Technical Note: Compact three-tesla magnetic resonance imager with high-performance gradients passes ACR image quality and acoustic noise tests

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Purpose: A compact, three-tesla magnetic resonance imaging (MRI) system has been developed. It features a 37 cm patient aperture, allowing the use of commercial receiver coils. Its design allows simultaneously for gradient amplitudes of 85 millitesla per meter (mT/m) sustained and 700 tesla per meter per second (T/m/s) slew rates. The size of the gradient system allows for these simultaneous performance targets to be achieved with little or no peripheral nerve stimulation, but also raises a concern about the geometric distortion as much of the imaging will be done near the system's maximum 26 cm field-of-view. Additionally, the fast switching capability raises acoustic noise concerns. This work evaluates the system for both the American College of Radiology's (ACR) MRI image quality protocol and the Food and Drug Administration's (FDA) nonsignificant risk (NSR) acoustic noise limits for MR. Passing these two tests is critical for clinical acceptance.

Methods: In this work, the gradient system was operated at the maximum amplitude and slew rate of 80 mT/m and 500 T/m/s, respectively. The geometric distortion correction was accomplished by iteratively determining up to the tenth order spherical harmonic coefficients using a fiducial phantom and position-tracking software, with seventh order correction utilized in the ACR test. Acoustic noise was measured with several standard clinical pulse sequences.

Results: The system passes all the ACR image quality tests. The acoustic noise as measured when the gradient coil was inserted into a whole-body MRI system conforms to the FDA NSR limits.

Conclusions: The compact system simultaneously allows for high gradient amplitude and high slew rate. Geometric distortion concerns have been mitigated by extending the spherical harmonic correction to higher orders. Acoustic noise is within the FDA limits. © 2016 American Association of Physicists in Medicine. [<http://dx.doi.org/10.1118/1.4941362>]

Key words: gradient nonlinearity, acoustic noise, compact three-tesla, high-performance gradients

1. INTRODUCTION

Recently, a compact three-tesla (3 T) magnetic resonance imaging (MRI) system has been developed. This system features a compact low-cryogen 3 T superconducting magnet and a gradient coil with a 42 cm inside diameter.¹⁻³ When fitted with birdcage radio frequency (RF) transmitter (also capable of reception), a 37 cm bore size results. The diameter of the spherical volume (DSV) provided by the gradients permits up

to a 26 cm imaging field-of-view (FoV) with approximately 15% gradient nonlinearity before correction. This provides full head imaging capability with a 32-channel phased array receive coil, with coverage as far down as the interspace between the fourth and fifth cervical vertebrae for a normal-size patient.¹ The 37 cm patient aperture is also compatible with imaging of adult extremities as well as infants.

Patient positioning is a major challenge when employing small diameter gradient coils. Anatomic constraints dictate the

brain will be positioned closer to the patient entry end relative to the center of the magnet. To address this problem, an asymmetric gradient design²⁻⁹ is employed for the transverse (i.e., the *X* and *Y*) gradients. For these gradient coils, the transverse gradient isocenter is shifted toward the patient-end of the bore. This design allows the brain to be positioned at gradient isocenter, but does incur some additional engineering challenges including zeroth and first order concomitant field terms¹⁰ and even-order terms in the gradient nonlinearity expansion (i.e., second and fourth order, in addition to the standard third and fifth order). As in a conventional, whole-body MRI, stronger gradient nonlinearity effects are more pronounced when imaging close to the maximum FoV, which in our case is 26 cm. Our previous work has characterized and corrected for these effects.^{11,12}

A major advantage of the compact system is the increased gradient performance for a given amplifier specification and a higher peripheral nerve stimulation threshold. When compared to commercially available, high-performance 60-cm diameter gradient systems operating at an amplitude of up to 80 millitesla per meter (mT/m) and slewing at up to 200 T/m/s, the compact system is capable of operating at 85 mT/m and 700 T/m/s, simultaneously. Moreover, the compact gradient requires only half the gradient power: 1 megavolt-amp (MVA) per axis rather than 2 MVA. This high slew rate and amplitude are concurrently achievable in this system due to the compact size of the gradient coil reducing peripheral nerve stimulation. With a standard whole-body system, a large current loop can form in the chest area of the imaged subject, as it is also in the presence of the strong gradient field even when only the head is being imaged. With

the compact gradient set, much of the anatomy distal to the head is mainly outside the gradient field, which reduces the size of the largest current loop to be excited, restricting it to the head.¹³ The PNS characteristics of the compact head gradient coil were studied in detail in Ref. 3, and shown to be within acceptable limits for the simultaneous high gradient amplitude and slew rate demonstrated here. With such a high, PNS-free gradient switching rate, acoustic noise as generated from the Lorentz forces acting on the resistive electromagnetic coils while spatial encoding gradients are active can also be a potential concern.

The purpose of this work is to demonstrate that (1) the head-only gradient system passes the American College of Radiology's (ACR) quality control (QC) testing and (2) the system adheres to the Food and Drug Administration's (FDA) nonsignificant risk (NSR) acoustic noise limits.

2. METHODS

In this work, the head-only gradient system was operated at a reduced maximum amplitude and slew rate of 80 mT/m and 500 T/m/s, respectively, for initial testing with conservative safety margins as compared to the maximum 85 mT/m and 700 T/m/s. These performance settings are consistent with the initial *in vivo* imaging studies on the compact gradient system^{1,3} and are still higher than those used in typical clinical whole-body gradient systems.

A standard ACR QC protocol for MRI, consisting of a sagittal localizer scan, and a T1 and T2 weighted 11-slice axial spin-echo acquisition with 5 mm slice thickness was acquired on the integrated system (compact head gradient + compact,

TABLE I. ACR annual test.

ACR weekly test	Test	Pass/fail	Note
	Geometric accuracy	Pass	Measure (pass values in mm)
	Several measurements of the ACR phantom are taken on the source images and compared to known truth values. These measurements are taken on the slice shown in Fig. 2. Ax T1 image is shown in Fig. 2(a), Sag loc in Fig. 2(c)		Sag loc
			Ax T1 image 1
			Ax T1 image 2
	High-contrast spatial resolution	Pass	1.0 mm (1.0 mm to pass)
	Central "pips" must be visually resolved—these are the 1.0 mm inclusions		
	Low-contrast detectability	Pass	All 40 spokes detectable in each of T1, T2 weighted imaging series (≥ 37 to pass)
	Slice thickness	Pass	T1: 5.33 mm T2: 5.21 mm (5.0 ± 0.7 mm to pass)
	Slice position accuracy	Pass	Bar length difference, ± 5.00 mm or less to pass
	Difference in bar length corresponds to slice position accuracy		T1: img 1: -3.40 mm, img 11: -3.20 mm
			T2: img 2: -2.50 mm, img 22: -2.70 mm
	Image intensity uniformity	Pass	Percent image uniformity T1: 88.1% T2: 85.1% (82.0% or higher to pass)
	ROIs in several parts of the image are compared for signal intensity		
	Percent-signal ghosting	Pass	Ghosting ratio: 0.0048 (pass is ≤ 0.025)
	ROI of the phantom and surrounding regions are compared to measure ghosting

FIG. 1. High-contrast spatial resolution from the (a) T1w and (b) T2w second echo. Note that central pips are resolved according to ACR criteria, constituting an image quality pass.

low-cryogen 3 T magnet) using an eight-channel receive only head coil (*In vivo* Diagnostics, Gainesville, FL, USA). The image quality was assessed by a board certified medical physicist (Y.S.). These tests assess the geometric distortion, high-contrast spatial resolution, slice position and accuracy, signal uniformity, low-contrast detectability, and signal ghosting (<http://www.acr.org/~media/ACR/Documents/Accreditation/MRI/LargePhantomGuidance.pdf>).¹⁴ An additional test above and beyond the ACR guidelines to assess geometric distortion was performed using a 3D IR-FSPGR (Ref. 15) sequence and the Alzheimer's Neuroimaging Disease Initiative (ADNI) phantom and evaluation framework.¹⁶

After acquisition, the images were corrected using standard image-domain interpolation for gradient nonlinearity distortion¹⁷ using a higher-order spherical harmonic model which includes even- and odd-order terms up to the seventh order, as derived from an iterative gradient nonlinearity characterization scheme.¹² Briefly, the iterative gradient nonlinearity characterization scheme works by utilizing the ADNI phantom¹⁶ and accompanying fiducial tracking software. The phantom is scanned with a 3D gradient-echo sequence, and the imaged fiducial positions are compared to the known phantom schematics. From this position information, spherical harmonic coefficients for gradient nonlinearity correction can be determined. The image is updated based on the new coefficients, and the process is then repeated until it converges, usually in less than ten iterations. Surface coil intensity correction was also used to minimize receiver-coil induced intensity nonuniformity.¹⁸

The torque-balanced gradient was designed to both fit in a specialized system in its final form, as well as to be mounted in a standard whole-body MR system. Acoustic noise (A-weighted and peak) measurements were assessed with the gradient inserted into a whole-body 3 T magnet

(MR750w, GE Healthcare, Waukesha, WI, USA). A variety of pulse sequences were tested, including sagittal inversion-recovery fast spoiled gradient echo (Sag IR-FSPGR),¹⁵ axial 3D fast imaging employing steady-state acquisition (Ax 3D FIESTA),^{19,20} sagittal T2 weighted fast spin-echo (Sag T2w FSE),²¹ axial susceptibility-weighted angiography (Ax SWAN),²² functional MRI (fMRI,^{23,24} EPI-readout based), and diffusion tensor imaging^{25–29} (DTI, EPI-readout based), by placing the microphone of a sound meter (Brüel Kjaer 2250, Nærum, Denmark) in the bore at the position of a typical patient ear near the edge of the receiver coil. Acoustic noise for the same sequences was measured in the same manner on a standard clinical GE scanner (MR750w) as a control. In each case, the sequences were run to take full advantage of the gradient capabilities available.

3. RESULTS

A summary of the ACR test is available in Table I.

Figure 1 shows the high-contrast spatial resolution insert, with the middle set of objects representing a 1.0 mm spatial resolution. These are visually resolved as per ACR criteria in both the T1w and T2w images. In contrast, the 0.9 mm spatial resolution inserts to the far right would be unresolved as per ACR criteria. Figure 2 shows the geometric accuracy grid and the sagittal localizer, each relatively unremarkable. The geometric accuracy results were measured on these images, and the results show acceptable geometric accuracy according to ACR criteria. Additionally, we report a root-mean-squared-error of less than 0.4 mm for the ADNI phantom fiducial test, which is within ADNI study guidelines for serial neuroimaging studies.

Figure 3 shows magnified and window/level appropriate views of each of the slice thickness and slice position accuracy

FIG. 2. Geometric distortion for (a) T1, (b) T2 second echos, and (c) sagittal localizer. Lines in the phantom are straight. Any gross distortion would be detected by measurements reported in Table I: geometric accuracy. Green lines denote measurements taken.

FIG. 3. Magnified slice 1 of ACR test with (a) T1 and (b) T2 second echos displayed at a high window/level setting to demonstrate that the ramps are of similar length, showing good slice thickness and slice position accuracy. Arrows denote measurements taken.

ramps, demonstrating small differences in the width of the ramps, corresponding to the passing scores in the ACR tests.

Figure 4 shows magnified views of one of the low-contrast detectability slices, demonstrating the ability to count all 40 of the low-contrast objects. Note that all of Figs. 1–4 are sinc-interpolated 2× for display.

The acoustic noise measurements were all below FDA NSR limits of 99 dBA with hearing protection in place and 140 dB peak. Figure 5 shows the acoustic noise results compared to the standard whole-body system.

4. DISCUSSION

The ACR QC imaging took place in the fully integrated magnet/gradient system, demonstrating the compliance of the feature complete imaging system. However, the acoustic noise results were acquired when the gradient system was inserted into the standard whole-body 3 T magnet, as the patient handling systems of the integrated magnet/gradient system were not completed at the time of the test. Once these systems are completed, ACR QC tests such as the slice position accuracy are expected to further improve due to increased rigidity of the receiver coil/patient table interface.

There are some assumptions made regarding the acoustic noise compliance results. It is assumed that the patient

FIG. 5. A-weighted and peak acoustic noise comparison between the compact 3 T and a whole-body 3 T MRI systems for a variety of clinical pulse sequences, with each scanner running at gradient strength and slew rate identified in the legend. Note that when assuming a conservative 20 dBA noise reduction from hearing protection, the A-weighted noise falls below the 99 dBA FDA limit for NSR operation.

wears hearing protection (e.g., earplugs) with at least a noise reduction rating (NRR) of 27, assuming that their effective dBA attenuation can be estimated from NRR-7. Under these assumptions, a measured noise level of 119 dBA still satisfies the 99 dBA figure specified by the FDA for NSR operation.

For future work, a model-based image reconstruction incorporating gradient nonlinearity compensation¹¹ can be used instead of the standard Fourier transform plus interpolative image-domain shifting. This model-based technique utilizes a nonuniform fast Fourier transform³⁰ to account for discrete vs continuous MRI sampling in the presence of gradient nonlinearity distortion. This technique has demonstrated reduced blurring, especially away from gradient isocenter.

The additional first and zeroth order concomitant field terms³¹ do not affect the results of the ACR test battery, but

FIG. 4. First low-contrast detectability slice showing (a) T1 and (b) T2 second echo images. Each low-contrast object is easily identifiable. Gibbs ringing in this image is unavoidable due to the specified matrix size of the acquisition and the sinc-interpolation for display.

mostly manifest in sequences with strong gradient activity such as phase contrast³¹ and echo-planar imaging.³² These terms were identified and partially addressed by Meier *et al.*,¹⁰ but recently, a closed form solution to exactly correct for the first-order concomitant field terms amenable to gradient preemphasis has been devised.³³

The ACR imaging tests are a relatively low duty cycle task for the gradient hardware, and as such are not designed to test the gradient system for advanced application performance³⁴ such as EPI, EPI-DTI, or non-Cartesian sequences, but rather to ensure clinical image quality standards are met on a weekly and yearly basis. These additional tests of advanced applications are important and will be done and reported on in the near future. However, for these advanced applications, one substantial advantage of this head-only gradient coil is a demonstrated sustained cooling performance, approaching 25 kW, built in to the system.² This allows for the gradient system to run at full power for sustained periods of time without requiring either “no-scan” breaks or lengthening of repetition times. Note, additionally, that this high cooling power should provide a stable platform for long-duration high duty cycle imaging, with a low percent-ghosting value reported in Table I indicating a stable k -space trajectory.

5. CONCLUSIONS

A high-performance small diameter gradient set and matching 3 T superconducting magnet have been designed and fabricated. This system passes the full ACR battery of tests when gradient nonlinearity in the images is corrected utilizing a higher-order spherical harmonic model incorporating both odd- and even-order terms, in distinction to odd-order-only terms for a whole-body system which employs symmetric gradient design on all three axes. This shows potential for comparable spatial distortion correction to whole-body systems. The higher performance gradient hardware is also shown to conform to FDA acoustic noise NSR limits, despite a small increase in sound pressure level when compared to a whole-body gradient system. Future work will highlight advantages in clinical image quality given the improved gradient system performance.

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