TOPICAL REVIEW

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MRI-only treatment planning: benefits and challenges

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Abstract

Over the past decade, the application of magnetic resonance imaging (MRI) has increased, and there is growing evidence to suggest that improvements in the accuracy of target delineation in MRI-guided radiation therapy may improve clinical outcomes in a variety of cancer types. However, some considerations should be recognized including patient motion during image acquisition and geometric accuracy of images. Moreover, MR-compatible immobilization devices need to be used when acquiring images in the treatment position while minimizing patient motion during the scan time. Finally, synthetic CT images (i.e. electron density maps) and digitally reconstructed radiograph images should be generated from MRI images for dose calculation and image guidance prior to treatment. A short review of the concepts and techniques that have been developed for implementation of MRI-only workflows in radiation therapy is provided in this document.

1. Introduction

Treatment planning in modern radiation therapy procedures involves the use of both computed tomography (CT) and magnetic resonance imaging (MRI) for patients in many disease sites, with the former providing electron density values that are necessary for treatment planning, and the latter providing superior soft tissue contrast and for tumor and soft tissue delineation. Examples of soft tissue contrast superiority of MRI in comparison with CT for brain, prostate and cervical cancer are shown in figures 1–3.

Delineating the gross tumor volume (GTV) has been called ‘the weakest link’ in the chain of factors affecting radiotherapy accuracy (Njeh 2008). Current computed tomography (CT) variability in GTV delineation introduces more error than daily setup uncertainties (Weiss and Hess 2003, Weiss et al 2003, Van Herk 2004, Rasch et al 2005, Vorwerk et al 2009, van Mourik et al 2010). In an era of intensity modulated radiation therapy, where steep dose gradients sculpt dose away from organs at risk (OARs), accurate delineation becomes of paramount importance to avoid geometric misses and prevent recurrences. Importantly, no level of on-board image guidance will eliminate these systematic delineation errors (Njeh 2008). The consequences of these systematic uncertainties can be great; inadequate target coverage has been linked to significant reductions in tumor control (Kim et al 2014) and clear patterns of failure (Chen et al 2011).

Importantly, incorporating MRI in treatment planning significantly reduces inter- and intra-observer contouring variability for many disease sites (Rasch et al 1999, 2005, Jolicoeur et al 2011, Giezen et al 2012). In the brain, MRI can resolve tumor boundaries not resolvable on CT (Just et al 1991) and identifies peritumoral edema (Chang et al 2007). For prostate, MRI is extremely beneficial for accurately identifying the prostate, areas of high tumor burden, sensitive erectile tissues, and the prostatic apex, which cannot be identified on CT as suggested by ACR Appropriateness Criteria (Debois et al 1999, Rasch et al 1999, Wachter et al 2002, Hentschel et al 2011, Nguyen et al 2014). Accurate delineation of this region is critical: high tumor incidence occurs in the apical posterior region (Chen et al 1997). For female pelvis, RTOG consensus atlas states that MRI provides precise delineation of the uterus and cervix and identifies the superior/inferior bladder extent (Gay et al 2012). GEC ESTRO guidelines conclude that MRI provides the most reliable delineation for gynecological cancer (Pötter et al 2006).
The benefits of using MRI for delineation include improved dosimetry and the potential to increase the therapeutic ratio. Prostate delineation on MRI has enabled dose escalation of 2–7 Gy while maintaining the same rectal wall dose (Steenbakkers et al 2003). Likewise, an MRI-assisted dose volume escalation study for cervical cancer revealed ~10–20% survival gains while reducing gastrointestinal and urinary late morbidity (Pötter et al 2007).

Figure 1. Axial brain images of patient with a metastatic tumor in the brain. (A) and (C) CT image. No contrast between the tumor and the surrounding normal tissue. (B) and (D) T2-weighted fluid-attenuated inversion recovery image (FLAIR) MR image. Higher soft tissue contrast of the MR image leads to more accurate delineation of the tumor.

Figure 2. Comparison of the transverse view of CT (A) and T2-weighted (B) images of a patient with prostate cancer. The volumes are as follows: prostate (magenta) and dominant intraprostatic lesion (cyan).
Furthermore, MRI is a multi-parametric imaging modality that not only can provide anatomical information with high soft-tissue contrast, but also can provide valuable functional information that can be used for assessment of disease progression and treatment response evaluation (Khoo and Joon 2006, Maikusa et al 2013). In brain, functional diffusion-weighted MRI images can be used to reliably evaluate treatment response at various time points during and post-treatment (Mardor et al 2003, Hamstra et al 2008). Functional information of surrounding normal structures can also be considered during treatment planning to improve target coverage while minimizing the dose to the adjacent functioning tissues (Garcia-Alvarez et al 2006, Kovacs et al 2011). Likewise in cervix, diffusion-weighted imaging has been considered as a potential tool for monitoring treatment response (McVeigh et al 2008, Levy et al 2011). In prostate, dynamic contrast enhanced (DCE) MRI scans can be used for detection and localization of recurrent prostate cancer after radiotherapy and such functional information can be helpful for planning of potential salvage treatment (Haider et al 2008). By incorporating functional information acquired from MRI images, patients’ quality of life has improved via minimizing the dose to the surrounding nerves, vessels and other normal structures without compromising target coverage (McLaughlin et al 2005).

Currently, the existing CT-based treatment planning workflow relies on target and OAR definition on MRI and a transfer of contours to CT via image registration. MRI-CT co-registration introduces geometrical uncertainties of ~2 mm for the brain (van Herk and Kooy 1994, Ulin et al 2010) and 2–3 mm for prostate and gynecological patients (Wang and Doddrell 2005). Importantly, these errors are systematic, persist throughout treatment, shift high dose regions away from the target (Van Herk 2004) and could lead to a geometric miss that compromises tumor control. Recently, MRI-simulation platforms have emerged as attractive alternatives to CT-simulation (Devic 2012, Kapanen et al 2013, Paulson et al 2015, Glide-Hurst et al 2015b). These differ from diagnostic MRI by including larger bore size, flat tabletops to accommodate immobilization devices, external laser systems, and dedicated imaging protocols. By acquiring MRI-simulation data in the treatment position, combined with modern low distortion techniques an accurate MRI-based anatomical patient model can be generated that minimizes the variation in patient positioning between the time of simulation and the time of treatment (Devic 2012). This capability has recently led to the concept of MRI-only based treatment planning, where artificial or synthetic CT data for dose calculation is generated directly from the MRI scan. Many groups have shown a strong interest to move toward MRI-only treatment planning (Doemer et al 2015, Glide-Hurst et al 2015b, Kim et al 2015a, Price et al 2015, Zheng et al 2015, McGee et al 2016, Kapanen et al 2013, Pöttér et al 2007, Hsu et al 2013, Hsu et al 2015, Dowling et al 2012, Lambert et al 2011, Rivest-Hénault et al 2015).

MRI-only treatment planning will reduce CT scanning (reducing radiation dose, patient time, and imaging costs), streamline clinical efficiency, and will fully exploit the benefits of MRI for high-precision treatment planning. It will enable more efficient uses of resources and a reduction in duplicated effort (i.e. between diagnostic radiology and radiation oncology). Importantly, MRI-only planning removes systematic CT-MRI registration uncertainties to facilitate improved geometric treatment accuracy. However there are several challenges to be
overcome to introduce MRI-only planning into the clinic. These include the production of robust MRI-only patient models and synthetic CT scans with accurate geometry and electron densities.

The purpose of this article is to review the current state-of-the-art, including potential benefits and challenges remaining for MRI only treatment planning for external beam radiotherapy. We will also highlight unmet needs and future directions.

Artifacts and geometric distortions

Geometric distortions in MRI consist of two major components: system-level (arising from gradient nonlinearity (GNL) in the spatial encoding gradients (Chen et al 2004b, Baldwin et al 2007) and B0 field inhomogeneities) and patient-level (chemical shift artifacts and susceptibility) (Wang et al 2013). System-level distortions are magnetc-specific and not sequence or object dependent. Currently, GNL distortion corrections are built into the MRI reconstruction software and have been shown to be the dominant source of geometric distortion (Baldwin et al 2007). In a perfect situation the main magnetic field is uniform and magnetic fields from gradients are linear; however, in reality the gradients are not perfect and gradient fields are non-linear. This non-linearity becomes more noticeable away from the scanner isocenter toward the edges of the scanner maximum field of view (Doran et al 2005, Wang and Doddrell 2005, Price et al 2015). Thus, one approach to reducing the impact of GNL is to localize the object of interest as close to isocenter as possible. Geometric distortion not only occurs in the phase and readout encoding direction, it can also occur in the slice selection direction where the slice thicknesses may change as a function of position. To reduce the effect of GNL, one can increase the gradient amplitude, but at the same time the bandwidth needs also to be increased. However, increasing the radiofrequency (RF) receiver bandwidth reduces the signal to noise ratio (SNR) and increasing the RF transmitter bandwidth will increase the transmitting RF power and may go over the allowable specific absorption ratio (SAR). Having images with acceptable quality and high geometric integrity is necessary for radiotherapy treatment planning (Jovicich et al 2006, Tavares et al 2014) and therefore, all necessary steps need to be taken to minimize potential distortions in MRI images.

Another source of image distortion may arise from eddy currents that are generated by rapidly pulsed gradients. According to the Faraday–Lenz Law of electromagnetism, changing magnetic field induce electrical currents in nearby conductors. Since MRI uses rapidly changing gradient magnetic fields, eddy currents are always produced; however, the magnitude of eddy currents depends on the rate of change of the magnetic field. Therefore, fast imaging sequences such as echo-planar imaging, diffusion-weighted imaging and MR spectroscopy produce the largest and most severe eddy current problems. In earlier generation magnets with unshielded gradients, distortions up to 1.3 mm have been reported in a 1.5 T cylindrical bore magnet (Tanner et al 2000). More modern hardware and shielded gradients have been shown to compensate for eddy currents, with distortion differences of <0.2 mm over several echo time settings reported for a 1.0 T Open MR-SIM (Price et al 2015) and <0.3 mm for a 3.0 T cylindrical magnet (Baldwin et al 2007).

Another method to reduce geometric distortion is to select pulse sequences with appropriate parameters. Sequences with fast gradient switching are more prone to gradient distortion and using them should be considered with caution. The other effective way of reducing system-related geometric distortion that have been described in literature (Caramanos et al 2010, Janke et al 2004, Maikusa et al 2013, Chen et al 2004b) is to apply distortion correction matrices before the final image is generated. This technique can be implemented once the distortion map of the MRI machine is characterized and it is currently included in the reconstruction for many MRI systems. It is important for the end-user to quantitize the residual GNL (i.e. after vendor corrections) to determine if they are negligible, and if not, to understand their magnitude and location. If residual GNL after 3D distortion corrections is non-negligible, additional post-processing corrections can be implemented (Price et al 2015). Closed bore magnets have shown clinically acceptable GNL characterized within the clinically usable field of view (FOV) (Torfeh et al 2016, Huang et al 2016). Thus, GNL needs to be measured for each magnet platform to deduce if further corrections are necessary before MRI-only radiotherapy is implemented. This can be done by using known test objects and phantoms with known landmarks, typically at a large field of view that at least encompass the clinical scanning volumes (Caramanos et al 2010, Doran et al 2005, Sun et al 2015, Chen et al 2004b, Huang et al 2016, Price et al 2017). MR images can then be compared to CT or a schematic of the expected phantom configuration as the gold standard.

Metal artifacts are one of the most common types of artifact in MRI images. These are due to susceptibility-related inhomogeneities in which metal-tissue or air-tissue interface in the presence of strong polarizing (B0) magnetic field will lead to strong susceptibility transitions which will be the source of large magnetic field distortions. Even though many implants are MRI safe, the artifacts induced by metal implants may distort the geometric integrity of the image and alter intensity values of the tissue voxels around the implant (Schenck 1996, Hargreaves et al 2011, Schmidt and Payne 2015, McGee et al 2016). For example metal artifact management is important for MRI-guided brachytherapy in which metallic objects, such as cervical applicators and titanium needles are used, which may adversely impact image quality and clinical usability (Tanderup et al 2008, 2013, 2014, Hellebust et al 2010, Kirisits et al 2014).
Some metal artefact reduction techniques are available although they may have limitations for radiation therapy purposes (Butts et al 2005, Lu et al 2011, Choi et al 2015, Reichert et al 2015). For example, some of these artefact reduction techniques have been developed for two-dimensional (2D) MRI images, whereas three-dimensional (3D) acquisition of MRI images and their use in radiotherapy planning are becoming more prevalent. Recently, advanced reconstruction methods for metal artifact reduction have been introduced that combine view-angle tilting (VAT) (Butts et al 2005, Choi et al 2015) and slice-encoding metal artifact correction (SEMAC) (Lu et al 2009). Other strategies include using spin-echo based pulse sequences instead of gradient echoes and increasing the receiver and excitation bandwidth. Since metal induced inhomogeneity of magnetic fields is much larger than tissue based inhomogeneity (Hargreaves et al 2011), these strategies may not completely eliminate artifacts. Some of the techniques applied by different vendors include imaging with high gradients and increased encoding to compensate the artefact in both in-plane and through-plane directions (Koch et al 2009, Lu et al 2009, Koch et al 2011). Figure 4 illustrates a case where a metal artifact correction technique was implemented in a patient with bilateral hip implants for both CT-simulation and MRI-simulation, highlighting the potential of mitigating metal artifacts while enabling the powerful soft tissue contrast of MRI to be utilized.

Patient-level distortions \( (B_0) \) are object and field-strength dependent, requiring patient-specific corrections. Effects resulting from susceptibility differences are most apparent near tissue/air interfaces due to local variations in the induced magnetic field and have been reported to be up to 4 mm at the sinus/tissue interface in the brain at 3.0 T (Wang et al 2013). To measure these distortions, field maps, or a map of the off-resonance frequency at each voxel, are obtained. These are typically performed with a double-echo gradient echo based sequence and calculating the field map based on the difference in phase between two different echoes. Recently, Wang et al performed repeat acquisition of field maps for 17 brain subjects and found a within-subject standard deviation of \( \sim 0.2 \) mm displacement in the frequency-encoding direction of 3D T1-weighted images (Wang et al 2013). Recent work by Tyagi et al evaluated patient-induced susceptibility distortion in the pelvis in 20 patients acquired at 3.0 T and measured the mean distortion within the prostate for a single time point as \( -0.2 \) mm (range: \( -0.62 \)– \( 0.35 \) mm) while the voxels within the body contour ranged from \( -0.73 \) to 0.56 mm (Tyagi et al 2017). While the overall magnitude of patient-specific distortions appears to be low, it can be further decreased by increasing the bandwidth. This suggests that anatomical site-specific recommendations may be advantageous. Overall, robust quantification and mitigation (either by increasing the bandwidth to minimize patient-specific distortions or developing a post-processing correction for system-level distortions), particularly for high precision MRI-only treatment planning, is imperative.

Production of synthetic CT

Unlike CT, electron densities of different tissues are not uniquely related to the image intensities in MRI. Therefore, direct mapping of MRI intensity to electron density using calibration phantoms cannot be used. However, several methods have been developed to estimate Hounsfield Units (HU) and hence electron density based on the intensity of MRI images. The estimation of HU is necessary as these are currently the input variable accepted by treatment planning systems, although in future systems electron density or tissue class/material may be more directly used. The HU map from MRI can therefore currently be considered as a scan from a ‘virtual’ CT scanner. These mapping methods can be used for radiation therapy purposes and have also been developed for positron emission tomography (PET)-MRI scanner attenuation correction algorithms (Zaidi et al 2003, Martinez-Möller and Nekolla 2012). Various terminologies have been used for the resulting synthetic CT including substitute CT, pseudo-CT, MRCAT and MRCT. The methods to generate synthetic CT images from MRI scans are classified here for clarity into: voxel based methods; atlas based methods; and hybrid methods. Systematic overviews of method types have recently been reported (Edmund and Nyholm 2017, Johnstone et al 2017).
Voxel based methods use classification or calibration type approaches to determine the HU values from MRI data. These use a voxel by voxel mapping based on the intensity and/or spatial location of the MRI image voxel or combinations of intensities from different sequences.

Classification techniques separate MRI into discrete tissue classes and assign a bulk HU value to a voxel identified to belong to a particular class or a weighted average of HU values according to class probabilities. The simplest approach is to assume the patient is water-equivalent and assign a single density such as is done for conventional brachytherapy planning. This was introduced for MRI-only dose calculations for brain initially (Beavis et al 1998) and then for prostate (Chen et al 2004, 2004b). However it has been found that dose calculations are not clinically acceptable compared to heterogeneous density (Eilertsen et al 2008, Karotki et al 2011). The addition of a bone class improves dose calculation to mostly within 2% of the dose calculations on CT (Eilertsen et al 2008, Karlsson et al 2009, Jonsson et al 2010, Karotki et al 2011, Kim et al 2015, Paradis et al 2015), falling within a range that has since been suggested as clinically acceptable (Korsholm et al 2014). For anatomical sites where air cavities are present the assignment of an air class will be required (Hsu et al 2015). Separation of bone and air has been performed with UTE pulse sequences (Robson et al 2003, Catana et al 2010, Keereman et al 2010, Johansson et al 2011, Johansson et al 2012, Edmund et al 2014). Since bone has a very short T2*, using UTE pulse sequences can improve the contrast between bone and surrounding air or soft tissue. Fat and water classes can also be separated and segmented in MRI images acquired with Dixon pulse sequences, however whether this improves dose calculation accuracy significantly is unclear. Dixon uses the chemical shift difference between water and fat to separate the signals. Solutions in the pelvis have been proposed for combinations of T1-weighted, T2-weighted, and balanced turbo field echo (BTFE) (Kim et al 2015c) or T1-weighted and T2-weighted imaging (Kim et al 2015a). For the pelvis, image voxels were sorted into five material classifica-

In atlas-based methods, both single and multi-atlas techniques have been developed. The single atlas represents an average patient anatomy and is registered to the acquired MRI images using deformable image registration to produce an estimation of the HU. A CT atlas can be used however this requires CT to MRI registration where the image signals are very different (Burgos et al 2013, Uh et al 2014). To overcome this problem a CT-MRI atlas-pair can be used where the CT and MRI atlas scan pair correspond anatomically (Dowling et al 2012). The MRI atlas scan is deformably registered to the acquired MRI image to derive deformation vectors that are then used to deform the conjugate CT atlas scan to produce the synthetic CT. The outcome of this method is dependent on how well the image registration can be performed which will depend on the differences in size and shape between the atlas and patient scans. A multi-atlas technique can similarly be performed to overcome the problem of variations in patient size and shape and the most similar atlas scan selected based on registration metrics. Applying image intensity and uniformity correction on the MRI images prior to CT-MRI image registration has been shown to improve image registration accuracy (Burgos et al 2013). It has also been shown that increasing the number CT cases that has been used to generate the atlas will improve the performance of atlas based registra-
The time to generate a synthetic CT using a multi-atlas approach has been quoted as at least 20 min using a Matlab with MEX code implementation of their algorithm (Farjam et al. 2017).

Hybrid techniques combine atlas methods and voxel based approaches. A combination of atlas based deformable registration and local patch pattern recognition was proposed for brain MRI based attenuation correction (Hofmann et al. 2008). An atlas database of MRI and CT scans was used with each MRI scan deformedly registered to the acquired MRI scan. For a voxel in the acquired MRI the neighboring patches in the database scans are then found. A gaussian distributed predictive distribution for the voxel HU value is derived from the differences in patch intensities and positions to the voxel combined with the mean atlas CT value. A study examined atlas based techniques for brain compared single atlas, multi-atlas and multi-atlas followed by pattern recognition using gaussian process regression. They found that multi-atlas performed better than single however the gaussian regression did not improve over the use of a mean CT value from the atlas (Uh et al. 2014).

More recently a patch-based method has also been applied, where cubic patches of MRI images are compared to patches in an atlas database of co-registered MRI/CT scan pairs following affine registration. The most similar patches from a local neighborhood search of the atlas scans are used to produce the HU value (Andreasen et al. 2016). For generation of prostate synthetic CT a multi-atlas registration to the acquired MRI from a large MRI/CT database has been employed. This was followed by voxel based weighting of atlas HU values according to the acquired voxel MRI similarities to the registered atlas voxels (Dowling et al. 2015). The applicability of this method to 1.5 T has recently been demonstrated (Wyatt et al. 2017). A similar approach termed ‘statistical decomposition algorithm’ uses atlas scans that are deformedly registered to the acquired MRI with a first registration used to drive segmentation of tissue classes on the acquired MRI and then a second structure-guided registration using the segmentations. HU values are then assigned by weighting the atlas HU values according to MRI similarities to the corresponding MRI atlas voxels (Siversson et al. 2015). A multi-atlas method for head and neck synthetic CT has been developed that registers the atlas MRI scans (12 patients) to the target MRI and uses a generalized registration error (GRE) metric. The final synthetic CT value at each point is a non-linear GRE-weighted average of the atlas CTs (Farjam et al. 2017). A method that combines regression based assignment of HU to soft tissue classes and atlas based bone HU assignment has been developed. The bone in the target MRI scan is segmented and the most similar bone in the atlas is deformedly registered to the MRI and combined with the tissue-specific HU maps to generate the synthetic CT (Ghose et al. 2017b). Recently a method for brain used a probability density function to estimate HU from the MRI signal value for the acquired voxel in T1 and T2 weighted images along with the voxel location in a reference anatomy derived from deformable registration to an atlas scan (Ren et al. 2017). Synthetic CT for cervix and lung have also been developed by a combination of atlas based bone registration and soft-tissue classification (Dowling et al. 2012, Edmund et al. 2014, Dowling et al. 2015, Rivest-Hénault et al. 2015, Kim et al. 2015c, Zheng et al. 2015, Andreasen et al. 2016, Liu et al. 2017, Ren et al. 2017, Wang et al. 2017).

To date, two clinically released MR-only packages are clinically available for prostate cancer. One solution, Philips MR-CAT, is FDA-approved and integrated inline with the MRI reconstruction software to generate synthetic CT images immediately after the acquired MRI images have been reconstructed. This software employs a dual echo 3D mDIXON fast field echo sequence to generate synthetic CT using assigned bulk HU values for air, adipose, water, trabecular/spongy bone and compact/cortical bone (Tyagi et al. 2016). Another commercially available product, Spectronic’s MriPlanner, is regulatory approved (CE-marked) and requires end-users to upload a T2-weighted dataset for generation of synthetic CT. A statistical decomposition algorithm (SDA) is used as described above (Siversson et al. 2015). Examples of preclinical and clinical implementations of synthetic CT for brain, prostate and female pelvis are provided below.

**MRI-only planning: brain**

One of the challenges with MRI-only planning of the brain is the presence of small, intricate air cavities and thin bones. Because bone has very short T2*, UTE pulse sequences may be employed to improve contrast between bone and surrounding air or soft tissue (Johansson et al. 2011, Robson et al. 2003). One such solution has been implemented at Henry Ford Health System, where a combined UTE-mDIXON sequence (TE1/TE2/TE3 = 0.144/3.4/6.9 ms) was implemented on a 1.0 T MR simulator (Zheng et al. 2015). A hybrid MRI phase/magnitude UTE image processing pipeline was developed consisting of two major workflows: (1) generation of a bone-enhanced image that significantly improved bone and air contrast in MRI and (2) segmenting air regions of interest from UTE phase data combined with Gaussian mixture modeling (GMM). A previously developed synthetic CT pipeline for the pelvis was modified by incorporating derived bone-enhanced images and air masks into the workflow with bone-enhanced, FLAIR, and UTE images. Images were semi-automatically segmented into five categories (air, bone, fat, brain matter, and CSF) using a 5-kernel GMM before generating synthetic CTs using a region-specific, voxel-based weighted summation method described previously. Overall, results agree well with clinical CTVs for treatment planning with mean absolute errors (MAE) between synthetic CT and CT-SIM of 147.5 ± 8.3 HU which was consistent with literature. The MAE in the brain tends to be higher than that...
in the pelvis because of challenging segmentation yielding larger errors near bone-air interfaces. A voxel-based comparison in the brain by Johansson et al yielded an average MAE of 137 HU (Johansson et al 2011). Similarly, atlas-based methods implemented in the brain have reported a wide range of MAE values, with differences of up to 600 HU found in bone (Demol et al 2016). Typical synthetic CT results are shown in figure 5 for a post-surgical subject and a corresponding radiosurgery plan (18 Gy, 1 fraction, 8.8 cc planning target volume) calculated in Eclipse TPS using the anisotropic analytical algorithm (Varian, Palo Alto, CA) on both reference datasets, illustrating excellent dosimetric agreement between plans (negligible difference in target volume coverage) despite the lesion being situated near the bone and sinuses.

For an atlas-based approach, DVH differences between an atlas-based synCT and CT were ~3% except for cases where the tumors were located within the sphenoid bone and dose differences were observed up to 5–7% (Demol et al 2016). Recent work by Paradis et al evaluated VMAT treatment plans in a 12 patient cohort and found no significant differences between calculated doses and planning constraints (OARs had an average D(max) differences of 0.0 Gy (−2.2 to 1.9 Gy) (Paradis et al 2015).

MRI-only planning: prostate

A site that has received considerable attention for MRI-only workflows due to the large patient numbers is prostate. Figure 6 shows an example of a synthetic CT generated for prostate using a voxel based calibration method (Korhonen et al 2014). This method has been implemented clinically at Helsinki University Central Hospital with nearly 400 prostate cancer patients treated with the MRI-only workflow since 2012. Bones are auto-segmented using at atlas based algorithm and separate mappings of MRI signal intensity to HU number performed for within bone and outside bone voxels. Recently the method has been generalized to other institutions and scanner types with similar results (Koivula et al 2017).
MRI-only planning: female pelvis

The development of synthetic CT for female pelvis has been more limited than for male, although preliminary results are emerging, often consisting of incorporating a bone shape model built from CT data (Liu et al 2015, Liu et al 2017). Volumetric modulated arc therapy (VMAT) plans between synthetic CT and CT were found to have similar dosimetric agreement. Recent work has been performed translating male pelvis solutions to female anatomy for a voxel-based weighted summation technique (Kim et al 2017). Overall, synthetic CT weighting produced small changes for MAE and calculated dose distributions, suggesting that male pelvis weights are good approximations of female data. However, 3D treatment plans were found to be slightly more sensitive than VMAT patients, likely due to the attenuation through the femoral bones and need for more robust bone solutions.

Dose calculation accuracy in synthetic CT images

Methods of generating synthetic images need to be evaluated for HU accuracy and by comparing the calculated dose distribution in the generated synthetic CT image set and corresponding registered CT image set. Dose calculation around tissue boundaries where there is electron density alteration can be challenging and therefore, accurate representation of tissue inhomogeneity in synthetic CT images is important. It has been shown in previous studies that accurate representation of tissue inhomogeneity in synthetic CT images can improve dose calculation accuracy (Lee et al 2003, Chen et al 2004a, Pasquier et al 2006, Stanescu et al 2006, Eilertsen et al 2008, Jonsson et al 2010, Greer et al 2011, Johansson et al 2011, 2012, 2013, Karotki et al 2011, Dowling et al 2012, Kapanen and Tenhunen 2013, Nyholm and Jonsson 2014, Yu et al 2014). For example, treatment plans generated for prostate cancer patients show some differences in calculated dose performed in synthetic CT images compared to standard CT images. It has been shown that the dose difference for synthetic CT images assuming the whole body as water equivalent is within 4% and for synthetic CT images using bulk density assignment and atlas-based electron density mapping, the dose differences were within 3% and 2%, respectively (Lee et al 2003, Chen et al 2004a, Pasquier et al 2006, Jonsson et al 2010, Greer et al 2011, Lambert et al 2011, Dowling et al 2012, Kapanen and Tenhunen 2013). More sophisticated voxel and hybrid methods can yield dose differences to CT calculations typically less than 1% (Dowling et al 2012, Korhonen et al 2014). Some evaluations of the accuracy of the commercial methods have been recently reported (Tyagi et al 2016, Christiansen et al 2017, Persson et al 2017) showing dose differences generally within 1%. Overall, the dose calculation differences are larger for organ-at-risks (OARs) compared to targets.

Bone is an OAR where the synthetic CT representation is important and at same time challenging. Accurate representation of boundaries is important for bones since their misrepresentation will results in some dose calculation inaccuracies. However, delineating the bone boundaries is also challenging mainly because bony tissues pose a significant susceptibility artifact and this may cause boundary perturbation and shift. The effect of bony tissues boundary distortion on dose calculation accuracy especially for surrounding tissues has not been studied yet and more related studies are necessary as MRI-only treatment planning workflow is become more prevalent in routine clinical practice.

Position verification with MRI images

Position verification using MRI reference images presents some challenges due to the need for multi-modality image registration. One solution is to use MR sequences that can identify implanted fiducial markers. One such example is using a 3D balanced-FFE sequence to elucidate implanted gold fiducial markers in the pelvis that can then be contoured for use in IGRT (Ghose et al 2016, Tyagi et al 2016, Maspero et al 2017). In comparisons of 3D matching (i.e. CBCT to MR-CAT or standard CT-SIM) for 5 SBRT prostate cases (5 fractions/patient), mean differences were less than 1 mm for left–right and anterior–posterior while the superior–inferior direction was <0.5 mm. A few cases showed registration differences of >2 mm, mostly due to rotations. For 2D matching in 20 patients (planar KV images matched to MR-CAT bony anatomy and fiducials found on MRI), mean differences were <0.6 mm along each axis. In another study of 20 prostate cancer cases
with 400 CBCTs evaluated with MRI as the reference dataset and daily shifts compared against CBCT-to-CT registration, shift positions for the cohort between CBCT-to-CT registration and CBCT-to-MRI registration are $-0.15 \pm 0.25 \text{ cm}$ (anterior–posterior), $0.05 \pm 0.19 \text{ cm}$ (superior–inferior), and $-0.01 \pm 0.14 \text{ cm}$ (left–right) (Doemer et al 2015).

Other previous studies have demonstrated the possibility of position verification between MRI images that have been used for treatment planning and orthogonal 2D verification images that have been acquired prior to treatment (Ramsey and Oliver 1998, Chen et al 2004a, Chen et al 2007, Yu et al 2014, Kapanen and Tenhunen 2013). Position verification using 2D images can be achieved by generating digitally reconstructed radiograph (synthetic-DRRs) from synthetic-CT (shown in figure 7) and register them to the orthogonal projection images acquired prior to treatment. In a whole-brain IGRT study conducted by Yang et al using orthogonal kV pairs in 7 patients found that all registrations were within 1 mm and 1 degree when aligned to their synthetic CT DRRs (Yang et al 2016). Yu et al calculated differences between bony landmarks in MR-DRRs (derived from manually contoured T1-weighted datasets with CT number mapping) and CT-DRRs and found good general agreement although cases with differences of up to 1.9 mm in landmarks were observed (Yu et al 2014). In recent work by Price et al studying registrations for 34–37 patient fractions, planar registrations had a mean shift differences were $0.4 \pm 0.5 \text{ mm}$ (range, $-0.6$ to $1.6 \text{ mm}$), $0.0 \pm 0.5 \text{ mm}$ (range, $-0.9$ to $1.2 \text{ mm}$), and $0.1 \pm 0.3 \text{ mm}$ (range, $-0.7$ to $0.6 \text{ mm}$) for the superior–inferior (S–I), left–right (L–R), and anterior–posterior (A–P) axes, respectively. For CBCT registrations, the mean shift differences in volumetric registrations were $0.6 \pm 0.4 \text{ mm}$ (range, $-0.2$ to $1.6 \text{ mm}$), $0.2 \pm 0.4 \text{ mm}$ (range, $-0.3$ to $1.2 \text{ mm}$), and $0.2 \pm 0.3 \text{ mm}$ (range, $-0.2$ to $1.2 \text{ mm}$) for the S–I, L–R, and A–P axes, respectively. The CT-SIM and synthetic CT derived margins were $<0.3 \text{ mm}$ different (Price et al 2016). Challenges have been realized in post-surgical areas, where resection cavities are not well characterized by synthetic CT solutions, which remains an area of potential opportunity.

**Advanced applications and future directions**

MRI has the capabilities of providing comprehensive anatomical and functional information regarding the tumour and its surrounding normal structure with respect to tumour burden and response assessment. For example, MRI spectroscopy (Harry et al 2010, Glunde et al 2011, Arias-Mendoza et al 2013), diffusion-weighted imaging (DWI) (Padhani et al 2009, Tsien et al 2014), dynamic contrast-enhanced (DCE) perfusion imaging (Harry et al 2010) and MRI elastography (Pepin et al 2014) are among the techniques that can be used to acquire functional information regarding tumour treatment response. Recently, a dose painting treatment planning pipeline was developed that incorporates functional multi-parametric MRI (including DWI and DCE) into an MR-only treatment planning workflow in the prostate with acceptable plan quality and excellent reproducibility (van Schie et al 2017).

Another challenge for synthetic CT includes MR-only planning for treatment sites that require motion management. Recent work evaluated MR-only planning for liver SBRT using a synthetic CT derived from a 3D gradient dual-echo Dixon sequence acquired at end-exhale (Bredfeldt et al 2017). Overall, excellent agreement was found between dose calculations on conventional CT-SIM data and synthetic CT, yielding $<0.5 \text{ Gy}$ difference for all metrics studied. A natural extension of this early work would be to incorporate four-dimensional MRI (4D-MRI) synthetic CT. 4D-MRI is becoming increasingly available in clinical and research prototypes including T2-weighted prospective acquisitions using external surrogates (Du et al 2015, Glide-Hurst et al 2015a), T1 and T2-weighted prospective acquisitions using internal navigators placed at the diaphragm/lung interface, and offline sorting using self-gating signal obtained from the k-space center (Freedman et al 2017) or body area for a respiratory surrogate (Liu et al 2014). As 4D-MRI becomes more commercially available, it is expected that MR-only planning will undergo further development.
Conclusion

The accessibility to functional imaging and more accurate structural information leads to improved tumor delineation in MRI images compared with CT. As MRI scanners become more widely available with radiation therapy platforms, their use in radiation oncology clinical practice is slowly increasing. A wide range of approaches for generating synthetic CT images exist which seem to provide promising results applicable for clinical use. Most of these methods are based on currently available standard clinical MRI sequences. As the field transitions toward MRI-only treatment planning, CT images will not be required and all imaging data required for delineation and dose calculation will be provided by MRI. To fully implement MR-only workflows in radiation therapy, patients must be set up in their treatment positions in the MRI simulator with appropriate immobilization devices.

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