Dosimetry needs for MRI-linacs

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Abstract. The pursuit of real-time image-guided radiotherapy has prompted the development of hybrid devices coupling MRI scanners with radiotherapy treatment units, usually linear accelerators (linacs). One of the challenges in MRI-linac technology is the magnetic field impact on the dose deposition. Dose deposition effects have to be considered in the radiation therapy chain as they alter the dose distribution in patients and influence the response of many commonly used radiation detectors. In this presentation specific issues of dosimetry for MRI-linacs will be reviewed and illustrated with examples from the Australian MRI-linac commissioning process.

1. Introduction
The pursuit of real-time image-guided radiotherapy has prompted the development of hybrid devices coupling MRI scanners with radiotherapy treatment units, usually linear accelerators (linacs) [1]. One of the challenges in MRI-linac technology is the magnetic field impact on charged particles, which affects both the radiation beam generation in the linear accelerator as well as the dose deposition in the irradiated volume. While the former can be remedied with magnetic shielding and is not discussed further, the latter alters the dose distribution in patients and in radiation detectors and is the focus of this article: specific issues of dosimetry for MRI-linacs will be reviewed and illustrated with examples from the Australian MRI-linac commissioning process.

2. MRI-Linac Systems
Four MRI-linac designs exist to date, covering a range of magnetic field strengths and employing two beam-to-magnetic-field orientations [2]: perpendicular (also referred to in the literature as transverse) or parallel (also referred to in the literature as inline). In terms of the interference between the magnetic field and the dose deposition, each system therefore has different characteristics, which are reviewed below and summarised in Table 1 (starting from lowest field strength warranting least interference). Discussion of the imaging properties of these systems is beyond the scope of this manuscript, however, it should be mentioned that all systems have been demonstrated to provide image quality sufficient for MR guided radiotherapy (MRgRT) [3-6].

The MRIdian (ViewRay, Oakwood Village, OH, USA) system uses a 0.35 T split superconducting magnet in a perpendicular configuration with a radiation beam from, originally three cobalt sources, or more recently a 6 MV linear accelerator. The accelerator rotates around the patient couch on a gantry
installed in the magnet split. The split design ensures an unobstructed beam path [7] and the relatively low magnetic field of this system results in minimal influence on dose deposition.

The Aurora-RT system (MagnetTx Oncology Solutions, Edmonton, Alberta, Canada) was developed at the University of Alberta and consists of a biplanar 0.5 T magnet and a 6 MV linac, which are both installed on a gantry and rotate around the patient couch. Whilst both beam-to-magnetic-field orientations are possible, in the current design the radiation beam axis and the magnetic field are parallel. Two features of this system should be highlighted from the dosimetry point of view. Firstly, the high temperature superconducting magnet allows the system to be turned on and off relatively easily allowing zero field measurements [2]. Secondly, an iron yoke used within the bore to restrict the extent of the fringe field reduces electron contamination [8]). (see section 3.1.2.).

The Australian MRI-linac consists of an open bore 1.0 T magnet and a linac which can produce photon beams of two energies 4 & 6 MV. While the versatile design permits radiation beam entry (and patient positioning) in either orientation, the beam is parallel to the magnetic field in the current system. Uniquely, the linac is mounted on rails, allowing variation of the source-to-isocentre distance (SID) and therefore enabling dosimetric measurements at different magnetic field strengths [2,6].

The system commercialised as Elekta Unity, has been developed at the University Medical Centre Utrecht (Utrecht, the Netherlands) in collaboration with Elekta AB (Stockholm, Sweden) and Philips Medical Systems (Best, the Netherlands). It consists of a modified closed bore 1.5 T Ingenia MR scanner and an Elekta 6 MV linac, which rotates around the patient couch. The radiation beam is perpendicular to the magnetic field and passes through the magnet cryostat. Although the magnet has been modified to minimise radiation beam perturbation, it attenuates the beam, introduces additional scatter [1,9] and prohibits the use of a (small) range of beam angles [9].

| System (vendor) | Magnet properties | Orientation | Radiation beam properties |
|-----------------|-------------------|-------------|--------------------------|
| MRIdian (Viewray) [7] | split, superconducting | 0.35 T, 70 cm bore | perpendicular | 6 MV, 0.9 m, 27.4×24.1 cm / 4.15 mm |
| Aurora-RT (MagnetTx) [10,11] | biplanar, high temp. superconducting | 0.5 T, 110(W) × 60(H) cm | parallel (perpendicular possible) | 6 MV, 1.22 m, 30 cm / 5 mm |
| Australian MRI-linac | open, superconducting | 1.0 T, 62 cm bore, 50 cm gap | parallel (perpendicular possible) | 4 & 6 MV, 1.9–3.3 m, 30–50 cm / 9.5–16.5 mm |
| Unity (Elekta) [9,12] | closed, superconducting | 1.5 T, 70 cm bore | perpendicular | 6 MV, 1.435 m, 22×57.4 cm / 7.2 mm |

SID – source-to-isocentre

3. Dose deposition in presence of the magnetic field
While the photons produced in the accelerator are not affected, both the contaminant electrons (originating in the accelerator and in other materials/air upstream from the patient or phantom) as well as the secondary electrons released in patient (or phantom) are affected by the magnetic field. The Lorenz force acts orthogonally to the magnetic field and to the direction of the electron motion, therefore its effects on the dose distribution depend on the orientation of the radiation beam relative to the magnetic field. It should be emphasised that, both the dose deposition in matter as well as the response of the dosimeters are affected by the magnetic field.

3.1. Influence on dose deposition by radiation beams
3.1.1. **Transverse configuration.** In MRI-linacs employing a transverse configuration, the radiation beam axis, and hence to the predominant component of electron motion direction, is perpendicular to the magnetic field. The Lorentz force causes the electron paths between the collisions to become curved, resulting with trajectories with a preferential direction change and a shorter overall range. Macroscopically, this results in a shifted and asymmetric beam penumbra and in a decreased build-up distance [13]. Furthermore, upon entering low-density materials (such as lung or air cavity), depending on the radiological thickness of this region relative to their trajectory radius, electrons may be curved back, towards the interface, in a phenomenon termed the electron-return effect (ERE) [14]. ERE leads to localised dose increases at such tissue interfaces, including at the beam exit surface [14]. Additionally, in transverse MRI-linac systems, the Lorentz force sweeps contaminant electrons away from the incident beam axis. This leads to the reduction of the skin dose within the primary beam [8], however these electrons can potentially reach the patient surfaces outside the primary beam [15].

3.1.2. **Inline configuration.** In inline MRI-linacs the radiation beam axis is parallel to the magnetic field, however, due to scattering, the electrons have motion components orthogonal to the magnetic field. The Lorentz force causes the electrons to spiral around the magnetic field direction and successive energy losses in collisions lead the shrinkage of their helical orbits [16]. Macroscopically, this results in the reduction of the beam penumbra and in dose enhancement on the beam central axis, especially pronounced in low density materials. Furthermore, it reduces the dose deposition perturbations due to density heterogeneities [16]. On the other hand, strong inline magnetic field reduces the divergence of the contaminate electrons and concentrates them around radiation beam axis, which can lead to increased skin doses if not counteracted [17].

3.2. **Influence on treatment plan dose distributions.**
Non-negligible modifications of dose distributions, especially at density interfaces, have been reported for clinical treatment plans for various tumour locations, if the influence of magnetic field was not accounted for [18-20]. Conversely, Oborn et al [21] showed a case of tumour surrounded by low density tissue (lung) in inline field, where these effects could be beneficial and lead to target dose enhancement. A full review of studies investigating the effects of magnetic fields on treatment plans is beyond the scope of this manuscript. However, the need for dose computation algorithms that model magnetic field influence accurately, allowing to compensate or to even exploit these effects in plan optimisation [22], should be emphasised. This in turn requires high resolution 3D dose measurement systems, such as radiochromic or polymer materials, for the benchmarking of the dose computation algorithms as well as for the treatment plan verification.

3.3. **Influence on detector response**
The response of radiation detectors is related to the paths of the electrons traversing their active volume. In the magnetic field the electron trajectories change, however this change may be different in the materials constituting the detector (e.g. air cavities in ion chambers, silicon wafers in diode detectors etc.) than the trajectory change in the surrounding medium (phantom). As a result, the reading of the detector may not represent the dose that would be deposited in the medium in its absence. Furthermore, many detectors are not symmetric, therefore the change in their response is dependent on their orientation in the magnetic field.

4. **Dosimetry for MRI-linacs: current status and open issues**
The general dosimetry framework for MRI-linac systems, similarly to other radiotherapy modalities, aims at ensuring that the therapeutic dose distributions are delivered with sufficient accuracy. In this section, dosimetry issues for MRI-linacs and related technical and clinical aspects different to conventional radiotherapy, will be discussed and illustrated with examples from the Australian MRI-linac commissioning process.
4.1. Dose measurements

4.1.1. Detector response. Detector response to radiation may be modified by different factors depending on the principle of operation of the dosimeter. Ideally, detectors used in MRI-linac dosimetry should have no or a well-characterised dependency of response on these factors and their linearity and reproducibility should not be affected. While the most appropriate detectors for dosimetry in high magnetic fields are still subject of research, investigations of the characteristics of standard detectors show that the change in reading can be in the order of several percent. The current state of knowledge on the use of various types of detectors in the magnetic field, grouped by their principle of operation, is reviewed below.

Several groups investigated the responses of common types of air-filled ion chambers in the presence of magnetic fields either experimentally or by employing Monte Carlo (MC) simulations. Negligible response changes of about 1–2% have been reported in inline magnetic fields of up to 2 T [23,24]. Both over- and under-responses, peaking at ca. 10% at field strengths of about 1 T, have been observed in transverse fields [23,25-26]. It should be noted that in these studies long axis of the ion chambers was typically oriented perpendicular to the magnetic field. O’Brien et al [27] described a difference between two possible perpendicular orientations, resulting from the electrons generated in the chamber stem being curved either towards or away from the sensitive volume. Additionally, they showed that the response change for chambers positioned parallel to the field is less than 1%. These observations were confirmed in a systematic MC study by Reynolds et al [28].

Recognising the fact that these effects should be included in the formalism, several authors, experimentally or using MC, derived correction factors ($k_B$), rather than of response changes reported earlier, for a number of standard cylindrical [27,29-32] and parallel-plate chambers [31]. Parallel-plate chambers were shown to require larger $k_B$ values (up to 8.9%) in both the parallel and perpendicular orientations to the magnetic field and to have larger angular sensitivity [31]. At the Australian MRI-linac a unique approach for $k_B$ measurements is being pursued, which exploits the linac-on-rail system enabling dosimetric measurements at different magnetic field strengths including ~0 T [33]. It should be noted that the $k_B$ factors obtained in these studies are contingent on the definitions introduced by the formalism and that they were obtained mostly for transverse MRI-linacs. Finally, it should be mentioned that recently a mobile water calorimeter has been developed and characterised for use in MRI-linacs allowing the direct determination of the correction factors [34].

Reynolds et al [35] simulated the response of two detectors relying on ionisation in solids, diamond detector PTW60003 (PTW, Freiburg, Germany) and PFD diode (IBA Dosimetry, Schwarzenbruck, Germany), in various orientations in transverse and inline magnetic fields up to 1.5 T. Similar to the case of ion chambers, the response change was more pronounced in transverse (up to ~20% at 1.5 T) as compared to in inline magnetic field (~2% at 1.5 T). In transverse fields however, unlike for ion chambers, the change in response was monotonically increasing in the examined range field strengths. The effect was strongly dependent on the detector rotation around the beam and magnetic field directions and it was most pronounced with the detector axis perpendicular to the beam. In a follow up paper the same group confirmed some of these results experimentally in inline magnetic fields up to 0.2 T [24].

Among the dosimeters relying on chemical transitions induced by radiation, two common types are radiochromic and polymer materials. Dose under-responses were reported for Gafchromic EBT2 films (Ashland Advanced Materials, Bridgewater NJ, USA) exposed to 0.35 T [36] and 1.5 T fields [37,38]. These effects have been attributed to the occurrence of magnetokinetic changes in polymer orientation observed microscopically [36, 38]. Additional changes in response were observed for EBT2 films that underwent concurrent imaging and correlated with the radiofrequency exposure and associated temperature increase [38]. However, two conference contributions on similar experiments with EBT3 films reported no changes in dose response in the presence of 0.35 T [39] and 1.5 T fields [40]. Only minimal effect of magnetic field on dose response (up to 5%) and its linearity (up to 1.6%) were observed for three types of 3D chemical dosimeters, each using a different response mechanism: radiochromic
plastic (PRESAGE, in-house made) in 1.5 T field, radiochromic gel (FOX, in-house made) in 1.5 T field and polymer gel (BANG, MGS Research, Madison, CT, USA) in 1 T field [41].

Wen et al [40] and Wang et al [42] reported no response modification for thermoluminescence detectors (TLDs) in a 1.5 T transverse field. Spindeldreier et al [43] observed a relatively small dose response decrease with increasing magnetic (-1.3%/T) and no influence on the angular dependence for optically stimulated luminescent dosimeters (OSLDs). Therefore, luminescence based detectors are considered promising candidate for dosimetry in magnetic fields.

Plastic scintillation detectors (PSDs) are non-magnetic and have density similar to water, which potentially renders their dose response independent of magnetic field. Stefanowicz et al [44] investigated this in fields up to 1 T and observed unexplained response increase by up to 7%. More recently, Therriault-Proulx et al [45] reported 2.4% dose response increase in fields up to 1.5 T, which they interpreted as the change in the inherent dose deposition within the phantom, and concluded that PSDs show great promise for use in magnetic fields. However, other works suggest that optimal noise-subtraction methods for PSDs are still to be established [46].

4.1.2. Air gaps. Air gaps might be present between the dosimeter and the surrounding material e.g. when using plastic phantoms or even in water phantoms when using non-waterproof ion chambers with sleeves. It has been demonstrated experimentally [47] and through MC simulations [48] that in a 1.5 T transverse magnetic field this may lead to the reduction of a Farmer-type chamber response by 0.7–1.2% in a standard solid water phantom. This effect was attributed to a reduction in ionisation produced in the immediate vicinity of the chamber wall rather than to ERE [48]. Agnew et al [49] investigated this systematically in a phantom with a recess manufactured within the chamber holder allowing the placement of the air gap at a controlled angle relative to the radiation beam. They observed variability of the Farmer-type chamber response, depending on the air bubble position, of 0.25% at 0.25 T and 3.8% at 1.5 T. To the best of the authors’ knowledge, these effects have not been investigated for inline MRI-linacs. Filling the chamber holder with water has been proposed as a method to mitigate air gap effects. However, it is difficult to apply in transverse MRI-linacs, where chamber orientation parallel to the magnetic field is preferred in order to minimise the dose response perturbation [27,31].

Small air volumes may also be present in detector arrays. This was investigated for a monolithic silicon array detector, Magic Plate (M512), developed at the University of Wollongong. A small air gap above the silicon chip in this device compensates for different scattering and electron spectra in silicon compared to water and was optimised in absence of magnetic field. Measured response changes of 2% in an inline and of 10% in a transverse ~1 T field as well as profile shift and penumbra width overestimation in MC simulations have been attributed to the presence of this gap [50].

4.1.3. Absolute dosimetry. For the MRgRT systems employing a low magnetic field strength, reference dosimetry based on TG-51 code of practice has been shown feasible [51]. However, given the significant response changes in higher magnetic fields for air-filled ion chambers [23-32] commonly used for reference dosimetry, the need to introduce a new correction factor and formalism has been recognised [32]. First approaches defined it as a ratio of the measured signal of the chamber without the magnetic field present under the same conditions [26], however, several aspects hinder such a simple expansion of existing codes of practice.

Due to the construction characteristics of MRI-linacs (see Table 1) the reference conditions (e.g. specific SSD or field size) cannot be always fulfilled, requiring extension of the formalism [27]. The effect of the magnetic field on the dose distribution may change the relationship between the beam quality specifiers and the Spencer-Attix water-to-air restricted stopping power ratios for which they are surrogates [27,32]. Finally, it had been suggested that the correction for the change in reading of the ion chamber should be separated from the change in dose distribution [30].

Under these considerations, several methods have been proposed to include the response modification in the formalism for reference dosimetry [27,30-31], however the establishment of the code of practice, in order to warrant the same level of uncertainty as for conventional radiotherapy modalities,
is still among the open issues in MRgRT. It should be complemented by a systematic characterisation of various detectors in terms of their response in the magnetic field and by recommendations on their suitability for different dosimetry tasks. To this end, a standardised experimental methodology and/or benchmarks of the MC codes for radiation transport simulations used in detector response modelling are required.

4.1.4. Relative dosimetry. Constancy of detector response with the off-axis position, depth or field size is a prerequisite in relative dosimetry.

Many ion chambers were shown to exhibit stronger angular response dependence in the magnetic field [23-31]. Through MC simulations Reynolds et al [35] showed response changes of 10-20% for the PTW0003 diamond and the PFD diode detector in a water tank at the periphery of the beam, when oriented perpendicular to a transverse magnetic field of 0.5 T. The response variation was much smaller with detectors positioned parallel to a transvers magnetic field and in an inline magnetic field. Woodings et al [52] observed increase of the angular dependence for the diamond detector (PTW60019) in a transverse 1.5 T field, which was deemed relevant for relative dosimetry at distant off-axis positions or at different gantry angles. At the Australian MRI-linac the PTW60019 detector has been recently used for percentage depth dose (PDDs) profiles measurements at near zero and 1.0 T fields [53]. Furthermore, MOSkin, MOSFET type dosimeters with the effective point of measurement at 70 μm (as recommended by ICRP for skin dosimetry) developed at the University of Wollongong, are currently employed for acquisition of high-resolution PDDs.

Different orientation of individual detectors in detector arrays leads to non-negligible artefacts in profile measurements [54] for transverse MRI-linacs. These effects have not been observed for the Australian MRI-linac employing an inline configuration [55].

The suitability of various detectors for specific relative dosimetry tasks in a transverse magnetic field of 1.5 T has been investigated recently by O’Brien et al [56]. They observed discrepancies between the results obtained with ion chambers of various volumes and with solid state detectors for PDD and output factor measurements for small fields and in the estimation of the lateral shift in the dose distribution. The latter has been simulated using MC and attributed to the detector density; the detectors containing dense materials dampen the magnetic field effects while the low density detectors accentuate it [57]. Finally, the effective point of measurement (EPOM) for air-filled ion chambers has been shown to change in the magnetic field, whereas for solid state detectors (diamond, diode) it remains unaffected [56].

Last but not least, the risk of detector response perturbation during beam scanning, due to motion in the magnetic field, has been considered, however no such effects have been observed for ion chambers [58] and for diamond detector [52].

4.2. Hybrid tests

The hybrid character of the MRI-linac treatment unit requires the assessment of its concurrent functionalities, for instance dose deposition during imaging, congruence of the imaging and radiation isocentres and end-to-end testing. The issues relevant to the verification of the dose deposition during imaging [6] have been addressed in other sections. In the context of non-dosimetric hybrid tests, there is a need for development and/or characterisation of dual MR/CT-visible synthetic tissue equivalent materials [59]. It is also interesting to note that even alignment tests pose additional challenges when executed on an MRI-linac. For example, in transverse MRI-linacs the result of the star shot test is altered by the magnetic field, due to the lateral shift of the deposited dose. To overcome this, QA phantoms using electron dense materials that reduce the magnetic field effect on the dose deposition [60] and the use of polymer gels to visualize the intersecting beams with MR [61] have been proposed.
4.3. Technical aspects

4.3.1. Magnetic field compatibility. The foremost technical requirement for the dosimetric equipment (e.g. phantoms, detectors) to be used in MRI-linacs is compatibility with the magnetic field. The two main compatibility issues are the safety and image quality aspects due to the presence of ferrous materials or unscreened mechanical or electrical components respectively.

Many authors reported on the use of standard point dose detectors (ion chambers, diodes, etc.) in the magnetic field [23,25-27,30,32,33,52], however some vendors introduced additional MR safety certification for these models. Other types of equipment exist now in MR-conditional versions, modified for use in MRI-linacs, perhaps most notably the detector arrays like Starcheck maxi MR and Octavius Detector 1500 MR (PTW, Freiburg, Germany), ArcCHECK-MR and IC PROFILER-MR (Sun Nuclear Corporation, Melbourne, FL USA) or MR-Delta4 (Scandidos AB, Uppsala Sweden). It should be underlined that these devices still require characterisation of their dose response in the magnetic field (discussed in section 4.1.1) [54,62,63].

In the context of magnetic field compatibility other equipment should also be mentioned. In-house designed or manual 1D tanks MR-compatible water tanks are currently used, while a motorised 3D plotting tank PTW-MP3T (PTW, Freiburg, Germany), developed in collaboration with the University Medical Centre Utrecht (Utrecht, the Netherlands) [58], is scheduled to be released in Q4/2018. It’s worth mentioning that the scanning speed of this tank is lower than that of the conventional ones due to the use of different motors. Finally, MR-compatible motion phantoms e.g. MRI-LINAC Dynamic Phantom (CIRS, Norfolk, VA, USA) or QUASAR MRI-Compatible Respiratory Motion Phantom (Oncology Systems Limited, Shrewsbury, UK) are available, addressing the fact that treatment of moving targets is seen as one of the most promising applications of MR-linacs.

4.3.2. Other compatibility aspects. MRI-linacs have restrictions in terms of accessibility and space around isocentre when compared to conventional linacs. Most notably, the standard full-size water tanks cannot be used in the magnet bore. This necessitates the development of dedicated water tanks, like the above mentioned PTW-MP3T scheduled to be released with two vendor specific sizes and scanning ranges. Furthermore, not all MRI-systems feature a full set of isocentric lasers for positioning. Fiducial markers for alignment based on portal imaging or, ideally, MR-visibility are desirable in the design of dosimeters and of auxiliary dosimetric equipment [59,64]. Finally, concurrent dose measurement and MR image acquisition scenarios require detectors to cause minimal RF interference (e.g. from the power supplies and readout system) and image distortion.

4.4. Clinical aspects

The greater targeting accuracy provided by MRI-Linacs lends their use in the most challenging cases involving small fields, steep dose gradients, non-uniform target doses, high fraction doses and employing dynamic delivery techniques, ultimately, with concurrent real-time tracking. Additionally, planned dose distributions must account for the effects of the magnetic field, like the dose perturbations at tissue interfaces, which may add to their complexity. These properties determine the requirements for dosimetry systems with special emphasis on high spatial resolution and volumetric information. Ion chambers, detector arrays or even films, commonly used for dose distribution verification in radiation therapy, may be insufficient for verification of dose distributions in MRgRT. One promising alternative are 3D gel or plastic dosimeters [65,66]. Some of these dosimeters are read-out using MR, which enables their use at MRI-linacs in the treatment set-up and even for real-time dosimetry if required [67].

Another aspect of MRgRT, which sets new requirements for dosimetry, is the online adaptive therapy workflow, where the patient remains on the table during the plan adjustment process. This has two consequences for the patient specific QA process: the original treatment plan may never be applied and the adapted plan cannot be measured before application. This may render obsolete the patient specific pre-treatment QA, but at the same time it increases the requirements for dose computation accuracy and prompts the need for online verification tools e.g. based on transient dosimetry using electronic portal
imaging devices (EPIDs). For application on MRI-linacs, EPID dosimetry requires some modifications
to account for: the influence of the magnetic field on the functioning of EPID and on the dose deposition
in the material traversed by the beam (patient or phantom) and, in some designs, for the presence of
additional material in the beam. A back projection algorithm accounting for the latter has been recently
developed for the Unity system [68]. Implementation of EPID dosimetry is ongoing within the
Australian MRI-linac program.

5. Summary
Several areas have been reviewed, where improvements or new developments in dosimetry are needed
to facilitate safe clinical implementation and to support further development of MRgRT. With the
expected rapid adoption of MRI-linacs, the need for a new formalism for dosimetry protocols and for
recommendations on the suitability of various detectors for different dosimetry tasks was discussed. The
limitations of current dosimeters were indicated and several possible developments that will benefit
MRg were highlighted, such as validation and further optimisation of 3D dosimeters, dosimetric
benchmarking of MC radiation transport codes for independent dose calculation and implementation of
EPID dosimetry in the presence of magnetic fields. Finally, the need for the development of phantoms
specific to the hybrid nature of the MRI-linac system was underlined.

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