Magnetic resonance imaging in the radiation treatment planning of localized prostate cancer using intra-prostatic fiducial markers for computed tomography co-registration

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Abstract

Purpose: To assess the feasibility, and potential implications, of using intra-prostatic fiducial markers, rather than bony landmarks, for the co-registration of computed tomography (CT) and magnetic resonance (MR) images in the radiation treatment planning of localized prostate cancer.

Methods: All men treated with conformal therapy for localized prostate cancer underwent routine pre-treatment insertion of prostatic fiducial markers to assist with gross target volume (GTV) delineation and to identify prostate positioning during therapy. Six of these men were selected for investigation. Phantom MRI measurements were obtained to quantify image distortion, to determine the most suitable gold alloy marker composition, and to identify the spin-echo sequences that optimized both marker identification and the contrast between the prostate and the surrounding tissues. The GTV for each patient was contoured independently by three radiation oncologists on axial planning CT slices, and on axial MRI slices fused to the CT slices by matching the implanted fiducial markers. From each set of contours the scan common volume (SCV), and the scan encompassing volume (SEV), were obtained. The ratio SEV/SCV for a given scan is a measure of inter-observer variation in contouring. For each of the 18 patient–observer combinations the observer common volume (OCV) and the observer encompassing volume (OEV) was obtained. The ratio OEV/OCV for a given patient–observer combination is a measure of the inter-modality variation in contouring. The distance from the treatment planning isocenter to the prostate contours was measured and the discrepancy between the CT- and the MR-defined contour recorded. The discrepancies between the CT- and MR-defined contours of the posterior prostate were recorded in the sagittal plane at 1-cm intervals above and below the isocenter.

Results: Phantom measurements demonstrated trivial image distortion within the required field of view, and an 18K Au/Cu alloy to be the marker composition most suitable for CT–MRI image fusion purposes. Inter-observer variation in prostate contouring was significantly less for MR compared to CT. The mean SEV/SCV ratio was 1.58 (confidence interval (CI): 1.47–1.69) for CT scans and 1.37 (CI: 1.33–1.41) for MR scans (paired t-test; P = 0.036). The overall magnitude of contoured GTV was similar for MR and CT; however, there were spatial discrepancies in contouring between the two modalities. The greatest systematic discrepancy was at the posterior apical prostate border, which was defined 3.6 mm (SD 3.5 mm) more posterior on MR- than CT-defined contouring.

Conclusions: Prostate contouring on MR is associated with less inter-observer variation than on CT. In addition, we have demonstrated the feasibility of using intra-prostatic fiducial markers, rather than bony landmarks, for the co-registration of CT and MR images in the radiation treatment planning of localized prostate cancer. This technique, together with on-line correction of treatment set-up according to the fiducial marker position on electronic portal imaging, may enable a reduction in the planning target volume (PTV) margin needed to account for inter-observer error in target delineation, and for prostate motion.

Keywords: Prostate cancer; Fiducial markers; Computed tomography; Magnetic resonance imaging; Prostate contouring

1. Introduction

Radical radiotherapy for localized prostate cancer relies on the delivery of a tumoricidal radiation dose to the prostate, while limiting the dose received by organs at risk. The adoption of conformal planning techniques has been shown to reduce rectal and bladder toxicity for a given prescribed dose, without reducing treatment efficacy [4]. This has permitted dose escalation, with both improved biochemical control rates and acceptable treatment-related morbidity.
The use of intensity modulated radiotherapy (IMRT) should enable even greater sparing of organs at risk, making further dose escalation a realistic option. As the trend towards more conformal treatment continues, so the accurate localization of the target volume becomes increasingly important.

The clinical target volume (CTV) typically consists of the entire prostate gland which is localized by contours drawn on axial computed tomography (CT) images. The planning target volume (PTV) is then generated by adding a margin to the CTV to account for prostate motion (internal margin) and for spatial uncertainty in dose delivery (set-up margin). Compared with CT, magnetic resonance (MR) imaging can provide better definition of the prostate gland with respect to the surrounding tissues (Fig. 1a,b), and the use of multi-planar reconstruction avoids the problem of partial volume averaging. Using the MR-defined prostate volume as the gold standard, CT has been shown to significantly overestimate the volume of the gland [5,16,19]. Co-registration of MR and CT datasets, matched on fixed bony landmarks, has enabled radiation planning using a MR-defined prostate volume, combined with CT-based electron density information [6]. This MR fusion technique, while reducing the size of the CTV in comparison with conventional CT planning, has no effect on the size of the margin needed to define the PTV. Furthermore, bony boundaries are not well seen on MRI, and the large field of view required for imaging of the bony pelvis may be associated with significant distortion of MR images, making precise image registration difficult [10].

The use of implanted intra-prostatic markers, rather than fixed bony landmarks, for co-registration of MR and CT datasets offers several advantages: First, it would overcome the problem of accurately identifying bony boundaries on MR. Second, the smaller field of view required would reduce system-related image distortion [10]. Third, and most importantly, knowledge of the position of the prostate gland with respect to the implanted markers would enable on-line alteration of treatment set-up, in response to the marker position observed on megavoltage electronic portal imaging [11] (Fig. 2). This would potentially allow a reduction in the internal margin needed to account for prostate motion, thus increasing the potential for target dose escalation while maintaining an acceptable risk of normal tissue complications. We have devised a technique using implanted intra-prostatic markers for the co-registration of CT and MR images of the prostate, and here describe its application to an initial series of patients.

2. Methods

2.1. Phantom studies

Men who receive conformal radiation for localized pros-

![Fig. 1. (a) Axial CT image through mid-prostate. The fiducial marker is well visualized, but the prostate gland is not easily distinguished from the surrounding soft tissues. (b) The corresponding FSE sequence axial MR image from the same patient. The fiducial marker appears as a signal void (arrowed). The prostate outline is clearly demarcated from surrounding tissues. In addition, a dominant intra-prostatic lesion is seen in the right peripheral zone.]

![Fig. 2. Amorphous silicone electronic portal image demonstrating intra-prostatic fiducial marker seeds at the base, midgland and apex of the prostate.]

induced image distortion caused by the

'green' gold seeds were compared with 'red' gold, 'green' gold (silver) and 'white' gold (palladium). Eighteen-karat gold was used in all cases. The MRI-phantom used to assess susceptibility differences between the different types of gold seeds consisted of a 2.54-cm thick 'slab' of Acquafoam pressure-fitted lengthwise inside a 1-l beaker filled with distilled water. This solid, sponge-like material was particularly suited to this application as it became easily saturated with water for the purpose of MR imaging, while at the same time providing a solid matrix into which the seeds could be mounted. The seeds were inserted longitudinally into the saturated foam in the following linear orientation (top to bottom, respectively): two 18-karat reference seeds, 'red' gold, 'green' gold, 'white' gold. Scans were acquired using the fast spin echo imaging sequence (FOV: 18 cm; 16 kHz bandwidth; 256 × 192 matrix; 4 NEX; TR/TE 4700/40 and 40 ms; echo-train length 10; slices thickness/gap 3/0 mm). The positional displacements at the intersections of the grid elements were calculated by measuring the coordinates of each intersection using GE image-analysis tools on the scanner console. MR geometrical distortion was assessed by taking the difference between the MR-image distance and the expected (known) distance for each measurement, and averaging these differences over ten measurements. Three axial slices were done at z = 0 (isocenter) and at +10 and −10 cm from isocenter; for the sagittal and coronal planes, all distances were measured in the central slice (x = 0 and y = 0, respectively).

Finally, potential marker-seed distortion of MR images was assessed using a specially built spherical grid-phantom (eliminating asymmetry-induced susceptibility artifacts). This consisted of an acrylic plastic sphere (7.9 cm o.d.) fitted with a single square lattice plastic grid (16.5 mm on a side per square) in the equatorial plane. Eighteen-karat gold seeds were fixed at several grid intersection points using cyanoacrylic adhesive, with the following orientations: one seed each along the X, Y and Z axes of the grid; one seed each along the XY, XZ and YZ grid planes (at 45°); and one seed pointing in the XYZ direction (i.e. at −45° with respect to the three orthogonal axes). Once assembled, the phantom was filled with a 5 mmol/l copper sulfate solution. MR imaging was again done using the standard prostate imaging sequences (FSE-xl: FOV 24 cm; 20.83 kHz bandwidth; 256 × 192 matrix; 3 NEX; TR/TE 4700/90 and 40 ms; echo-train length 10; slices thickness/gap 3/0 mm; GRE: FOV 24 cm; 31.25 kHz bandwidth; 256 × 192 matrix; 8 NEX; TR/TE 50/4 ms; flip angle 20°; slice thickness/gap 3/0 mm) and

Fig. 3. Effect of marker seed composition on MR appearance in a phantom, using a gradient recalled echo (GRE) sequence (left), and a fast spin echo (FSE) sequence (right). From top to bottom of the image: 18K Au/Ag/Cu (gold); 18K Au/Ag/Cu; 18K Au/Cu (‘red’ gold); 18K Au/Ag (‘green’ gold); 18K Au/Pd (‘white’ gold). Note the ‘halo’ MR-susceptibility effect around the ‘red’ gold seed in the fast spin-echo sequence, as compared to the substantial susceptibility-induced image distortion caused by the ‘white’ gold seed. The enhanced sensitivity of the GRE sequence to metal-tissue susceptibility is evidenced by the markedly increased image distortion in the image of the palladium-containing ‘white’ gold seed at the bottom of the left-hand figure.

Phantom measurements were also conducted to assess the extent of MRI image distortion. The MRI-phantom used in this part of the study consisted of an acrylic resin cylinder (18 cm o.d. × 25 cm long) fitted with a series of 13, square lattice-work plastic grids, and filled with a 5 mmol/l copper sulfate solution. Each 13-mm thick slab consisted of square elements, 14 mm on a side; adjacent slabs were separated by 3-mm thick plastic spacers. MR imaging was done in three planes using the standard prostate imaging sequence (Pulse sequence: FSE-xl; FOV 24 cm; 20.83 kHz bandwidth; 256 × 192 matrix; 3 NEX; TR/TE 4700/90 and 40 ms; echo-train length 10; slices thickness/gap 3/0 mm).
repeated with reduced bandwidths (7.8 kHz for FSE and 2.02 kHz for the GRE). Potential seed-induced positional misregistration/chemical shift artifacts were then evaluated by comparing MR-imaging determined seed positions with their actual geometrical positions with respect to grid vertices.

2.2. Clinical study

The clinical study involved six patients with localized prostate cancer receiving conformal radiation between October 2000 and February 2001. Following marker insertion, patients underwent a planning CT scan and a pelvic MR scan. Prior to each scan (and subsequently before each fraction of radiotherapy) patients were asked to evacuate their bowels, and to have a comfortably full bladder. The planning CT scan was obtained with the patient supine, using helical acquisition of 3-mm thick slices with a table index of 3 mm. Data were acquired from the superior border of the fifth lumbar vertebra to a point 2 cm inferior to the lesser trochanters. A high mAs scanning protocol was used to produce images with enhanced soft tissue contrast. The CTV was defined as the entire prostate gland only, and was contoured independently by three radiation oncologists (CP, AB and CC). The MR scan was performed within two weeks of the CT. When using our routine T2-weighted prostate pulse sequence with a TE of 90 ms and a slice thickness of 3 mm there was inconsistent visualization of the seeds. We attempted scans with a shorter TE of 40 ms and found this improved contrast between the dark seeds and the surrounding prostatic tissue. We also noted that gradient echo imaging, which has more susceptibility than T2-weighted images, allowed for better visualization of the seeds than either the TE 40 ms or TE 90 ms T2-weighted sequences but did not allow for precise definition of the prostatic margins. As a result of these observations our routine protocol for gold seed visualization consisted of a gradient echo pulse sequence (pulse sequence GRE: FOV 24 cm; 31.25 kHz bandwidth; 256×192 matrix; 8 NEX; TR/TE 50/4 ms; flip angle 20°; slice thickness/gap 3/0 mm) to visualize seeds and an intermediate T2 pulse sequence (pulse sequence FSE-xl: FOV 24 cm; 20.83 kHz bandwidth; 256×192 matrix; 3 NEX; TR/TE 4700/40 ms; echo-train length 10; slice thickness/gap 3/0 mm) to visualize prostatic margins and seeds. Both sequences were performed at the same slice locations in the axial plane with the patient in a supine position. All images were obtained using a 1.5 T MRI scanner with a torso phased array coil (GE Medical Systems, Milwaukee, WI).

The CTV was contoured on the axial MR images independently by each of the three oncologists, without reference to the CT scan (Fig. 4a,b). For contouring in both modalities, window settings were optimized at the discretion of the individual observers. The patients’ radiation treatment was based on the CT-defined prostate contours. The MR-defined contours were used for study purposes only.

For each patient the axial MR scan was fused with the CT by matching the intra-prostatic markers. Three pairs of conjugate location points, corresponding to the midpoint of each fiducial marker, were identified (by TH) within the two image sets. These points were then co-registered by the AcQSim multi-modality registration package, using the ‘mark and link’ method, to align the two studies. The degree of misalignment between the three points marked on CT and the corresponding three points marked on MR was calculated in terms of the total conjugate deviation, defined

Fig. 4. The same axial CT (a) and fused MR images (b) as shown in Fig. 1a and b, demonstrating contours defined by three observers at the same level in the prostate, initially on the CT images and subsequently on the MRI images without reference to the CT images. Less agreement about the location of the posterior border of the prostate is seen on the axial CT image. In addition, the postero-lateral CT contours are encroaching on the peripheral zone tumor that is clearly visible on the MR image.
as the square root of the sum of the squares of the deviation between each pair of points. For each fusion, it was ensured that the total conjugate deviation was less than 3 mm, which is consistent with a mean misalignment between scan modalities of less than 1.74 mm for each seed.

The contoured volumes were measured using the CADPLAN treatment planning system. For each scan the following volumes were obtained: the scan common volume (SCV), the largest volume common to the contours of all three observers; and the scan encompassing volume (SEV), the smallest volume encompassing the contours of all three observers. The ratio SEV/SCV for a given scan, is a measure of inter-observer variation in contouring. For each of the 18 patient–observer combinations the following volumes were obtained: the observer common volume (OCV), the largest volume common to the contours drawn on both CT and MR; and the observer encompassing volume (OEV), the smallest volume encompassing the contours drawn on both CT and MR. The ratio OEV/OCV for a given patient–observer combination, is a measure of the inter-modality variation in contouring. In addition, the distance from the treatment planning isocenter to the prostate contours was measured in the superior, inferior, anterior, posterior, and both lateral directions, and the discrepancy between the CT-defined and the MR-defined contour recorded in each case. The discrepancy between the CT- and MR-defined contours of the posterior prostate were also recorded in the sagittal plane at 1-cm intervals above and below the isocenter.

3. Results

3.1. Phantom studies

MRI geometrical distortion using both FSE and GRE sequences was less than 1% in each of the three planes. The average errors were found to be: axial (z = 0, isocenter), error = 0.807 mm (SD ±0.528) (0.5%); axial (z = 10 cm), error = 0.978 mm (SD ±0.359) (0.6%); axial (z = -10 cm), error = 0.802 mm (SD ±0.538) (0.5%); sagittal (x = 0), error = 1.774 mm (SD ±1.3) (0.82%); coronal (y = 0), error = 1.271 mm (SD ±1.22) (0.6%).

No detectable marker-seed induced MR-image distortions/chemical shift artifacts were found with either the FSE or GRE sequences. Decreasing the bandwidth (from 20.83 to 7.8 kHz for FSE and from 31.25 to 2.02 kHz for the GRE) produced a small (≈1 mm) positional misregistration in the X- and Y-oriented fiducial seed FSE images, but (other than the expected overall reduction in image quality) evidenced no changes in the GRE images.

Marker seed composition had a significant effect on seed visualization in the phantom (Fig. 3). On the basis of the phantom study, ‘red’ gold marker seeds, were chosen for the subsequent clinical investigation. An example of the appearance, on both MR and CT, of the marker seeds after insertion into the prostate is shown in Fig. 1.

3.2. Effect of scan modality on contoured prostate volume

Overall, the magnitude of the prostate volume contoured on CT did not differ significantly from that contoured on CT and MRI for six cases, contoured by three observers. The dotted line corresponds to equal volumes for both modalities.

Table 1

<table>
<thead>
<tr>
<th>Direction in relation to isocenter</th>
<th>Mean discrepancy (mm)*</th>
<th>Standard deviation (mm)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Superior</td>
<td>1.6</td>
<td>4.7</td>
</tr>
<tr>
<td>Inferior</td>
<td>-0.2</td>
<td>4.6</td>
</tr>
<tr>
<td>Anterior</td>
<td>-0.7</td>
<td>2.8</td>
</tr>
<tr>
<td>Posterior</td>
<td>1.1</td>
<td>4.9</td>
</tr>
<tr>
<td>Right</td>
<td>0.4</td>
<td>3.1</td>
</tr>
<tr>
<td>Left</td>
<td>2.7</td>
<td>2.3</td>
</tr>
<tr>
<td>Posterior (2 cm above)</td>
<td>-2.8</td>
<td>3.6</td>
</tr>
<tr>
<td>Posterior (1 cm above)</td>
<td>1.7</td>
<td>4.0</td>
</tr>
<tr>
<td>Posterior (1 cm below)</td>
<td>0.7</td>
<td>3.8</td>
</tr>
<tr>
<td>Posterior (2 cm below)</td>
<td>3.6</td>
<td>3.5</td>
</tr>
</tbody>
</table>

* A negative value indicates that the distance from the isocenter to the MR contour was smaller than that to the CT contour in the specified direction.
MR ($P = 0.20$) (Fig. 5). In three patients the mean CT-defined prostate volume was larger than, and in three patients smaller than, the mean MR-defined prostate volume. The mean OEV/OCV ratio was 1.44 (95% confidence interval (CI): 1.37–1.51), 1.44 (CI: 1.31–1.57), and 1.51 (CI: 1.39–1.63), for observers A, B and C, respectively, demonstrating that the magnitude of inter-scan variation was similar for each observer. However, the OEV/OCV ratio is a measure of the magnitude of the difference between contours, and does not take into account the direction of that difference. The mean ratio of the CT-defined prostate volume to the MR defined volume for observers A, B and C was 0.87 (CI: 0.80–0.94), 0.97 (CI: 0.82–1.12) and 1.21 (CI: 1.05–1.37), respectively, demonstrating a statistically significant difference between observers A and C (paired $t$-test, $P = 0.003$). The observer-dependency of the effect of scan modality on contoured prostate volume is illustrated in Fig. 5.

3.3. Spatial discrepancies between contoured prostate volume on CT and MR

The spatial discrepancies between the CT- and MR-defined prostate contours are shown in Table 1. The greatest systematic discrepancy was seen at the apical posterior border of the prostate, as defined by the measurements made in the sagittal plane 2 cm below the isocenter. The posterior border of the MR-defined contour was 3.6 mm (SD 3.5 mm) more posterior than the CT-defined contour at this site. In contrast, the posterior border of the MR-defined contour was 2.8 mm (SD 3.6 mm) more anterior than the CT-defined contour 2 cm above the isocenter. The location of the prostatic apex, as defined by the distance from the isocenter in the inferior direction, did not differ significantly between the two imaging modalities, with a mean discrepancy of 0.2 mm (SD 4.6 mm).

3.4. Effect of scan modality on inter-observer contouring variation

Inter-observer variation in prostate contouring was significantly less for MR rather than CT. The mean SEV/SCV ratio was 1.58 (CI: 1.47–1.69) for CT scans and 1.37 (CI: 1.33–1.41) for MR scans (paired $t$-test; $P = 0.036$).

4. Discussion

We have demonstrated the feasibility of using intra-prostatic markers for co-registration of MR and CT datasets in the radiation treatment planning of localized prostate cancer. We also observed that the volume of the prostate contoured on MR did not differ significantly from that contoured on CT, but that the use of MR was associated with less inter-observer variation.

What are the potential clinical implications? The uncertainty associated with inter-fraction prostate motion is significantly greater than that due to isocenter displacement [22]. In a portal imaging study of 33 men undergoing conformal radiation to the prostate, we previously found that the greatest spatial uncertainty was in the antero-posterior direction, with standard deviations for isocenter setup error of 1.8 mm, and for inter-fraction prostate motion of 5.8 mm. Defining the PTV so as to include points on the edge of the CTV with 95% probability, gives a total margin (setup + internal) of 10.0 mm [1]. Online alteration of treatment setup in response to fiducial marker position seen on electronic portal imaging (Fig. 2) could reduce the uncertainty associated with prostate motion, and hence the internal margin. Some uncertainty will remain due to the effects of intra- and inter-observer variation in prostate contouring [7], intra-fraction prostate movement [12,13,21,22] and any changes in prostate shape or volume during the course of treatment [2,18]. These uncertainties have not been well quantified, but if one assumes that the prostate positional uncertainty may be reduced from a standard deviation of 5.8 mm to, say, 3 mm, then using the method which we have previously described [1] this would give a total margin (setup + internal) of 5.8 mm. Given that there is a volume effect for late rectal morbidity [8], this reduced margin may improve the therapeutic ratio for prostate radiotherapy.

This is the first study to examine the spatial discrepancies between CT and MRI in terms of the definition of the prostate, while controlling for the effect of organ motion between the two scans. Although the prostate contour did not differ in volume between the two modalities, we identified two areas of systematic discrepancy. Considering the MR-defined contour as the gold standard, CT overestimated the posterior margin towards the base and underestimated the posterior margin towards the apex. The discrepancy at the posterior base likely reflects differences in defining the border between the prostate and the seminal vesicles. Rasch et al. identified the same effect in a study of 18 patients contoured by three observers, using chamfer matching on bony landmarks to co-register the two modalities [16]. Using the center of mass of the prostate as a reference point, the distance to the base of the seminal vesicles was 7 mm (SD 6 mm) shorter on MR than CT. The discrepancy in the posterior border towards the prostate apex has not previously been reported, and may be clinically important. Cancer is frequently found at the posterior apex in radical prostatectomy specimens [3] and our findings suggest that conventional CT planning may underestimate the prostate contour at this site, increasing the possibility of a geographic miss.

The finding that the volume of the prostate did not differ significantly between CT and MR contrasts with earlier reports in which the CT-defined prostate was found to be larger than the MR-defined prostate by a mean ratio of 1.2 to 1.4 [5,9,16,17,19]. Several factors may explain this difference. First, contouring on CT in the current study was done in the knowledge of these previous reports. Second, the CT contouring was done with the aid of reconstructed images in
the sagittal and coronal planes, which help to distinguish prostate from surrounding soft tissues. Third, we routinely measure the distance from the apical fiducial marker to the prostatic apex at TRUS, and this information is used to determine the inferior extent of the prostate.

The evidence that CT overestimated the actual volume of the prostate has previously been considered one argument for the use of MR imaging for defining the prostate for radiation planning purposes. Debois et al., in a study of ten patients, analyzed the difference between CT and MRI, both in terms of the contoured prostate volume and in terms of the rectal dose volume histogram for the resulting treatment plans [5]. They found that the prostate volume on MR was significantly smaller than the CT prostate volume in nine of the ten cases. Using a 10-mm margin around the prostate to define the PTV, and a three-field plan with an anterior and two wedged lateral fields, they found that the rectal volume receiving greater than 80% of the prescribed dose was reduced by a mean of 24% using the MR, rather than the CT, data. If, as we have shown, the prostate volume does not differ significantly between CT and MR, it is likely that this dosimetric advantage for MR will be lost. However, MR offers several other advantages over CT for target volume definition. First, it has been demonstrated [5,6], and our results confirm, that there is a significant difference between modalities in favor of MR, in terms of the inter-observer variation in prostate contouring. Second, MR has the potential to identify dominant intraprostatic lesions (Fig. 1b), which may be planned to receive boost treatment. For example, Pickett et al. have described the use of IMRT to treat the whole prostate to 70 Gy and a dominant intraprostatic lesion, defined by a combination of MR imaging and MR spectroscopy, to 90 Gy [14].

The current study, while demonstrating the feasibility of using intra-prostatic markers for co-registration of CT and MR, has several limitations. The accuracy of image co-registration depends on the degree of uncertainty associated with the identification of the center of each fiducial marker. We have not assessed the inter-observer and intra-observer variation for seed-center identification. At present the development of an automated process for marker localization is limited by the poor definition of the markers on MR. Further modification of marker specification or MR protocol may enable automated marker localization and image fusion, with improved accuracy.

Although we have shown that system-related image distortion is within acceptable limits, the use of fiducial markers for co-registration purposes raises the possibility of marker-induced distortion of the MR images. Object-induced distortions can be marked at the boundaries of structures of different intrinsic magnetic susceptibilities, such as at air/tissue interfaces [10]. In vivo measurements were beyond the scope of the current investigation, and this issue warrants further study. If seed-induced image distortion is a clinically significant problem, it could be corrected by the use of a gradient reversal MR protocol [20].

5. Conclusions

Our results confirm that contouring the prostate on MR, rather than CT, is associated with less inter-observer variation. Furthermore, our results suggest that the use of intra-prostatic fiducial markers, rather than bony landmarks, may be an effective method for the co-registration of CT and MR images in the radiation treatment planning of localized prostate cancer. This technique, together with on-line correction of treatment set-up according to the fiducial marker position on electronic portal imaging, might enable a reduction in the planning target volume (PTV) margin needed to account for prostate motion and for error in target delineation, and may improve the therapeutic ratio of radical radiotherapy for localized prostate cancer. This hypothesis should be tested with a larger scale clinical trial.

References


