

Characterising gradient non-linearities of a split gradient coil in a hybrid MRI-linear accelerator

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Purpose

At the University Medical Centre Utrecht, a hybrid MRI-linear accelerator prototype is constructed that facilitates MR imaging during radiotherapy treatment of cancer. In such treatment, ionizing radiation is delivered to a tumour that may move while a patient is breathing. Using online imaging protocols is widely accepted as a means of decreasing treatment uncertainties and reducing margins. So far, cone-beam computed tomography (CBCT) is the most advanced imaging modality that can be used online. Although CBCT provides geometrically very accurate images, the soft-tissue contrast is poor compared to MRI.

Real-time diagnostic quality MR imaging during treatment provides us with the soft-tissue contrast to further reduce margins and investigate stereotactic boosting and ablation strategies for macroscopic tumours. However, gradient non-linearities and field inhomogeneities lead to distorted images that must be used cautiously as a basis for such interventions.

Although all of our prototype MRI electronics is regular Philips Achieva equipment, adaptations have been made to the MRI hardware designs to accommodate passage of a therapeutic photon beam [2]. These adaptations include a split gradient coil. Here we characterise the performance of this gradient coil in terms of geometric accuracy.

Methods

In 2DFT spin echo (SE) images, gradient non-linearity and field inhomogeneity can be inferred from two images that would otherwise be identical, but where the phase and frequency encoding directions are interchanged from one image to the other [3]. SE images are acquired of a phantom consisting of 77 rods spaced equally in a grid, containing MnCl₂ doped water (Figure 1). Away from the ends, the rods can be considered infinite cylinders so that field inhomogeneity effects in the mid-phantom planes cannot be confused with susceptibility artefacts.

Images are acquired of 5mm thick slices with an in-plane acquisition resolution of 1.6x1.6mm. The TE/TR is 30/500ms and a flip-angle of 90° is employed. The read-out gradient strength is kept at 3mT/m. The imaging experiments are done with and without the standard gradient non-linearity compensation that makes use of description of the gradient field in spherical harmonics expansion.

Rod positions are extracted from the images using Matlab (The Mathworks, Natwick USA). Now we solve for each of 77 known rod coordinates

(x, y, z)

$$\Delta G_x(x, y, z) = G_x(x_2 - x),$$

$$\Delta G_y(x, y, z) = G_y(y_1 - y) \text{ and}$$

$$\Delta B_0(x, y, z) = G_x(x_2 - x_1),$$

where (x_1, y_1) is the coordinate of rods in the image where the frequency

encoding is along x, (x_2, y_2) is the coordinate of rods in the image where the frequency encoding is along y and the ΔG_i and ΔB_0 denote the gradient non-linearities and field inhomogeneities [3].

Results

Images are acquired from the mid-coronal, -sagittal and -transverse planes. Rod position are obtained and gradient non-linearity maps are computed according to the above equations. Below, the gradient non-linearities from the mid-coronal images are shown for the x and z gradient coils (with standard non-linearity compensation switched on). Non-linearities caused a discrepancy of rod positions of 4 mm at most.

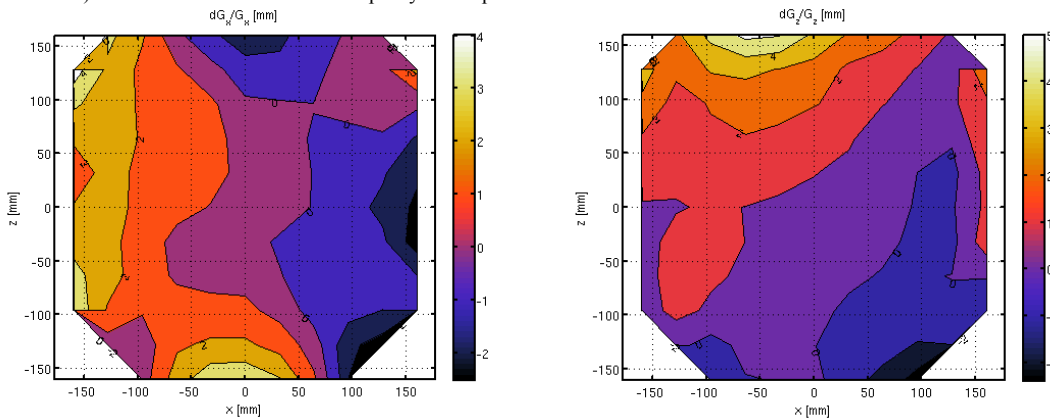


Figure 1: two views of the phantom (rod spacing 42mm in either direction, width/height/depth 336mm)

Conclusion

The performance of our split gradient coil in terms of geometrical accuracy has been presented. We found that, when using standard gradient non-linearity compensation present in the scanner software, the residual error due to gradient non-linearity was no more than 4mm in a field of view of 336x336mm of the coronal, sagittal and transverse mid-planes. Although this is a promising result, other methods have to be applied before images made with our gradient coil can be applied in an interventional setting.

References

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