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NEMA STANDARDS PUBLICATION

MS 7

MEASUREMENT PROCEDURE FOR TIME-VARYING GRADIENT FIELDS (dB/dt) FOR MAGNETIC RESONANCE IMAGING DEVICES

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NEMA Standards Publication MS 7-1993

(Revision 1998)

Measurement Procedure for Time-Varying Gradient Fields (dB/dt) for Magnetic Resonance Imaging Systems

Published by

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Foreword

Unless otherwise noted, this NEMA Standards Publication has been approved as a NEMA Standard. It describes two independent characterization methods with test conditions and parameters which ensure that worst case dB/dt values of the time varying gradients fields used for patient exposure regions in magnetic resonance imaging (MR) are measured or calculated. The individual characterization methods are based on the maximum peak dB/dt that the gradient system is capable of producing, when using pulsed waveforms. If the static magnetic field lies along z, then the time-varying gradient fields necessary to MR are the dB_z/dt components produced by the x, y, and z gradients.

Method 1 is applicable only to gradient systems which are incapable of producing peak dB_z/dt levels greater than 20 T/sec. This method will measure magnetic field vectors associated with the gradients that are useful to MR, which are parallel to the static magnetic field. Since the divergence of the time-varying B fields must be zero, components other than z must exist. However, it is assumed that in the region most occupied by the patient dB_x/dt and dB_y/dt components are not relevant with respect to patient safety at this peak dB/dt level. Method 1 applies to the measurement of dB_z/dt. Unless otherwise noted, dB/dt in this method will be taken to mean dB_z/dt.

Method 2 is applicable to all gradient systems and is the required method of characterizing gradient systems where dB_z/dt exceeds 20 T/sec. All magnetic field vectors associated with the gradients are characterized or measured, since magnetic field vectors not parallel to the static magnetic field may become appreciable. For this level of dB_z/dt, the peak magnitude of all magnetic field vectors are measured in the spatial region likely to be occupied by the patient. Method 2 applies to the measurement of $|dB/dt|$, the magnitude of all magnetic field vectors. Unless otherwise noted, dB/dt in this method will be taken to mean $|dB/dt|$, regardless of coordinate system convention used to characterize the gradient magnetic vectors (e.g., cartesian or cylindrical coordinates).

This document does not address time-varying fields produced by radio frequency coils.

This Standards Publication was developed by the NEMA Magnetic Resonance Section, which periodically reviews it for any revisions necessary to keep it up to date with changes in technology. Comments or proposed revisions are welcome and should be submitted to:

Vice President, Engineering
National Electrical Manufacturers Association
1300 North 17th Street, Suite 1847
Rosslyn, VA 22209

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 Magnetic Resonance Section was composed of the following members:

Bruker Medical, Inc.—Billerica, MA
Caprius—Wilmington, MA
Elscint MR, Inc.—Fort Collins, CO
Fonar Corporation—Melville, NY
GE Medical Systems, Inc.—Milwaukee, WI
Greatbatch Scientific—Clarence, NY
Hitachi Medical Systems America, Inc.—Twinsburg, OH
Medical Advances, Inc.—Milwaukee, WI
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Toshiba America MRI, Inc.—San Francisco, CA

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Section 1
GENERAL

1.1 PREAMBLE

This is one of a series of test standards developed by the medical diagnostic industry for the measurement of parameters governing the safety of Magnetic Resonance (MR) Imaging systems. These test standards are intended for the use of equipment manufacturers, 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, and the parameters supplied with each NEMA measurement serve as a guide to those factors which 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 assure stable test conditions necessary for reliable measurements.

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

The NEMA Technical Committee of the MR Section has identified a set of key magnetic resonance safety parameters. This standard describes the measurement of one of these parameters.

1.1.1 Equivalence

It is intended and expected that manufacturers or others who claim compliance with these NEMA standard test procedures for the determination of safety 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 to be equivalent to the NEMA standard. The validity or equivalence of the modified procedure will be determined by each reader.

1.1.2 Uncertainty of the Measurements

The measurement uncertainty of the safety parameter determined using this standard 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.

1.2 INTRODUCTION

In magnetic resonance (MR) imaging, magnetic field gradients are employed for spatial encoding. These time-varying magnetic fields induce currents in the body of the patient. Such currents could stimulate nerve and muscle cells and present a hazard (e.g., ventricular fibrillation) if the current densities and pulse widths exceeded certain thresholds. It is imperative to ensure that the induced currents do not compromise patient safety. The largest currents induced in the body result from gradient waveforms whose time-rates of change are greatest. The Maximum Gradient Rate (MGR) is defined as the greatest dB/dt that a gradient coil and gradient amplifier set are capable of producing using trapezoidal pulsed waveforms. For those systems

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employing sinusoidal gradients, such as echo-planar imaging, a Maximum Sinusoidal Rate (MSR) measurement procedure is defined. These two rates are not likely conditions for most clinical imaging sequences but are representative of a worst case. The worst case dB/dt produced under clinical imaging conditions is defined as the Maximum Clinical Rate (MCR). The MGR, MSR, or the MCR shall be measured and reported for a single gradient coil and gradient amplifier set. For gradient systems capable of simultaneously driving multiple gradient coils using gradient amplifier sets with either trapezoidal or sinusoidal waveforms, the MGR, MSR, or the MCR shall be measured and reported for this simultaneous excitation condition along with the individual single gradient coil excitation conditions. The choice of using MGR, MSR, or the MCR is left to the discretion of the user.

1.3 RATIONALE

Currents induced in the body are proportional to the derivative of the gradient waveform (dB/dt). A conservative threshold dB/dt level of 20 T/sec has been established with a sufficient safety margin to insure that patient exposure to this level of dB/dt cannot cause nerve stimulation and is not of concern. For systems incapable of producing a peak dB_z/dt greater than 20 T/sec, it is assumed that dB_z/dt sufficiently represents the peak magnitude dB/dt, and the dB_x/dt and dB_y/dt components are neglected. Therefore, two independent characterization methods of calculating or measuring peak dB/dt have been developed based on this dB/dt threshold limit.

Method 1 is a simple measurement method which is based on historical techniques, and can be used where there is no reason to be concerned for patient safety. Method 1 is applicable only to gradient systems which are incapable of producing peak dB_z/dt levels greater than 20 T/sec. This method will measure magnetic field vectors associated with the gradients that are useful to MR, which are parallel to the static magnetic field. If the static magnetic field lies along z, then the time-varying gradient fields necessary to MR are the dB_z/dt components produced by the x, y, and z gradients. Therefore, dB_x/dt and dB_y/dt components are neglected. The terms dB/dt and dB_z/dt will be taken to be equivalent for safety purposes in this method.

Method 2 is applicable to all gradient systems, regardless of peak dB_z/dt levels, however for this NEMA standard, this method is the only applicable method of characterizing gradient systems capable of exceeding 20T/sec. In this method, the dB_x/dt, dB_y/dt, and dB_z/dt components are measured in the region occupied by the patient for the maximum peak magnitude dB/dt. The terms dB/dt and |dB/dt| will be taken to be equivalent for safety purposes in this method.

Nerve and cardiac stimulation depend upon induced electric field amplitudes and on the orientation of the electric field. Electric fields in turn depend upon dB/dt and patient geometry (cross sectional area). Smaller patients should have higher dB/dt stimulation thresholds. The pulse width, τ, is defined as the full width at half maximum in microseconds of a single dB/dt pulse or the half period of a sinusoidal dB/dt waveform. The parameters required to characterize dB/dt safety include the amplitude of the dB/dt waveform and the pulse width. Both cardiac and peripheral nerve stimulation thresholds follow a strength-duration curve.

The dB/dt amplitude, R, required for threshold stimulation may for example be expressed as:

$$R = \frac{R_o}{1 - e^{-\frac{\tau}{k}}}$$

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Here, R_o is the rheobase or infinite pulse width stimulation threshold and K is the time constant. Time constant and rheobase vary as functions of pulse shape and number of pulses in relation to induced membrane charge. Induced current and electric fields also vary as functions of patient orientation with respect to the direction of the dB/dt vector, normally parallel to the static field. If the long axis of the patient is parallel to the static field, then currents induced by the gradients flow in axial cross sections. If the static field is parallel to the patient's anterior-posterior direction, then currents induced by the gradients flow in coronal cross sections. If the static field is lateral, then the currents induced by the gradients flow in sagittal cross sections. In each case, eddy current loop areas and circumferences differ, resulting in different threshold values for stimulation. The result is that depending on the orientation of the static field relative to the body, similar dB/dt levels may not provide an equal margin of safety.

Two independent methods (Method 1 and Method 2) utilizing three gradient waveform shapes for measuring dB/dt (MSR, MGR, and MCR) are specified in this NEMA Standards Publication. Values of dB/dt measured by the MGR and MSR waveforms are not typical for most clinical imaging sequences but are representative of the worst case that could be produced by the MR system in the region occupied by the patient (patient space). For clinical imaging conditions, MCR represents the worst case dB/dt experienced by the patient. MGR and MSR dB/dt measurements shall be made using single gradient coil and amplifier excitation, along with multiple gradient coil and amplifier excitations for systems capable of multiple MSR and MGR excitations. Typically, MGR and MSR measurements are easier to perform, while MCR measurements are more representative of clinical imaging. A search technique, explained in Sections 2.2.4, 2.3.5, or 4.3.1 is used to find the approximate position of maximum dB_z/dt for Method 1, and the maximum |dB/dt| for Method 2 in the patient space of the imager. Such an algorithm simplifies the search procedure and defines the region which may be occupied by the patient.

1.4 SCOPE

This NEMA Standards Publication describes two independent methods (Method 1 and Method 2), and three gradient waveform definitions (MGR, MSR, and MCR), used to calculate or measure the maximum dB/dt generated by whole-body gradient coils in diagnostic magnetic resonance imaging systems: the Maximum Gradient Rate (MGR), the Maximum Sinusoidal Rate (MSR), and the Maximum Clinical Rate (MCR). This document does not address time-varying fields produced by radio frequency coils.

1.5 REFERENCES

No standards are referenced herein.

1.6 DEFINITIONS

1.6.1 Search Coil

The search coil is a small diameter coil used to measure gradient field rates of change.

1.6.2 Magnet Isocenter

Magnet isocenter, normally defined by magnet geometry, is the zero crossing for all three gradients in this document.

1.6.3 Patient Space Radius, ρ

The patient space radius is the distance from magnet isocenter to the patient table (without pads) along the lines (see Figure 1-1).

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1.6.4 dB/dt PULSE WIDTH, τ

The pulse width, τ, is defined as the full width at half maximum in microseconds of a single dB/dt pulse or the half period of a sinusoidal dB/dt waveform (see Figure 1-2).

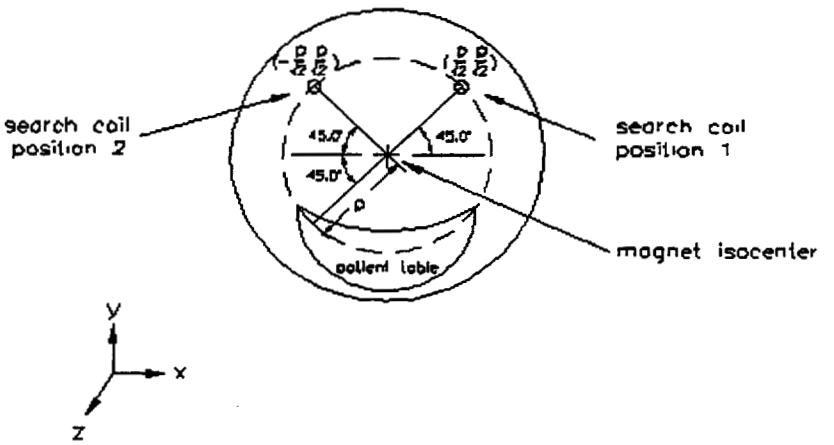
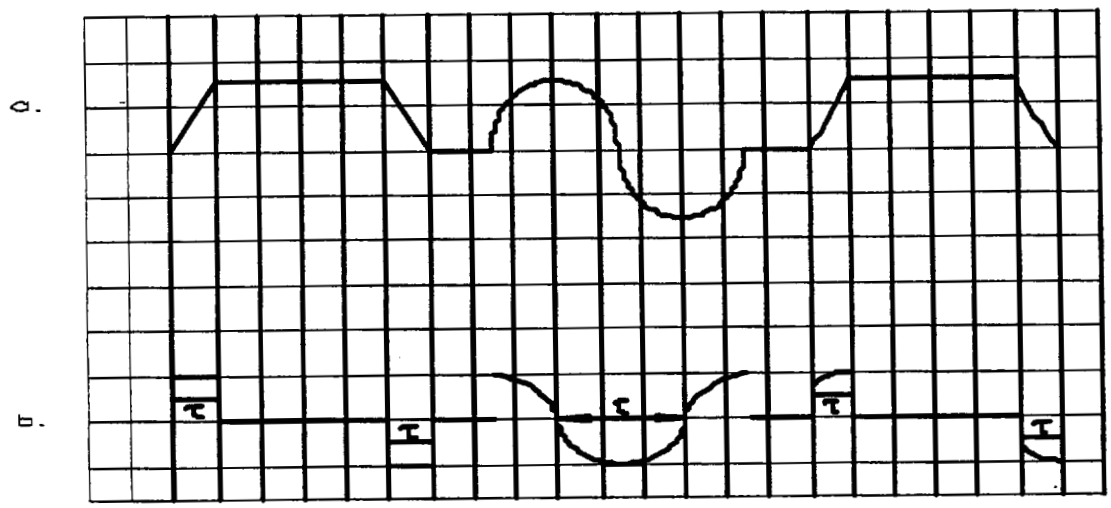


Figure 1-1

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a. Gradient Waveforms
b. dB/dt Waveforms

Figure 1-2

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Section 2
CHARACTERIZATION METHOD 1–METHOD OF MEASUREMENT

2.1 TEST HARDWARE

2.1.1 Search Coil

The search coil shall be circular and shall be small with respect to the gradient coils under test to ensure accuracy. The search coil consists of n turns of wire with a radius of r meters. The axial length of the coil shall be less than 20% of its diameter. The search coil shall be no more than 0.05 meters in diameter. The response of the search coil shall be determined by calculation. The instantaneous magnitude of the component of dB/dt (in tesla/second) coaxial with the search coil shall be determined from the peak voltage, V, induced in the coil by the time-varying magnetic flux:

$$\left|\frac{dB}{dt}\right| = \left|\frac{V}{n\pi r^2}\right|$$

For example, a typical coil would consist of 15 turns of #20 AWG copper wire wound on a form of 0.025 meters radius resulting in a circular coil approximately 0.008 meters long. An induced voltage of 0.200 volts would be indicative of a dB/dt = 6.79 tesla/sec coaxial with the search coil.

2.1.2 Voltage Measurement Device

The device used to measure voltage induced in the search coil shall have a high input impedance and sufficient bandwidth to prevent signal attenuation, e.g., a storage oscilloscope.

The search coil should be connected to the voltage-measurement device by means of a high inductance cable, e.g., a twisted pair to avoid ringing on the waveform that may be experienced with coaxial cables.

2.1.3 Positioning Device

A means shall be provided for fastening and aligning the search coil in the magnet in a stable and reproducible manner. The device shall permit positioning of the search coil throughout the region normally occupied by the patient while maintaining the search coil coaxial with the static magnetic field.

2.2 MGR MEASUREMENT

2.2.1 Turn off or maximally attenuate the radio frequency (RF) transmitter to prevent interference.

2.2.2 Position the search coil in the MR scanner using the positioning device so the search coil axis is aligned with that of the static magnetic field, regardless of patient orientation.

2.2.3 Begin simultaneous pulsing of all three gradients using the maximum amplitude bipolar gradient waveforms (Figure 2-1). The transition times, t_1, t_2, t_3, and t_4 shall be the minimum that the system can generate using the standard pulse sequence generation software and hardware used by the manufacturer. The amplitudes, A_{+max} and A_{-max}, are defined as the maximum gradient strength that can be generated by the system under normal operating conditions as stated above.

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2.2.4 Move the search coil to locate the position of the maximum induced voltage that can be found within the region of the machine ($y = \pm x$) normally occupied by a patient. Once this position is established, the coil shall be immobilized to prevent movement-induced voltage from interfering with the measurement. For example, the following procedure locates the approximate position of maximum dB_z/dt in the patient space of the imager with an axial field magnet and whole-body gradient coils. Consider a cartesian coordinate system with the Z axis pointing in the direction of the static field and the Y axis pointing vertically (e.g., as in Figure 1-1). For the MGR procedure, the points of maximum dB/dt will occur where each of the gradients is near maximum and all the gradient fields add. Typically, this is near the maximum winding density of the coil and near the wall of the bore of the gradients. Assume that the distance, ρ, from the magnet isocenter to the patient table along the lines $y = \pm x$ is the same as the maximum distance the patient may extend above magnet isocenter along the lines. Then the maximum dB/dt value is located at the points:

$$\left(\pm\frac{\rho}{\sqrt{2}}, \frac{\rho}{\sqrt{2}}, Z_m\right)$$

Where:

Z_m = coordinate of maximum dB_z/dt

Assuming all three gradients are applied simultaneously and with equal amplitudes, the highest induced voltages will be found near the walls of the magnet shroud near the ends of the gradient coils along the lines $y = \pm x$ (see Figure 1-1). Search coil position 1 in Figure 1-1 is located at:

$$Y = +X = \frac{\rho}{\sqrt{2}}$$

Similarly, search coil position 2 is located at:

$$Y = -X = \frac{\rho}{\sqrt{2}}$$

Place the search coil coaxial with the static magnetic field at position 1 and constrain the search coil so it may move only in the ± Z directions. Move the coil in the ± Z directions to the position where the maximum voltage is induced in the search coil. Repeat this procedure for position 2. The results from the position which yields the greatest search coil voltage shall be reported as described in Section 3.

2.3 MAXIMUM SINUSOIDAL RATE (MSR) MEASUREMENT

2.3.1 Turn off or maximally attenuate the radio frequency (RF) transmitter to prevent interference.

2.3.2 Position the search coil in the MR scanner using the positioning device so the search coil axis is aligned with that of the static magnetic field.

2.3.3 Turn off all gradients other than the sinusoidal gradient of interest. Apply the maximum amplitude and frequency waveform that the system can generate using the standard pulse sequence generation software and hardware used by the manufacturer.

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2.3.4 Move the search coil to locate the position of the maximum induced voltage that can be found within the region of the machine normally occupied by the patient. Once the position is established, the coil shall be immobilized to prevent movement-induced voltage from interfering with the measurement.

For most coils, a suitable search coil procedure is as follows. For x gradients, fix the coil along the $X = \rho$, $Y = 0$. For Y gradients, fix the coil along the $X = 0$, $Y = \rho$. For Z gradients, fix the coil along $X = 0$, $Y = \rho$. Then search along z to locate the position of the maximum induced voltage that can be found within the region of the machine normally occupied by the patient.

2.3.5 Measure the peak voltage which corresponds to the maximum dB/dt.

2.3.6 Record the pulse width, τ, of the dB/dt waveform.

2.3.7 Repeat the above procedure for any other sinusoidally driven gradient.

2.4 MCR MEASUREMENT

2.4.1 Turn off or maximally attenuate the radio frequency (RF) transmitter to prevent interference.

2.4.2 Position the search coil in the MR scanner using the positioning device so the search coil axis is aligned with that of the static magnetic field.

2.4.3 Select a protocol from each major sequence type and maximize dB/dt by adjusting the following parameters:

a. Maximize gradient amplitude (e.g., minimum field of view, minimum slice thickness, maximum bandwidth, use of spoiler pulses)

b. Minimize gradient transition times (e.g., minimum rise and fall times for trapezoidal pulses, minimum echo time, the highest frequencies and greatest amplitudes for sinusoids)

c. Maximize the number of gradients on simultaneously (double oblique scans)

In general, the rise times and fall times of the gradient waveform should be minimal while the gradient amplitudes shall be maximal. With sinusoidal gradient waveforms, the highest frequencies and greatest amplitudes usually lead to the highest values of dB/dt. Pulse sequences employing 45°, 45° double obliques with spoiler pulses produce higher dB/dt values than pulse sequences which only permit cartesian slices. In addition, the use of double oblique protocols simplifies the search coil procedure for the spatial peak dB/dt position. Therefore, double oblique sequences are recommended.

2.4.4 Run the selected clinical pulse sequence, modified to repeat at the maximum phase encoding value.

2.4.5 Move the search coil to locate the position of the maximum induced voltage that can be found within the region of the machine normally occupied by the patient. Once this position is established, the coil shall be immobilized to prevent movement-induced voltage from interfering with the measurement.

For 45° double oblique sequences, the search procedure described in Section 2.2.4 may be used.

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2.5 ADDITIONAL MEASUREMENTS

When a measurement of dB/dt is used to assess performance relative to a dB/dt safety level which depends on pulse width (or frequency), then measurements shall be made and reported for all applicable values of τ. If the τ which produces the largest dB/dt falls in the region where safety limits are relaxed from the strictest values, then that could result in largest dB/dt meeting the safety level, while a lesser dB/dt value associated with a larger τ may not. In this case, the dB/dt values measured under the test protocol having larger τ values shall be measured and reported.

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Section 3
CHARACTERIZATION METHOD 1–RESULTS

3.1 REPORTING dB/dt RESULTS

3.1.1 State the method used (MGR, MSR, or MCR).

3.1.2 State the calculated values of peak dB/dt for the gradient coils.

3.1.3 Report the radius and number of turns of the search coil.

3.1.4 State the position of the search coil with respect to the magnet isocenter.

3.1.5 List the data acquisition parameters as indicated in the following subsection (3.1.5.1). Report the pulse sequence used to produce each of the dB/dt waveforms.

3.1.5.1 Data Acquisition Parameters

3.1.5.1.1 Maximum Gradient Rate (MGR) Method

The following parameters shall be reported for MGR:

Parameter	Dimensions
Maximum gradient pulse amplitude	millitesla/meter
Patient space radius, ρ	meters
t_1	microseconds
t_2	microseconds
t_3	microseconds
t_4	microseconds

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3.1.5.1.2 Maximum Sinusoidal Rate (MSR)

The following parameters shall be reported for the MSR method:

Parameter	Dimensions
Pulse width, τ	microseconds
Measured dB/dt	tesla/second
Maximum gradient pulse amplitude	millitesla/meter
Patient space radius, ρ	meters

3.1.5.1.3 Maximum Clinical Rate (MCR)

The following parameters shall be reported for the MCR method:

Parameter	Dimension
Sequence type/name	--
Scan plane	--
Pulse width, τ	microseconds
Measured dB/dt	tesla/second
Echo time, TE	milliseconds
Field-of-view, FOV	centimeters
Slice thickness	millimeter
Angles for oblique scans	degrees
Slice Orientation	--
Patient space radius, ρ	meters

3.1.6 List any other parameters required to ensure repeatability.

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Section 4
CHARACTERIZATION METHOD 2—APPLICABLE TO GRADIENT SYSTEMS CAPABLE OF PEAK dB/dt GREATER THAN OR EQUAL TO 20 t/sec

The following characterization method describes the procedure for measuring peak |dB/dt| levels for gradient systems capable of producing peak |dB/dt| levels ≥ 20 T/sec. The procedure determines |dB/dt| from physical measurements of dB/dt components using either a cartesian (x,y,z), or cylindrical (ρ,ϕ,z) coordinate convention. The measurement procedure is very sensitive to positions of the three search coils and to timing offsets between the gradient axes. For this level of dB/dt, the peak magnitude of all magnetic field vectors are measured in the spatial region likely to be occupied by the patient. Characterization Method 2 applies to the measurement of |dB/dt|. Unless otherwise noted, dB/dt in this method will be taken to mean |dB/dt|, regardless of coordinate system convention used to characterize the gradient magnetic vectors.

4.1 TEST HARDWARE

4.1.1 Search Coil

Search coil(s) shall be constructed so that the components (either cartesian or cylindrical) of dB/dt may be measured. (Refer to Section 2.1.1 Search Coil and Annex A.)

4.2 VOLTAGE MEASUREMENT DEVICE

The device used to measure voltage induced in the search coil shall have a high input impedance and sufficient bandwidth to prevent signal attenuation, e.g., a storage oscilloscope.

Individual search coil outputs (x,y,z or ρ,ϕ,z) should be connected to the voltage measurement device by means of a high inductance cable, e.g., twisted pair to avoid ringing on the waveform that may be experienced with coaxial cables. Search coil outputs may be connected to the oscilloscope by means of an analog filtering device, to maximally attenuate switching frequency components, but not attenuate gradient waveform frequency components for gradient amplifiers utilizing switching power supplies.

4.3 SEARCH TECHNIQUE

4.3.1 The maximum |dB/dt| in the space normally occupied by the patient will occur in two specific octants. These octants depend on the gradient geometry and geometric convention, and excitation current polarities and phases. The octants can be determined by either a numerical characterization of the combined fields of all gradient coils (Authorized Engineering Section—Field Plot Procedure), or by an iterative search technique with the search coils.

4.4 MEASUREMENTS

4.4.1 Measurements are made using gradient waveforms MGR, MCR, or MSR (Sections 2.2, 2.3, and 2.4) within the region of the patient space which results in the maximum |dB/dt|. If the gradient coils and amplifiers permit simultaneous multiple axis gradient excitations, then measurements should be taken using both the single independent gradient axis, and multiple simultaneous gradient excitation capabilities.

4.4.2 Move the search coil to the location determined to produce the maximum |dB/dt| within the patient space (Section 4.3.1). Once this position is established, the coil shall be immobilized to prevent movement-induced voltage from interfering with the measurement.

4.4.3 Measure the peak voltage of each orthogonal dB/dt at the location determined to produce the maximum |dB/dt| within the patient space (Section 4.3.1).

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4.4.4 Record the pulse widths, τ, of the dB/dt waveform (see Figure 1-2).

4.5 DETERMINATION OF |dB/dt|

4.5.1 The magnitude of the time-varying gradient field (|dB/dt|) shall be determined as:

$$|\,dB/dt\,| = \sqrt{(dB_x/dt)^2 + dB_y/dt)^2 + (dB_z/dt)^2}$$

or

$$|\,dB/dt\,| = \sqrt{(dB_\rho/dt)^2 + (dB_\phi/dt)^2 + dB_z/dt)^2}$$

4.5.2 Note that if all three gradient waveforms do not peak simultaneously, then the values for the cartesian or cylindrical components of dB/dt used in the equations above are the instantaneous peak values which result in the maximum |dB/dt|.

4.5.3 Reporting Results

Copy the Report Results in Section 3.1 and include them for each search coil element. In addition, report |dB/dt|.

4.5.4 Numerical Field Plot Procedure

A computer program may be developed that is capable of calculating magnetic field components and magnetic field magnitude from pulsing independent or multiple gradient axes simultaneously. A static calculation using Biot Savart is acceptable.

The maximum |dB/dt| in the space normally occupied by the patient will occur in two specific octants. These octants depend on the gradient geometry convention, and excitation current polarities and phasing presented to each coil. The field components and magnitude shall be calculated for these regions.

Computer calculations simulating the appropriate type of gradient waveforms (MGR, MCR, or MSR) shall be made following the procedures analogous to those in Section 2. Magnitude for cartesian or cylindrical component data from all three gradient axes shall be calculated. The magnitude of the time-varying gradient field (|dB/dt|) shall be determined as:

$$|\,dB/dt\,| = \sqrt{(dB_x/dt)^2 + (dB_y/dt)^2 + (dB_z/dt)^2}$$

or

$$|\,dB/dt\,| = \sqrt{(dB_\rho/dt)^2 + (dB_\phi/dt)^2 + dB_z/dt)^2}$$

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ANNEX A
(Informative)

A.1 Sample Search Coil Design and Calibration

For example, three independent orthogonal search coils may be constructed about a common center using a 2 cm diameter spherical former. This search coil design should allow measurement of each independent component of dB/dt without requiring repositioning of the search coil during the measurement procedure.

A recommended calibration of this search coil design is proposed to measure the sensitivity factor of the individual search coil elements, and this factor relates the output voltage to dB/dt (tesla/sec). This calibration will validate the search coil sensitivity against the theoretically calculated sensitivity based on coil geometry and number of turns. A minimum sensitivity factor of $\geq$ 0.01 volts per tesla/sec is recommended to measure signal amplitudes in regions of small dB/dt, (e.g., 1T/sec dB/dt).

Calibration of the search coils is done by positioning them near isocenter within the linear range of the gradient of a single body gradient coil, which can be excited with a sinusoidal current at a defined frequency and peak-to-peak amplitude. The gradient characteristics must be known, specifically gradient strength (mT/meter) per unit ampere G_o, the frequency f_o, the current amplitude I, the measured search coil voltages (V2,V1) at two positions (μ2, μ1), and the distance between these positions (μ2-μ1). The search coil sensitivity factor S, is calculated as:

$$S = \frac{(V2 - V1)}{G_O * I * (\mu 2 - \mu 1) * 2 * \pi * f_O}$$

For this calculation, $\mu 2 > \mu 1$ and the measurements must not cross through the zero of the gradient field. The calibration measurements are to be made in the region near the isocenter of the gradient coil, along a primary axis, to insure a linear field variation. Absolute positioning is not critical, however differential positioning is, and preferably $\mu 2 \leq$ 10cm.

A sensitivity factor for each of the search coil elements is determined from the calibration procedure, namely S_x, S_y, and S_z (for cartesian components), or $S\rho$, $S\phi$, and S_z (for cylindrical coordinates). The voltage measurements made using the voltage measurement device are then multiplied by these sensitivity factors to report the dB/dt results.

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