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Reduced FOV imaging using a static second-order gradient for fMRI applications

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Abstract

Purpose—Imaging using reduced FOV excitation allows higher resolution or SNR per scan time, but often requires long RF pulses. Here, a recent reduced FOV method that uses a second-order shim gradient to decrease the pulse length, was improved and evaluated for fMRI applications.

Theory and Methods—The method, initially limited to excite thin disc-shaped regions at isocenter, was extended to excite thicker regions off isocenter, and produced accurate excitation profiles on a grid phantom. Visual stimulation fMRI scans were performed with full and reduced FOV. The resolution of the time-series images and functional activation maps were assessed by the full-width half-maxima of the autocorrelation functions (FACFs) of the noise images and the activation map values, respectively.

Results—The resolution was higher in the reduced FOV time-series images ($4.1 \pm 3.7\%$ FACF reduction, $P < 0.02$) and functional activation maps ($3.1 \pm 3.4\%$ FACF reduction, $P < 0.01$), but the SNR was lower (by $26.5 \pm 16.9\%$). However, for a few subjects the targeted region could not be localized to the reduced FOV due to the low Z2 gradient strength.

Conclusion—Given these results, the authors conclude that the proposed method is feasible, but would benefit from a stronger gradient coil.

Keywords

reduced FOV; selective excitation; Z2 gradient

Introduction

In imaging, the region of interest may be smaller than the object, allowing the field of view (FOV) to be reduced to decrease scan time or increase resolution (1–3). This is particularly beneficial in dynamic imaging applications, such as cardiac imaging (4), interventional treatment (5, 6), contrast-uptake (7), or functional brain imaging (8), where temporal resolution is important but often traded off for spatial resolution or signal-to-noise ratio (SNR). Thus, methods to reduce the required amount of encoding may be helpful.

This may be achieved by reducing the FOV to only the region of interest, but due to potential aliasing from signal possibly outside the FOV, the full FOV is typically encoded, which requires greater scan time. Approaches to avoid aliasing with a reduced FOV include applying a pre-saturation pulse to the unwanted regions (2) or a refocusing pulse to the region of interest (3). However, these methods require multiple radiofrequency (RF) pulses, which increases the specific absorption rate (SAR) deposition, and are sensitive to RF amplitude (B_1) inhomogeneity and off-resonance, which leads to residual signal outside the region of interest. Because these approaches are excitation-based, they can be combined with acquisition-based methods, such as rapid k-space trajectories (9, 10), keyhole techniques (11), and parallel imaging (12–14), or reconstruction-based methods, such as compressed sensing (15, 16), partial Fourier sampling (17), and low-rank approximations (18, 19). Such acceleration methods may obviate the need for reduced FOV imaging, but for some applications, the increase in temporal resolution or SNR afforded by excitation-based methods may be useful.

Another approach is to excite only the region of interest with a multidimensional selective RF pulse (20). Such pulses, especially three-dimensional pulses, are typically quite long and suffer from off-resonance and T_2^* decay. Parallel excitation can decrease the pulse length (21, 22) and excite localized regions with high spatial selectivity (23), but because most MRI sites lack the necessary hardware, the method has found limited use to date. Selective excitation has also been performed with non-linear gradients (24), which may increase the excitation efficiency by encoding along more than one spatial dimension simultaneously, i.e. when exciting a radially symmetric region using a radially symmetric gradient field (25). Since pulse length increases substantially with the number of dimensions that need to be encoded when using linear gradients, this increase may be crucial, but comes at the cost of restricting the shape of the excited region. In reduced FOV applications, however, the exact shape is often not important.

Recently, Ma *et al.* used a second-order gradient concurrently with a spatial-spectral (SPSP) pulse (26) to excite a thin disc-shaped region (1), utilizing the fact that the field generated is circularly symmetric in the x - y plane. The field can be produced by a second-order resistive shim gradient, the Z2 gradient, found in many MR scanners, so no additional hardware is necessary. In addition, because both the field and the targeted region are circularly symmetric in the x - y plane, the method is efficient.

Ma *et al.* demonstrated their method on a phantom, and excited a thin disc-shaped region at isocenter. Here, we extend the method to excite disc-shaped regions of arbitrary thickness, i.e. cylinders, at arbitrary positions. We explicitly take into account the effect of off-resonance on the excitation process due to the limited strength of the available Z2 gradient. We assess the feasibility of the method for functional MRI (fMRI) applications by testing on subjects performing a visual activation task. Finally, we discuss the advantages and limitations of the method, and ways to address these limitations. This work was presented as a conference abstract (27).

Theory

Here we provide a brief overview here (see Ma *et al.* and Supplemental Information online for a more detailed review). In the method, the Z2 gradient is used to produce a unique bandwidth within a confined region of an axial slice, permitting excitation of a fully localized region with a SPSP pulse. The field produced by the Z2 gradient is given by:

$$B_z(x, y, z) = G_{z2} \left(z^2 - \frac{x^2 + y^2}{2} \right) \quad [1]$$

where G_{z2} is the amplitude of the Z2 gradient, with a maximum value of $G_{z2} \approx \pm 1.75$ mT/m² on the scanner utilized here. Thus, in the plane $z = 0$, a SPSP pulse on-resonance at the center of the FOV with a bandwidth

$$BW = \gamma G_{z2} a^2 \quad [2]$$

where γ is the gyromagnetic ratio divided by 2π , excites a circular region of radius $a = \sqrt{BW/(\gamma G_{z2})}$ (see Fig. 1). With a standard SPSP pulse, the excited region is approximately cylindrical if the slice thickness Δz is small compared to the radius, i.e. $\Delta z \ll a\sqrt{2}$. However, a SPSP pulse whose response's center frequency f_c varies with z , more specifically $f_c(z) = \gamma B_z(0, 0, z)$ for $|z| \leq z/2$, can excite cylindrical regions of arbitrary thickness. Designing such a pulse is more complicated since the pulse is not separable along the frequency and slice dimensions. Thus, a numerical approach may be required, i.e. by applying the Fourier Transform to the desired SPSP response and sampling along the k_z - t trajectory. This would also allow B_1 inhomogeneity compensation to be built into the design, but it would only be selective along frequency and slice-position, i.e. spins precessing at the same frequency in a plane will be compensated equally. The design and response of a standard and a non-separable SPSP pulse are shown in Fig. 2. Note that to avoid exciting spins outside the targeted FOV, the opposed null at frequency f_{rep} (or $-f_{\text{rep}}$) must fall outside the object for a given plane.

Off-center regions can be excited by shifting the second-order Z2 field by the addition of linear gradients and an RF frequency offset (see Eq. 3) (1, 28–29), and if necessary, i.e. when shifting in z , modulating the RF frequency (or phase). Note that when shifting the field, it is necessary to ensure that the opposed-nulls still fall outside the object.

$$B_0(x - x_0, y - y_0, z - z_0) = G_{z2} \left(z^2 - (x^2 + y^2)/2 \right) - 2G_{z2}z_0z + G_{z2}x_0x + G_{z2}y_0y + G_{z2} \left(z_0^2 - (x_0^2 + y_0^2)/2 \right) \quad [3]$$

Off-resonance can distort the shape of the Z2 field. Thus, when shifting the field, it may be necessary to fit separate second-order coefficients along x , y , and z since the field may vary independently along each dimension. The shape of the region excited can be determined through simulation.

Methods

Field map

Field maps were acquired using a 2DFT sequence at two echo times $TE_1 = 0.1$ ms and $TE_2 = 2.1$ ms, with matrix size = 64×64 , repetition time $TR = 10$ ms, readout BW = 125 kHz, and FOV = 24 cm for the phantom and 22 cm for the brain.

Phantom and brain scans

Scans were performed on a grid phantom and on human brains. For the grid phantom, images were acquired using a 2DFT sequence with matrix size = 64×64 cm, FOV = 24 cm, $z = 5$ mm, $TE = 30$ ms, $TR = 1$ s, and readout BW = 125 kHz. Second-order coefficients of polynomial fits to the field map along the radial (here, along x) and slice dimensions were calculated and used to design SPSP pulses for exciting cylindrical regions of various radii and slice thicknesses at different center positions (see Results).

For the human brains, functional scans were performed using a block design visual activation task with a flashing checkerboard. Two types of scans were conducted: a full FOV scan ($TR = 2$ s) using a standard sinc pulse and a reduced FOV scan of the visual cortex ($TR = 1.5$ s) using a SPSP pulse and Z2 gradient. The TRs were set based on the minimum achievable with each method for a typical 20 slice acquisition, though only one slice was acquired. The minimum TRs for the reduced and full FOV scans were 70 ms and 92 ms per slice, equivalent to 1.84 s and 1.40 s per volume, respectively. These were rounded to 2 s and 1.5 s to fit evenly in each block. Both scans were 3 min with a block length of 18 s, and used a single-shot spiral-out sequence with $z = 5$ mm, $TE = 30$ ms, and readout BW = 125 kHz. For the full FOV scan, the matrix size was 128×128 with a 22 cm FOV, slightly larger than the head size in the anterior-posterior direction, which (along with resolution) determines the readout length in both spiral and echo-planar imaging. For the reduced FOV scan, the matrix size was 64×64 with an 11 cm FOV that encompassed the excited region. Due to off-resonance, the linear gradient amplitudes and frequency offsets were experimentally modified to excite the targeted region. To assess repeatability, two runs of each scan type (full and reduced FOV) were performed. A total of 9 healthy human subjects was scanned in concordance with IRB guidelines. The study was performed on a 3T GE Discovery 750 scanner (GE Healthcare) using a single channel T/R quadrature head coil.

fMRI analysis

The fMRI data were reconstructed with time-segmented off-resonance correction (30), and the zeroth and first order trend components were removed from the time-series. The theoretical resolution for each scan type was assessed by the full-width at half-maximum of the point-spread function (FPSF). The resolution of the time-series images was assessed by the full-width at half-maximum of the autocorrelation function (FACF) of the noise image, obtained by subtracting the mean of the even time frames from that of the odd time frames, with the number of time frames averaged kept constant between the scans. The image SNR was measured by dividing the mean intensity in a certain brain region by the standard deviation of the same region in the noise image.

Correlation analysis was performed using a convolution of the block design with a standard hemodynamic response function. This produced activation maps (31), which were thresholded at $P < 0.05$ and smoothed with an adaptive filter that used the local standard deviation (32) to preserve regions of larger activation while removing spuriously activated pixels. The numbers of activated pixels were calculated to compare the extent of activation for each method, and the FACFs of the activation maps were calculated to compare the functional spatial resolution. The experimental values were compared with theoretical calculations.

Results

Grid phantom

Localized regions of a grid phantom were excited using SPSP pulses and the Z2 gradient. Due to off-resonance, two second-order coefficients of polynomial fits to the Z2 field were calculated, one along the radial (here, along x) ($G_{Z2,r}$) and one along the slice ($G_{Z2,z}$) dimension. We obtained $G_{Z2,r} = 3.24 \text{ mT/m}^2$ and $G_{Z2,z} = 2.86 \text{ mT/m}^2$. The bandwidth of the pulse was adjusted based on the target radius (see Eq. 2), and the linear gradients and frequency offset were adjusted based on the desired shift of the excited region. Fig. 3 shows the phantom images and cross-sections of interest for different excitation geometries and positions. As shown, the sizes and positions of the excited regions correspond well with the target designs.

Noise and resolution in fMRI images

Figs. 4a–b show one frame of the full and reduced FOV fMRI time-series images for one subject. The mean FACFs of the noise images for the full and reduced FOV scans were 1.418 ± 0.064 and 1.358 ± 0.014 , respectively ($P < 0.02$), indicating higher spatial resolution in the reduced FOV scan. Figs. 4c–f show the activation maps of both runs of the full and reduced FOV fMRI data for one subject, thresholded at $P < 0.05$, which show similar spatial distribution. The full FOV scan has slightly greater number of activated pixels, but the activation maps also appear less sharp, consistent with the resolution estimates using the FACFs of the noise images.

Table 1 shows the number of activated pixels and the FACFs of the activation maps. The reduced FOV scan had on average a lower FACF ($P < 0.01$), corresponding to increased functional spatial resolution, but also a fewer number of activated pixels ($P < 0.03$), though the results were not consistent across subjects. In addition, data for two subjects were discarded because the excited region could not be localized to the reduced FOV (see Discussion).

Discussion

Spatial resolution

The theoretical resolution of the full and reduced FOV scans was calculated using the point spread function of each scan, obtained by reconstructing the weighting term $\exp(-t(R_2^* + i\omega))$ along the spiral trajectory, where R_2^* is the effective transverse relaxation

rate and ω is the off-resonance angular frequency, taking into account the difference in FOV (22 cm versus 11 cm) and readout length (59.8 ms versus 28.6 ms). Since off-resonance correction was performed, only R_2^* effects were considered. The FPSF of the reduced FOV scan was 9% smaller than that of the full FOV scan, which corresponded with the decreased FACFs of the noise images in the reduced FOV scans ($4.1 \pm 3.7\%$ reduction), as well as the decreased FACFs of the activation maps ($3.1 \pm 3.4\%$ reduction). The decreases in the measured values were smaller than in the theoretical calculations because brain images and activation maps have finite sharpness, and thus intrinsically wider autocorrelation functions than the delta function used to calculate the point spread functions. The increased functional resolution afforded by the reduced FOV may be a crucial reason to use high-field spin-echo fMRI (33).

Temporal SNR

The temporal SNR of a reduced FOV time-series image relative to that of the full FOV time-series image is determined by inter-related scan parameters, and given by:

$$\frac{SNR_r}{SNR_f} \approx \sqrt{\frac{(1 - E_r)/(1 + E_r)N_r T_r}{(1 - E_f)/(1 + E_f)N_f T_f}} \quad [4]$$

where $E_i = e^{-TR_i/T_1}$ with T_1 being the longitudinal relaxation time, N is the number of temporal frames, T is the readout length, and the subscripts f and r correspond to the full and reduced FOV scans.

Given the scan parameters used, the reduced FOV time-series images should have theoretically 28.6% less temporal SNR, which can explain the fewer numbers of activated pixels. The image SNR ratio, given by Eq. 4 without the N terms, was experimentally calculated. The reduced FOV scan showed $26.5 \pm 16.9\%$ less image SNR, slightly lower than the theoretical 38.2%. The reduction also varied considerably across subjects, possibly from aliasing due to the long transition band of the excitation profile or from shifting of regions outside the FOV into the passband frequency due to off-resonance. In both cases, the noise variation increases. If the additional noise contains a physiological component, the results may be biased in a particular way. Furthermore, if the noise fluctuations affect the static magnetic field, the aliasing may also be inconsistent across frames, adding further bias. However, this may be mitigated using methods such as RETROICOR (34) that remove the effects of physiological noise, i.e. from breathing or cardiac pulsatility.

Limitations

The major limitation of the method, implemented with the resistive shim gradient, is the low bandwidth of the SPSP pulse, which increases the excitation's sensitivity to background off-resonance and reliability in exciting regions localized to the reduced FOV. With spiral trajectories, aliasing results in a spiral-like streaking artifact. This occurred for two subjects, whose data were thus discarded. The low bandwidth also increases the pulse length, and thus the minimum TE. In applications with relatively long TEs, such as fMRI, this may be acceptable, but may still require the use a low time-bandwidth pulse, which produces long transition regions in the excitation profile, and thus a larger reduced FOV to avoid aliasing.

These problems can be substantially mitigated with a higher strength gradient, which would likely need to be custom-built. This would allow the region of interest to more reliably be excited and localized to the reduced FOV. However, this also requires the aliases of the passband to be farther out in frequency to avoid unwanted excitation, which decreases the maximum subpulse length and hence limits the minimum slice thickness. Thus, the ideal gradient strength gives a reasonable compromise between insensitivity to off-resonance and a small minimum slice thickness.

There are other considerations of the method. The shim gradient is static, and generates off-resonance during the readout. This could be addressed with a pulsed gradient, but with standard hardware, off-resonance correction is typically necessary. Standard motion correction methods are also affected, as with any reduced FOV method, but should work as long as there are contrasting features in the image. Oblique imaging is limited since the isocontours of the Z2 field become more elliptical as the plane is tilted away from axial, allowing the FOV to be reduced less for a given bandwidth pulse. SAR is greater with the SPSP pulse than with a standard sinc pulse for a given flip angle and duration. The difference depends on the specific pulse parameters, but the SAR of the SPSP pulse is still low and well within safety limits.

Conclusions

A second-order shim gradient was used to excite disc-shaped regions using a previously developed method (1), which was extended and demonstrated on a grid phantom and human brains. The phantom results showed high correspondence with simulation. The functional brain activation maps showed higher spatial resolution in the reduced FOV scans than in the full FOV scans, at the cost of reduced SNR. The major limitation of the method, as implemented here, is the low strength of the Z2 gradient, and thus a practical implementation will require a custom gradient coil.

Supplementary Material

Refer to Web version on PubMed Central for supplementary material.

Acknowledgments

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Fig. 1.

Z2 field and excitation pattern with a SPSP pulse. **a:** The target region (with radius a) is highlighted in white in the center, and the ring-shaped region excited by the opposed null is highlighted in white near the edge of the object (with radius R). To prevent aliasing with a reduced FOV, the SPSP pulse should be designed so that the "ring" falls outside the object.

b: Both regions are confined to the slice thickness Δz .

Fig. 2.

SPSP pulse designs and their responses. **a:** Standard SPSP pulse with parameters: $\alpha = 90^\circ$, $z = 1$ cm, $BW = \pm 50$ Hz, $f_{rep} = 800$ Hz, and spatial and spectral $TB = 2$. **b:** SPSP response of pulse in (a). **c:** SPSP pulse with spatially varying frequency response, designed with $G_{z2} = 0.002$ mT/m² with parameters: $\alpha = 90^\circ$, $z = 10$ cm, $r = 4$ cm, spectral $TB = 2$, and spatial $TB = 4$. **d:** SPSP response of pulse in (c).

Fig. 3.

Full and reduced FOV images of a grid phantom. **a:** full FOV (24 cm). **b:** $a = 4$ cm. **c:** $a = 6$ cm. **d:** $a = 4$ cm at $x_0 = y_0 = -4$ cm. **e:** side-view projection, $a = 4$ cm, $z = 10$ cm. **f:** $a = 4$ cm, $FOV = 8$ cm. **g:** $a = 4$ cm at $z_0 = 4$ cm. **h:** profile in (c). **i:** profiles in (d). **j:** profile in (e). The abscissa in (h–j) are the position along the FOV, and range from -12 cm to $+12$ cm.

Fig. 4. Full and reduced FOV brain images (subject 3). **a–b:** Time-series image from full (a) and reduced FOV (b) fMRI scan. The reduced FOV region in the full FOV image is shown in the white rectangle. **c–f:** Single subject activation maps ($P < 0.05$) of both runs of full (c–d) and reduced (e–f) FOV fMRI data.

Table 1

Number of activated pixels N_{act} and FACFs of the activation maps of the full FOV (fFOV) and reduced FOV (rFOV) fMRI data. The two numbers in each column for a given subject correspond to the two runs performed.

Subject	fFOV N_{act}	rFOV N_{act}	fFOV FACF	rFOV FACF				
1	577	675	375	434	1.49	1.50	1.48	1.49
2	531	612	531	566	1.47	1.49	1.50	1.47
3	577	577	601	739	1.50	1.48	1.51	1.50
4	445	359	137	153	1.52	1.51	1.42	1.49
5	397	384	396	407	1.58	1.53	1.46	1.45
6	499	684	488	578	1.46	1.50	1.43	1.41
7	576	662	478	447	1.59	1.56	1.44	1.46
Mean $\pm$ Std	540 $\pm$ 108	452 $\pm$ 162	1.51 $\pm$.040	1.47 $\pm$.032				