A Regularly Structured 3D Printed Grid Phantom for Quantification of MRI Image Distortion

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Abstract Radiotherapy treatment planning (RTTP) necessitates accurate delineation of tumor volume and adjacent structures at risk. Magnetic resonance imaging (MRI) offers superior soft-tissue contrast compared with CT but suffers from inherent geometric distortions. The problem is exacerbated at higher field strength where there is an increase in inhomogeneties in the main B_0 magnetic field as well as the B_1 RF field [1,2]. Furthermore, non-linearities in the applied gradients add a further element to these distortions. With 3T scanners becoming increasingly popular in RTTP due to the improvement in signal-to-noise ratio (SNR) [2], accurate quantification of distortion, and correction for it, is becoming more important. In this study, we propose a cost-effective regularly structured three-dimensional 3D printed grid phantom, which enables one to quantify machine-related MR distortion by comparing the locations of corresponding features in both MR and CT data sets.

Materials and Methods A rigid 170mm×170mm×250mm mesh cuboid insert with a 2mm wireframe was designed in SketchUp (Trimble Navigation Limited, Sunnyvale, CA). This rigid 3D grid was printed externally (Shapeways BV, Eindhoven) with a spatial error margin of ±0.15mm using a strong white plastic material and was fixated inside a 240mmϕ × 250mm cylinder. The 3D printed insert and the leak-tight phantom container are shown in *Fig. 1a* and *Fig. 1b* respectively. Each point of intersection between the mesh lines in 3D space is treated as a reference feature or a control point. Control point spacing is approximately 20×20×20.0mm^3 in the *x, y* and *z* dimensions. The volumetric mesh cuboid insert shown in *Fig.1a* contains a matrix of 9×9 mesh lines in the *xy* plane and 9×11 mesh lines in both *xz* and *yz* planes yielding a total of 9×9×11 (891) control points. The phantom was then filled with off-the-shelf baby oil (Johnson&Johnson, New Brunswick, NJ) with T_1=289ms and T_2=93ms. This is a readily available perfumed mineral oil to avoid wavelength-induced artefacts within the imaged structure. The American Association of Physicists in Medicine thus recommends oil over water to reduce such artefacts [3,4]. With the axis of the cylinder aligned along the B0 field, MR images were acquired on a 3.0T Philips Achieva system (Philips Healthcare, Best, the Netherlands) using the body coil and a 3D gradient echo sequence with TE, TR and flip angle of 5.2ms, 11ms and 30° respectively [3]. A 361mm field of view was used and a total of 289 slices contiguous slices were acquired with a voxel size of 0.71mm×0.71mm in-plane and 1.25mm through-plane. A single *k*-space line was encoded per TR. In order to define the true, undistorted control point positions, a corresponding CT scan of the phantom was generated using a GE LightspeedRT16 system (GE Healthcare, Milwaukee,WI, USA) using the same MR parameters.

All of the 891 control points were detected using 3D normalized cross correlation using a MATLAB (The Mathworks, Natick, MA) program. This was achieved by defining two templates of 4.97×4.97×8.75mm^3 (7×7×7 voxels) extracted from the iso-center (origin) of the CT and MR acquisitions. We assumed that this point in space was free of distortions in the MR acquisition as it is in the magnet iso-center. Each template was then convolved with its originating volume yielding a volume of correlation coefficients for both the CT and MR acquisitions. Thresholding was then used to determine the control points corresponding to the highest correlating points. Connected components were then computed of the thresholded correlation coefficients and the centroids of each of the control points were determined. A distance map was then calculated from the iso-center to each of the control points. We used L2 norm or the Euclidean distance metric to determine the distance between the two distance maps corresponding to MR and CT control points as the CT distortion map was treated as the ground truth.

Results Axial and sagittal CT and MR acquisitions of the 3D printed grid are shown in *Fig. 2*. Measured distances between the control points in the CT dataset were accurate to within 0.05mm, which suggests that the printing accuracy was better than the manufacturer stated value of 0.15mm. The overall mean error across the entire MR volume was found to be 1.6mm with a standard deviation of 0.65mm. The majority of distortion was observed in the through-plane. In *Fig. 3a* in-plane and *Fig. 3b* through-plane L2 norm errors are shown. In the axial slices the error oscillated between 1.2mm and 1.9mm for different slice positions. In the sagittal slices the error was 2.4mm in the right direction and 1.7mm in the left direction at the edge slices 80mm away from the central slice. This error reduced to a minimum of 1.2mm in the central slices.

Discussion & Conclusions We have demonstrated that an accurate, cost-effective and simple distortion phantom can be constructed using 3D printing technology and readily available baby oil. We measured this error in our 3.0T clinical scanner across the entire imaged volume and was found to be 1.6mm with a standard deviation of 0.65mm.

Fig. 1: a) 3D printed insert b) leak-tight phantom container

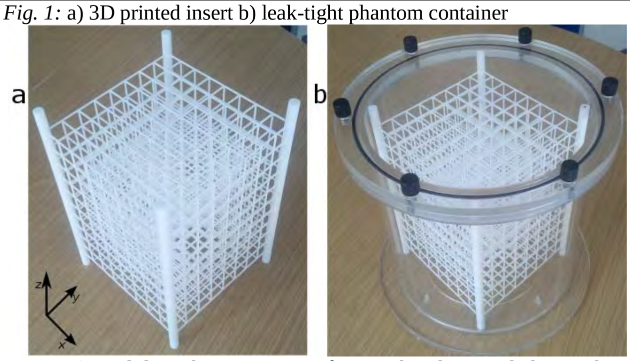


Fig. 3: a) Euclidean distance in mm for Axial and Sagittal slices. The vertical line in each figure represents the middle slice in both orientations.

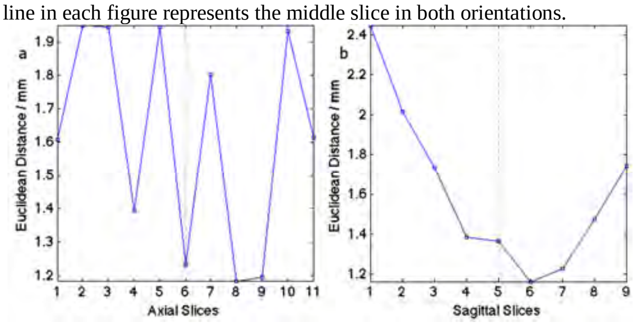
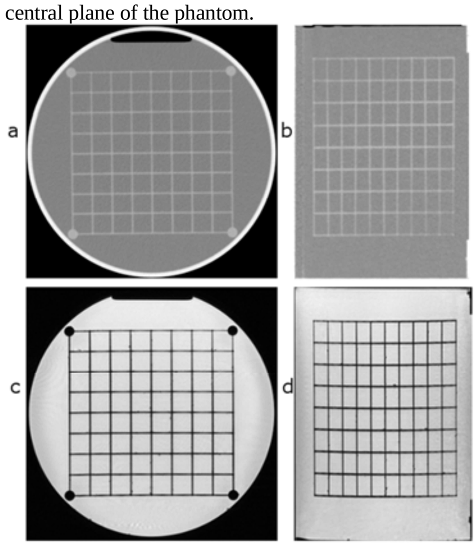


Fig. 2: (a) Axial CT, (b) Sagittal CT, (c) Axial MR and (d) Sagittal MR images through the central plane of the phantom.



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