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From static to dynamic 1.5T MRI-linac prototype: impact of gantry position related magnetic field variation on image fidelity

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Abstract

Recently, the MRI-linac prototype at UMC Utrecht was upgraded with a ring-based gantry, where all linac components are mounted on a ring around the MR scanner. Although adaptations have been made to both linac and MR magnets in order to account for mutual magnetic interference, interference problems cannot be ruled out completely. Therefore, in this paper the impact of gantry position dependent magnetic field inhomogeneity variation on the geometrical accuracy of acquired MR images is quantified. Magnetic field maps were acquired in a large field of view for static gantry positions in shimmed and un-shimmed conditions. Reproducibility of the shim settings was assessed. From the fieldmaps, a minimum gradient strength needed to acquire images with geometric distortions of, at most, 1 mm was derived. Moreover, imaging during gantry rotation was performed for a range of imaging parameters and rotation speeds. From the measurements we conclude that images with good geometric fidelity can be obtained for all static gantry positions, provided that shimming is performed for each new gantry position. This indicates that the present prototype is suitable for static IMRT scenarios. Shim settings are highly reproducible, suggesting that shimming via look-up tables is feasible. Finally, imaging during gantry rotation may produce severely distorted images at present and is likely to require advanced compensation methods such as dynamic shimming or higher order reconstruction.

Keywords: MRI-guided radiotherapy, MRI-linac, magnetic field inhomogeneity

(Some figures may appear in colour only in the online journal)
1. Introduction

MRI can provide high soft-tissue contrast anatomical data with both high spatial and temporal resolution. The use of MRI in radiotherapy is explored for all current phases, e.g., tumour (motion) characterization, delineation and response assessment, but also for on-line treatment guidance. Several groups are working on integration of MRI into the treatment machine (Dempsey et al 2005, Lagendijk et al 2002, Fallone et al 2009). At UMC Utrecht, in collaboration with Elekta (Eleta AB, Stockholm, Sweden) and Philips (Best, The Netherlands), a prototype was constructed to proof this concept for a combination of a 1.5T MR scanner and a 6 MV linear accelerator (Raaymakers et al 2009). The prototype was recently upgraded with a ring-based gantry, where all components needed for radiation production were mounted on this gantry around the magnet (figure 1(b)), to facilitate modern IMRT type treatments.

One of the obstacles in integration of the two machines is the magnetic field interference. A linear accelerator cannot function in a magnetic field without suitable adaptations. At the same time, bringing all the components of an accelerator close to the MR scanner may influence the homogeneity of the magnetic field in the scanner, which could potentially lead to image distortions. Hence, the fringe-field of our scanner magnet was shaped such as to create a toroidal magnetic field-free zone around the scanner cryostat for the gantry and the equipment it carries (indicated by the light green torus in figure 1(a)). Still, this zone is too small to contain the whole ring and all the components it carries leading to potential magnetization of some hardware. Also, there are active magnetic components on the gantry which might affect MR imaging irrespective of their location relative to this zone. These active magnets are for example necessary for functioning of magnetron and RF circulator.

An MR scanner is usually able to account for external magnetic fields by means of various types of shimming. The static distortion from the surroundings, e.g., magnetized iron reinforcements in the floor, can be accounted for by passive shimming. During the installation process, a number of small ferro-magnetic plates is placed at the inside of the bore of the magnet to compensate the external static disturbances. For further fine tuning and correction of for
instance patient induced distortions, active shim coils can be used prior to data acquisition.
Finally, if time-varying distortions need to be accounted for, dynamic shimming can be applied.

As active shimming is a fairly slow process, a multi-beam IMRT scenario would require
shimming prior to each beam, which potentially leads to timing problems. In addition, our
MRI-linac prototype is well-prepared to be used in dynamic delivery techniques such as
VMAT. When the gantry is rotated while imaging is performed, time-varying magnetic field
variations may lead to image artefacts which might require dynamic shimming.

In this paper, the impact of gantry position dependent magnetic field inhomogeneity
variation on the geometrical accuracy of acquired images is quantified. We do this for both
static gantry positions in shimmed and un-shimmed scenarios, as well as with a moving
gantry, and derive conditions under which undisturbed images can be produced with acceptable
geometric fidelity.

2. Methods

2.1. MRI-linac prototype

The upgrade to our MRI-linac prototype concerns the accelerator components as well as the
gantry. The 1.5T MR scanner hardware is unchanged. The accelerator is a 8 MV standing wave
accelerator (Elekta AB, Stockholm, Sweden). All its components have been mounted on a 4 m
diameter steel ring, supported by four wheels. Slipping technology is used to transfer power
onto the rotating structure and a closed-circuit on-board cooling system dumps heat into the
air in the room, hence making continuous rotation at a maximum speed of 6 rpm possible. The
collimator is upgraded from a cast block system to a 160-leaf multi-leaf collimator derived
from Agility (Elekta AB, Stockholm, Sweden). To cope with the magnet fringe-field, more
powerful dc motors were fitted to this MLC.

2.2. MR imaging

A large, disc-shaped homogeneous phantom (diameter 40 cm) was used for field mapping
experiments. A dual gradient echo sequence was used to measure magnetic field maps
in mid-transverse, sagittal and coronal planes, respectively. Sequence parameters were
\( T_E/T_R = 7/20 \) ms, \( \Delta T_E = 1 \) ms, flip angle 70°, FOV 45 × 45 cm², in-plane resolution
2 × 2 mm², bandwidth 78 Hz per pixel. The gantry was rotated in 5° steps and a magnetic field
map was acquired at each position while keeping the gantry static during each measurement.
All experiments took place on our MRI-linac prototype.

To investigate the importance of shimming, the above experiment was performed without
and with shimming turned on. For each of the 72 examined gantry angles, optimal shim settings
were determined by the MR scanner for the volume of the imaging phantom and the settings
were recorded via the scanner logfiles. Moreover, to investigate the repeatability of the shim
settings the experiment with shimming turned on was repeated ten times.

Finally, the effect of gantry rotation during image acquisition was assessed for varying
gantry speeds and imaging parameters. The gantry speed in this experiment ranged from 0 to
5 rpm. For each speed setting the echo time of a gradient echo sequence (TR/\( \alpha \) = 100 ms/50°,
FOV 250 × 250 mm², in-plane resolution 2 × 2 mm², bandwidth 76 Hz per pixel) was varied
from 10 to 50 ms. Transverse images were acquired of a phantom bottle with a diameter of
18 cm positioned such that the long axis of the bottle was aligned with the scanner bore.
In each experiment, the gantry position was first set to 0° and optimal shim settings were
Figure 2. Minimum needed read-out gradient strength for $\Delta x_{\text{acceptable}} = 1$ mm (a) without and (b) with shimming (maximum intensity projections over all measured gantry angles).

determined by the MR scanner, before the image acquisition and gantry rotation were started simultaneously.

2.3. Data processing

Before further examination, field maps were unwrapped using an in-house implementation of Jenkinson’s phase unwrapping algorithm (Jenkinson 2003). For the canonical gradient echo or spin echo sequence, the read-out gradient strength $G_{\text{read}}$ needed to stay within an acceptable signal positioning error $\Delta x$ is given by

$$G_{\text{read}} = \frac{\Delta B_0}{\Delta x},$$

with $\Delta B_0$ the deviation of the magnetic field strength from the nominal value (in our case 1.5T). The set of 72 read-out gradient strength maps derived from the field maps were stacked and a maximum intensity projection was taken along the gantry position direction to derive a presentable result.

3. Results and discussion

Figure 2 details the results obtained from the field mapping experiments. From figure 2(a) we observe that a minimum read-out gradient strength of 15 mT m$^{-1}$ would be needed to satisfy a geometrical accuracy of 1 mm. This is almost independent of the location of a region of interest where the geometrical accuracy should be high.

Figure 2(b) displays the needed gradient strength when shimming is taken into account. Accurate shimming substantially reduces the gradient strength needed by almost half the initial value, to below 10 mT m$^{-1}$, if the geometrical errors are to be kept within 1 mm. Field disturbances that result in this value for the gradient strength occur predominantly at the edge of the 40 cm diameter FOV. Other than in the un-shimmed situation, the effect is much more benign for a region of interest closer to the centre of the scanner. In fact, in a 30 cm diameter spherical volume the minimum needed gradient strength is already less than 5 mT m$^{-1}$. One should keep in mind that this result does not take into account gradient nonlinearity effects or field inhomogeneity caused by patient specific susceptibility, which both have to be accounted for separately.
An overview of the shim settings obtained from the repeated shimmed experiments is given in figure 3. We observe approximate sine and cosine patterns for the x- and y-shims as a function of gantry position; these are the shims along principle axes orthogonal to the gantry structure rotation axis. The sine and cosine functions together describe a trajectory close to a circle, from which we can deduce that the main magnetic field contribution we account for in the shimming process is likely to come from specific components on the gantry. The active magnets in the RF circulator and the magnetron are suspected to be the main contributors to this field. Note that shim values for the z-direction (i.e., the feet–head direction in the MR scanner and parallel to the rotation axis of the gantry) hardly vary with gantry position.

Finally we perform the dynamic experiment where images are acquired during gantry rotation. Results are shown in figure 4. Depending on gantry speed and echo time, artefacts may comprise of scaling of the apparent phantom size as well as (severe) image warping and signal loss. It can be readily observed that the artefact level increases when gantry rotation speed is set to larger values and the echo time is increased (from upper left to lower right corner). This observation can be clarified in terms of increased measurement inconsistencies; the phase contribution coming from each part of the phantom changes during the measurement as the field inhomogeneity changes.

The spread for the shim settings per gantry position is only small, indicating good reproducibility. This suggests that shimming for the gantry position dependent fields could in principle be done through the use of a look-up table. In an on-line imaging scenario where images are acquired before and during delivery of each radiation field of a step-and-shoot IMRT plan to (re-)verify the patient set-up and monitor delivered dose, this saves valuable time needed after rotating to a next beam.

The field maps themselves where very reproducible as well. Similar look-up table approaches could thus be implemented for use with retrospective image distortion correction method such as proposed by Crijns et al (2011), for those cases where one wishes to use lower than adequate gradient strengths due to requirements on SNR or even higher geometrical fidelity.
Our experiments show that the magnetic field variations due to gantry rotation are too large to produce good quality images while the gantry is moving. This puts a constraint on motion of the gantry and hence on treatment scenarios that rely on a dynamic image feedback. This constraint may be removed if dynamic magnetic field adaptations become available, such as proposed by Blamire et al (1996). In such scenario, for instance tabulated shim settings can be taken as function of the gantry position. On the canonical 1.5T MR magnets, however, only zeroth and first order (i.e., linear) magnetic field components can be shimmed and thus experimental validation is required to confirm the suitability of such approach given the magnetic field patterns measured. Alternatively, magnetic field variations can be taken into account during reconstruction (Wilm et al 2011). Such reconstruction schemes rely on higher order spatiotemporal information of the magnetic field and ideally receive high temporal resolution information from, for example, magnetic field probes (Barmet et al 2008), which could be integrated in the MR scanner bore.

4. Conclusion

In the upgraded MRI-linac prototype, imaging with acceptable geometric fidelity is possible for static gantry angles provided that the read-out gradient strength used is at least 10 mT m$^{-1}$ and the magnetic field is shimmed adequately. Imaging during gantry rotation results in artefacts, the severity of which depends on both gantry speed and imaging parameters. In the near future, we will carry out a re-evaluation for a new and improved magnet with a lower stray field.
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