Measurement of isocenter alignment accuracy and image distortion of an 0.35 T MR-Linac system

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Abstract

For hybrid devices combining magnetic resonance (MR) imaging and a linac for radiation treatment, the isocenter accuracy as well as image distortions have to be checked. This study presents a new phantom to investigate MR-Linacs in a single measurement in terms of (i) isocentricity of the irradiation and (ii) alignment of the irradiation and imaging isocenter relative to each other using polymer dosimetry gel as well as (iii) 3-dimensional (3D) geometric MR image distortions. The evaluation of the irradiated gel was performed immediately after irradiation with the imaging component of the 0.35 T MR-Linac using a T2-weighted turbo spin-echo sequence. Eight plastic grid sheets within the phantom allow for measurement of geometric distortions in 3D by comparing the positions of the grid intersections (control points) within the MR-image with their nominal position obtained from a CT-scan. The distance of irradiation and imaging isocenter in 3D was found to be (0.8 ± 0.9) mm for measurements with 32 image acquisitions. The mean distortion over the whole phantom was (0.60 ± 0.28) mm and 99.8% of the evaluated control points had distortions below 1.5 mm. These geometrical uncertainties have to be considered by additional safety margins.

1. Introduction

Image-guidance is a key element of modern radiotherapy. Independent of the method, image-guidance may assist in patient setup, detection of anatomical changes during treatment or real-time motion management, e.g. by plan adaptions (Martinez et al 2001, Kontaxis et al 2017), gated or tracked treatments (Kubo et al 1996, Heerkens et al 2014). While most modern linear accelerators are equipped with on-board imaging systems to acquire images of the patient anatomy by kilo voltage Cone Beam Computed Tomography (kV-CBCT) (Jaffray et al 2002), hybrid devices for magnetic resonance (MR)-guided radiotherapy (MRgRT) have recently attracted much attention and became increasingly important (Lagendijk et al 2008, Fallone et al 2009, Keall et al 2014, Mutic and Dempsey 2014). In contrast to x-ray imaging, MR-imaging (MRI) is not linked to additional dose to the patient and provides superior soft tissue contrast (Reiser et al 2008). All radiotherapy machines require tests to assure accurate alignment of the irradiation isocenter to a certain point in space.

Determination of the irradiation isocenter is usually performed with films by simple star shot measurements (Treuer et al 2000) or the so-called Winston–Lutz test (Lutz et al 1988). These well-known quality assurance (QA) tests compare the measured irradiation isocenter position with a reference point in space that was formerly marked by accurately aligned room lasers. While MRgRT-devices may still be equipped with lasers their accurate alignment is less critical as the final setup is based on the acquired MR-images in which the isocenter is defined at a certain position (termed as imaging isocenter, which is identical to the nominal irradiation isocenter). The alignment of imaging and the actual irradiation isocenter is of great relevance across all devices dedicated to image-guided radiotherapy and is not restricted to MRgRT-devices.
The measurement of the isocenter-alignment in MRgRT-devices involves several requirements and problems regarding the phantom design as well as the evaluation procedure: (i) to allow for image- rather than laser-based positioning in the MRgRT-device, at least part of the phantom has to be visible in the MR-images. (ii) To visualize the irradiation isocenter, a 2D or 3D radiation detector is needed. (iii) To evaluate the isocenter alignment, the position of the imaging isocenter has to be either transferred to the detector or the measured position of the irradiation isocenter has to be transferred to the MR-image. While requirement (i) prevents solid state phantoms without any liquid structures, (ii) is usually realized by films. However, as the irradiation of films or radiochromic 3D dosimeters like PRESAGE® (Adamovics and Maryanski 2006, Brown et al 2008, Thomas et al 2013, Costa et al 2018) are not visible in the MR-images, the imaging isocenter has to be transferred to the resulting image of the dosimeter. As an alternative, polymer gel (PG) dosimeters (De Deene et al 1998) may be used. Irradiation of PG leads to local changes in mass density and relaxation rate due to polymerization of monomers (Baldock et al 2010), making unirradiated and irradiated parts distinguishable in CT (Hilts et al 2000) but also in MRI (Venning et al 2005). 3D Polymer gel dosimeters are a useful tool to verify motion compensation concepts in photon radiotherapy (Mann et al 2017) and their radiation response is only minimally influenced by magnetic fields (Lee et al 2017). In measurements of the isocenter accuracy, a significant advantage of PG is that it can be evaluated by the MRI unit of the MRgRT-device immediately after irradiation, if only geometric rather than dosimetric aspects are of interest. It has been shown previously, that results of geometrical measurements are well comparable to those of radiochromic films (Dorsch et al 2018).

A recent study demonstrated the coincidence of radiation and imaging isocenter in 3D for a conventional Linac with an onboard kV-CBCT (Adamson et al 2019). A first isocenter alignment measurement at a 0.35 T MR-Linac-System was presented by Dorsch et al (2019), however, this study used a suboptimal 2D sequence with a slice thickness of 20 mm to generate a sufficiently high signal at this low field strength. The 2D image had a low signal-to-noise ratio and did not allow for isocenter alignment measurements in 3D. In the present study, we investigate the alignment accuracy of irradiation and imaging isocenter of a MR-Linac system in 3D using an isotropic spatial image resolution of 1 mm.

The intention of MRgRT, however, is not only to perform accurate image-based setup corrections, but also to detect anatomical changes and motion within the patients and to adapt the treatment plan as well as the delivery accordingly. Besides accurate isocenter alignment, this also requires distortion-minimized images, which is more difficult to achieve for MR- than for x-ray images. MRI distortions result from inhomogeneities of the static magnetic field, susceptibility effects originating from the scanned object as well as gradient non-linearities (Schad et al 1992, Janke et al 2004, Wang et al 2004b, 2004b, Doran et al 2005, Reinsberg et al 2005, Baldwin et al 2007, Tadic et al 2014) and increase with increasing distance from the center of the magnet. While this is less critical for purely diagnostic MRI-applications, these distortions may have impact on the delivered radiation dose in MRgRT (Yan et al 2018) and are required not to exceed certain thresholds. Distortions in MRI can be measured by employing a stack of regularly arranged grids and by comparing the positions of the grid intersections in the image with their nominal positions (Wang et al 2004a, Stanescu et al 2010).

In general any regular geometric structure can be used and various commercial phantom solutions exist (for 2D-measurements: the ACR-phantom (American College Of Radiology 2005) and the spatial integrity phantom (Fluke Biomedical, Everett, WA); for 3D-measurements: the MAGPHAN® phantom series (The Phantom Laboratory, Greenwich, NY, USA)). Also spherical harmonic analysis (Janke et al 2004, Tadic et al 2014) has been used to quantify geometric distortions in 3D (Phantom: ModusQA MRI3D (Modus Medical Devices Inc., London, Canada)).

In addition to the measurement of the isocenter alignment by PG also geometric image distortions are performed in this study. For this, a new phantom was developed allowing both measurements to be performed either simultaneously while disregarding distortions near to the isocenter or sequentially, which provides the distortions over the whole phantom volume.

2. Material and methods

2.1. Experimental setup

For this study a special QA phantom was developed, to investigate a 0.35 T MR-Linac (MRIdian, ViewRay, Inc., Oakwood Village, OH, USA) (Klüter 2019) in terms of (i) isocentricity of the irradiation, (ii) alignment of the irradiation isocenter and imaging isocenter to each other, and (iii) geometric image distortions using the clinically applied MR sequences.

2.1.1. Phantom

The phantom (figure 1) consists of a polymethyl-methacrylate (PMMA) cylinder (height 23.5 cm, diameter 20 cm) containing a mounting for a spherical glass flask (borosilicate glass with outer diameter 8.5 cm, volume 250 ml, wall thickness: 1 mm) at the center, which can be filled with PG. The gel container is surrounded by
eight uniformly spaced, regular plastic grids (thickness 12.8 mm), which are used to measure geometric image distortions in the \( \text{zx} \)-plane. By means of eight rods, the grids are aligned within the phantom as well as with respect to each other. The distance between the grids was maintained by spacers on the rods (height 12.8 mm). In each plane, the grid intersections serve as control points and the distortions are obtained by comparing the control point positions in the image with their nominal positions obtained from a CT scan. In the \( \text{zx} \)-plane, the spacing of the grid is 14.2 mm \( \times \) 14.2 mm and distance of two neighbouring control points in \( \text{y} \)-direction is 25.6 mm (coordinate system according to ViewRay & IEC61217 system (International Electrotechnical Commission 2011), see figure 1). In total, the phantom contains 994 control points. According to a recommendation of AAPM Report 100 for MR QA (American Association of Physicists in Medicine 2010), the phantom is filled with a 3.6 g l\(^{-1}\) NaCl- and 1.25 g l\(^{-1}\) CuSO\(_4\)-solution to enhance the conductivity and reduce the T1-relaxation time constant.

To fine-tune the alignment of the phantom in the MR-Linac, four pairs of wedges marking the center of the spherical flask are placed in the phantom (figure 1). The wedges (12 mm \( \times \) 12 mm \( \times \) 4 mm, 45° slope) are realized by 3D printing (VeroClear™-material and Objet30 pro 3D-printer, StrataSys, Eden Prairie, USA) and are visualized in the MR-image due to their signal extinction in contrast to the surrounding solution. Only if the image slice is centered with respect to the four wedges, the contributing pixels of the fiducials in the image appear with the same length (figure 1(c)). A similar technique is used to measure the slice position accuracy for the standard ACR phantom measurement. In addition, these markers are used to define the nominal irradiation isocenter in the treatment planning system (TPS). With this phantom configuration, isocenter accuracy and image distortions can be measured simultaneously disregarding potential distortions in the isocenter region. By replacing the combination of the gel container and grid by a continuous grid that fills the whole phantom (1330 control points), the distortions can be measured also in the isocenter region. In this case, isocenter accuracy and image distortions have to be measured sequentially.

For additional comparison with our distortion measurements, the commercial 2D spatial integrity phantom (Fluke Biomedical, Everett, WA) was employed. This phantom contains a single PMMA slab with contrast

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**Figure 1.** (a) Side view of the in-house developed phantom with the inserted gel container (overlay of images of the complete phantom and the phantom where 4 of the 8 grids were removed). (b) Top view showing the upper grid. (c) Corresponding MR-images of the container together with the wedge-shaped fiducials in coronal (left) and transversal (right) planes. (1) For a slice position centered exactly at the intersection of the wedge pair, two equally long dark bars are seen in the image resulting from an extinction of the MR signal by the solid material of the wedges. (2) If the slice position is off-center, the bars will have different lengths.
agent-filled bores. A detailed description of this phantom and the related measurement method can be found elsewhere (Ginn et al 2017, Green et al 2018).

2.1.2. Polymer gel

For the polymer gel measurements the PAGAT (PolyAcrylamide Gelatin gel fabricated at ATmospheric conditions) polymer gel was used due to its visibility in MRI, low dose rate dependence (De Deene et al 2006) and low in-house manufacturing costs. The gel consists of two different monomers (2.5% w/w acrylamide and 2.5% w/w N,N′-methylene-bis-acrylamide) as active components embedded within a gelatin matrix (6% w/w Gelatin, 300 bloom, SIGMA Aldrich). Additionally, 5 mM bis(tetrais(hydroxymethyl)phosphonium) chloride (THPC) was used as radical scavenger to reduce the influence of dissolved oxygen. In addition, the gel was flushed with nitrogen for 1–2 min directly before adding the antioxidant resulting in a reduced amount of dissolved oxygen (De Deene et al 2002). As a next step, the PG was filled in a spherical flask (also previously flushed with nitrogen), sealed with a plug, wrapped in aluminum foil, and placed in a desiccator. The desiccator was completely flooded with nitrogen to store the flask in a low oxygen atmosphere until irradiation. Additionally, it was stored in a refrigerator at 4 °C for 20–24 h. Four hours before irradiation the desiccator was removed from the refrigerator and stored at room temperature (Vandecasteele and De Deene 2013, Mann et al 2017) such that the gel was in a temperature equilibrium with its surrounding at the time of irradiation.

In general, irradiation of the PG leads to polymerization and a change of the R2-relaxation rate of the PG measured quantitatively in MRI by means of a multi-spin-echo sequence (Baldock et al 2010). After proper calibration, changes in R2 can be converted into dose (Venning et al 2005). As this study investigates only geometrical aspects, a calibration and conversion into dose was not performed. It has been shown that uncalibrated PAGAT polymer gels lead to comparable results in terms of geometric parameters in irradiation isocenter measurements as the gold standard radiochromic film (Dorsch et al 2018).

2.2. Measurements

2.2.1. Isocentricity and alignment of irradiation and imaging isocenter

The developed phantom was inserted in an additional PMMA tube at the clinical 0.35 T MR-Linac, allowing the complete enclosure of the phantom with the pair of surface flex coils of the MR-Linac system (12 receiver channels) without moving the phantom by the coil positioning. Then, the phantom was aligned by aligning the crosshairs on the phantom surface to the lasers of the device. The longitudinal axis of the phantom was oriented along the main magnetic field. In a second step, MR-images were acquired, using a TrueFISP sequence (Balanced Steady State Free Precession (bSSFP) (Bieri and Scheffler 2013) used for clinical routine measurements as the gold standard radiochromic film (Dorsch et al 2018).

2.2.2. Geometric distortions

To measure image distortions, the phantom was scanned in two configurations: (i) using the combination of gel container and grid inserts (simultaneous measurement of isocenter accuracy and image distortion), and (ii) using the continuous grid (distortion measurement only, including the isocenter region). For this, a clinically applied bSSFP MR imaging-sequence implemented on the MR-Linac was used, as described in section 2.2.1. To validate the implemented distortion correction of the MR-Linac device, each measurement was performed with and

7 Research sequence not available on the commercial MRIdian.
without the correction. As ground truth, a CT scan of the phantom was performed using a Somatom Definition Flash (Siemens Healthineers, Forchheim, Germany) scanner with the following parameters: voltage 120 kVp, current 600 mAs, slice thickness = 0.6 mm, and a resolution of $1 \times 1 \text{mm}^2$. Additionally, sequence-related distortions caused by the turbo spin-echo sequence, which was used for PG measurement (see section 2.2.1), were also quantified by measuring the central grid of the phantom but with an increased FOV of $186 \times 186 \text{mm}^2$ to scan the whole grid structure of the phantom. For comparison, all distortion measurements were repeated with the commercial 2D Phantom. For this, the applied bSSFP sequence as described in section 2.2.1 was slightly modified by employing a larger FOV and slice thickness of 3 mm resulting in a resolution of $1.5 \times 1.5 \times 3.0 \text{mm}^3$. The 2D phantom was scanned in one central coronal and transversal as well as in five sagittal planes (at $-12.5 \text{cm}$, $-7 \text{cm}$, $0 \text{cm}$, $7 \text{cm}$, $12.5 \text{cm}$ distance from the isocenter) with enabled distortion correction. All measurements were performed at a gantry angle of 0°.

NEMA SNR (National Electrical Manufacturers Association 2001) and ACR measurements are regularly performed on the MR-Linac device. To additionally investigate the influence of susceptibility effects induced by the glass a $B_0$-mapping (dual-echo method) was performed (Schneider and Glover 1991, Webb and Macovski 1991).

### 2.3. Evaluation

#### 2.3.1. Isocentricity and alignment with imaging isocenter

The acquired images were transferred to a personal computer and processed by an in-house developed Matlab (The Mathworks Inc., Natick, USA)-based PG evaluation tool (Mann et al 2017). To determine the position of the irradiation isocenter in axial direction, the beam profile of each individual beam in $y$-direction was investigated. To compensate for the low signal-to-noise ratio (SNR) (figure 2(a)), the profiles were averaged over a 15 mm area along the beam axis. The profiles were averaged separately for the beam entry (figure 2(a) red area) and exit (figure 2(a) blue area) side.

This averaged signal profile was then plotted against the corresponding slice number (see figure 4(c)). To determine $y$-coordinate of the isocenter position, the middle of the full width at half maximum (FWHM) of the averaged profile was determined. The difference of the $y$-positions on the entry and exit sides of the beams may also be used together with the trigonometric relationship $\alpha = \arctan(d_1/d_2)$ (figure 2(b)) to detect a potential inclination of the beam originating from a gantry tilt.

To determine the irradiation center in the transversal plane, the commercial software Mephisto (Version mccc 1.8, PTW, Freiburg, Germany) was used. This software reconstructs the individual beams in each image slice by performing regression over the maximum positions of the lateral profiles along the pre-estimated beam axis. As a result, the so-called isocircle (IC) is determined, defined as the smallest circle touching or intersecting all of the reconstructed beam axes. The radius of this isocircle (ICr) is a quality indicator of the beam alignment for different beam angles. The center of this isocircle is then defined as the radiation isocenter. Comparing this position with the position of the imaging isocenter, reconstructed by means of the wedge-shaped fiducials in the images, the distance between the irradiation and imaging isocenter can be calculated for each individual slice. Together with the beam position in axial direction, the distance of imaging and irradiation isocenter can be determined in 3D. In addition, an SNR analysis was performed using the mean and SD of a region of interest (ROI), which is comparable to the SNRmult-method described elsewhere (Dietrich et al 2007).

#### 2.3.2. Geometric distortions

The automatic detection of the control points was performed using the trainable Weka segmentation (Version 3.3.9.2, University of Waikato, Hamilton, New Zealand) for Image J (Version 1.52h National Institute of Health, Bethesda, USA). First, a classifier with two different classes (figure 3(a)) was created by manually selecting both, control points of the grid and characteristic points of the phantom (e.g. the phantom wall or the fixation rods). After this step, the classifier was able to perform a full segmentation of all control points on the selected images. This was controlled by eye and falsely detected control points were removed manually. To derive the coordinates of each control point, which include between 1 and 9 pixels, a center of mass was determined by measuring the maximum positions of the lateral profiles along the pre-estimated beam axis. As a result, the so-called isocircle (IC) is determined, defined as the smallest circle touching or intersecting all of the reconstructed beam axes. The radius of this isocircle (ICr) is a quality indicator of the beam alignment for different beam angles.

The 2D phantom was scanned in one central coronal and transversal as well as in five sagittal planes (at $-12.5 \text{cm}$, $-7 \text{cm}$, $0 \text{cm}$, $7 \text{cm}$, $12.5 \text{cm}$ distance from the isocenter) with enabled distortion correction. All measurements were performed at a gantry angle of 0°.

#### 3. Results

##### 3.1. Irradiation isocenter accuracy and alignment of irradiation and imaging isocenter

The results of the beam position in axial direction for NSA = 12, 24 or 32 are displayed in table 1 and figure 4. Although no significant differences were found between the measurements with different averages, a lower number of acquisition numbers resulted in larger uncertainties.

The nominal radiation isocenter position in axial direction was located at the interface between slice #10 and #11. The mean distance between the irradiation...
Figure 2. (a) Transversal slice (\#10) of the measurements with 32 signal averages with the visible star shot in the PG and the reconstructed beam axes (yellow lines) by the Mephisto tool. The profiles for the entering and exiting beams are averaged over the red and blue areas, respectively. (b) Schematic representation of a sagittal slice including 4 axial slices (\#9–\#12). The axial position difference of entry and exit area, \(d_1\) and \(d_2\), and the inclination angle \(\alpha\) can be determined.

Figure 3. (a) The two different Weka Segmentation classes are indicated in green (control point) and red (no control point) to train the classifier. (b) Fully segmented image of one representative slice with all grid intersections detected (green). Within the green areas, a center of mass-analysis is performed to determine the final control point position.

Table 1. 1D Distance (mean ± SEM) of the beam center for each beam in axial direction averaged over the profiles \((n = 15)\) in the areas shown in figure 2(a) (for NSA of 12, 24 and 32) relative to the position of the imaging isocenter. Additionally, the average distance over all beams and both areas (entry and exit) ± the total uncertainty (statistical error plus positioning uncertainty) is displayed.

|        | 0° [mm] | 72° [mm] | 144° [mm] | 216° [mm] | 288° [mm] | Mean [mm] |
|--------|---------|----------|-----------|-----------|-----------|-----------|
|        |         |          |           |           |           |           |
| 12 averages |         |          |           |           |           |           |
| Entry   | 0.3 ± 0.2 | 0.5 ± 0.2 | 0.2 ± 0.3 | 0.4 ± 0.3 | 0.6 ± 0.3 | 0.4 ± 0.6 |
| Exit    | 0.6 ± 0.2 | 0.0 ± 0.3 | 1.1 ± 0.3 | −0.4 ± 0.2 | 0.7 ± 0.3 |           |
| 24 averages |         |          |           |           |           |           |
| Entry   | 0.2 ± 0.1 | 0.1 ± 0.2 | 0.9 ± 0.2 | 0.2 ± 0.2 | 0.3 ± 0.2 | 0.3 ± 0.5 |
| Exit    | 0.3 ± 0.2 | 0.7 ± 0.2 | 0.2 ± 0.3 | −0.2 ± 0.2 | 0.7 ± 0.3 |           |
| 32 averages |         |          |           |           |           |           |
| Entry   | 0.2 ± 0.1 | 0.1 ± 0.1 | 0.4 ± 0.1 | −0.1 ± 0.2 | 0.7 ± 0.2 | 0.3 ± 0.5 |
| Exit    | 0.5 ± 0.1 | 0.2 ± 0.2 | 0.7 ± 0.2 | −0.2 ± 0.2 | 0.5 ± 0.2 |           |
and imaging isocenter in axial direction (y-direction) was found to be (0.4 ± 0.3) mm, (0.3 ± 0.2) mm and (0.3 ± 0.2) mm for 12, 24 and 32 NSA, respectively (mean ± standard error of the mean (SEM)).

The isocenter position in the transversal plane was evaluated for the slices closest to the previously determined axial isocenter position and the average distances from the imaging isocenter, (ICd), are displayed in table 2. Table 2 additionally shows the isocircle radius, (ICr). Accounting for a positioning uncertainty of the phantom of ±0.5 mm (one third of a voxel size) in each direction and applying quadratic error propagation results in a 3D shift between irradiation and imaging isocenter of (0.6 ± 0.9) mm, (0.9 ± 0.9) mm, and (0.8 ± 0.9) mm for 12, 24, and 32 NSA, respectively. The comparison of the axial positions determined for the entry and exit part of the beams using the data for the 32 averages (highest SNR) revealed inclinations of (−0.24 ± 0.24)°, (−0.12 ± 0.32)°, (−0.37 ± 0.27)°, (0.19 ± 0.25)° and (−0.24 ± 0.35)° for gantry angles of 0°, 72°, 144°, 216° and 288°, respectively.

3.2. Geometric distortions
The geometric image distortions evaluated for the standard imaging sequence with and without distortion correction for the simultaneous and sequential measurements are shown in figures 5(a) and (b). The tolerances recommended by the manufacturer of <1 mm for all control points within a sphere of 100 mm radius and <2 mm for 90% control points within 175 mm radius around the isocenter, were exceeded by a few points for the setup with the PG-filled glass flask inside of the phantom. With distortion correction, the passing rate was 87.7% for the first criterion and 100% for the second one. The respective overall mean distortion was (0.62 ± 0.32) mm with a maximum distortion of 1.72 mm. Without distortion correction, the passing rates were 59.8% and 68.9%, respectively. The respective overall mean distortion was (1.93 ± 1.61) mm within a 140 mm sphere with a maximum distortion of 7.32 mm.
For the continuously grid inserts with distortion correction the passing rate was 95.0% for the first criterion and 100% for the second one. The respective overall mean distortion was (0.60 ± 0.28) mm with a maximum distortion of 1.6 mm and 99.8% of the control points had distortions below 1.5 mm. Without distortion correction, the passing rates were 69.5% and 77.2%, respectively. The respective overall mean distortion was (1.50 ± 1.56) mm within a 140 mm sphere with a maximum distortion of 4.9 mm.

The measurements with the commercial phantom met both tolerance criteria of the manufacturer (figure 6) for all orientations of the phantom. The overall mean distortion was (0.57 ± 0.25) mm with a maximum distortion of 1.35 mm.

The mean distortions of the T2w-TSE sequence used for PG evaluation within a single plane in the setup with the inserted PG container was (0.59 ± 0.28) mm with maximum distortion of 1.4 mm. 91% and 100% of the control points fulfilled the first and the second tolerance criterion, respectively. The mean distortions of the inner part was determined with the continuous grid structure to (0.55 ± 0.19) mm with maximum distortion of 0.8 mm. Measuring the T2w-TSE sequence with the commercial 2D phantom and evaluation with the ViewRay
software tool revealed mean distortions of \((0.32 \pm 0.23)\) mm and pass rates of 100% for both tolerance criteria (figure 6).

4. Discussion

This study demonstrated the feasibility of a polymer gel-based isocenter alignment measurement in 3D at a 0.35 T MR-Linac. According to the manufacturer’s recommended workflow, this is currently achieved by the following procedure: first, the laser/radiation isocenter coincidence has to be established by separately checking the isocenter position in the \(xy\)- and \(xz\)-plane by a film or ionization chamber array measurements. Then, the lasers have to be adjusted to the measured irradiation isocenter as good as possible, and finally an MR scan of a cylindrical phantom positioned exactly by means of the lasers has to be performed. In contrast, the method presented here is not relying on the laser system as the required accurate positioning of the phantom is achieved directly by the MR-images. In principle, accurate positioning of the phantom is not necessarily required for isocenter accuracy measurements, if the actual irradiation isocenter is visualized directly in the MR-image. However, as the star shot was evaluated in the separate Mephisto-Software rather than the MR-console, where the imaging isocenter is known, the nominal isocenter position in the phantom had to be reconstructed by means of the wedge-fiducials. To assure that this point is actually located at the nominal irradiation isocenter of the MR-Linac, accurate positioning is necessary. This type of nominal irradiation isocenter reconstruction is also a standard procedure when using films.

The other commercially available System (Elekta Unity, Elekta AB, Stockholm, Schweden) uses a different solution to check the isocenter accuracy. As this system is additionally equipped with an electronic portal imaging device (EPID), a phantom with seven \(\text{ZrO}_2\) spheres, which are visible on the EPID, is used (Hanson et al 2019). This method offers a high accuracy, however, due to the lack of an EPID-device, it is not applicable at the MRIdian system.

The gel evaluation was performed with the same device immediately after irradiation employing a T2w-TSE sequence with an isotropic resolution of 1 mm\(^3\) in 3D. In addition, image distortions were measured up to a distance of 140 mm from the isocenter. This can be achieved by a single measurement with a newly developed phantom.

No significant difference for the central position of each of the five individual beams was determined for different NSA (table 1), however the measurement with lower NSA showed a slightly higher uncertainty due to a lower SNR. It is known that the polymerization of the gel continues up to 48 h after irradiation leading to an increased signal. Although this would not improve the SNR itself, an improved contrast-to-noise ratio (CNR) can be expected over this time period. In principle, this offers the possibility to improve the beam center determination. This, however, was not performed in the present study as we aimed to evaluate the measurement directly after the measurement without repositioning the phantom at the MR-Linac.

The accuracy of the irradiation isocenter in the transversal plane (table 2) revealed no significant difference between NSA of 12, 24 and 32 in terms of the radius of the isocircle (IC\(_r\)). In addition, the tolerance limit of 0.5 mm for a star-shot measurement as recommended by the report of Task Group 142 (Klein et al 2009) was met for the average over all evaluated image slices of all NSA. Exceeding this tolerance limit normally indicates a misalignment or instability of the gantry, however, in presence of a magnetic field, an increase of IC\(_r\) is expected as the Lorentz-force systematically deflects the secondary electrons to the same direction with respect to the beam axis. This results in asymmetric beam profiles with a laterally shifted maximum (Raaijmakers et al 2008) leading to an increased IC\(_r\)-value (van Zijp et al 2016, Dorsch et al 2018). Without compensating the effect of the magnetic field, e.g. by using high-density materials (van Zijp et al 2016), an increased isocircle radius is therefore not solely an indication of machine inaccuracies. However, for equiangular distributed beams, the profile shift of each beam induced by the magnetic field is the same and therefore also the center of the isocircle can be expected to be independent of the magnetic field. Thus, this point can be used to define the actual irradiation isocenter. Its distance to the imaging isocenter (IC\(_d\)) is an important alignment parameter and the mean values for IC\(_d\) were \(<1\) mm for all measurements.

Also the 3D-shift between irradiation and imaging isocenter resulting from the combined radial and axial shifts, was \(<1\) mm for all NSA, however with an increased uncertainty. This increased uncertainty is dominated by the image-based positioning uncertainty of the phantom, which is estimated to be \(\sim 0.5\) mm (a third of a voxel size) in each direction.

To determine potential inclinations of the beams relative to the transversal plane, the beams were separated into entry and exit areas (figure 2(a), red and blue boxes). It was found that these areas differ slightly in their axial beam center position, however, the deviations are still within the experimental uncertainties. No significant tilt of the beams against the transversal plane could be detected.

All values shown here (IC\(_a\), IC\(_d\) and inclination angles) could be reproduced in further independent measurements within the uncertainty limits. Furthermore, no significant geometric differences for IC\(_a\) and IC\(_d\) between
a single contrast T2w-TSE and a quantitative T2 acquisition (Dorsch et al 2019) were identified. However, in this study the slice thickness of the single contrast measurement could be reduced from 20 mm to 1 mm, which resulted in a significantly better resolution in the y-direction. Finally, the T2w-TSE used in this work provides seven-fold higher SNR within a comparable acquisition time. However, as the TSE is a research sequence, it is not yet available to the general user of the MRIdian system.

The distortion measurements using the newly developed phantom without distortion correction clearly showed the necessity of the correction. While spatial errors of up to 7.32 mm may occur without correction, they were reduced to values < 1.72 mm within 140 mm and < 1.7 mm within 100 mm distance from the isocenter when the correction was applied. In contrast to the commercial phantom, where the recommendations of ViewRay were met for all phantom orientations, some points exceeded these tolerances when using the developed phantom. This could originate from the fact that the control points in our phantom contained significantly less pixels leading to a larger variability of the positions. While up to 70 pixels were used in the commercial phantom to determine the center of mass of one control point, the grid intersections of our phantom was segmented by only 1–9 pixel. Therefore falsely segmented pixels have a higher impact on the center of mass position. It has to be noted, however, that we measured the distortions in 3D while the commercial phantom provides only 2D distortions. This may also contribute to the slightly larger distortions measured with our phantom. A further advantage of our phantom is that a single measurement is sufficient to measure the distortions in all three directions while the commercial 2D phantom requires 7 independent measurements with different phantom orientations.

In general, the isocenter accuracy measurement presented in this study may also be affected by image distortions. With the phantom equipped with the combination of gel container and grid, distortions within radial distances between 44.5 mm and 140 mm can be measured, covering the size, e.g. of head and neck treatment areas. This disregards the isocenter region, where distortions are expected to be smaller. To verify this assumption, the phantom was also equipped with a continuous grid covering radial distances of up to 140 mm including the isocenter region. It could be shown that the distortions of 85% of the control points in the region of the PG container were below 0.55 mm confirming the underlying assumption of the simultaneous isocenter accuracy and image distortion measurement.

Nevertheless, the distortion may be significant at the position of the wedges used to setup the phantom. However, as the wedges are located within the main planes of the phantom symmetrically to its center and since the wedges are also aligned to the main planes of the MR-Linac symmetrically to the isocenter, the distortions at opposing wedge locations are likely to point in radial but opposite directions. As a result, the effects of distortions at opposing wedge locations are expected to largely compensate each other leading to a small impact on the determination of the isocenter position. For this study this was confirmed by the distortion map. Also B0-inhomogeneities due to susceptibility jumps in the region of the PG were found to be negligible (average in the order of 10 Hz). Solely in the immediate border of the glass flask shifts of up to 60 Hz were detected. The use of either simultaneous of sequential measurement of isocenter accuracy and image distortions with the new phantom may then be used based on the size and long-term stability of the distortions near the isocenter. For larger anatomical regions (e.g. abdomen or pelvis), it is likely that the anatomical structures relevant for registration purpose in adaptive procedures are still located within the presented distance to the isocenter. If this is not the case, distortions have to be checked with larger phantoms.

5. Conclusion

This study investigated the alignment of irradiation and imaging isocenter of a 0.35 T MRI-Linac as well as the spatial distribution of MR-image distortions in a single measurement using a newly developed phantom. The method was evaluated at the MR-unit of the MR-Linac immediately after irradiation using an isotropic spatial image resolution of 1 mm. Isocenter accuracy was found to be (0.6 ± 0.9) mm, (0.9 ± 0.9) mm and (0.8 ± 0.9) mm for 12, 24 and 32 NSA, respectively. After 3D correction, image distortion was significantly reduced and showed a mean distortion of (0.60 ± 0.28) mm and distortions below 1.5 mm for 99.82% of the evaluated control points with a distance of 140 mm. These geometrical uncertainties have to be considered by additional safety margins.

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References

Adamovics J and Maryanski M J 2006 Characterisation of PRESAGE™: a new 3D radiochromic solid polymer dosimeter for use in radiation
Radiat. Prot. Dosimetry 120 107–12
Adamson J, Carroll J, Trager M, Sood S, Kordja J, Maynard E, Hils M, Oldham M and Jirasek A 2019 Delivered dose distribution visualized
directly with onboard kV-CBCT: proof of principle Int. J. Radiat. Oncol. 103 1271–9
American Association of Physicists in Medicine 2010 Acceptance testing and quality assurance procedures for magnetic resonance imaging
facilities AAPM Report No 100, American Association of Physicists in Medicine
American College Of Radiology 2005 Phantom Test Guidance for the ACR MRI Accreditation Program (Reston, VA: American College of
Radiology)
Ballock C, De Deene Y, Doran S, Ibbott G, Jirasek A, Lepegey M, McAuley K B, Oldham M and Schreiner L J 2010 Polymer gel dosimetry
Phys. Med. Biol. 55 R1–63
Baldini L N, Wachowicz K, Thomas S D S, Rivest R and Fallone B G 2007 Characterization, prediction, and correction of geometric
distortion in 3 T MR images Med. Phys. 34 388
Bieri O and Scheffler K 2013 Fundamentals of balanced steady state free precession MRI Mag. Reson. Imaging 38 2–11
Brown S, Venning A, De Deene Y, Vial P, Oliver L, Adamovics J and Baldock C 2008 Radiological properties of the PRESAGE and PAGAT
polymer dosimeters Appl. Radiat. Isot. 66 1970–4
Costa F, Doran S J, Hansen I M, Nill S, Billas I, Shipley D, Duane S, Adamovics J and Oelfke U 2018 Investigating the effect of a magnetic
field on dose distributions at phantom-air interfaces using PRESAGE® 3D dosimeter and Monte Carlo simulations Phys. Med. Biol. 63 05NT01
De Deene Y, De Weger C, Van Duyse B, Derycke S, De Neve W and Achten E 1998 Three- dimensional dosimetry using polymer gel and
magnetic resonance imaging applied to the verification of conformal radiation therapy in head-and-neck cancer Radiother. Oncol. 48 283–91
De Deene Y, Hurley C, Venning A, Vogtke K, Mather M, Healy B J and Baldock C 2002 A basic study of some normoxic polymer gel dosimeters
Phys. Med. Biol. 47 3441–63
De Deene Y, Venning A, Claey S C and De Weger C 2006 The fundamental radiation properties of normoxic polymer gel dosimeters: a
comparison between a methacrylic acid based gel and acrylamide based gels Phys. Med. Biol. 51 653–73
Dietrich O, Raya J G, Reeder S B, Reiser M F and Schoenberg S O 2007 Measurement of signal-to-noise ratios in MR images: influence of
multichannel coils, parallel imaging, and reconstruction filters J. Magn. Reson. Imaging 26 755–85
Doran S J, Charles-Edwards L, Reeder S B, Reiser M F and Schoenberg S O 2007 Characterization, prediction, and correction of geometric
distortion in 3 T MR images Phys. Med. Biol. 50 1343–61
Dorsch S, Mann P, Elter A, Runz A, Kluter S and Karger C P 2019 Polymer gel-based measurements of the isocenter accuracy in an
MR-LINAC. I Phys. Conf. Ser. 1305 012007
Dorsch S, Mann P, Lang C, Haering P, Runz A and Karger C P 2018 Feasibility of polymer gel-based measurements of radiation isocenter
accuracy in magnetic fields Phys. Med. Biol. 63 11NT02
Fallone B G, Murray B, Rathe S, Stanescu T, Steciw S, Vidakovic S, Blosser E and Tymofichuk D 2009 First MR images obtained during
megavoltage photon irradiation from a prototype integrated linac-MR system Med. Phys. 36 2084–8
Ginn J S, Agaryazan N, Gao M, Baharom U, Low D A, Yang Y, Hu P, Lee P and Lamb J M 2017 Characterization of spatial distortion in a
0.35 T MRI-guided radiotherapy system Phys. Med. Biol. 62 4525–40
Green O L et al 2017 First clinical implementation of real-time, real anatomy tracking and radiation beam control Med. Phys. 45 3728–40
Hils M, Auer C, Duzenli C and Jirasek A 2000 Polymer gel dosimetry using x-ray computed tomography: a feasibility study Phys. Med. Biol. 45 2559–71
International Electrotechnical Commission 2011 IEC 61217: Radiotherapy equipment—coordinates, movements and scales Document
Number IEC 61217 Ed 2.0 (International Electrotechnical Commission)
Jaffray D A, Siewerden J H, Wong J W and Martinez A A 2002 Flat-panel cone-beam computed tomography for image-guided radiation
therapy Int. J. Radiat. Oncol. 53 1337–49
Janke A, Zhao H, Cowin G J, Galloway G J and Dodds D M 2004 Use of spherical harmonic deconvolution methods to compensate for
nonlinear gradient effects on MRI images Magn. Reson. Med. 52 115–22
Keall P J, Barton M and Crozier S 2014 The Australian magnetic resonance imaging—linac program Semin. Radiat. Oncol. 24 203–6
Klein E E et al 2009 Task Group 42 report: Quality assurance of medical accelerators Med. Phys. 36 4197–212
Kluter S 2019 Technical design and concept of a 0.35 T MR-Linac Clin. Transl. Radiat. Oncol. 18 98–101
Kontaxis C, Bol G H, Stenkens B, Giltmier M, Prins F M, Kerkenieer J G W, Lagendijk J J W and Raaymakers B W 2017 Towards fast online
intrafraction replanning for free-breathing stereotactic body radiation therapy with the MR-linac Phys. Med. Biol. 62 7233–48
Kubo H D, Hill B C, Moli N, Crijns S P M, Van Santvoort H C, van Vulpen M, Van Den Berg C A T, Reerink O and Meijer G J 2010 Task Group 142 report: Quality assurance of medical accelerators Med. Phys. 36 4197–212
Lee H J, Roed Y, Venkataraman S, Carroll M and Ibbott G S 2017 Investigation of magnetic field effects on the dose—response of 3D
dosimeters for magnetic resonance—image guided radiation therapy applications Radiother. Oncol. 125 426–32
Lutz W, Winston K R and Maleki N 1988 A system for stereotactic radiosurgery with a linear accelerator Int. J. Radiat. Oncol. Biol. Phys. 14 373–81

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Mann P, Witte M, Moser T, Lang C, Runz A, Johnen W, Berger M, Biederer J and Karger C P 2017 3D dosimetric validation of motion compensation concepts in radiotherapy using an anthropomorphic dynamic lung phantom Phys. Med. Biol. 62 573–95
Martinez A A, Yan D, Lockman D, Brabblins D, Kota K, Sharpe M, Jaffray D A, Vicini F and Wong J 2001 Improvement in dose escalation using the process of adaptive radiotherapy combined with three-dimensional conformal or intensity-modulated beams for prostate cancer Int. J. Radiat. Oncol. 50 1226–34
Mutic S and Dempsey J F 2014 The ViewRay system: magnetic resonance-guided and controlled radiotherapy Semin. Radiat. Oncol. 24 196–9
National Electrical Manufacturers Association 2001 Determination of signal-to-noise ratio (SNR) in diagnostic magnetic resonance imaging NEMA Standards Publication MS 1–2001 NEMA
Raaijmakers A J E, Raaymakers B W and Lagendijk J J W 2008 Magnetic-field–induced dose effects in MR-guided radiotherapy systems: dependence on the magnetic field strength Phys. Med. Biol. 53 909–23
Reinsberg S A, Doran S J, Charles-Edwards E M and Leach M O 2005 A complete distortion correction for MR images: II. Rectification of static-field inhomogeneities by similarity-based profile mapping Phys. Med. Biol. 50 2631–61
Reiser M F, Semmler W and Hricak H 2008 Magnetic Resonance Tomography (Berlin: Springer-Verlag)
Schad L R, Ehrlich H-H, Wowra B, Layer G, Engenhart R, Kauczor H-U, Zabel H-J, Brix G and Lorenz W J 1992 Correction of spatial distortion in magnetic resonance angiography for radiosurgical treatment planning of cerebral arteriovenous malformations Magn. Reson. Imaging 10 609–21
Schneider E and Glover G 1991 Rapid in vivo proton shimming Magn. Reson. Med. 18 335–47
Stanescu T, Jans H S, Wachowicz K and Gino Fallone B 2010 Investigation of a 3D system distortion correction method for MR images J. Appl. Clin. Med. Phys. 11 200–16
Tadic T, Jaffray D A and Stanescu T 2014 Harmonic analysis for the characterization and correction of geometric distortion in MRI Med. Phys. 41 112303
Thomas A, Niebanck M, Juang T, Wang Z and Oldham M 2013 A comprehensive investigation of the accuracy and reproducibility of a multitarget single isocenter VMAT radiosurgery technique Med. Phys. 40 121725
Treuer H, Hoevels M, Luyken K, Gierlich A, Kocher M, Muller R-P and Sturm V 2000 On isocentre adjustment and quality control in linear accelerator based radiosurgery with circular collimators and room lasers Phys. Med. Biol. 45 2331–42
van Zijp H M, van Asselen B, Wolthus F J W H, Kok J M G, de Vries J H W, Ishakoglu K, Beld E, Lagendijk J J W and Raaymakers B W 2016 Minimizing the magnetic field effect in MR-linac specific QA-tests: the use of electron dense materials Phys. Med. Biol. 61 N56–9
Vandecasteele J and De Deene Y 2013 On the validity of 3D polymer gel dosimetry: II. Physico-chemical effects Phys. Med. Biol. 58 19–42
Venning a J, Hill B, Brindha S, Healy B J and Baldock C 2005 Investigation of the PAGAT polymer gel dosimeter using magnetic resonance imaging Phys. Med. Biol. 50 3875–88
Wang D, Doddrell D M and Cowin G 2004a A novel phantom and method for comprehensive 3-dimensional measurement and correction of geometric distortion in magnetic resonance imaging Magn. Reson. Imaging 22 529–42
Wang D, Strugnell W, Cowin G, Doddrell D M and Slaughter R 2004b Geometric distortion in clinical MRI systems: Part II: correction using a 3D phantom Magn. Reson. Imaging 22 1223–32
Webb P and Macovski A 1991 Rapid, fully automatic, arbitrary-volume in vivo shimming Magn. Reson. Med. 20 113–22
Yan Y et al 2018 A methodology to investigate the impact of image distortions on the radiation dose when using magnetic resonance images for planning Phys. Med. Biol. 63 085005