On the Artifact of a Subvoxel Susceptibility Deviation in Spoiled Gradient-Echo Imaging

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In MRI, susceptibility-based negative contrast amplifies the effect of objects that are too small to be detected by water displacement or intrinsic contrast properties. In this work, a simplified description of the susceptibility artifact of a subvoxel object in spoiled gradient-echo imaging is presented that focuses on the elimination of signal in its vicinity: the dephased-volume. The size and position of the dephased-volume are investigated using 3D time-domain simulations and in vitro experiments in which scan parameters and object magnetic moment are systematically varied. Overall signal loss is found to be linearly related to a dephasing parameter that contains the susceptibility difference with tissue, object volume, and echo time (TE), and thus allows the magnetic moment of the object to be assessed. Gradient strength, in-plane resolution, fractional echo, and slice orientation have limited influence. For the settings used, the center of mass of the artifact was always within 0.5 mm of the object’s in-plane position. Magn Reson Med 50: 400–404, 2003. © 2003 Wiley-Liss, Inc.

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In MRI, the negative contrast generated by susceptibility effects is widely used to visualize objects and to study processes. In this context, the relaxation effects of systems of infinite cylinders or microscopic spheres have been studied to model susceptibility effects in a capillary bed (1,2) and a suspension of blood cells (3,4). Until now, little attention has been given to the appearance of a small individual object of deviating susceptibility. However, insights into mesoscopic susceptibility deviations would be relevant to the presentation of, e.g., iron deposits (ferritin), embolic metal fragments (5), paramagnetic markers on interventional devices (6), seeds used in brachytherapy (7), clusters of magnetically labeled cells (8,9), and air bubbles.

In this work, the susceptibility artifact of a subvoxel object in gradient-echo imaging, assuming an otherwise homogeneous environment, is studied. We introduce the dephased-volume model, which simplifies the effect of a subvoxel susceptibility deviation to a localized suppression of signal. The size and positional accuracy of the dephased volume in relation to scan parameters and object composition are determined by 3D time-domain simulation of the imaging process and by in vitro experiments.

THEORY

The magnetic volume susceptibility $\chi$ of a material (henceforth termed “susceptibility”) is a dimensionless quantitative measure of the material’s response to an applied magnetic field. Susceptibility differences are a major source of $B_0$ inhomogeneities. These field inhomogeneities $\Delta B_0(r)$ lead to slice distortion and signal displacements in the frequency-encoding direction, intensity distortions and, in gradient-echo imaging, signal loss due to intravoxel dephasing and echo shift from the center of the acquisition window (10,11).

The phase dispersion at the echo time (TE) of the magnetization $M(r)$ that is mapped into the same voxel changes the signal $I$ according to:

$$ I \propto \int M(r) e^{i\Delta B_0(r) / \gamma TE} d^3r. \tag{1} $$

Define a subvoxel object as one that will fit in a sphere with radius $d$, the in-plane pixel size. Then, for a subvoxel object with volume $V$, with a susceptibility difference with the surrounding tissue $\Delta \chi \ll 1$, the inhomogeneous part of the field disturbance for $\rho_d > 1$ corresponds to the field of a magnetic dipole moment $m = \Delta \chi V B_0 / \mu_0$ (12), given by:

$$ \Delta B_0(\rho_d, \vartheta, \varphi) = \frac{B_0 \Delta \chi V}{4\pi d^3} \left( 1 - 3\cos^2 \vartheta \right), \tag{2} $$

using a polar coordinate frame. Here, $\rho_d$ denotes the radial coordinate normalized to the pixel size. From Eqs. [1] and [2] it is directly apparent that the intravoxel dephasing at TE is determined by:

$$ K = \frac{\gamma TE B_0 \Delta \chi V}{4\pi d^3}. \tag{3} $$

Physically, $K$ is equal to the phase at the TE for $\rho_d = 1$, in the equatorial plane ($\varphi = 0$).

Dephased-Volume Model

The premise of the dephased-volume model is that intravoxel dephasing caused by the phase accumulation at the TE [see Eq. [1]] overshadows geometrical distortions. This
condition is largely met for gradient-echo imaging of a subvoxel object in a thick imaging slice. In that case, the dispersion of the frequency distribution is primarily attributable to the voxel’s extent in the slice direction (13, 14), and in-plane variations can be neglected. All of the elongated voxels will contain on-resonance spins, because $\Delta B_z(r)$ falls off rapidly to zero. Thus, any magnetization with a phase accumulation at TE, $\Delta \Phi_{TE}$, such that $|\Delta \Phi_{TE}| = |\gamma \Delta B_z(r) TE| > 2\pi$ will find some magnetization somewhere in the slice direction that cancels its signal, and is part of the dephased volume (Fig. 1). Since the dephased volume will contain the near field, $p_d = 1$, of the object, the exact shape of $\Delta B_z(r)$ near the object is unimportant. Then, since only regions with $|\Delta \Phi_{TE}| < 2\pi$ contribute to the voxel signal, geometric distortions are limited to $|\Delta r| < 2\pi/\gamma TE G_r$, with $G_r$ being the readout gradient strength. This limit is less than a pixel for full-echo acquisitions. In summary, total signal loss is expected to scale with $K$, to be insensitive to in-plane resolution, and to be independent of geometrical distortions. In addition, the position of the artifact is expected to accurately reflect object position.

**METHODS**

**Time-Domain Simulations**

Simulations were performed to study the susceptibility artifact in the absence of system nonidealities, and to study the dependence on parameters that cannot easily be varied. A 3D supersampled time-domain simulation program, similar to that described in Ref. 15, was implemented. The field disturbance was introduced using the analytical expression of Eq. [2].

Proton imaging, $\gamma = 2\pi \cdot 42.576 \cdot 10^6$ rad/s/T, at 1.5 T, was simulated using the following scan parameters: field of view (FOV) = 256 mm, imaging matrix = 256, a full echo centered at 9.2 ms, $G_R = 5.2$ mT/m, slice thickness = 20 mm, and a slice-selection pulse with a bandwidth of 2000 Hz. In the default slice orientation, the readout direction was parallel to $B_0$. The object was a sphere with a radius of 0.5 mm, with $\Delta \chi = 860$ ppm, giving $\Delta \chi V = 450 \cdot 10^{-6}$ mm$^3$.

Starting from these settings, the parameters were systematically varied: TE ranged from 1.5 to 25 ms for a full echo, and from 0.5 to 9.2 ms for a 62.5% fractional echo, increasing $G_R$ when necessary. Three slice orientations were examined: a coronal slice ($B_0$ in-plane) with readout respectively parallel or perpendicular to $B_0$, and an axial slice ($B_0$ through-plane). For these orientations the effects of readout gradient polarity and fractional echo were studied. Slice thickness was varied from 5 to 50 mm, FOV from 128 to 512 mm, and asymmetric echo fraction from 62.5% to 100%. $G_R$ was varied between 1.7 and 10.4 mT/m. In addition, a fully phase-encoded image was simulated (i.e., with no geometric distortion).

The effect of increasing the susceptibility of the sphere was examined, with $\Delta \chi V$ ranging from 100 to 4000 $\cdot 10^{-6}$ mm$^3$. Finally, the dephasing parameter $K$ was varied over four decades (0.01–100), assuming an infinitesimally small object.

**In Vitro Experiments**

Imaging was performed on a 1.5-T system (Intera; Philips Medical Systems, Best, The Netherlands), using a 22-cm circular coil for signal reception. A small Dy$_2$O$_3$ ring marker from a 1.7-mm diameter catheter, $\Delta \chi V = 450 \cdot 10^{-6}$ mm$^3$, was embedded in agar. Scan parameters were systematically varied as described in the Simulations section, if this was permitted by scanner software and hardware. The slice was centered at the marker. We used a flip angle of 10° and a default repetition time (TR) of 20 ms. In addition, the orientation of the imaging slice was varied in steps of 15° from coronal ($B_0$ in-plane) to axial ($B_0$ through-plane).

**Image Analysis**

In the images, integrated signal loss and artifact center of mass were evaluated. In the simulations, the signal of the undisturbed object was known, and in the in vitro experiments the average signal of an undisturbed area was used as a reference. Integrated signal loss, $V_{loss}$, was obtained by summing the signal loss over the region of interest (ROI). The center of mass of the artifact was calculated by integrating the product of position and loss over the ROI; only pixels with a negative signal change were included in the calculation.

The relation between $V_{loss}$ and $K$ was determined by linear regression through the origin. To this end we only used data with $K < 300$, obtained in the simulations that varied $\Delta \chi$, $V$, $K$, and TE, and in the experiments that varied TE, all using full echoes. Coefficients of proportionality are given $\pm 95\%$ confidence limits.

**RESULTS**

**Scan Parameters**

Simulations and experiments showed that artifact size increases with TE (Fig. 2a): $V_{loss}$ was nearly linearly related to TE, and thus to the dephasing parameter $K$. For TE longer than 10 ms, $V_{loss}$ is smaller than expected from a linear relation. Simulated and experimentally obtained $V_{loss}$ correlated well.

Figure 3 shows how the artifact changes with gradient polarity and using fractional echo for three different orientations. For the full-echo acquisitions, the images are highly similar to the projections of the dephased volume in Fig. 1. For fractional-echo acquisitions, however, the shape of the artifact changes by echo shifting of the signal from regions where the gradient of the field disturbance has the same polarity as the readout gradient. This one-sided signal loss slightly shifted the center of mass of the artifact and contributed to $V_{loss}$, which was up to 20% larger for fractional-echo acquisitions.

Experimentally, $V_{loss}$ was independent of the orientation of the readout gradient within 3%, except that for fractional echo a 7% increase was caused by swapping readout and phase-encoding directions. Likewise, for the settings examined, $V_{loss}$ was independent of the readout gradient strength (3% variation for full echoes). Although displacements within the artifact were apparent at reduced gradient strengths (Fig. 3), the shift of the center of mass remained well within 0.5 mm. $V_{loss}$ was hardly reduced for higher in-plane resolution (12% variation from 0.5 to 2 mm).
Signal loss was slightly larger for a transverse slice than for coronal/sagittal slices (Fig. 2). A peak is apparent for intermediate angles. This peak is related to an increased overlap of the poles and the toroid, which contain phases of opposite sign, resulting in increased intravoxel dephasing.

The shape of the artifact varied with slice thickness (Fig. 4): for slices over 10 mm, the shape of the artifact resembled the projection of the dephased volume in Fig. 1. Poles and toroid were unrecognizable when the thick-slice assumption was violated (e.g., for a 2.5-mm slice). $V_{\text{loss}}$ increased until the slice thickness was, in this case, about 20 mm and then leveled off (Fig. 2c). Thus, for thin slices the dephased volume is only partly contained in the imaging slice, and hence increasing the slice thickness will pick up more dephasing until the dephased volume is completely contained in the

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**FIG. 1.** Dephased volume. Surfaces of $2\pi$ (poles) and $-2\pi$ (toroid) phase accumulation during TE delimiting the dephased volume of a magnetic moment positioned at the origin (a). Dephasing parameter $K = 130$. Projection of the dephased volume on the $xz$-plane (b) and the $xy$-plane (c), respectively.

**FIG. 2.** a: $V_{\text{loss}}$ as a function of TE. On the top axis the dephasing parameter $K$ is indicated, multiplied by the pixel size cubed. The solid line is a fit to the results for TEs up to 9.2 ms, forced through the origin. b: $V_{\text{loss}}$ as a function of the angle of the imaging plane with $B_0$: $0^\circ$ corresponds to a coronal/sagittal slice, $90^\circ$ to a transverse slice. c: $V_{\text{loss}}$ vs. slice thickness. d: Simulated $V_{\text{loss}}$ vs. $\Delta V$. The solid line is a linear fit through the origin to the first five data points.
Increasing the slice thickness beyond this point will not improve the detectable effect of the object.

**Magnetic Moment**

The simulation allowed us to evaluate the extent of the artifact for a range of $\Delta \chi V$, i.e., the magnetic content of the susceptibility deviation. Figure 2d shows that $V_{\text{loss}}$, the volume of dephased signal (in mm$^3$) was nearly linearly proportional to $K d^3$. In addition, the signal loss was well localized, and shown to be within 0.5 mm (0.5 $d$) of the object for practical settings. The shape of the artifact, for full-echo acquisitions, resembles the projection of the dephased-volume in the slice direction (Fig. 1).

$V_{\text{loss}}$, being proportional to $K$, the most important parameters are the para-/diamagnetic content of the susceptibility deviation, $\Delta \chi V$, and the TE. The increase of artifact size with TE in gradient-echo sequences has long been recognized (13,14,17). Neither phase-encoding direction nor readout gradient strength (scan parameters that have been shown to strongly influence artifact appearance in spin-echo imaging (10,18)) had an effect on the size of the artifact in this case. Likewise, the dependence of signal loss on in-plane resolution, which has a strong influence in nearly isotropic 3D gradient-echo scans, is almost absent. The independence of $V_{\text{loss}}$ on these parameters bears on the thick slice condition that makes intravoxel dephasing dominant over geometrical distortion.

The exact shape of the object with deviating susceptibility appears to be immaterial when $|\Delta \chi| \ll 1$, as long as the susceptibility deviation is small compared to the pixel size. This is supported by the close correspondence between the simulations and in vitro experiments that were performed using a 0.5-mm sphere and a thin 1.7-mm diameter ring.

**DISCUSSION**

Susceptibility-based negative contrast provides a way to amplify the effect of objects that are too small to be detected easily using MR—for example, clusters of cells (8,9), lymph nodes (16), and interventional devices (6). For these applications, reliable negative contrast and positional accuracy are desired.

In this study, the dephased-volume model is used to describe the artifact of a subvoxel susceptibility deviation in gradient-echo imaging using a thick slice. In this model the artifact is viewed simply as a localized removal of the signal from the vicinity of the object. The size of the artifact was demonstrated in simulations and in in vitro experiments to scale with the dephasing parameter $K$:

$$V_{\text{loss}} \propto (3.4 \pm 0.2) K d^3, \ r = 0.998 \ (N = 14).$$

The experimental results obtained by varying TE were characterized by $V_{\text{loss}} = (3.1 \pm 0.3) K d^3, \ r = 0.997 \ (N = 6)$. 

![FIG. 3. Simulated (odd rows) and measured (even rows) susceptibility artifacts for different imaging geometries: coronal/sagittal slice with readout parallel to $B_0$ (top rows), coronal/sagittal slice with readout perpendicular to $B_0$ (middle rows), and transverse slice (bottom rows). Results for a readout gradient strength of 5.2 mT/m (first column), fractional echo (second column), and a lower gradient strength of 1.7 mT/m with opposite readout gradient polarities (third and fourth columns) are shown. Pluses and minuses indicate polarity and direction of the readout gradient. Tiles are 32 mm.](image)

![FIG. 4. Simulated (top row) and measured (bottom row) susceptibility artifacts for varying slice thickness. Slice thickness is indicated in mm. Tiles are 32 mm.](image)
Three parameters, which are not included in K, may significantly change $V_{\text{loss}}$, i.e., slice thickness, fractional-echo acquisition, and slice orientation. When the thick-slice condition is violated, the dephased-volume model no longer applies. Phase dispersion in-plane becomes important, and the dephased volume will only be partly contained in the imaging slice. Fractional-echo acquisitions change the shape and slightly displace the center of mass of the artifact. The signal that is preserved at the zeros of the dipole field for full-echo acquisitions is shifted outside the acquisition window by the nonzero local gradient (11). Similar effects are expected for half-matrix acquisition in the phase-encoding direction.

The orientation of the imaging slice affects the artifact shape, which can be imagined as taking different projections of the dephased volume (Fig. 1). In going from coronal to transverse, the toroid becomes wider and finally becomes a circle. The poles will increasingly overlap the toroid, and because poles and toroid are of opposite polarity, this overlap will result in a slightly increased signal loss.

Visualization methods that employ negative contrast benefit from a high background signal. Therefore, scan parameters should be optimized so as to yield the maximum possible signal from the tissues surrounding the object. Thus, although TR, flip angle, readout gradient strength, gradient moment nulling, etc., do not influence the overall signal loss in mm³ that relates to the extent of the region void of signal, they do affect the conspicuity of the object by changing the signal yield of background tissue.

**Magnetic Moment**

A quantitative relation between object composition, i.e., $\Delta \chi$, $V$, scan parameters, and artifact size, can be used to narrow down the possible sources of artifacts encountered (5). In addition, such a relation allows one to design susceptibility markers that provide the desired artifact size.

The constant of proportionality relating $K d^3$ and $V_{\text{loss}}$ has been determined to be between 2.8 and 3.6. This value is considerably larger than the volume bounded by the $|\Delta \phi_{\text{TE}}| = 2\pi$ surfaces used in introducing the dephased volume, which is $8\sqrt{3/27} K d^3 \approx 0.51 K d^3$. Clearly, magnetization with $|\Delta \phi_{\text{TE}}| < 2\pi$ contributes significantly to the signal loss. On the other hand, $V_{\text{loss}}$ is significantly smaller than if the magnetization in a dipole field were summed over all of space—that is, as if the image consisted of one big voxel (4,19): $8\pi^2\sqrt{3/27} K d^3 \approx 5.06 K d^3$. Integration of the complex signal change over the ROI in the simulations obtained by varying $\Delta \chi$, $V$, and TE resulted in $V_{\text{loss}} = (4.9 \pm 0.1) K d^3$. The difference between $V_{\text{loss}}$ on magnitude images and the loss predicted by Yablonskiy and Haacke (4) is related to the spatial separation of positive and negative frequencies in the dipole field. This separation reduces intravoxel phase dispersion, especially in slices parallel to $B_0$. Note that for slices at an angle of 45°–60° with $B_0$, frequency separation is reduced and $V_{\text{loss}}$ comes close to the value predicted in Ref. 4. For long TEs and high $\Delta \chi$ (see Fig. 2a and d), $V_{\text{loss}}$ is noticeably less than proportional to $K d^3$; apart from the spatial separation of frequencies, the slice in those cases will not completely encompass the dephased volume. The nearly linear dependence of $V_{\text{loss}}$ on $K$, with the constant of proportionality determined, allows one to estimate $\Delta \chi$ from the artifact. However, because the object volume is not easily determined, one cannot distinguish between objects with the same $\Delta \chi$ product.

**REFERENCES**