JOURNAL OF MAGNETIC RESONANCE IMAGING 12:795–797 (2000)

Technical Note

Elimination of Magnetic Field Foldover Artifacts in MR Images

David J. Larkman, PhD,* Amy H. Herlihy, PhD, Glyn A. Coutts, PhD, and Joseph V. Hajnal, PhD

Foldover artifacts arise when the same imaging frequency occurs both at a desired location within a slice and at another location within the sensitive region of the radiofrequency (RF) coil. Foldover artifacts can be caused by nonlinearity in the gradient system and by inhomogeneity in B_0. This study investigates an approach in which an extra RF receiver coil and a postprocessing method are used to identify and remove foldover artifacts. J. Magn. Reson. Imaging 2000;12:795–797. © 2000 Wiley-Liss, Inc.

Index terms: foldover artifact; cusp artifact; short bore

CONVENTIONAL SLICE SELECTION is achieved by the combination of a linear field gradient and a frequency-selective radio frequency (RF) pulse such that spins are only excited in a slab-shaped spatial region where the local Larmour frequency matches the range of frequencies ($\omega - \Delta\omega/2$ to $\omega + \Delta\omega/2$) of the RF pulse. However, in certain situations it is possible that within the sensitive region of the RF coil, gradient inhomogeneity and static field inhomogeneity can result in additional spins that satisfy the Larmour condition. These additional signals generally arise from the fringe regions of gradient and main fields and are mapped into the image in a spatially distorted manner to produce "foldover" artifacts. The result can be focal, high-intensity signals that overlap the desired image data.

These artifacts can occur on conventional scanners (1), but they frequently manifest more seriously on systems with short magnets and/or short gradient tubes. Foldover artifacts can also occur as a result of spatial degeneracy in the frequency or phase-encoding gradient directions. It may be possible to alter the scan parameters and/or increase the gradient amplitude to move the artifacts out of the field of view, but this cannot be achieved in all cases. Since the standard gradient localization schemes fail for foldover artifacts, these artifacts cannot normally be separated from the desired data. However, the fact that the signals arise from spatially distinct locations does allow them to be separated utilizing the spatial sensitivity of multiple RF receive coils. Previous work has used similar methods to reduce the number of phase-encode steps required in spin wrap imaging (SENSE) (2–4) and to process data obtained by simultaneously excited multiple slices using modified RF pulses (5).

THEORY

Figure 1 shows a schematic illustration of the spatial variation of magnetic field (B) produced by the combination of the static B_0 field and the gradient field applied during RF excitation. The Larmor frequency (ω) is proportional to B, and so when a tailored RF pulse is used to excite a chosen slice or slab at a in Fig. 1, it also excites the spins at b. The standard imaging coil (birdcage coil 1 in Fig. 1) detects signal from both regions, giving rise to an artifacted image. Coil 2 provides extra information about both regions a and b, but with different weightings. Each coil is connected to a separate receiver, and each results in separate image data. At a single pixel the signals from coil 1, C_1, and coil 2, C_2, are given by the matrix equation;

$$\begin{bmatrix} S_{1a} & S_{1b} \\ S_{2a} & S_{2b} \end{bmatrix} \cdot \begin{bmatrix} x_a \\ x_b \end{bmatrix} = \begin{bmatrix} C_1 \\ C_2 \end{bmatrix} \tag{1}$$

or

$$[S] \cdot [x] = [C] \tag{2}$$

where S_{ij} is the complex sensitivity of coil i to position j, and x_j is the spatially dependent complex signal from position j. Provided $[S]$ is nonsingular, solving Eq. [1] for the vector $[x]$ will then separate out the desired signals from the extra artifactual signals:

$$[x] = [S]^{-1} \cdot [\mathrm{C}] \tag{3}$$

Where $[S]^{-1}$ is the inverse of the complex sensitivity matrix $[S]$.

The elements of $[S]$ can be determined by imaging a uniform test object. Placing the object at the location of volume a, while leaving volume b empty yields S_{1a} from the image obtained from coil 1 and S_{2a} from the image obtained from coil 2. Repeating the procedure with the

The Robert Steiner Magnetic Resonance Imaging Unit, Imperial College School of Medicine, Hammersmith Hospital, London W12 0HS, United Kingdom.

*Address reprint requests to: D.J.L., The Robert Steiner MRI Unit, Imperial College, Hammersmith Hospital Campus, Ducane Road, London W12 OHS, UK.

Received February 23, 2000; Accepted June 26, 2000.

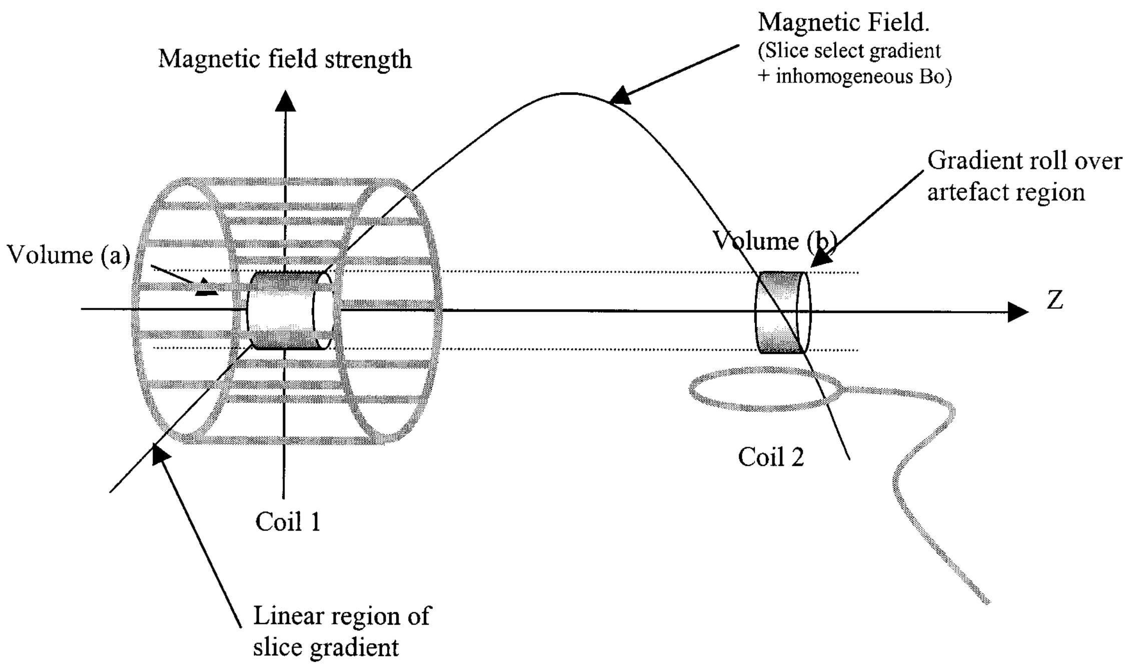

Figure 1. A sketch of the magnetic fields generating a fold-over artifact arising from region b in addition to signal from desired region a. Coil 1 is the primary imaging coil, and coil two is used to obtain additional information required to separate out the artifact. For clarity coil 1 is shown reduced in size; in reality coil 1 is much larger, partially enveloping coil 2.

object at volume b with volume a empty yields S_{1b} from coil 1 and S_{2b} from coil 2.

The process of producing a corrected image for region a employs information from coil 2 to cancel out the signals from region b. This process also introduces noise from coil 2, which is undesired. For pixels in which there is negligible contribution of signals from region b, application of Eq. [3] adds noise without changing the information content. For this reason is it advantageous to make use of data from coil 2 only for those pixels for which it is necessary to remove artifactual signals arising from region b. For pixels that contain low levels of signal from region b, S_{1b} and S_{2b} will be small, so that either may be used as a basis for inclusion or exclusion of data from coil 2. More generally, wherever coil 2 does not yield information that is significantly distinct from coil 1, the system of equations is ill conditioned. The modulus of the determinant of $[S](|\Delta S|)$ provides a measure for this, and so we employ it as a means of identifying pixels that either will or will not be modified with information from coil 2. The appropriate threshold value of $|\Delta S|$, above which correction is applied, depends on the details of the hardware used and the spatial distribution of the artifact signal. We have found a suitable value to be half the mean value of $|\Delta S|$ for the systems we have tested.

Application of Eq. [3] to correct pixel values also performs a signal normalization that compensates for the local coil sensitivities. In order to combine both corrected and original pixel data within a single image, this coil sensitivity compensation must be removed. To achieve this we multiply the corrected pixel value by $|S_{1a}|$. The result is the signal that would have been produced by coil 1 in the absence of any signal found in region b.

MATERIALS AND METHODS

Data were acquired using a 1.0-T prototype neonatal scanner (6) (Marconi Medical Systems, Cleveland, OH), which has a 38-cm-long bore and produces foldover artifacts at approximately 10 cm from isocenter for a conventional spin-echo sequence. This scanner is equipped with a birdcage transmit and receive system to which was added a 7-cm-diameter receive-only surface coil. The birdcage body coil was positioned at the magnet isocenter. The 7-cm-diameter surface coil was positioned adjacent to a location from where signal is folded into the main image. Sensitivity maps of the RF coils were obtained by imaging a uniform cylindrical 10-cm-diameter copper sulfate-doped phantom. The phantom was located at isocenter and imaged with both coils and was then moved to the location of the source of the artifact and again imaged with both RF coils. The moduli of the sensitivity maps are shown in Fig. 2a, and the modulus of the determinant of $[S]$ is shown in Fig. 2b.

Anatomic data were collected by imaging the left hand of a normal volunteer. A spin-echo sequence was used, with the following parameters: TR/TE 200/20 msec, field of view 20 cm, slice thickness 5 mm, and 128 × 256 data matrix. Complex image data were collected in all cases. The foldover field artifact was removed from the images by solving the complex linear system as described.

RESULTS

Figure 3a shows a transverse slice through the hand obtained from coil 1. Note the large, intense foldover artifact that appears near the bottom of the image. The corrected image, obtained by correcting pixels in Fig. 3a, wherever the modulus of the determinant of [S] was greater than 50% of its mean value in the image plane, is shown in Fig. 3b. The artifact has been largely removed.

DISCUSSION

It has been demonstrated that foldover artifacts can be removed from images by separating the foldover data from the desired data. The noise level at the corrected pixels is strongly dependent on $|\Delta S|$, which is a measure of the linear independence of the signals in the images being combined for the correction. This can be used to good effect to restrict correction of the data from the imaging coil to only those regions that are adversely affected by the foldover artifacts to minimize loss of signal-to-noise ratio. The values of the sensitivity matrix (S) required are a measure of basic field properties of a given scanner, so that determination of the elements of S need not be repeated for each examination for a given set of sequence parameters.

Correction for foldover artifacts using multiple receiver coils is simple to implement, avoids pulse sequence limitations, and can be applied to a wide variety of situations, maintaining its integrity with all slice orientations where foldover artifacts contaminate image data. More generally, this method could also be applied to any case where images overlap due to spatially repeating resonant frequencies. In addition to removing artifacts, similar principles can be applied to increase acquisition speed.

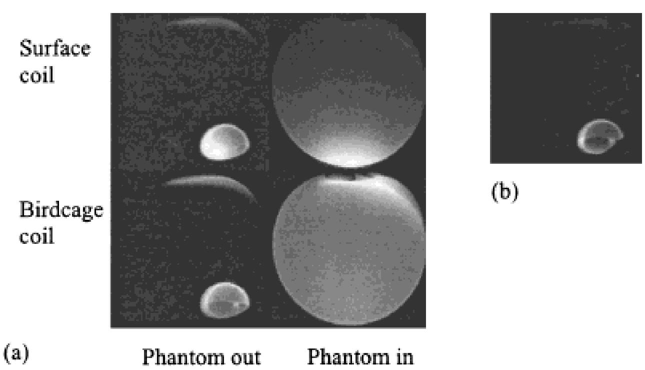

Figure 2. **a:** The magnitude of the sensitivity map, matrix [S] in Eq. [3]. **b:** The magnitude of the determinant of this matrix used to identify the image regions in which to apply the correction.

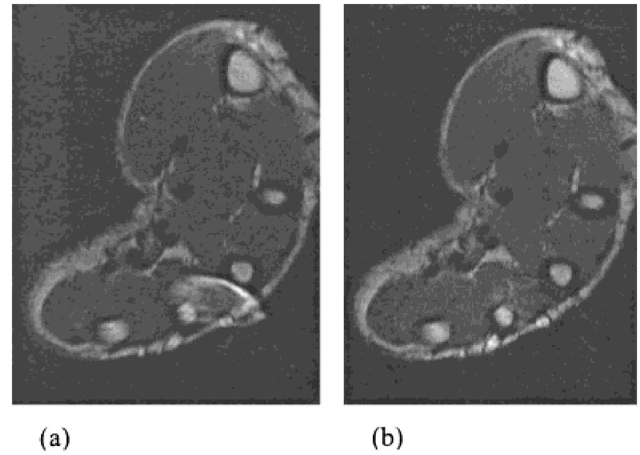

Figure 3. Uncorrected (**a**) and corrected (**b**) images of a volunteer's hand. The artifact (a, arrow) is removed in b, leaving a corrected area with a small local reduction in SNR.

REFERENCES

1. Kim JK, White LM, Hinks RS, King KF. The FSE cusp artifact: a phase wrap-in artifact seen on routine clinical MR images of the knee. In: Proceedings of the ISMRM Annual Meeting, 1999. p 1033.
2. Ra JB, Rim CY. Fast imaging using subencoding data sets from multiple detectors. Magn Reson Med 1993;30:142–145.
3. Hutchinson M, Raff U. Fast MRI data acquisition using multiple detectors. Magn Reson Med 1988;6:87–91.
4. Pruessmann KP, Weiger M, Scheidegger MB, Boesiger P. SENSE: sensitivity encoding for fast MRI. Magn Reson Med 1999;42:952–962.
5. Larkman DJ, Hajnal JV, Herlihy AH, Coutts GA, Young IR, Ehnholm G. Use of multi-coil arrays for separation of signal from multiple slices simultaneously excited. In Proceedings of the ISMRM Annual Meeting, 1999. p 91.
6. Hall AS, Young IR, Davies FJ, Mohapatra SN. A dedicated magnetic resonance system in a neonatal intensive therapy unit. In Bradley WG, Bydder GM, editors. Advanced MR imaging techniques. London: Martin Dunitz; 1997. p 281–290.