Published in final edited form as:

Magn Reson Med. 2014 July ; 72(1): 49–58. doi:10.1002/mrm.24888.

MRI by steering resonance through space

Angela L.S. Snyder, Curtis A. Corum, Steen Moeller, Nathaniel J. Powell, and Michael Garwood

Center for Magnetic Resonance Research and Department of Radiology, University of Minnesota School of Medicine, Minneapolis, Minnesota

Abstract

Purpose—This work introduces a technique to excite MR signals locally and to steer this localized region over the object in a spatiotemporal manner. The purpose is to demonstrate the feasibility of MRI with multi-dimensional spatiotemporal-encoding in a way that provides the ability to compensate extreme field inhomogeneity.

Methods—The method is called steering resonance over the object (STEREO). A modulated gradient is applied in concert with a frequency-modulated pulse to steer a resonant region through space and thus produce sequential excitation and echo formation. Images are reconstructed using exclusively an inverse problem solution.

Results—Images of phantoms and human brain were produced to demonstrate the feasibility of the STEREO sequence and image reconstruction. Simulations support the postulated capability to compensate for extreme field inhomogeneity.

Conclusion—STEREO represents a substantial departure from conventional MRI in which spins contained in the sample, slab, or slice are excited synchronously. By exciting spins sequentially along a curved spatial trajectory, STEREO in principle affords a unique opportunity to adjust for spatial variations in static and radiofrequency fields. By adjusting field amplitudes and frequencies in a temporal manner in STEREO, in future works it should be possible to perform MRI with highly inhomogeneous fields.

Keywords

MRI methodologies; spatial encoding; field inhomogeneities

INTRODUCTION

Spatiotemporally-encoded MRI using frequency-modulated (FM) pulses has been attracting much interest recently due to its unique features and capabilities. In some cases, the technique has demonstrated significant advantages over conventional frequency- and phase-encoded MRI (i.e., Fourier imaging). Early inspiration for spatiotemporally-encoded MRI

*Correspondence to: Michael Garwood, Ph.D., Center for Magnetic Resonance Research, 2021 Sixth Street SE, Minneapolis, MN 55455., gar@cmrr.umn.edu.

Presented in part at the 2013 ISMRM Annual Meeting, Salt Lake City, the 2011 ISMRM Annual Meeting, Montréal, and the 2007 ISMRM Annual Meeting, Berlin.

came from Pipe's analytical description of the spatial quadratic phase produced with FM excitation (1,2). During the same period, Meyerand and Wong proposed a spatiotemporally-encoded imaging technique they called "time encoding" (3). Subsequently, Frydman and colleagues expanded the theory of time encoding and demonstrated ways to drastically shorten the acquisition time for multidimensional NMR spectroscopy (4–6). Implementations of this concept for rapid imaging followed shortly (7,8).

Although echo-planar imaging is the most widely used single-shot imaging sequence, explorations into spatiotemporal-encoded MRI have been motivated in part by the need for fast imaging methods that offer greater tolerance to static magnetic field (B_0) inhomogeneity. For MRI, multiple variants of the time-encoding strategy have been proposed. Frydman's group has produced a body of work on the use of time encoding as a substitute for frequency encoding (7,9–12). Instead, Chamberlain et al. replaced phase encoding with time encoding and introduced the use of two radiofrequency (RF) π -pulses to obtain constant effective echo time (TE) (8). Recognition of the ability to tolerate field inhomogeneity makes spatiotemporal encoding appealing as an alternative to echo-planar imaging (11–13). Previous work has also demonstrated the robustness of spatiotemporal MRI to B_0 inhomogeneity caused by metallic objects (14). Finally, spatiotemporally-encoded MRI uses alternative image reconstruction algorithms from those used in Fourier imaging, which has led to the development of new methods to improve resolution (10,15).

Typical implementations of spatiotemporal MRI involve at least one dimension of frequency or phase encoding. It is conceivable, however, that variants of these sequences could be implemented in an entirely spatiotemporal manner such that the use of Fourier transformation (FT) to reconstruct an image becomes unnecessary. Indeed, a method for fully spatiotemporally-encoded MRI (without FT) was described by Shrot, Tal, and Frydman (7,12). In that method, a standard (i.e., constant frequency) excitation pulse is followed by two chirped π -pulses applied with gradients along different spatial axes, which essentially creates a spatial grid that can be "walked-through" using shaped gradients during acquisition, thereby obtaining the image without explicit k -space encoding. With that sequence, clear advantages were demonstrated for compensating B_0 inhomogeneity along one dimension by modulating the pulse for encoding that dimension of the grid. However, expanding the approach to more dimensions proved difficult (12). The potential to compensate inhomogeneity in the RF field (B_1) was also discussed.

Described herein is a new approach for spatiotemporally-encoded MRI that remedies some of the issues of the earlier methods. This new MRI method called STEREO is based on steering resonance over the object in a time-dependent manner. STEREO produces images without Fourier encoding by moving a localized resonant region along a curved trajectory in space such that the excitation itself is spatially selective. Each region can then be refocused sequentially during acquisition, retaining the unique spatiotemporal dependence that was established in the excitation. By exciting spins sequentially in time along tailored traversals in space, temporal adjustments (e.g., B_1 amplitude and frequency) can in principle be made to compensate for B_1 and B_0 inhomogeneity. Because the resonance region is not sufficiently compact (i.e. not a delta function), reconstruction by geometrically-driven coordinate transformation is not ideal, whereas an inverse problem solution, in this case a

Bloch-simulation-derived encoding matrix, is well suited to reconstruct images. Importantly, the spatiotemporal domain allows direct treatment of spatial problems. This spatial isolation is particularly unusual in MR imaging and opens opportunities for numerous novel applications that could not be considered with standard Fourier-based techniques.

THEORY

Spatiotemporal encoding on a curved spatial trajectory

In previous implementations, spatiotemporal encoding was accomplished by the application of a constant gradient in the desired encoding direction and concurrently applying a chirped frequency sweep. [Note: In practice, the frequency sweep is usually produced by applying a quadratically varying phase of the pulse.] As the pulse frequency is swept, magnetization rotates toward the transverse plane sequentially in time in the direction of the gradient ($\mathbf{G}$). By the end of the pulse, these isochromats have acquired a phase that is a function of the quadratic phase acquired from the RF pulse at the time of excitation, plus the phase that accumulates due to free precession following excitation; hence, the phase of isochromats has a quadratic plus linear dependence on position. Upon inverting the gradient direction and acquiring signal, the linear dependence of the phase is unwound, which means the vertex of the quadratic phase function sweeps in the reverse direction across the sample during acquisition. At a given time, only the spins located in the vicinity of the moving vertex contribute the majority of the signal and thus are temporally localized in the direction of $\mathbf{G}$.

Another familiar spatially-selective MR technique of relevance is outer-volume-suppression (OVS). In this method, a slice of magnetization outside the region of interest is selectively excited (by proper choice of $\mathbf{G}$ and B_1 frequency) and signal from inside the slice is subsequently suppressed (spoiled) by gradients. In most implementations of OVS, signals are sequentially excited and suppressed in a series of slices surrounding a region of interest. In the limit of infinite number of slice-selective pulses applied with infinitely small incremental change in gradient orientation, the method produces a curved, continuous suppression region. In this case, instead of using a constant $\mathbf{G}$ for each slice, the driving functions of the gradient components, $G_x(t)$ and $G_y(t)$, might be sine and cosine functions to yield a continuous ring of signal outside the region of interest (16,17). With a constant offset frequency, the ring has a constant radius. However, if the B_1 -frequency is swept from a maximum frequency offset toward zero, then the continuously decreasing radius in the presence of the sine- and cosine-modulated gradients forms a spiral. STEREO is designed to preserve signal along such a spiral of excitation, not suppress it as in continuous OVS.

Many possible curved trajectories can be designed to work with STEREO. For the sake of simplicity in this proof-of-principle study, the method is implemented with an Archimedean spiral having constant radial and angular velocity over 2π radians. In STEREO, magnetization is excited by a chirp pulse having a linear frequency sweep terminating at zero frequency offset. This chirp pulse is applied in concert with gradient modulation on two axes using a sinusoidal function that is shifted by one-quarter cycle between axes. The resultant trajectory is a spiral from the maximum radial distance to the center, with constant angular and radial velocity components. Alternatively, the frequency offset may similarly be swept from zero to a maximum value to create a spiral starting at the center and proceeding

outward. With a chirp pulse, which has a constant B_1 amplitude, the flip angle α is proportional to $(b_w/T_p)^{1/2}$, where b_w is the frequency-swept bandwidth and T_p is the duration of the pulse (7). The frequency offset $\omega(t) \equiv \omega_{RF}(t) - \omega_0$ varies linearly in time, and thus, the rate of change of the frequency offset is b_w/T_p . Accordingly, α is proportional to $(d(\omega(t))/dt)^{1/2}$, which is constant, showing that the flip angle along the trajectory is constant.

The modulated gradients in STEREO may initially appear to complicate many of the calculations shared by other time-encoded imaging methods, especially if attempting to consider each Cartesian axis separately; however, the description in polar coordinates is informative. In STEREO, the resonance condition is satisfied when

$$\Delta\omega(t) - \gamma \mathbf{G}(t) \bullet \mathbf{r}(t) = 0, \quad [1]$$

where γ is the gyromagnetic ratio and $\mathbf{r}$ is the position vector in polar coordinates. For the Archimedean spiral form of STEREO, let θ be the spiral rotation angle, such that

$$G_x(t) = A_x \cos(\theta(t)) \quad \text{and} \quad G_y(t) = A_y \sin(\theta(t)), \quad [2]$$

where

$$\theta = 2\pi t/T_p, \quad t \in [0, T_p] \quad [3]$$

In this work, the gradients are chosen such that the maximum amplitudes of the x and y gradients are equivalent, $A_x = A_y \equiv A_{xy}$. By expressing $\mathbf{G}$ and $\mathbf{r}$ in polar coordinates and considering only consecutive centroids on the trajectory, the centroids expressed in polar coordinates are described by

$$|r(t)| = \Delta\omega(t) / (\gamma A_{xy}) \quad [4]$$

with θ described by eq. [3]. Therefore, when using a chirp pulse with an Archimedean spiral trajectory having constant angular and radial velocities,

$$\frac{d^2\theta}{dt^2} = \frac{d^2r}{dt^2} = \frac{d^2\Delta\omega}{dt^2} = 0, \quad [5]$$

which is helpful for understanding the sequence behavior since $\theta \propto r \propto \omega \propto t$.

Imaging with STEREO

Three possible implementations of STEREO are shown in Figure 1 in the form of a gradient-echo (GRE), spin-echo, and double-spin-echo sequence. Acquisition gradients are chosen to sequentially refocus and then defocus the magnetization phase along the spiral trajectory. The acquired signal is still an echo, but rather than a simultaneous refocusing of all spins, the echo is a superposition of signals arising from the spatially and temporally continuous

nature of the frequency-swept excitation (i.e., in so-called pseudo echoes (18)). These 2π sinusoidal gradients are inherently self-refocusing, which allows encoding gradients to proceed in forward or reverse order. However, with the addition of ramps which are not self-refocusing, the gradient shapes require more careful consideration of the directionality and polarity for proper signal refocusing. In the case of a gradient echo, the acquisition gradient shapes can refocus spins in the order that they were excited, taking advantage of the self-refocusing nature of the sinusoidal gradients, which is to say that the excitation and acquisition gradient shapes are the same. Consequently, TE across the image is constant. For spin-echo sequences, the odd echoes are generated by refocusing spins in the reverse order of excitation using a temporal mirror image of the excitation shapes. Accordingly, in odd echoes, spins excited first will be refocused last, and vice versa, which means TE varies along the trajectory. This problem is eliminated when acquiring even-numbered echoes (i.e., TE is constant across the image) because the acquisition gradients are identical to the excitation gradients. In the spin-echo sequences, the π -pulse(s) can be applied in the presence of a gradient to restrict signal to a slice for accomplishing two-dimensional (2D) STEREO. In the present study, slice-selective π -pulses were used in the third dimension, although in principle three-dimensional STEREO should be possible by modulating the gradient in the third dimension during excitation and acquisition.

Repetition of the spiral trajectory with different initial angles of the gradients permits coverage of a 2D area, which requires an additional term describing the degree of initial rotation θ_i . As such, Eq. 3 becomes

$$\theta = (2\pi t/T_p) + \theta_i. \quad [6]$$

Excitation, evolution, and refocusing of spins take place sequentially for neighboring isochromats on the spiral trajectory. During signal acquisition, phase coherence occurs only near the moving vertex of the quadratic phase profile, and this is sufficient to achieve coarse localization of signal along the spiral trajectory. If the spatial profile of the localized region could be approximated by a 2D delta function, image reconstruction could simply be done by performing a geometric coordinate transformation and Jacobian weighting. However, as shown below, the assumption of a delta function is not valid, and therefore, an inverse problem solution must be used. The inverse problem solution also affords a theoretical sensitivity advantage over the geometric coordinate transformation, since all data points from the time-domain signals are used in reconstructing each image pixel, rather than a single data point.

The spatial profile of the localized region is governed by two types of frequency sweep which together provide two degrees of localization. As in other time-encoding methods, a sweep of the B_1 frequency provides one degree of localization. In STEREO, the second degree of localization is produced by modulating the resonance frequency using the temporally- and spatially-dependent B_0 component from the gradient field, $\mathbf{G}(t)$. These principles of STEREO can be better understood by considering the illustrations in Figure 2.

It is instructive to start by considering what happens during the sweep of the B_1 frequency. A constant B_1 frequency excites a slice perpendicular to the gradient direction ($\mathbf{G} = \text{constant}$); on the other hand, a chirp pulse excites not a slice but rather a continuous superposition of slices, each having a different excitation time and corresponding phase, which can be considered as a translating plane, or wavefront, perpendicular to $\mathbf{G}$. The frequency sweep function, $\omega(t)$, and the gradient amplitude, A_{xy} , determine the velocity of the wavefront. The traveling wavefront creates the time encoding of the other spatiotemporal methods mentioned earlier and provides one degree of localization in STEREO. In polar coordinates, this process represents a sweep in the direction of the radial vector $\mathbf{r}$ while θ is constant.

Next, consider a slice excited by a pulse with zero frequency offset and no frequency sweep. The gradient direction determines the orientation of the slice, and with sequential incremental changes in gradient orientation, the slice orientation rotates. In the latter rotating gradient case (Figure 2a), the spins at the center will experience the majority of the resonant RF energy while more distal spins, which fall within a slice only for a limited period of time, experience a smaller amount of resonant RF energy. With a constant non-zero offset, this process is like continuously varying outer-volume suppression before spoiling (Figure 2b). In polar coordinates, this can be viewed as a sweeping θ with constant r .

Finally consider a slice which is both translating and rotating (Figure 2c). Imagine exciting a series of sequentially rotated slices with the B_1 -frequency offset ($\omega(t)$) sweeping from a maximum value to zero. At any given moment the slice being formed is perpendicular to $\mathbf{G}(t)$ (Figure 2d). At this time, the whole slice is subject to the same radial velocity; however, the effect of the instantaneous angular velocity is least at the centroid (the point along the slice that is on the spiral trajectory and indicated in Figure 2d as point A), but increases at distal points in the slice (point B) by an amount proportional to the distance from the centroid. Therefore, the path defined by the sequential centroids experiences the greatest resonant RF energy, and points more distal in the slices experience reduced amounts of resonant RF energy. If the gradients are continuous, the excitation defines a scroll.

Simulation of excitation profiles

Upon conversion to polar-coordinate space, it is readily observed how the temporal localization of STEREO resembles that produced by a B_1 -frequency sweep using a constant gradient. As such, the transverse magnetization (M_{xy}) profile perpendicular to the trajectory (i.e., the radial component) is similar to that produced in the time-encoded dimension of previous spatiotemporal methods (8). To investigate this, Bloch simulations were performed with a combination of Matlab and C programs developed in-house. The simulated image matrix size was 51×51 elements, each consisting of 10×10 isochromats that were averaged to yield transverse magnetization for each matrix element, $M_{xy} = (M_x^2 + M_y^2)^{1/2}$. Relaxation effects were ignored. As shown in Figure 3b (blue), the excitation is localized in the radial direction.

The rotational velocity is the dominant factor dictating the M_{xy} profile tangential to the trajectory. Excitation near the centroid is determined mainly by the instantaneous radial velocity, while every other point in the plane perpendicular to the radial vector $\mathbf{r}$ is further

from the axis of rotation and thus experiences an increasing rotational frequency-sweep velocity, as well as the radial frequency-sweep velocity. The degree of excitation of any point in the tangential profile is therefore inversely related to its distance from the centroid. As the centroid of excitation travels along the defined trajectory, the isochromats falling within the slice at a point nearer the centroid experience lower instantaneous velocity, and thus, experience more resonant RF energy and subsequently contribute proportionally more signal during acquisition than the isochromats in the excitation slice further from the centroid. In this manner, the excitation in the tangential direction is coarsely localized, as shown in Figure 3b (red).

Bloch simulations in 2D space provide a clear picture of the spatiotemporal properties of the coarsely localized region generated by STEREO. A single Archimedean spiral trajectory was simulated using a B_1 -frequency sweep range ($\omega_{\max}$) of 40 kHz, $T_p = 6$ ms, and image matrix size of 129×129 . The simulated M_{xy} at specific time points during the excitation confirms the expectation that the greatest excitation is near the centroid, but also demonstrates clearly that the excitation is not limited to a single point but rather a coarsely defined region of resonance (Figures 3a–c). From the image showing the maximum M_{xy} over time (Figure 3d), it can be seen that the excitation is localized to the expected trajectory and the contributions from other regions, while proportionally much weaker than the signal from the trajectory of interest, constitute a substantial part of the total signal. As shown later, this degree of localization is insufficient to permit image reconstruction by simply assigning spatial coordinates, as can be done in one-dimensional spatiotemporal-encoding techniques.

The spatial nature of STEREO makes it well suited for compensating B_1 and B_0 inhomogeneity. Multiple degrees of freedom for compensating field inhomogeneity are afforded by modulating $\mathbf{G}$, as well as the B_1 amplitude and frequency. For example, by mapping B_1 inhomogeneity and knowing the desired trajectory, a uniform flip angle can be produced by simply increasing the B_1 amplitude proportionally when the resonance region is in an area of weak RF field (and vice versa). Because excitation occurs in sequential, coarsely localized regions rather than a single point, B_1 compensation will be tricky in situations of extreme B_1 variation. Nevertheless, some advantages are expected for STEREO as compared to or in combination with other methods of compensating B_1 inhomogeneity. Similarly, with a B_0 map and known trajectory, it may be possible in the future to compensate for B_0 inhomogeneity, either in post processing, as is possible with standard imaging methods, or by including temporal frequency adjustments into the sequence itself. Besides providing an alternative means to achieve B_0 compensation either in the acquisition or post-processing stage, STEREO may offer a degree of intrinsic tolerance to B_0 inhomogeneity similar to other spatiotemporal-encoding methods (11,14).

Reconstruction

As described above, the translational and rotational motion of the frequency sweeps provides a coarse localization of signal in time along the spiral trajectory; however, the point spread function of the localization at any given moment is a complex and spatially varying point spread function, which is insufficient for image reconstruction by geometric coordinate transformation (i.e., assigning signal along the trajectory according to the known

spatial-temporal relationship) as used in previous time-encoding methods. Thus, a more complete signal model is needed to tackle STEREO image reconstruction. Numerical inversion is a general decoding method for reconstructing the desired image (unknown) from the time-dependent magnetization evolution (known), and requires a representation for the encoding process. The general form of the inverse mapping problem comes from earlier literature (19,20). Such reconstruction methods have been employed with other imaging modalities, including electrocardiography (21). Inverse mapping has been used in MRI with standard encoding for spectral localization and spectroscopic imaging (19,20), as well as for addressing Gibb's ringing (22,23). Similar methods have been applied in MRI along one dimension for enhancing resolution, sometimes called super-resolution methods (10,15,24). For this initial demonstration of STEREO, the signal model is based on a Bloch description of the magnetization evolution, which produces an encoding matrix.

The inverse problem solution procedures are illustrated in Figure 4, where $\mathbf{E}$ is the encoding matrix and $\mathbf{E}^+$ is the pseudoinverse of that matrix. The pseudo inverse can be described as $\mathbf{E}^+ = (\mathbf{E}'\mathbf{E})^{-1}\mathbf{E}'$. It is worth mentioning that 2D STEREO reconstruction involves four variables: x , y , t , θ . Because 2D matrices are easily addressed, this representation was used to describe the dependence of the evolving magnetization on these four variables, by treating x and y as one dimension (range space) and t and θ as the other (domain space). A Bloch simulation was used to model the spin evolution in each pixel (defined by 10×10 isochromats) assuming constant spin density. The simulation was repeated for all 128 possible rotated trajectories. These simulated data were used to approximate the forward problem and provide an estimate of $\mathbf{E}$. Then a singular value decomposition solution was computed to obtain a pseudoinverse (25). Because the known values (128 spirals of 512 points each) greatly exceed the unknown values (51×51 pixel image), the problem is highly over-determined, and the matrix is well conditioned. A matrix size of 51×51 was used for all simulated or experimental image reconstructions described herein unless otherwise specified. Although 51×51 matrix size corresponds to low spatial resolution, it was considered sufficient for this first demonstration of STEREO MRI.

METHODS

Simulations

Simulations were also used to predict the behavior of the sequence and assess the quality of image reconstruction. To replicate STEREO acquisition, the transverse magnetization components (M_x and M_y) were summed over all pixels for the duration of the acquisition. As in experiments, data obtained with each of the rotated spiral acquisitions were combined to form an image. Simulations were performed using a chirp pulse operating in the rapid-passage, linear region (26,27).

Experiments

Experiments were performed with a research MRI instrument (Varian DirectDriveTM, Agilent Technologies, Santa Clara, CA) interfaced to a 4 T magnet having a 90 cm bore diameter (Oxford Magnet Technology, Oxfordshire, UK) and a clinical gradient system (Sonata, Siemens, Erlangen, Germany). STEREO data were collected with the double-spin-

echo sequence using π -pulses of the hyperbolic secant family (28). The number of complex points along each spiral trajectory (N_p) was 512, and the number of rotated spirals (M) collected was 128. The phantom was imaged with a quadrature surface coil consisting of two loops for RF transmission and signal detection (29). For imaging human brain, healthy volunteers were studied under a protocol approved by our institution's IRB. The RF coil was a 16-element transverse electromagnetic head coil (30). Standard GRE images were collected at the same low resolution for comparison.

RESULTS

Simulations

Figures 5a–d show results from a STEREO simulation. The modeled 2D object (Figure 5a) consisted of three small circles and one large circle, all with uniform spin density. The M_{xy} obtained from the simulation (Figure 5b) is displayed in the form of a spirogram, whereby time (t) from the beginning to the end of acquisition is represented along the horizontal axis and individual spirals are ordered according to initial angle θ_i along the vertical axis. A comparison of images reconstructed with the geometric coordinate transformation (Figure 5c) and the inverse problem solution (Figure 5d) confirms the expectation that the traversing localized region cannot be approximated as a 2D delta function. Reconstruction using the inverse problem solution produced images that closely resemble the objects, whereas the geometric transformation did not.

Simulations were also used to validate the ability to compensate B_1 inhomogeneity, as shown in Figure 6. For the simulations, the object shown in Figure 6a was modeled. The double-spin-echo STEREO sequence was simulated, assuming the inhomogeneous B_1 distribution of the transmitter coil shown in Figure 6b, while the receive B_1 distribution was assumed to be perfectly homogeneous. The simulated STEREO image without B_1 -inhomogeneity compensation is shown in Figure 6c. To demonstrate the ability of STEREO to produce constant flip angle in the presence of B_1 inhomogeneity, the chirp pulse was modified by superimposing a B_1 -amplitude modulation that was calculated based on a known B_1 inhomogeneity. Given the map of the B_1 inhomogeneity, new RF shapes were generated for each of the spirals, which were designed to compensate for the B_1 inhomogeneity along each spiral trajectory. These new RF shapes were used to generate an image (Figure 6d). The result demonstrates the ability to compensate image intensity variation using a B_1 -amplitude function tailored according to the known spatial B_1 distribution.

Experiments

The bottom row in Figure 5 shows results from a phantom experiment. The object and parameters used in these experiments were intentionally similar to those used in the simulations (Figure 5 top row). One 50 mL test tube and three 10 mL test tubes were positioned upright in the coil, and coronal images were obtained. As in the simulations, a comparison of images reconstructed with the geometric coordinate transformation (Figure 5g) and the inverse problem solution (Figure 5h) was performed and confirmed the expectation that the traversing localized region cannot be approximated as a 2D delta

function. The complex, spatially-varying point-spread function causes the geometric coordinate transformation to fail (Figure 5c and g). However, the inverse problem reconstruction effectively functions like a super-resolution method, not increasing the resolution (in this implementation) but compensating for spatial variations in point-spread function.

Finally, STEREO was compared with conventional (GRE) MRI in imaging human brain (Figure 7). The STEREO image (Figures 7b–c) shows signal dropout near the ventricles and misassigned signal outside the object, which were not observed in the GRE image (Figure 7a). Both images are shown with the same low resolution (51×51 matrix size) for fairness of comparison.

DISCUSSION AND CONCLUSION

In recent years, techniques based on spatiotemporal-encoding are providing new tools to accomplish rapid data acquisition and to overcome problems arising from non-uniform magnetic fields. Here, a spatiotemporal-encoding technique called STEREO has been introduced which offers the unique capability to steer resonance through time and multi-dimensional space. A demonstration of the feasibility of STEREO was shown by steering resonance along spiral trajectories in space, acquiring spatiotemporally-encoded data, and reconstructing images using only an inverse problem solution.

The STEREO images shown establish feasibility, but also exhibit artifacts including dropped and misassigned signals. A probable source of these problems is gradient infidelity, which distorts the spiral trajectory. Such imperfections were not considered in the present initial study of STEREO. Measurement and correction of the gradient non-linearities and timing errors in the future will likely improve image quality substantially. Further, the reconstruction method does not yet account for relaxation, nor is sensitivity to frequency shifts considered. Much work on improving STEREO image quality remains to be done.

STEREO has features in common with one of the earliest MRI techniques known as the sensitive-point method (31). As in STEREO, gradients modulate sinusoidally on multiple axes in the sensitive-point method to create a single region in the sample (i.e., the sensitive point) where resonance occurs. With these unconventional MRI methods, image formation requires measurement of the signal amplitude in “resonant regions” throughout the object. In the sensitive-point method, the resonant region is moved in discrete steps, whereas in STEREO, the resonant region is moved in a continuous manner to execute a rapid passage (in the linear region) through space. With STEREO and the sensitive-point method, images can be reconstructed by assigning each temporally measured signal intensity to a position in space, although this approach yielded highly distorted STEREO images and pays a price in terms of SNR. With the inverse problem solution used to reconstruct STEREO images, signal evolution from every pixel in the image is encoded and utilized in the image reconstruction to yield improved SNR efficiency. As demonstrated by Ben-Eliezer et al. (32), the SNR of spatiotemporal-encoded MRI can be comparable to that provided by k -space-based methods in optimal implementations.

In the demonstrated double-echo form, STEREO may be limited by SAR, and consequently not suitable for high-field imaging in its current implementation. When comparing to other images of similar resolution, this problem is exacerbated by the preliminary method of reconstruction through which significant resolution is lost. However, substantial improvements are expected when using a gradient-echo form, super-resolution, and an optimized number of spirals for the desired resolution.

The primary focus in this work has been on Archimedean spirals, but their selection has been for simplicity rather than ideal imaging properties. In the demonstrated form, spatial coverage was not a problem because we used a constant flip angle along the spiral, a long TR between spirals, and a high density of spirals relative to our reconstructed resolution. However, the spiral trajectory samples the center of the field of view more densely than the periphery, which makes T_1 -weighting non-uniform. In the future, the trajectory shape and rate of traversal may be altered to minimize such problems and increase efficiency. In particular, choosing a pattern that does not include the center point in every trajectory (e.g., concentric ellipses) may be beneficial. Trajectories can also be scanned in an interleaved fashion for more uniform flip angle, even with a short TR. More possible trajectories arise when expanding to three dimensions. In a straightforward expansion, STEREO becomes 3D with an additional dimension of rotational isolation by using a sinusoidal gradient on the third axis during excitation. In 3D STEREO, the most efficient trajectories might entail tracing the outside of various cones, spheres, or ellipses.

A frequency sweep can be produced either by varying the amplitude of B_0 or by varying the frequency of B_1 . In previous spatiotemporal-encoding methods, the frequency sweep is usually performed with a version of the classic chirp pulse, in which the frequency offset varies linearly in time during constant-amplitude irradiation. The latter produces sequential excitation of (infinite) planes stacked in the direction of the applied constant gradient field. Alternatively, with STEREO, magnetization is excited regionally and sequentially due to the modulating frequency-offset of the chirp pulse and the simultaneously modulated (temporally- and spatially-dependent) frequency-offset due to the rotating gradient field. This approach is different from previous spatiotemporal-encoding schemes that use only chirped frequency sweeps in the presence of a constant gradient. By frequency-modulating both B_1 and B_0 , STEREO extends the flexibility of spatiotemporally-encoded MRI to compensate for spatial variations in magnetic field during the excitation process, including multi-dimensional field non-uniformities.

In Fourier imaging, spatial resolution is determined by the acquisition parameters, whereas the resolution in spatiotemporal imaging is determined mainly by the excitation scheme. Consequently, the resolution for a given field of view is limited by the length of the pulse used for spatiotemporal encoding. However, Ben-Eliezer et al. recently showed an ability to recover this resolution in hybrid sequences with super-resolution techniques (10), and some related techniques have shown further benefits (15,33). The super-resolution reconstruction is in fact a one-dimensional inverse problem solution applied along the spatiotemporal-encoding direction. With these methods, reconstruction is based on an approximate analytic description of the spatiotemporal spin evolution used to reconstruct one-dimensional spatiotemporal data (10,15,33). This sort of analytic approach would similarly be beneficial

when used in two or three dimensions for higher resolution reconstruction of STEREO images. Alternatively, a similar effect could be achieved by increasing the number of pixels along the x and y coordinates of the Bloch-simulated transfer function, although that approach would demand more time and computational power. A combination of iterative inversion methods, advanced computing systems, and analytical approaches are being explored as these resolution gains could substantially enhance STEREO's imaging capabilities.

The capability to independently adjust for the main magnetic field and RF field actually experienced by each resonance region in principle gives this compensation method particular simplicity, flexibility, and robustness. Although not shown, STEREO should be capable of addressing B_0 inhomogeneity, however a B_0 map is needed (12). Also as a member of the family of spatiotemporally encoded sequences, STEREO has a particular ability to tolerate B_0 inhomogeneity (11,14). The ability to treat these problems from an entirely spatial perspective is expected to have value when used alone or in combination with existing methods. Image intensity correction is easily integrated into the inverse solution reconstruction. However, in the case of B_1 inhomogeneity, compensation during excitation is always preferred because it maximizes signal.

Acknowledgments

Grant sponsors: This work was supported by Biotechnology Research Center (BTRC) grant P41 RR008079 (NCRR), P41 EB015894 (NIBIB), R21 CA139688, and KL2 RR033182.

The authors thank Drs. Djaudat Idiyatullin, Ryan Chamberlain, and Jang-Yeon Park for helpful discussions over the course of this work. The authors also thank Michael Tesch for computational improvements and insight in simulation development.

References

1. Pipe JG. Spatial encoding and reconstruction in MRI with quadratic phase profiles. *Magn Reson Med.* 1995; 33(1):24–33. [PubMed: 7891532]
2. Pipe JG. Analysis of localized quadratic encoding and reconstruction. *Magn Reson Med.* 1996; 36(1):137–146. [PubMed: 8795032]
3. Meyerand ME, Wong EC. A time encoding method for single-shot imaging. *Magn Reson Med.* 1995; 34(4):618–622. [PubMed: 8524031]
4. Frydman L, Scherf T, Lupulescu A. The acquisition of multidimensional NMR spectra within a single scan. *Proc Natl Acad Sci U S A.* 2002; 99(25):15858–15862. [PubMed: 12461169]
5. Shrot Y, Frydman L. Single-scan NMR spectroscopy at arbitrary dimensions. *J Am Chem Soc.* 2003; 125(37):11385–11396. [PubMed: 16220962]
6. Shapira B, Frydman L. Spatially encoded pulse sequences for the acquisition of high resolution NMR spectra in inhomogeneous fields. *J Magn Reson.* 2006; 182(1):12–21. [PubMed: 16807020]
7. Shrot Y, Frydman L. Spatially encoded NMR and the acquisition of 2D magnetic resonance images within a single scan. *J Magn Reson.* 2005; 172(2):179–190. [PubMed: 15649744]
8. Chamberlain R, Park JY, Corum C, Yacoub E, Ugurbil K, Jack CR Jr, Garwood M. RASER: a new ultrafast magnetic resonance imaging method. *Magn Reson Med.* 2007; 58(4):794–799. [PubMed: 17899612]
9. Ben-Eliezer N, Frydman L. Spatiotemporal encoding as a robust basis for fast three-dimensional in vivo MRI. *NMR Biomed.* 2011; 24(10):1191–1201. [PubMed: 21360603]
10. Ben-Eliezer N, Irani M, Frydman L. Super-resolved spatially encoded single-scan 2D MRI. *Magn Reson Med.* 2010; 63(6):1594–1600. [PubMed: 20512863]

11. Ben-Eliezer N, Shrot Y, Frydman L. High-definition, single-scan 2D MRI in inhomogeneous fields using spatial encoding methods. *Magn Reson Imaging*. 2010; 28(1):77–86. [PubMed: 19608367]
12. Tal A, Frydman L. Spatial encoding and the single-scan acquisition of high definition MR images in inhomogeneous fields. *J Magn Reson*. 2006; 182(2):179–194. [PubMed: 16843690]
13. Goerke U, Garwood M, Ugurbil K. Functional magnetic resonance imaging using RASER. *Neuroimage*. 2011; 54:350–360. [PubMed: 20699123]
14. Ben-Eliezer N, Solomon E, Harel E, Nevo N, Frydman L. Fully refocused multi-shot spatiotemporally encoded MRI: robust imaging in the presence of metallic implants. *Magn Reson Mater Phy*. 2012; 25:433–442.
15. Shen J, Xiang Y. High fidelity magnetic resonance imaging by frequency sweep encoding and Fourier decoding. *J Magn Reson*. 2010; 204(2):200–207. [PubMed: 20223688]
16. Powell, N.; Marjanska, M.; Valette, J.; Henry, P-G.; Garwood, M. A New Method for Single-shot 2-D OVS. Toronto: 2008.
17. Wu, X.; Powell, N.; Marjanska, M.; Garwood, M.; Ugurbil, K.; Van de Moortele, P-F. A flexible design algorithm for single-shot 2d circular/elliptical OVS rf pulses. Honolulu, Hawaii: 2009.
18. Park JY, DelaBarre L, Garwood M. Improved gradient-echo 3D magnetic resonance imaging using pseudo-echoes created by frequency-swept pulses. *Magn Reson Med*. 2006; 55(4):848–857. [PubMed: 16506188]
19. Bertero M, De Mol C, Pike ER. Linear inverse problems with discrete data. I. General formulation and singular system analysis. *Inverse Problems*. 1985; 1:301–330.
20. Bertero M, De Mol C, Pike ER. Linear inverse problems with discrete data:II. Stability and regularization. *Inverse Problems*. 1988; 4:573–594.
21. He, B. Modeling & Imaging of Bioelectrical Activity: Principles and Applications. Springer; 2004.
22. Barrett, HH.; Myers, KJ. Foundations of Image Science. John Wiley and Sons; 2004.
23. Van de Walle R, Barrett HH, Myers KJ, Altbach MI, Desplanques B, Gmitro AF, Cornelis J, Lemahieu I. Reconstruction of MR images from data acquired on a general nonregular grid by pseudoinverse calculation. *IEEE Trans Med Imaging*. 2000; 19(12):1160–1167. [PubMed: 11212364]
24. Weiger M, Hennel F, Pruessmann KP. Sweep MRI with algebraic reconstruction. *Magn Reson Med*. 2010; 64(6):1685–1695. [PubMed: 20949600]
25. Hansen, PC. Regularization Tools. 4.1. 2008.
26. Ernst R. Sensitivity enhancement in magnetic resonance. *Adv Magn Reson*. 1966; 2:1–135.
27. Idiyatullin D, Corum C, Park JY, Garwood M. Fast and quiet MRI using a swept radiofrequency. *J Magn Reson*. 2006; 181(2):342–349. [PubMed: 16782371]
28. Silver M, Joseph R, Hoult D. Highly selective $\pi/2$ and π pulse generation. *J Magn Reson*. 1984; 59:347–351.
29. Merkle, H.; DelaBarre, LJ.; Bolan, PJ.; Baker, EH.; Everson, LI.; Yee, D.; Garwood, M. Transceive Quadrature Breast Coils and Applications at 4 Tesla. Glasgow: 2001.
30. Vaughan JT, Hetherington HP, Otu JO, Pan JW, Pohost GM. High frequency volume coils for clinical NMR imaging and spectroscopy. *Magn Reson Med*. 1994; 32(2):206–218. [PubMed: 7968443]
31. Hinshaw W. Image Formation by Nuclear Magnetic Resonance: The Sensitive-Point Method. *J Applied Physics*. 1976; 47(8)
32. Ben-Eliezer, N.; Frydman, L.; Sodickson, DK. Comprehensive Theoretical and Experimental Analysis of the Parametric Framework and SNR of Super-Resolved Spatiotemporally-Encoded (SPEN) MRI. Salt Lake City: 2013.
33. Cai C, Dong J, Cai S, Li J, Chen Y, Bao L, Chen Z. An efficient de-convolution reconstruction method for spatiotemporal-encoding single-scan 2D MRI. *J Magn Reson*. 2013

Figure 1.

STEREO can be implemented in a number of forms including GRE (a), spin-echo (b), and double-spin-echo (c). A chirp pulse is used for excitation in the presence of a modulated gradient. Signal is refocused and then acquired. Both GRE and double-spin-echo STEREO will have constant TE across the image. The gradient shapes within the dashed box are optional because being sinusoidal (if the ramps are omitted), they balance themselves.

Figure 2.

Illustration of rotating slices. a) A rotated set of slices with no B_1 -frequency offset will result in much greater excitation at the center than in the periphery. b) If a constant frequency offset is added, then something more akin to OVS bands is produced. The greatest excitation can be seen along the circle having a radius determined by the frequency offset and gradient amplitude. c) When the offset frequency is stepped with each gradient rotation, the greatest excitation is generated along a spiral trajectory. Any of these can be applied continuously with the use of sinusoidal gradients rather than repeating rotated slices. d) A diagram showing the centroid (A), which has a position described by the angle of rotation (θ) and the vector r , and a second point (B) within the plane located further from the centroid, as well as the planes tangential and perpendicular to a spiral trajectory and perpendicular to the gradient direction.

Figure 3.

Simulations showing the time-dependence of the signal localization in 2D physical space during excitation along a spiral trajectory. The magnitude of the transverse magnetization for three time points is shown (a–c) as well as the maximum magnitude of the transverse magnetization of each point from the duration of the excitation (d). The moving region of resonance in STEREO can be characterized by its tangential and perpendicular profiles (b). The M_{xy} profiles are as indicated: red (bottom): profile along the line tangential to the trajectory and blue (right): profile perpendicular to the trajectory. In the simulations, the initial longitudinal magnetization was set to 1. The total pulse length was 6 ms. These simulations show that the localized region is not isolated to a 2D delta function moving along the desired trajectory, but rather a moving region of resonance. However, excitation is clearly greatest along the trajectory.

Figure 4.

Schematic diagram of the inverse problem reconstruction. The forward problem is shown in (a). An encoding matrix is generated using a Bloch simulator with different gradient shapes for each rotated spiral. The rotations and time are combined and the spatial coordinates (x and y) are combined to generate a 2D matrix. The experimental acquisition is shown in (b). Signal acquired from the scanner is combined into one vector. The inverse problem reconstruction is shown in (c). The pseudoinverse of the encoding matrix can be used to generate spatial values from the time and rotation data.

Figure 5.

Simulations of STEREO (top row) showing the modeled object (a), spirogram (b), and images reconstructed with geometric (c) and inverse problem solutions (d). Simulations used excitation bandwidth (b_w) of 40 kHz, acquisition bandwidth (s_w) of 80 kHz, number of acquired points (N_p) chosen such that $N_p/s_w = 6.0$ ms, $T_p = 6.0$ ms, $TE = 21$ ms.

Experimental demonstration of STEREO (bottom row) showing a GRE image of the phantom (e), spirogram (f), and experimental images reconstructed with geometric (g) and inversion solutions (h). The phantom experiments were performed with $b_w = 40$ kHz, $s_w = 80$ kHz, $T_p = N_p/s_w = 6.4$ ms, $TE = 21$ ms, $TR = 4$ s, and field-of-view = 20 cm. In the spirograms (b and f), the horizontal axis is time and the vertical axis is initial gradient angle. It can be seen that the inverse problem solution yields images that closely resemble the object, whereas the geometric coordinate transformation does not.

Figure 6.

Inhomogeneity compensation demonstration by simulation. The intensity profile of the phantom used for simulation is shown in (a). The B_1 inhomogeneity is shown in (b). The scale is relative B_1 amplitude. The inhomogeneity profile was incorporated in the simulated transmit profile, producing the inhomogeneous image shown in (c). (A homogeneous receive profile was assumed). Inhomogeneity compensation by RF magnitude modulation was used to produce the image shown in (d).

Figure 7.

Comparison of STEREO with a standard imaging method. In the center (b) is an experimental image acquired at 4 T with double-spin-echo STEREO, $b_w = 40$ kHz, $s_w = 80$ kHz, $T_p = N_p/s_w = 12$ ms, $TE = 25$ ms, $TR = 4$ s, and field-of-view = 25 cm. The figure on the left (a) is a GRE image of the object collected at a comparable resolution with $T_p = N_p/s_w = 1.5$ ms, $TE = 25$ ms, $TR = 4$ s, and field-of-view = 25 cm. The figure on the right (c) is the STEREO image with the signal scaled up ten times to show the level of misassigned signal outside the object of interest.