transmitter channel with a switchable coil array, a single receiver channel, a console and a magnet. There is more freedom in magnet design as eddy currents are no longer a concern, space is no longer needed to accommodate B_0 gradient coils and, in some circumstances, homogeneity requirements are lower. RF phase gradient transmit coil array design is key as coil efficiency, phase gradient strength and B_1 homogeneity significantly influence all aspects of imaging performance including spatial resolution, SNR, scan duration and occurrence of artifacts. The RF phase gradient coil designs used in this paper are not optimized, so future coil designs should give improved performance.

We suggest three possible application areas for this new imaging method. Firstly, applications may arise from the novel k-space trajectories, made possible by the off-center k-space refocusing behavior, and which can be incorporated into conventional imaging

sequences. Since TRASE is a k-space method it is conceptually straightforward to design new hybrid pulse sequences combining B_0 gradient and RF phase gradient encoding. Secondly, the physical differences of TRASE encoding - the absence of undesirable eddy current effects, nerve stimulation (dB/dt), and acoustic noise during scanning - may prove useful features to exploit for some other applications. Thirdly, imaging entirely without a B_0 gradient system in a less homogeneous field may present opportunities for novel and simplified MR system designs in existing or new application areas.

In comparing RF imaging with state-of-the-art MRI, it is clear that the latter is a medical and research imaging modality with phenomenal levels of performance in imaging speed, spatial resolution and scan flexibility. However the high level of resources required to install and to operate state-of-the-art MRI inevitably limits the accessibility of MRI for many patient populations. So even without attaining the imaging speed and resolution performance levels of state-of-the-art MRI, TRASE arguably may have a role to play in broadening access to MRI, because, although in its infancy, the imaging speed, efficiency of k-space data acquisition (SNR), and spatial resolution (few mm) appear to promise sufficient performance for a diagnostic imaging device. One possible approach is the pairing of RF imaging with a low field permanent magnet. Relative to start-of-the-art MRI such an installation would offer simplified equipment (requiring less maintenance), lower capital and operating costs, and less demanding siting requirements, including low overall system power consumption and cooling requirements.

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Table 1

Comparison of three MRI encoding methods.

Encoding Method	Gradient Type	Static Field	RF Field
Frequency Encoding	Main Field	$B_0 + x.G_x$	B_1
Rotating Frame Imaging	RF Magnitude	B_0	$B_1 + x.G_{1x}$
TRASE	RF Phase	B_0	$B_1 \exp(i x.G_{1x})$

Figure Legends

Figure 1.

Basic k-space properties for RF phase gradient fields. **(a)** the phase gradient is represented as a point in k-space, $\mathbf{k}_1$, referred to as the field (or coil) ‘k-space origin’; **(b)** excitation using the phase gradient field results in excitation to the point $\mathbf{k}_1$; **(c)** refocusing an arbitrary spin-state S1 using the phase-gradient field results in refocusing about $\mathbf{k}_1$ according to Eqns. 4 & 5; **(d)** the effect of two successive refocusing pulses with different fields ($\mathbf{k}_A$, $\mathbf{k}_B$) is to translate any arbitrary k-space point by $\Delta\mathbf{k}_{BA} = 2(\mathbf{k}_B - \mathbf{k}_A)$, i.e. k-space traversal.

Figure 2.

A 1D phase gradient imaging experiment. **(a)** The basic 1D TRASE pulse sequence consists of a 90° excitation followed by a train of 180° refocusing pulses applied alternately to the A and B transmit coils. NMR signal is observed (1-4) after each refocusing pulse. **(b)** The corresponding k-space sampling pattern showing the k-space coil origins ($\mathbf{k}_{1A}$, $\mathbf{k}_{1B}$, crosses), and sampling locations (circles 1-4).

Figure 3

Example of a set of TRASE k-space trajectories for 2D imaging. **(a)** Detail of central square cluster of 4 points in each diagram representing the k-space origin of 4 fields: A-D, (D – unused). The A coil is used for excitation in all cases. **(b)** Sequence (AB)₅ scans a vertical k-space line; **(c)** a pair of sequences C-(AB)₅ and C-(BA)₅ together provide the 1st pair of vertical lines; **(d)** C-AC-(AB)₅ and C-AC-(BA)₅ provides the 2nd pair of lines; **(e)** C-(AC)₂-(AB)₅ and C-(AC)₂-(BA)₅ provide a 3rd pair.

Figure 4

TRASE receive acceleration. **(a)** k-space diagram showing three receiver coils, A,U,B, (crosses) and a single spin state (circle); **(b)** three received k-space data points as

obtained from the three coils. The coordinate of each received data point (b_1 , u_1 , a_1) corresponds to the difference between the spin state and the receiving coil k_1 coordinate (arrows in (a) and (b)); (c) 1D pulse sequence with transmitted pulses (upper case) and receive windows (lower case); A and B are RF phase gradient coils with equal and opposite gradients, U is a uniform phase coil; (d) k-space diagram for the data points received by each coil individually (columns 1-3) and in combination (4th column) showing increased density of k-space coverage. Asterisks denote data points generated by complex conjugation. The point 'a1' is at $k=0$ (signal fully in phase) because the excitation pulse and first refocusing pulse both use coil 'A'.

Figure 5

TRASE slice selection (sequence shortened for clarity). (a) Pulse sequence showing pulses transmitted on 3 coils (A,B – RF phase gradients, U – uniform). K-space traversal is effected by 180° refocusing pulses (A, B) while excitation is performed by interleaved small flip angle pulses (a_n , b_n , u_n), (refocusing and excitation pulses not shown to scale). The small flip angle pulses jointly define the excitation weighting function (sinc); (b) the corresponding k-t diagram showing the 11 excitation pulses (squares, circles), and 6 refocusing pulses (crosses). The k-t trajectory of each excitation is shown. The temporal excitation is transformed into a symmetrical k-space weighting, corresponding to rephased spins across the slice. The acceleration procedure of using multiple coils for excitation pulses allows higher k-space density and results in reduced aliasing.

Figure 6

The 1D RF phase gradient imaging experiment results. (a) $\pm 1.5\pi$ spiral birdcage coil array consisting of two coils wound with copper tape on a single 200mm long, 90mm diameter former. (b) B_1 magnitude (dB) and phase (deg) plots for the spiral birdcage array; (c) Experimental 1D profile obtained by reordering and Fourier transformation of acquired k-space data from TRASE (ETL=64), RF pulses: 90° (0.4ms), 180° (0.8ms), single train, pulse spacing 5ms; and (d) standard SE frequency encoding (TE = 22ms, 128 points). (e) The two-compartment water phantom (2.5cm diameter water-filled syringe) with graphic.

Figure 7

2D MRI experiment using 3 RF phase gradients (A,B,C). **(a)** 2D k-space trajectory for excitation with coil C followed by the refocusing sequence $(CB)_2-(CA)_3-C$ which using orthogonal encoding directions $(\mathbf{k}_C-\mathbf{k}_B)$ and $(\mathbf{k}_C-\mathbf{k}_A)$; **(b)** Photo of RF gradient coil array showing a spiral birdcage pair as in Fig.3 producing fields A & B. Field C was produced by the crossed combination of a Helmholtz coil (3 turns) and a Maxwell coil (1 turn); **(c)** Photo of phantom: 25mm diameter syringe, filled with tap water to 50mm depth with conical plunger and bottom. **(d)** Conventional 2D spin-echo image (TE=18ms, BW=10kHz; single RF coil). **(e)** 2D TRASE image (800 μ s square refocusing pulses; 5ms echo spacing, BW=3.3kHz (effective), acquired with B_0 gradient amplifiers powered down. Both images **d** & **e** were acquired as projections without slice selection, TR=1sec, 4 averages, 128x128 matrix, acquisition time 512s, and 800 μ s square excitation pulse.

Figure 8

Experimental 1D profiles from receiver coils A & B (spiral birdcages as in Fig.2) and the combined profile from the two datasets interleaved in k-space resulting in a doubled FOV.

Figure 9

TRASE slice selection results. **(a)** experimentally measured slice profiles using a sequence of the form shown in Fig.5a (excitation on coil A only) with sinc envelope excitation train lengths of NP= 32,40,50, corresponding respectively to slice widths of 16mm, 12.5mm, 10mm. Each excitation train was followed by NP/2 refocusing pulses in reversed order (BA). The phantom of Fig.7 was used. **(b)** Similarly measured experimental profiles showing slice shift (NP=32); **(c)** simulated single-shot interleaved multi-slice profiles, (slice excitations separated by 160ms, T1 = 800ms, pulse length = 1.6ms, TE=3ms, T2 = 200ms) with a 3-lobe sinc waveform of 50 excitation pulses using a uniform field, and refocusing with +/- 1deg/mm fields. **(d)** Simulated slice profiles as in (c) with excitation coil A (upper) and A & B coils (lower).