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To cite this article: Simon J Doran 2019 *J. Phys.: Conf. Ser.* **1305** 012006

View the article online for updates and enhancements.

Radiotherapy in the presence of magnetic fields: a brief review of detector response characteristics and the contribution of 3-D measurements to the study of dose distributions at interfaces

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Abstract. The combination of MRI and radiotherapy on a single platform has the potential to revolutionise image-guided delivery of radiation doses. However, in order to realise these ambitions, good dosimetry must be available. The electron return effect gives rise to significant perturbations of dose at interfaces between tissue and air within the body, and this might lead to difficulties in dose compensation if air cavities move during treatment. In this article, I review briefly the ways in which the available methods of dosimetry are affected by the presence of magnetic fields and discuss the contribution that three-dimensional measurements can make to studies in this area. The methods of MRI and optical computed tomography have well known issues in imaging close to interfaces. These are described together with progress so far in providing solutions.

1. Introduction

The creation of treatment platforms that can acquire Magnetic Resonance Imaging (MRI) data during radiotherapy has enormous potential benefits for real-time, image-guided therapy. Improved tumour visibility in MRI, as compared with cone-beam CT, has led to significant investment over the last decade and the development of a number of systems - see the Introduction section of ref. [1] for a useful summary - some of which are already available commercially and approved for treating patients in the clinic. Much has already been written (and will not be discussed here) about the ways in which these different designs have tackled the various engineering challenges inherent in combining the two modalities. Each has consequences for the treatments that can be delivered and the quality assurance procedures required.

This article considers the general problem of how we can make reliable measurements of radiation doses to tissue in the presence of magnetic fields. What is different about the way dose is deposited? How are the various detectors at our disposal affected by magnetic fields? And what effects must we be particularly alert for as we plan our treatments? I start by reviewing briefly the different areas of dosimetry research that have been undertaken and then focus in more detail on the problems of 3-D dosimetry at the interfaces between different materials.

2. Background

2.1. Why is radiation dose deposition different in a magnetic field?

Whilst the incident photons employed in radiotherapy are unaffected by the presence of a magnetic field, the dose deposited in tissue has large contributions from secondary electrons and these are subject to the Lorentz force, whose direction is perpendicular to both the applied magnetic field and the instantaneous direction of motion of the electron. Two effects are important:

(i) For those electrons that remain within the tissue, trajectories can be calculated using the formalism developed by Jette [2] and these lead to a modified dose deposition. Where lateral electron equilibrium exists, build-up distances are decreased; there is a small increase in the penumbra perpendicular to the magnetic field; and there is a small shift of the depth dose curve compared with the B = 0 case [3].

(ii) Electrons can exit the original tissue and travel through air, either externally or within a body cavity, in circular arcs relatively unimpeded by scattering. For the magnetic fields and electron energies typically found in combined MR-radiotherapy systems - see Table 1 of [4] - the radius of curvature is of order mm or cm. This means that the location in which the dose from these secondary electrons is finally deposited might be some distance from the point at which the electrons previously left the tissue. This is the so called electron return effect (ERE) [4, 5].

Magnetic field effects on dose deposition can be simulated using Monte Carlo methods [6].

2.2. Measurement using ionisation chambers
Typically, reference dosimetry is performed with ionisation chambers containing an air cavity. The paths of the ion pairs created are perturbed by the magnetic field and this leads to variations in the chamber response with model, strength of the magnetic field and orientation of the chamber with respect to the field [7-11]. For accurate dosimetry, it has been shown that it is necessary to place the ionisation chambers in water and that inserting them directly into solid phantoms is not advisable due to the likely presence of air gaps that may perturb the dose reading [12, 13].

2.3. Solid-state detectors
Reynolds *et al.* [14] simulated both a PTW60003 diamond detector and an IBA PFD diode detector and found that both required corrections for use with a magnetic field that depends on the relative orientations of magnetic field, detector long axis and incident photon beam. Gargett *et al.* [15] also found an orientation dependence in Monte Carlo simulations of a novel 2D silicon diode. Although it is generally difficult to perform identical tests *in situ* with and without a magnetic field present, several groups have compared the performance of clinical multi-diode devices mounted in an MR linac with the same tests carried out on a conventional linac at B = 0. Houweling *et al.* [16] investigated an MR-compatible ArcCHECK device in the presence of a 1.5 T magnetic field and found it to be similar within 1% to reference data of Li *et al.* [17]. De Vries *et al.* [18] performed a detailed comparison between the detector response characteristics of an MR-compatible Delta4 positioned in both MR- and conventional linacs and noted the same general agreement in properties, but an orientation dependence of between 5 and 10%.

2.4. EPID-based dosimetry
Although Raaymakers *et al.* [19] demonstrated some time ago the feasibility of using a megavoltage portal imaging device in conjunction with their MR linac, it is only recently that attempts have been made to adapt existing back-projection algorithms for quantitative portal dosimetry [1] and characterise EPID detectors in detail [20]. Issues making this difficult are:

(i)   the elements of the MR scanner situated between the patient and the EPID, which comprise a non-uniform attenuating medium that also alters the photon energy spectrum;

(ii) changes in dose deposition in the patient, as described in Sec. 2.1;

(iii) possible changes in the dose-response characteristics of the EPID in a magnetic field.

Torres-Xirau *et al.* have addressed issue (i) successfully in a "mock-up" of an MR linac [1]. There is also encouraging agreement in the EPID profiles of normalised pixel intensity between data for different field sizes acquired in both the presence and absence of a magnetic field [20].

2.5. Response of radiochromic film

Despite the ability to obtain excellent relative dosimetry using Gafchromic EBT2 film [21], there have been persistent concerns about the potential for low-level effects of magnetic fields on the absolute reading from radiochromic films. Raaijmakers *et al.* [22] noted that measurements in magnetic fields up to 1.3 T led to readings in the range 1 - 4% lower than at B = 0, and Reyhan *et al* [23] observed a similar decrease in optical density, whilst Reynoso *et al.* [24] observed dramatic reductions of up to 15%. Potential explanations proposed have included a change in the underlying yield of the radiochemical reaction and a localised temperature increase associated with thermal energy deposition by MR imaging sequences. Early results using the more recently introduced EBT3 film [25] suggest an effect of magnitude approximately 2%.

2.6. 3-D dosimeters

For ionisation chambers, diodes and EPIDs, energy deposition and quantitative readout happen contemporaneously. By contrast, for 3-D dosimeters and films, the dose detection process has *two stages*: first, the radiation dose leaves some form of "chemical imprint" on the detector, and then, at some later juncture, we read this out with one of several possible 3-D imaging techniques (typically, Magnetic Resonance Imaging (MRI), x-ray computed tomography (CT), optical CT or ultrasound imaging) [26]. The effect of magnetic field on the dose-imprinting process can thus be separated from the readout, the main exception being where the readout device is actually the MR linac itself and a "real-time" dose readout is required [26][27].

Given the close historical links between gel dosimetry and MRI, it is perhaps surprising that (to the best of my knowledge) no studies have been performed to measure the effect of magnetic field on the radiochemical yield (G-value) of ferric ions. Similarly, it is not clear whether comparisons have been performed to determine whether there is any magnetic field dependency of the polymerisation reaction in polymer gel dosimetry.

For the PRESAGE® dosimeter, which is designed for optical readout, three sets of measurements [27-29][28-30] all found only very small effects (~1%) at the edge of detectability when comparing the optical response of dosimeters irradiated in the presence and absence of a magnetic field.

3. Dosimetry at interfaces

Understanding the effects that occur at interfaces between different media is of great importance in many clinical dosimetry problems. Of particular concern for the MR linac are air-containing cavities, as found in the head and neck, lung and pelvic areas, where the ERE may modify dose significantly. Although strategies exist to optimise dose to compensate for these perturbations in static situations, the cases of air in the bowel and rectum represent considerable challenges for MR linac therapy because the position of the air cavity may move during treatment delivery. Precisely those cases that might benefit most from intra-fractional image-guided therapy are those where the electron return effect may present the greatest hazards [30][31].

Since body cavities often have irregular shapes for which prediction of dose in a magnetic field requires 3-D models, it is likely to be insufficient to use 2-D radiochromic film for patient-related dose verification. Similarly, the inhomogeneities involved and potentially high spatial resolution required suggest that neither ionisation chambers nor diode arrays will be suitable. 3-D materials moulded to relevant anthropomorphic shapes are thus attractive potential solutions.

However, as detailed below, there are a number of problems that must first be addressed.

3.1. MRI measurements at interfaces

The presence of interfaces is a concern in MRI because of the differences in magnetic susceptibility (or, equivalently, magnetic permeability) between materials, particularly tissue and air. The main magnetic field B_0 is perturbed and this leads to shifts in the Larmor frequencies of the imaged nuclei ("NMR spins"). Two effects are important:

(i)   Since the Larmor frequency of a spin relates directly to the position at which it appears in the final image, images can become distorted. In general, both positional information and measured image intensities are affected, with signal "pile-up" in some voxels and signal voids in others.

(ii) Distributions of precession frequencies within voxels can lead to the cancelling out of spin contributions (additional "dephasing", which is expressed as a dramatically decreased T_2*). This is a second cause of "signal dropout", i.e., apparent voids in the images.

However, despite these potential issues, robust experimental design, using the correct types of sequence allows quantitative dosimetric imaging near interfaces [31, 32][32, 33] An area where MRI may have a significant role to play is in the creation of anthropomorphic phantoms, with a range of different densities [33][34].

3.2. Inhibition of polymer gels

In the early days of polymer gel dosimetry, a major confounding factor preventing accurate dosimetry at interfaces was the failure of the gels to polymerise. Oxygen free radicals have an inhibitory effect on the radiation detection process, because they "mop up" the free radicals created by the incident photons, thus preventing them taking part in the polymerisation process [34][35]. Since many commonly used phantom construction materials are permeable to oxygen, early solutions to the inhibition problem involved the use of glass vials of various sizes. However, as noted in [32][33], the use of glass containers is not suitable for interface dosimetry because its density perturbs the very dose distributions it is aiming to measure. Other solutions have involved the use of the thermoplastic Barex® to form gel containers, or, starting with the work of Fong *et al.* [35][36], the use of dosimeter gels that are unaffected by oxygen.

3.3. Optical measurements at interfaces

Perhaps the most significant difference between optical and x-ray CT is that, whilst x-rays travel in straight lines through the sample, regardless of any interfaces present, light in the visible spectrum is refracted at boundaries between materials of different refractive indices. In order to minimise this effect, samples are almost always placed within some form of "aquarium" containing a so-called "matching liquid" of similar refractive index. An imperfect match leads to a prominent image artefact at the interface, with significant perturbations of image intensity extending both into and outside the sample. The situation has been modelled both for gel samples inside containers [36][37] and solid PRESAGE® dosimeters that do not require any external support [37][38].

Historically, many of the applications of optical CT have used large cylindrical samples, for which the effect typically manifests itself as a high intensity ring surrounding the dosimeter and occupying a small fraction of the analysed region that can often be excluded. However, the spatial extent of the artefact does not scale with sample radius [36][37] and this leads to potential problems whose aim is to make accurate measurements of dose next to small moulded air cavities.

As in the case of MRI, an understanding of the physics governing the measurement process has allowed useful measurements of dose to be obtained close to boundaries, by ensuring careful optimisation of refractive index and judicious choice of interface positions in experiments [27, 38, 39][28, 39, 40].

3.4. PRESAGE®

Recently, it has become apparent that the optical edge artefact discussed above may, in fact, mask a genuine change in physical and dosimetric properties at the outer edges of PRESAGE® samples. Anecdotal reports have noted that when machining PRESAGE® cylinders, some regions of the sample

are softer than others. During their work on remote dosimetry of the Viewray MR-guided IMRT platform, Rankine *et al*. [29][30] observed disagreements "in the periphery of the dosimeter" and these were similar to results seen and analysed in detail by Dekker *et al.* [40][41]. A radially non-uniform sensitivity was observed, which was batch dependent. At this IC3Ddose meeting, Costa *et al.* present studies specifically designed to measure and correct for this type of radial dependence. This is vital if credible measurements of dosimetry at interfaces are required.

4. Conclusion
The translation of combined MRI-radiotherapy platforms from the research laboratory to the clinic is now well underway. Together with intense technical development on the systems themselves has come a renewed focus on dosimetry in the presence of magnetic fields. Although most of the physical principles involved are now well established, issues still remain. As with other areas of radiotherapy, 3-D dosimetry has its part to play, and the study of interfaces between different tissues has the potential to be of great interest moving forward.

5. Acknowledgments

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