# NEMA Standards Publication MS 12-2016

*Quantification and Mapping of Geometric Distortion for Special Applications*

*Published by:*

**National Electrical Manufacturers Association**
1300 North 17th Street, Suite 900
Rosslyn, Virginia 22209

www.nema.org

## NOTICE AND DISCLAIMER

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# Preamble

This is one of a series of test standards developed by the medical diagnostic imaging industry for the measurement of performance parameters governing image quality of magnetic resonance (MR) imaging (MRI) systems. These test standards are intended for the use of equipment manufacturers, testing houses, prospective purchasers, and users alike.

Manufacturers are permitted to use these standards for the determination of system performance specifications. This standardization of performance specifications is of benefit to the prospective equipment purchaser. The parameters supplied with each NEMA measurement serve as a guide to those factors that can influence the measurement. These standards can also serve as reference procedures for acceptance testing and periodic quality assurance.

It must be recognized, however, that not all test standards lend themselves to measurement at the installation site. Some test standards require instrumentation better suited to factory measurements, while others require the facilities of an instrumentation laboratory to ensure stable test conditions necessary for reliable measurements.

The NEMA test procedures are carried out using the normal clinical operating mode of the system. For example, standard calibration procedures, standard clinical sequences, and standard reconstruction processes shall be used. No modifications to alter test results shall be used unless otherwise specified in these standards.

The NEMA Magnetic Resonance Section has identified a set of key magnetic resonance image quality parameters. This standards publication describes the measurement of one of these parameters.

*Equivalence*



It is intended and expected that manufacturers or others who claim compliance with these NEMA standard test procedures for the determination of image quality parameters shall have carried out the tests in accordance with the procedures specified in the published standards.

In those cases where it is impossible or impractical to follow the literal prescription of a NEMA test procedure, a complete description of any deviation from the published procedure must be included with any measurement claimed equivalent to the NEMA standard. The validity or equivalence of the modified procedure will be determined by the reader.

*Uncertainty of the Measurements*

The measurement uncertainty of the image quality parameter determined using this standards publication is to be reported, together with the value of the parameter. Justification for the claimed uncertainty limits shall also be provided by a listing and discussion of sources and magnitudes of error.

# Foreword

This standards publication is classified as a NEMA standard unless otherwise noted. It describes a method for evaluating the geometric distortion characteristics throughout a specified imaging volume of a Magnetic Resonance Imaging (MRI) system. The equipment contribution to geometric distortion in MRI systems is largely due to imperfections of the main magnetic field and the spatially encoding gradient subsystem. In addition, the object to be imaged by the MRI system may also induce magnetic field distortions that geometrically distort the image representation of the object to a lesser or greater extent than the MRI system imperfections, depending upon the object and scanning parameters. Since geometric distortion is spatially variable, it is important to understand the spatial distribution of error when MR images are used quantitatively.

The purpose of this procedure is to provide a standard means for measuring and reporting the geometric distortion characteristics of an MRI system. Clinically, this information is helpful in matching MR scanner characteristics to clinical requirements, when geometric accuracy is crucial (e.g., image-guided interventions.) This information is also helpful in evaluating the impact of system changes on performance, for quality control programs that seek to continually reaffirm system performance, or in demonstrating effectiveness for FDA applications.

The measurement methods have not been designed for compatibility with existing NEMA methods, but some of the methods for reporting described in this standard may be compatible with data acquired for MS 2, *Determination of Two-Dimensional Geometric Distortion in Diagnostic Magnetic Resonance Images*. Evaluations are performed on images generated using standard clinical scan protocols.

This standards publication is intended for use by MRI system manufacturers, testing houses, manufacturers of accessory equipment, and MRI end users.



This standards publication has been developed by the Magnetic Resonance Section of the National Electrical Manufacturers Association. User needs have been considered throughout the development of this publication. Proposed or recommended revisions should be submitted to:

Executive Director, Medical Imaging & Technology Alliance
National Electrical Manufacturers Association
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Section approval of the standard does not necessarily imply that all section members voted for its approval or participated in its development. At the time it was approved, the section was composed of the following members:

Computer Imaging Reference Systems—Norfolk, VA
GE Healthcare, Inc.—Milwaukee, WI
Hitachi Medical Systems America, Inc.—Twinsburg, OH
Medipattern Corp.—Toronto, Ontario
Modus QA—London, Ontario
Philips Healthcare—Bothell, WA
Siemens Healthcare, Inc.—Malvern, PA
Toshiba America Medical Systems—Tustin, CA

# Rationale

Magnetic Resonance (MR) image formation is based, in part, on the ability to impose varying magnetic field strengths at different locations within the imaging volume. These magnetic field variations change the precession frequency of the nuclei being imaged and are the basis of the MR image spatial localization process. The linear relationship of precession frequency with magnetic field strength permits the determination of signal source location. Any mechanism that distorts the magnetic field will, therefore, introduce a spatial location error in the final image.

The dominant equipment error sources are the inhomogeneity of the main magnetic field and the nonlinear characteristics of the spatially encoding gradient magnetic fields. In addition, the object to be imaged may also alter the magnetic field, thus creating spatial errors that may exceed the hardware-induced errors in certain situations. As the accuracy of spatial information in MR images becomes more important, e.g., for image guided procedures, quantification of tumor position and volume, co-registration of images from different modalities, it becomes necessary to quantify these errors. For example, the geometric accuracy of spatial information is important for image-guided procedures when the intervention is not based on real-time MR image guidance. Spatial geometric accuracy is also important if the MR images are being used to guide external beam radiation treatment planning because it is important to size the radiation beams appropriately and direct the radiation accurately. Additionally, if treatment progression is quantified by volume measurements, it is important to understand how geometric distortion changes the perceived volume. Lastly, co-registration of images from other modalities with MR images improves with decreased geometric distortion in the MR image.

This standard also has secondary benefits, such as quantifying the degree of gradient non-linearity and its impact on various quantitative measures, such as Apparent Diffusion Coefficient (ADC) measurements, where gradient non-linearity may introduce undesirable spatial non-uniformities in ADC images, and phase contrast MRI, where gradient non-linearities introduce flow velocity errors. Another secondary benefit of this standard is the ability to visualize the homogeneity of the main field by imaging the test phantom at extremely low imaging bandwidths when gradient non-linearity errors are dominated by main field inhomogeneity errors.

# Scope

This standards publication defines test methods for measuring the absolute spatial variation of geometric accuracy within MR images. This standard presents the absolute geometric accuracy as a map, graph, or table throughout the imaging region rather than as simple figures of merit such as average or worst case error. Specifying both the acquisition and data presentation methods is the key function of this standard because the results are not easily reduced to a few simple figures of merit; the results are spatial in nature. This standard deals exclusively with absolute error measurements because it is assumed the end user will need geometric distortion error measurements in absolute versus relative terms.

While the intent of this standard is to quantify equipment induced geometric errors only, the phantom used for these measurements will also introduce some geometric errors. It is not possible to remove the phantom-induced errors within the scope of this standard, and this standard assumes that the measured errors are exclusively equipment errors. Therefore, it is necessary for the user of this standard to be able to differentiate between geometric errors due to the MR imaging system and errors that arise from measuring geometric distortion with a test object. The user should attempt to estimate the error the phantom introduces for the specific test conditions used.

This standard also recognizes that these measurements are ideally performed with three-dimensional acquisitions and large volume phantoms, but the cost, weight, and size of the required phantom may be prohibitive in certain situations. Therefore, this standard permits the use of a substantially two-dimensional phantom in conjunction with a set of two-dimensional image acquisitions in different orientations. It is recognized that the use of a two-dimensional phantom will fundamentally undersample the three-dimensional spatial error map.

These procedures could also be helpful in evaluating the impact of system changes on performance, for quality control programs that seek to continually reaffirm system performance, or in demonstrating effectiveness for FDA applications. However, this standard does not supersede NEMA MS 2 *Determination of Two-Dimensional Geometric Distortion in Diagnostic Magnetic Resonance Images*. MS 2 is designed to produce simple figures of merit that describe basic geometric distortions, or image field of view errors, that could arise from imaging gradient amplitude scaling errors.

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# Section 1
# Definitions

## 1.1 Specification Volume

The specification volume is the imaging volume within which a manufacturer guarantees image performance specifications. Images or portions of images outside this volume will not necessarily meet performance specifications, but may still be useful for diagnostic or image guided purposes. For head scans, the specification volume must enclose, as a minimum, a 10 cm diameter spherical volume (dsv) centered in the RF head coil. For body scans, the specification volume must enclose a 20 cm dsv centered in the RF body coil.

## 1.2 Reference Position

The reference position is a well-defined point within the nearest measurable phantom element to magnet isocenter. The reference position must not be more than 1.5 cm from magnet isocenter (half of the element spacing). The reference position within the reference element will vary depending on the design of the phantom. The user may define the reference position as the middle of a discrete element, or the middle of the intersection of continuous elements, or any of the four corners created by intersecting continuous elements, or some other consistent, well-defined point.

## 1.3 Characterization Volume

The characterization volume is the intersection of the image volume and the specification volume and shall include the reference position. This implies that the phantom used to acquire the spatial distortion measurements is a three dimensional object.



## 1.4 Characterization Area

The characterization area is the intersection of the image plane and the specification volume, and it shall include the reference position, unless otherwise specifically noted. This implies that the phantom used to acquire the spatial distortion measurements is a planar two-dimensional object with sufficient size in the third dimension to fill no more than the thickness of the imaging plane. In the case of a three-dimensional object and threedimensional image acquisitions, an arbitrarily extracted slice can also be used.

## 1.5 Image Artifact

An image artifact is an image anomaly (excluding random noise) that is not representative of the structure or chemistry of the object being scanned, or, that is derived from the structure or chemistry of the object being scanned but which appears in the image at a location other than expected.

## 1.6 Image Distortion

Image distortion is the spatial deviation of an arbitrary point in the imaging volume from its expected true location. Image distortion will be specified as absolute (magnitude or signed magnitude) quantity.

## 1.7 Phantom

A combination of signal producing and non-signal producing materials, used for MR image testing purposes.

Two-dimensional phantoms have large extent in two dimensions (much larger than one pixel in plane) and have limited extent in the third dimension, approximately equal to the slice thickness.

Three-dimensional phantoms have large extent in all three dimensions (much larger than one pixel in plane and much larger than nominal thickness in the orthogonal direction).

# Section 2
# Methods of Measurement

## 2.1 Test Hardware

### 2.1.1 MR Characteristics of the Signal Producing Volume

The following are the MR characteristics of the signal producing volume:

$T_1$ < 1200 milliseconds (at operating field strength)
$T_2$ > 50 milliseconds (at operating field strength)
Single peak NMR spectrum

### 2.1.2 Signal Producing Element Volume

For each selected slice in a 2D phantom, the signal-producing element volume of the phantom in the slice thickness direction should be equal to or less than the slice thickness, and at a minimum must cover the characterization area for that slice. If the signal producing thickness of the phantom elements is thicker than the slice thickness, then signal from outside the slice of interest may be introduced due to distortions in the slice direction.

For 3D phantoms, the signal-producing element volume of the phantom should be as large as, or larger than, the voxel size, and large enough for adequate signal production based on field strength as recommended below:

Below 0.5T: $(5\ \text{mm})^3$ minimum control point size, 1.5 $\text{mm}^3$ minimum imaging voxel size
Above 0.5T: $(3\ \text{mm})^3$ maximum control point size, 1 $\text{mm}^3$ to 1.5 $\text{mm}^3$ minimum imaging voxel size



For each selected volume, the signal-producing volume of the phantom must at a minimum fill the entire characterization volume. The 3D field of view (FOV) should be selected to avoid having signal outside of the acquisition volume alias back into the FOV and should be larger than the signal-producing volume.

Phantom size and design are dictated by measurement accuracy requirements described below, as applied to the user requirements.

### 2.1.3 Construction of the Signal Producing Volume

There are several different possible designs for the signal producing volume, but all share two attributes in common:

a) MR-visible structures at known positions within the phantom
b) MR-visible structures are positioned in a two- or three-dimensional array

The first and second attribute may be a regular lattice of discrete, small MR visible structures or a grid of continuous, intersecting linear MR structures. Discrete structures may be more easily detected than continuous intersecting linear MR structures. For example, in regions of large distortion, perpendicular linear structures may be distorted into nearly parallel structures that obscure the precise location of the intersection point. The same distortions may obscure the exact position of the center of a small discrete MR visible structure.

The second attribute may be a stacked series of two-dimensional structures or a continuous three-dimensional structure for three-dimensional phantoms. Figure 1 demonstrates two different possible two-dimensional structures. These structures can be extended to three dimensions. There are other possible structures, such as a series of nested spherical or cubic shells, but such structures do not provide uniquely identifiable points, limiting the ability to measure regional rotations, unless uniquely identifiable points are provided.

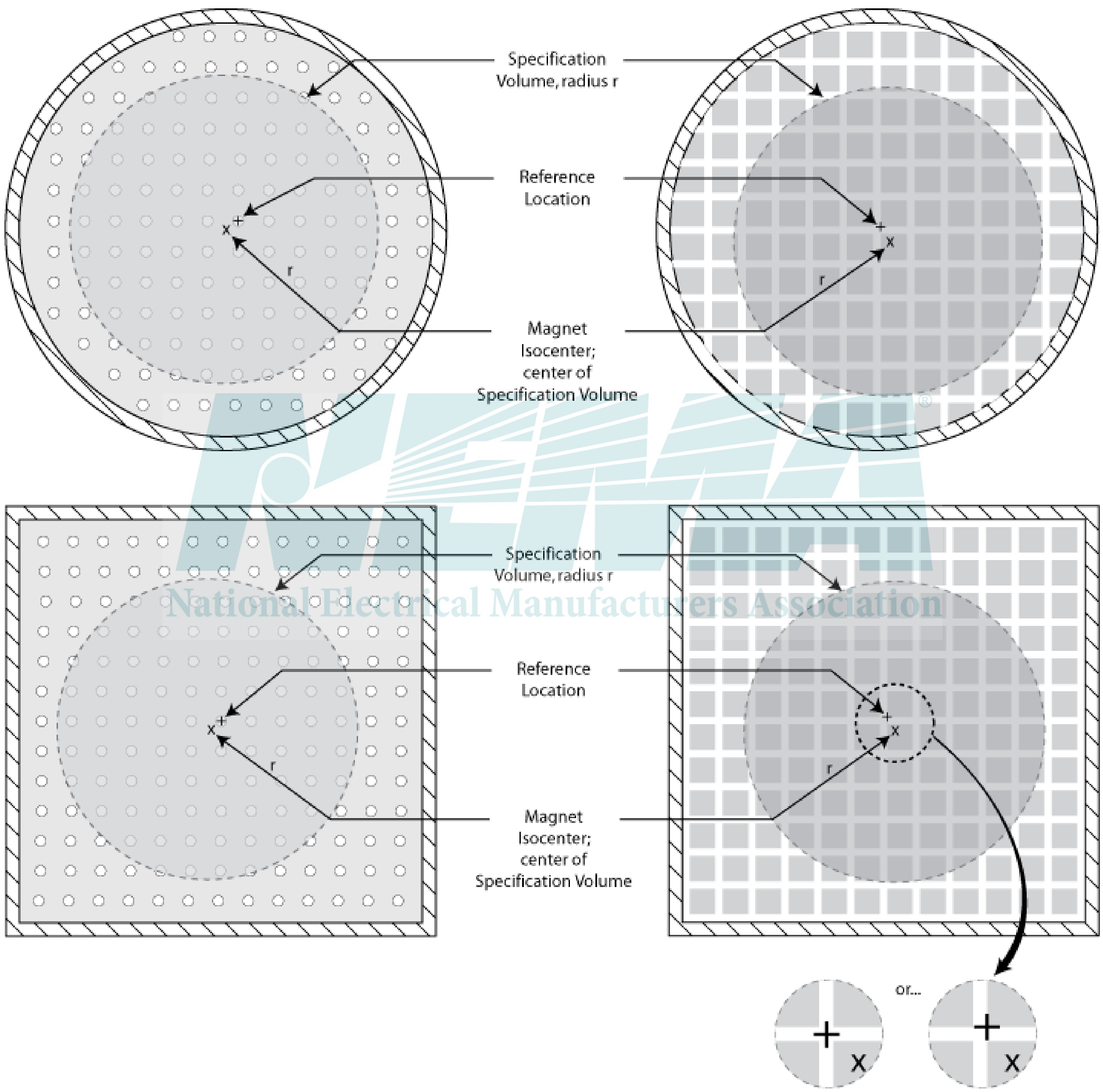


**Figure 1**
**Possible 2D Phantom Configurations for Measuring Geometric Distortion**

The phantom can be cylindrical or square with discrete elements or grids inside. The location of isocenter is indicated with an “x” and the nearest element becomes the reference location. Note that in the case of a grid the reference location could be any of the four corners of the grid or the center of a grid element.

This standard notes that three-dimensional phantoms are the best choice for measuring the geometric distortion characteristics at all points within the specification volume, but due to weight and manufacturing expense considerations, this standard permits measurements with two-dimensional phantoms. Two-dimensional phantoms have the added benefit of ensuring that all signal acquired is from a precise two-dimensional plane if the phantom is no thicker than the slice thickness. If the signal producing thickness of the phantom elements is thicker than the slice thickness, then signal from outside the slice of interest may be introduced due to distortions in the slice direction. However, phantom-air interfaces induce their own distortions, which should also be considered in the phantom design and image parameter selection, as noted below. Alternatively, to significantly reduce weight and manufacturing expense, hollow three-dimensional phantoms with an internal reference point and fiducials constrained to an outer layer which are analyzed via spherical harmonics in an automated fashion can be used.

The spacing requirements are dictated by the spatial accuracy requirements of the distortion maps. The higher the spacing density, the more precisely the location where the distortion contours transition from one level to the next will be defined. More data points will be required if the resolution and accuracy of the distortion maps need to be improved. For example, if the user plots contour bands at every 2 mm of error, the phantom element spacing should not be more than an order of magnitude larger (e.g., 20 mm).

Accuracy and rigidity of the phantom structure are critical features. Ensure no metal machining debris remains in the phantom. Alternatively, new three-dimensional fabrication methods may be sufficiently accurate for phantom construction. The construction accuracy of the phantom should be at least five times smaller than the minimum pixel size (e.g., 1 mm image pixels result in 0.2 mm phantom construction accuracy) and better accuracy (10x) is preferred.

The shape of the phantom and its structures partly control the degree of field distortion introduced by the phantom itself. Sharp corners are to be avoided where possible or removed as far as possible from the region of measurements. Spherical shapes are ideal because of the minimum interaction with the magnetic field, but may not be practical from a manufacturing perspective. Cylindrical shapes are useful for minimizing phantom induced field distortions [References 1, 2]. Various phantom designs are also shown in [References 3–6].



The thickness of phantom walls is another important design consideration. Air/phantom wall interfaces introduce susceptibility distortions of the magnetic field. Susceptibility-induced errors can be minimized by increasing the wall thickness, thereby increasing the air/phantom wall to MR signal producing volume distance. Ensure the phantom is always completely filled or else the air voids will introduce susceptibility artifacts around the air bubble.

Plastics have a range of susceptibility values. Minimizing the susceptibility differences between the phantom filler and plastic housing may be beneficial. [Ref 7 Magnetic properties of materials for MR engineering, micro-MR and beyond Matthias C. Wapler, Jochen Leupold, Iulius Dragonu, Dominik von Elverfeld, Maxim Zaitsev, Ulrike Wallrabe, Journal of Magnetic Resonance Volume 242, May 2014, Pages 233–242, http://dx.doi.org/10.1016/j.jmr.2014.02.005, Ref 8 Characterization, prediction, and correction of geometric distortion in 3T MR images, Lesley N. Baldwin, Keith Wachowicz, Steven D. Thomas, Ryan Rivest and B. Gino Fallone, Med. Phys. 34, 388 (2007); http://dx.doi.org/10.1118/1.2402331]

For the specific phantom design used, the errors in image distortion introduced by susceptibility should be estimated.

### 2.2 Scan Conditions

The following scan conditions are recommended:

a) Spin echo-based sequences are recommended where possible

b) Scan at the highest possible in-plane resolution consistent with condition #6
c) Maximum 3 mm slice thickness for 2D acquisitions, 1.5 mm for 3D acquisitions
d) Single-slice acquisition for two-dimensional phantoms, volume or multi-slice acquisitions for three-dimensional phantoms
e) Sufficient averaging for adequate signal-to-noise ratio (SNR) for good signal detection
f) Clinically appropriate imaging bandwidth(s)

Spin echo-based sequences are recommended to minimize extraneous sources of distortion for producing or checking image specifications. In clinical quality assurance situations, it will be appropriate to use the clinical sequences to the greatest possible extent.

Scanning at the highest possible resolution is recommended to improve the accuracy of the measurements. At worst case, the resolution of the image should be at least one-half of the contour spacing interval used in the spatial maps (e.g., 2 mm error contour intervals require 1 mm resolution images). Reconstruction based interpolation methods (zero padding) may be helpful, but all resolution and slice thickness statements shall be determined from un-interpolated data.

The slice thickness recommendation is based on reporting of results accuracy requirements and should reflect user requirements. Since two-dimensional phantoms cannot capture geometric distortion information in the slice direction, the user can minimize slice direction uncertainty errors by selecting the thinnest slice thickness possible. Signal regions in a 2D phantom that cannot be imaged because they fall outside the slice are not reported and thus represent unknown errors whereas excessively thick 2D slices capture all signals from the 2D phantom with no discrimination against excessive errors in the slice direction. Depending on the construction of the phantom, multi-slice 2D acquisitions may help resolve slice direction errors in a 2D phantom.

When using 3D phantoms, it is recommended that slice thickness matches in-plane resolution to maximize the accuracy of the distortion measurements, but not to exceed 1.5 mm. The choice of volume or multi-slice 2D acquisitions when using a 3D phantom is left to the user, but volume acquisitions are recommended for their SNR benefits.



Selection of image bandwidth has a direct impact on the level of geometric distortion. At low bandwidths (low-gradient amplitudes) main field inhomogeneity may be the dominant contribution to image distortion. (Imaging tests performed at very low bandwidths are a useful qualitative test of main field inhomogeneity.) Sufficiently higher bandwidths (higher gradient amplitudes) result in geometric distortions that arise substantially from the gradients. However, in a typical clinical imaging situation, a range of bandwidths are typically selected such that the relative contribution of the main field and gradient induced distortion will vary, and the total amount of distortion will vary as well. Therefore it is advised that images be collected using parameters that reflect the usage of the scanner for a specific application.

### 2.2.1 Reference Position and Acquisition Orientation

All image acquisitions (2D single-slice, 2D multi-slice, 3D) must include isocenter. Otherwise, it will be impossible to determine the absolute magnitude of the error at any location. The user can define the reference position as the middle of a discrete element, or the middle of the intersection of continuous elements, or any of the four corners created by intersecting elements or any other easily defined point.

Measuring an imaging volume with a two-dimensional phantom will require the user to collect several images in different orientations. At a minimum, images should be collected in each of the three primary orientations (Transverse, Sagittal, Coronal) at a minimum, and further ±45-degree rotations between any of the various primary orientations can also be acquired if necessary.

Three dimensional acquisitions with isotropic resolution can be post-processed to any orientation.

## 2.3 Measurement Procedure

Images are analyzed by first selecting a structure or element that is closest to the reference position. All subsequent position measurements are made relative to that element. Measurements from two-dimensional phantoms are indicated by (x,y) and measurements from three dimensional phantoms are indicated by (x,y,z). The $A_i(x,y)$ or $A_i(x,y,z)$ acquired position of the ith element is then recorded and compared against the known true location $T_i(x,y)$ or $T_i(x,y,z)$ of the ith element, and an error term is computed for the ith element $E_i(x,y)$ or $E_i(x,y,z)$. The sign of $E_i(x,y)$ is positive when the measured spacing between elements is larger than the expected spacing. Given the initial reference MR visible element, all $T_i(x,y)$ or $T_i(x,y,z)$ values are computed as integer multiples of the known MR element spacing. These three values are related as follows for the simple one-dimensional situation:

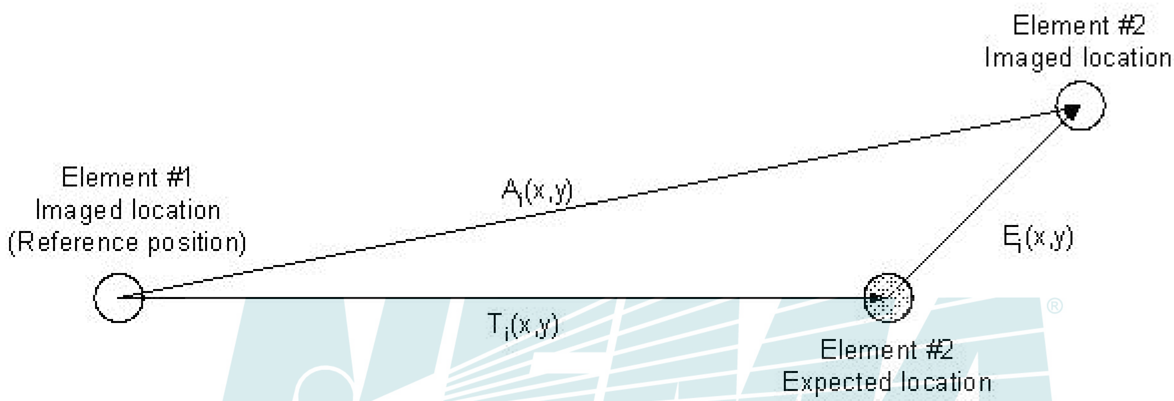


**Figure 2**
**Two Elements with an Apparent Spacing of $A_i(x,y)$ but a True Spacing of $T_i(x,y)$**



$T_i(x,y,z) + E_i(x,y,z) = A_i(x,y,z)$

Where $A_i(x,y,z)$ is the location of an arbitrary point in the Acquired image frame of reference

$T_i(x,y,z)$ is the True location of the corresponding arbitrary point in the phantom

$E_i(x,y,z)$ is the Error between the True and Acquired position

(x,y,z) is the coordinate system where the reference position is at (x,y,z) = (0,0,0) or (x,y) = (0,0)

While maximum geometric errors typically increase monotonically with radial distance from isocenter, there will still be small regions with little or no geometric error far away from isocenter. This implies that the range of geometric error as a function of radial distance from isocenter increases.

### 2.3.1 Measurement Procedure Hints and Tips

Ideally, image measurements will be performed by automated structure/element detecting algorithms, but it is possible to perform these measurements manually. It is critical that the images be acquired at the highest resolution so that the detection accuracy of the MR-visible elements is maximized.

If the measurement process is being done manually, and the MR console or workstation provided by the MR system vendor does not provide pixel location information, acquire a “screen grab” of the image in standard image format and use one of the many available consumer software packages that provide (x,y) cursor location information. This will require the user to create a linear scaling algorithm that converts cursor location information to a millimeter representation. This transformation process is greatly simplified if the MR visible elements track exactly along the pixel rows and columns. Some of the same consumer software packages also have tools for rotating the contents of the image by fractions of a degree.

Note that many MR images are stored with high grey-level fidelity (16 bits) whereas many of the common image file formats use only 8 bits. The loss of grey-level accuracy is immaterial for these measurements. Since grey-level information is immaterial for these measurements, image filtering is permitted, as long as the filters applied are documented and the filters do not reduce location measurement accuracy.

Alternatively, if the markers are placed in a configuration conducive to spherical harmonic reduction and analysis, then all of the necessary Ai(x,y,z) Ti(x,y,z) and associated Ei(x,y,z) can be derived.

# Section 3
# Reporting of Results

## 3.1 Data Reduction

A considerable amount of data is produced by this standard. This standard describes three different data presentations, of increasing degrees of simplification. These three methods are:

a) Spatial mapping (contour plots)
b) Scatter plots (a graph of error vs. radial distance from the isocenter)
c) Error Table (a table of maximum and average error in 5 cm radial increments)

All three of these methods work best when volume acquisitions and three dimensional phantoms are used because positional errors can be measured in all three directions simultaneously. When two-dimensional phantoms are used, the slice direction distortion error value is poorly quantified, if at all, and therefore the actual errors are always worse than the results suggest. Unfortunately, the relative contribution of errors produced by any of the gradients is a function of the gradient design, gradient/magnet geometry, and the spatial location within the imaging volume. Therefore, it is important when using two-dimensional phantoms to collect data in several different slice orientations so that the user has an appreciation of the size of the missing error term. It is also important that the clinical users of such information understand that results produced from two-dimensional phantoms understate the size of the geometric error.

### 3.1.1 Spatial Mapping

The spatial mapping method presents the magnitude error terms [sqrt(x*x + y*y)] or [sqrt(x*x + y*y + z*z)] as contour plots. This maps what degree of geometric accuracy is found and where. An example is shown in Figure 3. Alternative presentations may also be useful, e.g., a contour plot of each signed component of the error, to determine the direction and axis of the dominant error. However, it is assumed for all clinical work that only the magnitude of the error is important, not the direction of the error.

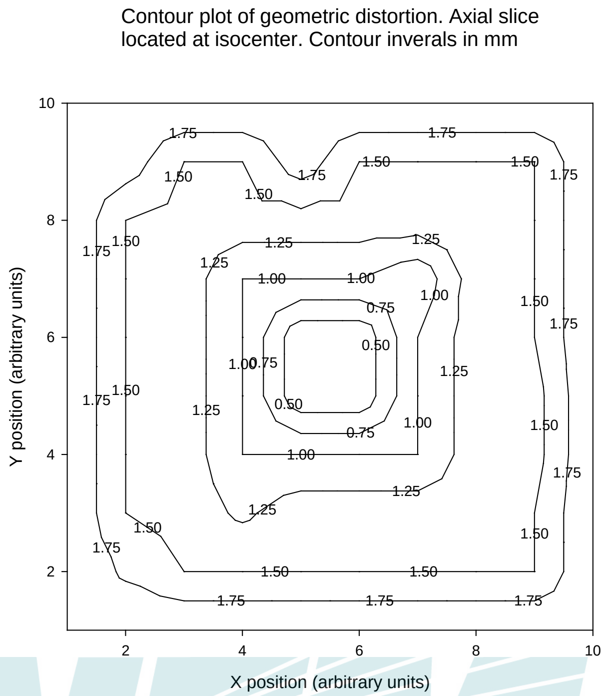


**Figure 3**
**A Schematic of a Spatial Mapping Geometric Distortion Plot**
**Absolute Deviations are Mapped with Contour Lines**

These spatial maps can be useful for clinical purposes, and they are also helpful for quality control purposes because small changes will result in changing contours.

All spatial maps will indicate whether the measurement technique captured two or all three dimensions of geometric error.

**Hints and Tips**
Two-millimeter contour intervals may be sufficient for most applications. Higher accuracy and resolution will require a denser array of MR-visible structures.

Several types of problems may arise with the spatial mapping method:

a) errors due to incorrect gradient scaling
b) errors due to imprecise MR visible structure/element position measurements

Errors due to incorrect gradient scaling are visually obvious. The contours will appear to be parallel lines in the direction perpendicular to the axis scaled incorrectly. For example, errors in the x-direction gradient-scale factor will appear as parallel lines in the y-direction. There is no ideal way to solve this problem. A least-squares method could be used to minimize such scaling errors over the entire field of view, but that may redistribute the location of errors throughout the field of view. For example, a smoothly degrading geometric error with distance from the isocenter will also produce approximately parallel contour lines. A least-squares fit will partly reduce the geometric errors farther from the reference position, but increase the errors closer to the reference position. The optimum solution to this problem is to ensure the gradients are accurately scaled on a small object over the most linear region near (0,0,0).

Errors due to imprecise MR visible structure/element position measurements manifest as erratic contour lines and/or islands of contour lines that otherwise disrupt the smooth progression of contour lines. Either the measurements must be made more accurately, or the step size between contours must be increased, or both.

### 3.1.2 Scatter Plots

Scatter plots present the magnitude error terms as a function of radial distance from isocenter. The radial distance used for each data point shall be the correct radial distance, not the measured radial distance. All data points collected are plotted. An example is shown in Figure 4. As with the spatial maps, the error terms may be given as signed values, or the error terms broken down into the constituent components.

These plots are useful for determining the degree of accuracy as a function of radial distance from the isocenter and are simpler to interpret than the spatial map technique. By using these graphs, it is possible to determine how large a spherical imaging volume will be for a given required level of geometric accuracy. The scatter plots are also useful for generating the Error Table.

All scatter plots will indicate whether the measurement technique captured two or all three dimensions of geometric error and whether the two-dimensional results are from one slice or all slices.

Scatter plot of geometric distortion error.
Axial slice located at isocenter.

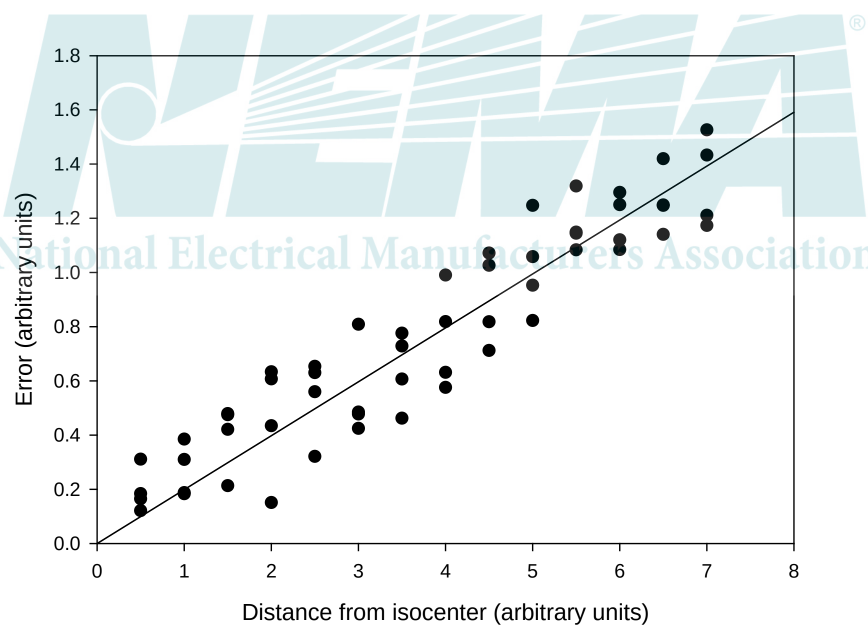


**Figure 4**
**A Schematic of a Geometric Distortion Scatter Plot.**
**Absolute Deviations are Mapped with Contour Lines**

### 3.1.3 Error Table

The Error Table condenses the scatter plots into a simple table of maximum and average absolute geometric errors within a given spherical volume in 5 cm radius increments. The average must be of absolute geometric errors; otherwise, the average will tend to be close to zero.

This table will answer the worst case probable error that will be encountered within a specified spherical volume. With the Error Table, clinical decisions can easily be made whether the imaging volume for a specific task is sufficiently accurate.

All Error Tables will indicate whether the measurement technique captured two or all three dimensions of geometric error.

| Distance from Isocenter (arbitrary units) | Maximum, Average Error (arbitrary units) | |
|---|---|---|
| 1 | .4 | .2 |
| 2 | .6 | .4 |
| 3 | .8 | .6 |
| 4 | 1.0 | .8 |
| 5 | 1.2 | 1.0 |
| 6 | 1.4 | 1.2 |
| 7 | 1.6 | 1.4 |

### 3.1.4 Data Acquisition Parameters

The following data acquisition parameters must accompany the statement of image-geometric distortion:

| Parameter | Dimension |
|---|---|
| Phantom filler $T_1$ | milliseconds |
| Phantom filler $T_2$ | milliseconds |
| Phantom filler composition | …. |
| Sequence type | SE, GRE, etc. |
| Pixel bandwidth | Hertz per pixel |
| Voxel dimensions | millimeters |
| Sequence repetition time (TR) | milliseconds |
| Echo delay time (TE) | milliseconds |
| Number of signals averaged (NSA) | …. |
| Data acquisition matrix size | …. |
| Image matrix size | …. |
| Field of view size | millimeters |
| Type of acquisition | 2D, 2D multi-slice or 3D |
| Number of slices | …. |
| Slice orientation | …. |
| Slice position | …. |
| Slice thickness | millimeters |
| Direction of phase encoding | …. |

The state (i.e., on or off) and purpose of all filters (i.e., temporal or image domain), image processing algorithms and especially geometric distortion-processing algorithms must be indicated. Note that this standard, unlike other NEMA standards, does not require all user selectable filters to be shut off. SNR enhancing filters may be used to the extent that the filter does not significantly alter the spatial accuracy of the measurements.

The results must also include a statement of the phantom dimensions, material composition, design, and shape, including the dimensions of the internal structures from which the position measurements are made. This description must be sufficient to allow the conditions to be reproduced and to determine whether the dimensions of the internal structures are appropriate for the measurements made (e.g., element thickness is less than or equal to slice thickness for 2D distortion measurements). In addition, the results must also include a statement of the shape and size of the characterization area or volume and specification volume.

### 3.2 Sources of Error

The sources of errors and their relationship to the clinical use of the results shall be understood and reported. In addition, it must be understood that the phantom based results describe equipment errors, and do not account for extra geometric distortions introduced by the clinical subject.

The largest source of error in this standard is the use of planar two dimensional phantoms to measure fundamentally three-dimensional errors. This error is important because it consistently results in underestimated geometric distortions. Therefore, all results will be accompanied by a statement of the number of dimensions of error measurements acquired.

It is anticipated that all other sources of error will increase the magnitude of the geometric distortions and thus result in a more conservative error estimate.

In addition, the user is expected to discuss errors caused by experimental setup errors, B0 drift (as determined by center frequency measurements before and after the image acquisition, if necessary), phantom design-induced errors, errors in performing the measurements, errors due to gradient scaling, errors due to sequence selection, and sequence parameters (e.g., bandwidth). In addition, the user will indicate the desired aim of the measurements (e.g., for clinically representative acquisitions, low bandwidth low-gradient amplitude acquisitions to emphasize main field distortion errors, or high bandwidth high gradient amplitude acquisitions to emphasize gradient non-linearity contributions to the total error.)

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# Annex A
# Polynomial Analysis of the Displacement Errors in Magnetic Resonance Imaging

The displacement errors measured using the procedures as described, $e_i(x_i, y_i)$ for 2D and $e_i(x_i, y_i, z_i)$ for 3D, can be modeled by polynomials:

$$e_i = \sum_{j=1}^{M} a_j x_i^{O(j)} y_i^{P(j)} z_i^{Q(j)}$$

Where M is the number of polynomial terms used in the model; $a_j$ is the coefficient of the polynomial term; *O(j), P(j),* and *Q(j)* are the power of the polynomial term corresponding to the respective spatial dimension. In case of 2D, all *Q(j)* should be set to zero.

In matrix notation, the measured values of displacement error can then be expressed by the polynomial model as

$$E = A \cdot R$$

where E is the displacement array collecting all the values, $e_i$, from the measurement; A is the coefficient array collecting all the polynomial coefficients, $a_j$; R is the geometry matrix having elements of $r_{i,j}$ calculated according to:

$$r_{i,j} = x_i^{O(j)} y_i^{P(j)} z_i^{Q(j)}$$

Where *i* is the index of measurement point and *j* is the index of the polynomial term. The polynomial coefficients can be determined by solving the linear equations, using established algorithms such as SVD (singular value decomposition):

$$A = E \cdot R^{-1}$$

Where $R^{-1}$ is the inverse of the geometry matrix that can be solved for by using the SVD methodology. See [for example]:

WH Press, SA Teukolsky, WT Vetterling, BP Flannery, *Numerical Recipes in C: The Art of Scientific Computing*, (ISBN 0-521-43108-5) copyright 1998-92, Cambridge University Press; pp. 59-70.

# Annex B
# Changes to Standard

## B.1 Changes to MS 12-2006 Resulting in MS 12-2010

1.7, 2.1.2, 2.1.3
Various changes to clarify the relationship between slice thickness of acquisition and 2D phantom thickness.

## B.2 Changes to MS 12-2010 Resulting in MS 12-2016

Various changes to consider spherical harmonic analysis based methods.

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