Summary

Geometric distortion in MRI comes from multiple sources. Some of these distortions are corrected by manufacturers’ image reconstruction software, and some are not. Gradient-induced distortions are typically the largest source of distortion, except in EPI sequences. Gradient-induced distortions are corrected in all three directions for some MR pulse sequences, but for another class of commonly-used sequences, distortion in the slice direction is not always corrected. The distortion in the slice direction for this large class of pulse sequences can be several millimeters.

Distortions that arise from disturbances in the background magnetic field (‘B,’ field) are also not corrected by manufacturers’ software. These B, distortions are further differentiated by those due to residual inhomogeneity in the magnetic field (‘hardware’ distortions), and disturbances induced by placing a patient into the scanner (‘patient-induced’ distortions).

A phantom-based technique can measure and correct much of the distortion left uncorrected by manufacturers’ software (Mallozzi, Blezek, Gunter, Jack, & Levy, 2006).

Details

Most MRI pulse sequences in clinical scanners yield images where distortion is less than a few millimeters. This is acceptable for many applications, particularly diagnosis based on qualitative visualization of the anatomy. However, some quantitative applications place higher demands on geometric fidelity. Most notable among these are radiation therapy planning, and structural studies of the brain for diagnosing and tracking neurological disorders.

Distortion in MRI arises from two general sources: nonlinearity in the gradient magnetic field, and inhomogeneity of the static magnetic field. Below each type is discussed.
Gradient-induced distortions

The gradient magnetic field is the component of the scanner that encodes spatial information. Imprecise calibration and the inherent nonlinearity in the gradient field lead to geometric distortion. The specifications adhered to for these distortions are designed primarily for non-quantitative diagnostic imaging, and may not be sufficient for quantitative applications.

For example, gradient calibration is performed with an accuracy of approximately 1%. To pass the accreditation for the American College of Radiology (ACR), a circular object of diameter 190 mm must be measured to be within plus or minus 2 mm of its true diameter (American College of Radiology, 2005). Thus anatomy that is only 10 cm from isocenter could be off by as much as one mm on an accredited scanner. This distortion is present even after the manufacturer’s distortion correction algorithm has been applied.

The nonlinearity of gradient fields also leads to distortion. It is commonly believed that the manufacturers’ gradient distortion correction algorithms correct all of this distortion. The component of distortion that is within the slice plane is indeed corrected in all routine clinical sequences. Distortion in the slice direction, however, is routinely corrected only in ‘3D acquisitions,’ in which the slice direction is encoded with a phase-encoding gradient rather than a slice-selection gradient. The remaining distortion in the slice direction for 2D sequences is commonly several millimeters, and is often uncorrected by the manufacturer’s distortion correction software.

The general difference between a ‘3D’ and a ‘2D’ acquisition is in how the spatial information is encoded in the slice direction. In a ‘3D’ sequence, the entire volume of anatomy imaged is excited during every repetition period (TR). The spatial information in the slice direction is encoded in a similar manner as it is for one of the in-plane directions: through a series of pulses of the ‘phase-encoding’ gradient. Because of the manner in which this data is collected, the gradient-induced distortions in all three directions can be corrected by the scanner reconstruction software. The underlying reason for this is that the entire raw data set is a single three-dimensional Fourier transform of the image volume, and so the interpolation can be done accurately during the distortion correction step.

In slice-selective (2D) pulse sequences, which represent a large class of commonly-used sequences, only one slice is excited during a single repetition period. The data from that slice is collected, and during the next repetition period a different slice is excited. Because the slice-selective gradient deviates from linear spatial dependence, the slice excited is warped, as depicted in Figure 1 below. If only that slice were acquired, then no method could correct this distortion, as the wrong part of the anatomy was excited for some of the regions of the slice. However, if an entire 3D volume is collected during the acquisition by acquiring multiple adjacent slices, then the distortion can be corrected within the limits of the ability to interpolate slices. However, this type of correction is not always implemented, and often not turned on for these 2D sequences. This uncorrected distortion in the slice direction for 2D
sequences is often several millimeters over parts of the slice. The phantom-based measurements performed by Image Owl detect this slice-direction distortion for the 2D sequences.

![Figure 1: depiction of slice distortion for 2D acquisitions](image)

**Figure 1: depiction of slice distortion for 2D acquisitions**

**$B_0$-induced distortions**

**Residual inhomogeneity** in the static magnetic field $B_0$ contributes to distortion. The magnitude of the distortion depends upon the receiver bandwidth setting and the slice thickness, along with some internal sequence implementation details selected by the vendor such as RF transmit bandwidth. Lower transmit and receive bandwidths exhibit higher distortion than do the higher bandwidths. For some sequences, notably Echo Planar Imaging (EPI), the distortions induced by the $B_0$ inhomogeneity can be very large. Several millimeters distortion in the phase encode direction is common.

For the purposes of understanding distortion, it is important to further distinguish $B_0$ inhomogeneity as either hardware-induced or patient-induced.

**Hardware-induced inhomogeneity** is caused by everything other than the patient’s tissue. These sources include residual inhomogeneity of the magnetic field after magnet shimming, as well as possible effects from hardware such as RF receive coils inside the bore. It could also be due to a slightly magnetic object that has inadvertently made it into the bore, or a magnetic field shim that has gone out of specification.

**Patient-induced inhomogeneity** occurs when a patient is placed in the scanner. The patient’s tissue disturbs the magnetic field. This disturbance is caused by non-uniformity of a quantity called the magnetic susceptibility, which characterizes the tendency of a material to magnetize in the presence of a magnetic field. Near regions of abrupt change in the magnetic susceptibility, such as interfaces between tissue and air, the magnetic field is disturbed. The distortion caused by this disturbance is particularly acute near internal air-tissue interfaces such as near the sinuses, lungs, and in smaller anatomical structures such as ankles, feet, and hands.
MR scanner software does not correct for distortions caused by static magnetic field inhomogeneity. A phantom-based technique, such as that used by Phantom Lab/Image Owl, is able to measure and correct those B_{o}-induced distortions arising from the scanner hardware. The measurement relies on using the same pulse sequence parameters for the phantom acquisition as is done for the clinical acquisition. Under these conditions, the hardware-induced distortions in the phantom will match those in the clinical scan, and can be measured. Patient-induced distortions are not corrected by any of the implemented techniques, though methods have been studied to make such a correction (Baldwin, 2009).

**Conclusions**

Although scanner reconstruction hardware corrects some of the geometric distortion, significant distortion remains. Imprecise gradient calibration, uncorrected nonlinear warping along the slice direction of 2D sequences, and B_{o}-induced distortion can lead to positional errors of several millimeters for an in-specification scanner with distortion correction applied.

Phantom-based techniques can provide accurate measurement and correction of such distortion, and have been shown to significantly reduce hardware-induced sources of distortion.

**References**

