Assessment of pelvic imaging for a static inline MRI-Linac

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ABSTRACT
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Purpose: The Australian MRI-Linac\(^1\) utilizes a 1 Tesla split bore magnet in conjunction with a fixed beam linac. To allow multiple beam angles, a patient rotation system is soon to be implemented. There are two sources of geometric inaccuracy in this system (1) B0/system distortion in the MRI magnet, which are worse than a diagnostic system due to the novel open design of the magnet (2) anatomical distortions caused by patient rotation\(^2\). The purpose of this work is to quantify and compare these two sources of geometric inaccuracy.

Materials & Methods: Pelvic (T2-weighted turbo spin echo) MR images were acquired from a healthy volunteer on a 3 Tesla (Magnetom Skyra, Siemens Healthcare, Erlangen, Germany) dedicated radiotherapy MRI system (MR-Sim) and on the Australian MRI Linac (MRL)\(^3\). Scans on the MRI-Sim were acquired in a pelvic radiotherapy treatment position (supine – head to gantry), with a high imaging receiver bandwidth (400 Hz/Px) and vendor supplied correction algorithm to provide a geometric ‘gold’ standard image. The MRL images were acquired both feet first and head first to simulate rotation of a patient to generate two treatment beam angles with a static radiotherapy beam. The images acquired on the MR-Sim were rigidly registered to the MRL image using the MilxView software package\(^3\). MRL images were then deformably registered to the MR-Sim images using the b-spline registration algorithm in niftyReg\(^4\). The two MR images, and the registration deformation are shown in Figure 1. The resultant deformation field was analysed in MATLAB (The MathWorks, Inc., Natick, Massachusetts, United States) to compute the maximum and mean deformations.
To allow the comparison of image related deformation with deformations associated with patient rotation, deformations of the prostate, rectum and bladder were assessed for a different volunteer on the MR-Sim using a bespoke patient rotation system. Images were acquired for couch angles from 0-360 degrees in 45-degree increments. A description of the system and method for contouring and registration of the images have been previously reported\(^2,5\). The workflow is summarized in Figure 2. Deformation was assessed using dice similarity coefficient (DSC), 95% Hausdorff Distance (HD) and centroid position (CP).

Results:

![MR images taken 8 cm inferior to the pelvis (from L-R) MR-Sim, MRL, difference image following the non-rigid registration and overlaid displacement map onto the MR-Sim image non-rigidly registered to the MRL image.](image1)

The maximum distortion in static pelvic imaging was measured to be 13.2 mm. The maximum distortions were caused by some hip rotation and bowel variation between the two images, with average distortions around the patient surface of 6-8 mm. Mean distortion and standard deviation was 3.65 mm and 1.77 mm respectively.

![Figure 2 (a) Raw MR image slice at the 90-degree couch position. (b) The MR image rigidly aligned to the zero-degree position using the couch markings visible on the bottom right. (c) The generated auto-contoured structures overlaid on the synthetic CT image. (d) The deformation vector field following the non-rigid registration.](image2)

For rotating acquisitions, average prostate deformation relative to the 0-degree position was 0.56 ± 0.15, 8.90 ± 5.30 mm and 13.00 ± 7.0 mm for DSC, HD and CP respectively. Average rectum deformations were 0.83 ± 0.03, 2.20 ± 0.84 mm and 2.90 ± 1.60 mm, and average bladder deformations 0.76 ± 0.05, 3.80 ± 1.50 mm and 3.70 ± 1.90 mm.

Conclusions: On average, system related distortions up to 13.2 mm were observed, and rotation related distortions up to 13.00 ± 7.0 mm in prostate centroid position. Both these effects must be corrected for to enable accurate treatment. Methods to correct for both sources of inaccuracy are being developed.

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