Analysis of magnetic resonance image quality using an in-house phantom: Gamma knife application

A R Fauzia¹, A R Setiadi² and S A Pawiro¹,*

¹ Department of Physics, Faculty of Mathematics and Natural Sciences, Universitas Indonesia, West Java, Indonesia.
² Gamma Knife Referral Center, Dr. Cipto Mangunkusumo General Hospital, Jakarta, Indonesia.

*supriyanto.p@sci.ui.ac.id

Abstract. Magnetic Resonance Imaging (MRI) is increasingly used for purposes of radiosurgery and radiotherapy planning. This imaging modality can explore the physical properties of tissue with great details, especially for the brain. However, the geometric distortion is reasonably occurred and can make significant differences in certain MR application such as, localization in Gamma Knife Radiosurgery. Therefore, the geometric distortion measurement and correction should be applied. In this study, an in-house phantom was developed to measure not only geometric distortion, but also uniformity and high contrast spatial resolution of MR image. The phantom attached to Leksell stereotactic frame was scanned using 1.5T MRI GE. T1-Bravo and T2-Fiesta scan protocol were applied. Standard MRI phantom, Magphan SMR100, was used as data comparison. The result shows that geometric distortion could not be found either in Magphan SMR 100 or in-house phantom images. PIU of the in-house phantom image generated by T1-Bravo and T2-Fiesta scan protocol was around of 79.57% and 56.39%, respectively. Furthermore, the high contrast spatial resolution of the in-house phantom images was 1.0 lp/cm lower compared to Magphan SMR100 phantom images. For this study, it was concluded that the in-house phantom could be employed to analyze MR image quality, yet still needs some improvements, especially for uniformity module.

1. Introduction
Gamma Knife (GK) stereotactic radiosurgery (SRS) is a non-invasive technique that allows precise delivery of high single-dose radiation to small intracranial targets, while minimizing dose to surrounding’s normal structure [1]. The efficacy of this technique depends on many factors. One in particular is imaging modality which is employed. Magnetic resonance imaging (MRI) with static magnetic field 1.5 Tesla (T) is usually applied for GK SRS treatment planning using a Leksell stereotactic frame [2]. MRI allows high-resolution non-invasive imaging of intracranial lesion and normal structures so it is essential to target definition and treatment planning in SRS [1].

However, although the extent of a target can be determined in great detail on MR images, the geometric accuracy of these images is limited by distortions. Geometric distortion in MRI can arise from a variety of sources. These sources can be classified as hardware-related and tissue-related. The main sources contributing to geometric distortion from MRI hardware are the inhomogeneity in the main magnet, the nonlinearity in the gradient fields, and the eddy currents associated with the...
switching of the gradient coils. The tissue-related sources mainly include susceptibility difference and chemical shift [3]. External devices such as localization frames utilized for patient immobilization and stereotactic space definition in GK SRS can also induce distortions and artifacts in MR images [4].

The magnitude of the MR image distortion is minimal at the center of a closed-bore magnet and increasing gradually towards the radial edges of the scanning volume. Previous studies recently showed that in SRS application, relatively small distortion of up to 1.3 mm in MR images may result in a significant underdosage (up to 30%) of specific very small targets [5]. Thus, it is crucial to measure the geometric distortion so that the correction can be done.

The geometric distortion of MR image has been evaluated using phantoms. Scanning with phantoms containing landmarks (rods, grids, spheres, etc.) at designed location provides a clear view of the distortion in MR images and also gives a way to analyze the distortion quantitatively by comparing the estimated coordinates of the landmarks to their designed values [6]. Damyanovich et al [7] used phantom with 3D grid arrays inside to study distortion. 3D phantoms with arrays of spheres were also used in the evaluation [8].

In this study, an in-house phantom compatible with the Leksell stereotactic frame was designed and constructed. The phantom can analyze not only geometric distortion but also uniformity and high contrast spatial resolution of MR images. A 3D array of grids served as landmarks in the phantom. The design of the in-house phantom followed report of AAPM nuclear magnetic resonance Task Group No 1: quality assurance methods and phantoms for magnetic resonance imaging. As data comparison, a standard MR phantom, Magphan SMR 100, was employed.

2. Materials and Methods

2.1. Phantom design

The external dimensions of the in-house phantom are 153 mm × 153 mm × 193 mm, which forms a box. It was constructed to fit into the Leksell stereotactic frame shown in figure 1a. The in-house phantom consists of three modules such as, geometric distortion, uniformity, and high contrast spatial resolution. The design of the in-house phantom followed report of AAPM nuclear magnetic resonance Task Group No 1. In terms of imaging, the phantom is compatible for both Computed Tomography (CT) and MRI. It is filled with standard copper sulfate pentahydrate (CuSO₄·5H₂O) 2 mM solution, commonly used in MR phantoms [5].

The geometric distortion module is composed of 3D grid cube with dimension 151 mm × 151 mm × 151 mm. It was made from resin which was formed using 3D printing technique. The grid design allows MR image distortion to be assessed in all three orthogonal image planes simultaneously, without moving or readjusting the phantom. The thickness of the grid line is 1.5 mm and the spacing among vertices are 11.5 mm. The grid line needs to be sufficiently narrow to ensure easily identifiable vertex-centers, yet not so narrow as to negatively affect precision and SNR [7]. The design of this module was displayed in figure 1b.

The second module of the in-house phantom is uniformity module. The dimension of this module is 114 mm × 114 mm × 10 mm. It is composed of a uniform signal producing material, which in this case is CuSO₄·5H₂O 2 mM solution. The last module of the in-house phantom is high contrast spatial resolution module shown in figure 1c. It was made from acrylic cut into square bar patterns. The patterns consist of alternating signal and non-signal producing areas set apart from each other by a width equal to the bar’s width. The bar’s width varies from 3, 2, 1.5, 1.25, 1.00, 0.75, and 0.50 mm, while their depth is 10 mm.

2.2. Image acquisition

The in-house phantom mounted in Leksell stereotactic frame was scanned using MRI GE Optima MR450w with static magnetic field 1.5 T. Images were acquired according to the standard protocol used for GK SRS. The MRI imaging protocol consists of a T1-weighted Bravo and a T2-weighted Fiesta (Fast Imaging Employing Steady-state Acquisition) acquisition. Details of MR acquisition
parameters are provided in Table 1. Magphan SMR 100 as a standard phantom was also scanned using same protocols.

**Figure 1.** The front view (left) of the in-house phantom that was mounted into the Leksell stereotactic frame and MRI adapter. The phantom consists of three modules, they are geometric distortion (middle), uniformity, and high contrast spatial resolution (right) module

**Table 1.** MRI acquisition parameters for T1-Bravo and T2-Fiesta images

| Parameters          | T1-Bravo                  | T2-Fiesta              |
|---------------------|---------------------------|------------------------|
| Scan Plane          | Oblique                   | Oblique                |
| Frequency Direction | Anterior/Posterior        | Anterior/Posterior      |
| TE (echo time) (ms) | 4.3                       | Minimum (1.6)          |
| TR (repetition time) (ms) | 10.6                  | 5.0                    |
| Slice thickness (mm) | 1.0                       | 1.0                    |
| Flip Angle (°)      | 11                        | 65                     |
| Matrix size         | 256 × 256                 | 256 × 256              |
| Bandwidth (kHz)     | 20.83                     | 62.50                  |
| NEX                 | 2                         | 2                      |
| Intensity Filter    | None                      | E                      |

The in-house phantom later was emptied and CT scanned in order to obtain the reference control point distribution. Control point (CP), in this case, is defined as intercepting point of the grid lines. CT image were acquired by a Philips Ingenuity CT and acquisition parameters were as follows: 120 kVp, 400 mAs/slice, FOV (Field of View) 250 mm, slice thickness 5 mm, rotation time 0.5 s, and helical acquisition.

2.3. **Image analysis**

MR images of both phantoms were analyzed using Image-J and RadiAnt Dicom Viewer 5.0.2. The geometric distortion was evaluated by comparing MR images with CT images of the in-house phantom. If there is no difference between the two images, it can be explained that geometric distortion might not occurred or its value is quite small. The uniformity of MR image was described as a percent image uniformity (PIU). For a volume head coil, the PIU measure should meet or exceed 90% for scanners operating at 2T or below [10]. A region of interest (ROI) which enclosed approximately 75% of the phantom area was created to evaluate the uniformity of MR image. Then, a small ROI is chosen in the area of maximum ($S_{\text{max}}$) and minimum ($S_{\text{min}}$) pixel intensity. The relationship for calculating PIU is

$$ \text{PIU} = \left[ 1 - \frac{S_{\text{max}} - S_{\text{min}}}{S_{\text{max}} + S_{\text{min}}} \right] \times 100\% \quad (1) $$

The high contrast spatial resolution is expressed in line pair (lp)/cm. ROIs were made at each square bar pattern in MR image. Later, the profile was plotted from them. The smallest square bar
pattern which still has some ripples in its profile is determined as the value of the high contrast spatial resolution of MR image.

3. Result

3.1. The geometric distortion
Using Magphan SMR 100 phantom, the geometric distortion can be measured in test plane 3 by measuring the distance between the center of the 3 mm holes which are spaced in pattern forming 2, 4, and 8 cm squares. Its images generated by T1-Bravo and T2-Fiesta scan protocol were shown in Figure 2. The distances between holes were acknowledged exactly as 8, 4, and 2 cm, same with the ones that written in phantom guide book. This result showed that geometric distortion might not occurred or its value is way too small in both of images generated by two MR scanning protocols employed.

![Figure 2. Magphan SMR 100 phantom images generated by T1-Bravo (right) and T2-Fiesta (left) scan protocol](image)

The in-house phantom was scanned by both MRI and CT. MR and CT images of that phantom were being compared to see whether or not there are any differences between them. Figure 3 showed MR and CT images of the in-house phantom in all three orthogonal planes. Control point (CP) was
described as intercepting point of the grid lines. The distance between CP in both MR and CT images was measured manually. The distance measured was around of 11.5 mm, that value was same in MR as well as CT images. This result indicated that geometric distortion might not entirely or slightly occurred in MR images.

3.2. The Uniformity

The uniformity test was performed using the second module of the in-house phantom and test plane 2 in Magphan SMR 100 phantom. The uniformity of MR images was described as PIU. PIUs of both phantom images are provided in Table 2. PIU of Magphan SMR 100 images exceeds the tolerance limit (90 %), it shows that the MRI is still in good condition.

![Figure 4. The high contrast spatial resolution of Magphan SMR 100 and in-house phantom generated by T1-Bravo and T2-Fiesta scan protocol.](image)

### Table 2. Percent image uniformity (PIU) of Magphan SMR 100 and in-house phantom images

| Phantom            | T1-Bravo | T2-Fiesta |
|--------------------|----------|-----------|
| Magphan SMR 100    | 90.06    | 86.57     |
| In-house phantom   | 79.57    | 56.39     |

3.3. The High Contrast Spatial Resolution

High contrast spatial resolution measurement was accomplished using test plane 3 in Magphan SMR 100 phantom. American College of Radiology (ACR) recommended that the tolerance limit of high contrast spatial resolution is 5 lp/cm or 1 mm. The high contrast spatial resolution of both Magphan SMR 100 and in-house phantom displayed in Figure 4. It shows that the high contrast spatial resolution of MR image generated by T2-Fiesta scan protocol is better than another protocol. According to Magphan SMR 100 images, the high contrast spatial resolution was in tolerance limit so the MRI used is still in good condition. The high contrast spatial resolution of the in-house phantom image is 1 lp/cm lower compared to Magphan SMR 100 image.

4. Discussion

Two MR scanning protocols (T1-Bravo and T2-Fiesta) applied in GK SRS treatment planning were evaluated in terms of geometric distortion in MR image. The geometric distortion wasn’t found in any MR images generated by both of T1-Bravo and T2-Fiesta scan protocol. The magnitude of geometric distortion in the relatively small FOV employed in intracranial applications is rather limited. Pappas et al reported that the mean absolute geometric distortion was found less than 0.5 mm for MR image used in GK SRS [5].
Percent image uniformity (PIU) of in-house phantom images generated by T2-Fiesta scan protocol was far away below the tolerance limit (90%). As shown in Figure 3, in-house phantom images generated by T2-Fiesta scan protocol had more artifacts than images generated by T1-Bravo scan protocol. The existence of those artifacts decreased the magnitude of PIU. However, PIU of Magphan SMR 100 images generated by T1-Bravo scan protocol was more than tolerance limit, so it can be said that the MRI is still in good condition.

The high contrast spatial resolution is affected by many factors include: FOV, acquisition matrix, and reconstruction filter [9]. T2-Fiesta scan protocol uses intensity filter type E, yet T1-Bravo scan protocol doesn’t use any filter. Because of it, MR images generated by T2-Fiesta scan protocol have better resolution than ones generated by T1-Bravo scan protocol. The high contrast spatial resolution of the in-house phantom images was lower than Magphan SMR 100 images. Nevertheless, the difference was still acceptable.

5. Conclusion
The in-house phantom was constructed to measure MR image qualities include geometric distortion, uniformity, and high contrast spatial resolution. The result of measurement using in-house phantom is proximity to the one using Magphan SMR 100. Hence, this study concluded that the in-house phantom could be employed to analyze MR image quality, yet still needs some improvements, especially for uniformity module.

Acknowledgments
This study was supported by a grant from Universitas Indonesia (PIT 9) with contact number NKB-0034/UN2.R3.1/HKP.05.00/2019. The authors would like to acknowledge some radiographers, dr. Reyhan Eddy Yunus, Sp.Rad, M.Sc, and Mrs. Yekti Nastiti for their technical support.

References
[1] Zhang B, MacFadden D, Damyanovich A Z, Rieker M, Stainsby J, Bernstein M, Jaffray D A, Mikulis D, & Menard C 2010 Development of a geometrically accurate imaging protocol at 3 Tesla MRI for stereotactic radiosurgery treatment planning Phys. Med. Biol. 55 6601-6615.
[2] Nakazawa H, Mori Y, Yamamuro O, Komori M, Shibamoto Y, Uchiyama Y, Tsugawa T, & Hagiwara M 2014 Geometric accuracy of 3D coordinates of the Leksell stereotactic skull frame in 1.5 Tesla- and 3.0 Tesla-magnetic resonance imaging: a comparison of three different fixation screw materials Journal of Radiation Research 55 1184–1191.
[3] Wang D & Doddrell D M 2005 Geometric distortion in structural magnetic resonance imaging Current Medical Imaging Reviews 1 49–60.
[4] Pappas E P, Seimenis I, Motsatsos A, Georgiou E, Nomikos P, & Karaiskos P 2016 Characterization of system-related geometric distortions in MR images employed in Gamma Knife radiosurgery applications Phys. Med. Biol. 61 6993-7011.
[5] Pappas E P, Alshanqity M, Moutsatsos A, Lababidi H, Alsafi K, Georgiou K, Karaiskos P, & Georgiou K 2017 MRI-Related geometric distortions in stereotactic radiotherapy treatment planning: Evaluation and dosimetric impact Technology in Cancer Research & Treatment 16 1120–1129.
[6] Huang K C, Cao Y, Baharom U, Balter J M 2016 Phantom-based characterization of distortion on a magnetic resonance imaging simulator for radiation oncology Phys. Med. Biol. 61 774-790.
[7] Damyanovich A Z, Rieker M, Zhang B, Bissonnette J P, & Jaffray D A 2018 Design and implementation of a 3D-MR/CT geometric image distortion phantom/analysis system for stereotactic radiosurgery Physics in Medicine & Biology 63.
[8] Vermandel M, Baert G, Reyns N, & Betrouni, N 2015 Phantom and non-rigid registration to tackle distortions from MRI in stereotactic conditions: Proof of concept and preliminary results Proceedings-International Symposium on Biomedical Imaging 1061–1064.
[9] Price R R, Axel L, Morgan T, Newman R, Perman W, Schneiders N, Selikson M, Wood M, & Thomas S R 1990 Quality assurance methods and phantoms for magnetic resonance imaging: Report of AAPM nuclear magnetic resonance task group no 1 Med. Phys. 17 287-295.

[10] Jackson E F, Bronskill M J, Drost D J, Och J, Pooley R A, Sobol W T, & Clarke G D 2010 AAPM Report No 100: Acceptance testing and quality assurance procedures for magnetic resonance imaging facilities (College Park: American Association of Physicist in Medicine One Physics Ellipse).