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Optical imaging method to quantify spatial dose variation due to the electron return effect in an MR-linac

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Abstract

Purpose: Treatment planning systems (TPS) for MR-linacs must employ Monte Carlo based simulations of dose deposition to model the effects of the primary magnetic field on dose. However, the accuracy of these simulations, especially for areas of tissue-air interfaces where the electron return effect (ERE) is expected, are difficult to validate due to physical constraints and magnetic field compatibility of available detectors. This study employs a novel dosimetric method based on remotely captured, real-time optical Cherenkov and scintillation imaging to visualize and quantify the ERE.

Methods: An intensified CMOS camera was used to image two phantoms with designed ERE cavities. Phantom A was a 40cm × 10cm × 10cm clear acrylic block drilled with 5 holes of increasing diameters (0.5cm, 1cm, 2cm, 3cm, 4cm). Phantom B was a clear acrylic block with 3 cavities of increasing diameter (3cm, 2cm, 1cm) split into two halves in the transverse plane to accommodate radiochromic film. Both phantoms were imaged while being irradiated by 6MV flattening filter free (FFF) beams within a MRIdian Viewray (Viewray, Cleveland, OH) MR-linac

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(0.34T primary field). Phantom A was imaged while being irradiated by 6MV FFF beams on a conventional linac (TrueBeam, Varian Medical Systems, San Jose, CA) to serve as a control. Images were post processed in Matlab (Mathworks Inc., Natick, MA) and compared to treatment planning system (TPS) dose volumes.

Results: Control imaging of Phantom A without the presence of a magnetic field supports the validity of the optical image data to a depth of 6cm. In the presence of the magnetic field, the optical data shows deviations from the commissioned TPS dose in both intensity and localization. The largest air cavity examined (3cm) indicated the largest dose differences, which were above 20% at some locations. Experiments with Phantom B illustrated similar agreement between optical and film dosimetry comparisons with TPS data in areas not affected by ERE.

Conclusion: There are some appreciable differences in dose intensity and spatial dose distribution observed between the novel experimental data set and the dose models produced by the current clinically implemented MR-IGRT TPS. Further validation of the proposed experimental methods, as well as the magnetic field effect models within the TPS, is recommended.

Keywords

Cherenkov; scintillation; electron return effect

1. Introduction

The field of radiotherapy has progressed towards the goal of visualizing internal anatomy in real time during irradiation in order to validate targeting, shrink treatment margins, and enable precise dose escalation. Tracking soft tissue motion has recently proven possible using integrated magnetic resonance image guided radiotherapy (MR-IGRT) systems, now adopted clinically at several institutions worldwide. Early approaches combined MRI with Co-60 radioactive sources, and more recent and complex systems incorporate MRI with medical linear accelerators (MR-linacs).¹⁻⁷

One of the primary technical challenges of MR-IGRT systems is the presence of a strong transverse magnetic field, which in the case of an MR-linac affects the electronic generation of x-rays, as well as the behavior of dose-depositing secondary electrons. These effects have been extensively studied throughout the development and implementation of MR-IGRT systems through Monte Carlo (MC) simulations.⁸⁻¹² It is also understood that for instances of highly contrasting physical density interfaces, such as tissue-air boundaries, the electron return effect (ERE) will generate localized areas of escalated or reduced dose.^{10, 12} However, experimental validation of these studies proves challenging due to magnetic field effects and resolution limits on many currently accessible detectors.

Diodes have been shown to produce incorrect dose profiles in the presence of a magnetic field.^{13, 14} Ionization chambers can be characterized and corrected with respect to magnetic field strength,¹⁵⁻¹⁸ but are inherently point measurements that lack spatial resolution required to characterize bulk dose field effects such as those introduced by the ERE.

Two-dimensional dose distributions can be measured with radiochromic films with reasonable accuracy, but require strict physical and temporal protocols for irradiation and readout; films have been used to measure ERE in limited scenarios.^{19–21} Gel-based 3-D dosimeters, such as PRESAGE® radiochromic plastic and FOX gels, are insensitive to magnetic field effects;^{22–24} however, they are not generally regarded as user-friendly, utilize toxic materials,²⁵ and limited literature is available in regards to using them to quantify ERE with respect to commissioning MR-IGRT treatment planning system (TPS) data.²⁶

To be prudent in clinical treatment with MR-IGRT systems, rigorous validation of simulated ERE results is necessary. This study employed a novel method to observe and quantify the ERE spatial patterns through optical imaging of scintillation and Cherenkov emission. The experimental results are compared to TPS and radiochromic film results in an attempt to substantiate the predicted scope and severity of dose variations due to this effect. Cherenkov emission has previously demonstrated its potential for radiotherapy beam dosimetry in an MR-IGRT system in water phantoms.²⁷

2. Materials and Methods

2.1. MR-Linac and TPS

The Viewray MRIdian (Viewray, Cleveland, OH) MR-linac and its commissioned treatment planning system (TPS) were studied in Washington University in St Louis Radiation Oncology. All irradiations from the Viewray were 6MV flattening filter free (FFF) beams, delivered with set dose rate of 650MU/min, for a total of 200MU; the only exception was the film Phantom B, which was irradiated with 400MU. The primary magnetic field of the system was measured to be 0.345T.

Each phantom was scanned in a CT simulator at a resolution of 0.90mm x0.90mm pixels with 0.8mm slice width. Individual treatment plans were created for each irradiation scheme using the Viewray TPS, and the associated dose volume distribution data were exported in DICOM format for analysis in MATLAB environment (Mathworks Inc., Natick, MA).

2.2. Phantom Design

The two phantoms, shown in Figure 1, were fabricated with air cavities of varying sizes to induce the ERE at different intensities, where the larger cavities are expected to be more severely affected. Acrylic was chosen as the material for the phantoms for the comparability of its density to water ($\rho=1.19 \text{ g/cm}^3$, $\rho_e=3.86 \times 10^{23} \text{ cm}^{-3}$, $Z_{eff}=6.47$),^{28, 29} the ease of machining air cavities in the solid material, and optical transparency. Irradiated acrylic emits an optical signal as a result of predominantly scintillation, with additional light contributions from the Cherenkov Effect, the latter often regarded as a challenging contaminant signal in the field of absolute dosimetry with plastic scintillating fiber dosimetry.^{30–32} Both light sources are known to be linearly proportional to deposited dose, although Cherenkov emission exhibits more energy-dependence.^{33–38} However, by using camera systems to image volumes as opposed to the point measurements of fiber-based dosimeter systems, a relative dosimetry approach is possible, diminishing the need to separate the two signals.

2.3. Phantom A: Five Cavity Phantom

The first phantom studied was a 40cm × 10cm × 10cm block of clear acrylic milled with 5 holes of increasing diameter (0.5cm, 1cm, 2cm, 3cm, and 4cm), shown in Figure 1(a). The top apex of each circular cavity was aligned 3cm below the surface, so that the beam attenuation would be equivalent in each case before reaching the hole. The interiors of all cavities were painted black to eliminate optical reflections or light channeling effects. All faces, except the front optical face, were covered in optical blackout tape (T137–2.0, Thorlabs, Inc., Newton, NJ, USA).

The air cavities were spaced so that each could be independently irradiated by an 8.3cm × 8.3cm beam, with the top face of the phantom set to SSD=95cm. The inner 3 cavities were individually aligned so that the center of the circular holes were coincident with the lateral treatment isocenter. Due to the physical constraints of the MRI bore, the outer two cavities (4cm and 0.5cm diameter) could not be aligned with the treatment isocenter. The remaining offset between the center of the outer two cavities and the optical imaging axis lead to partial obstruction of the optical image, therefore, we excluded the outer two cavities from our analysis.

2.4. Phantom B: Three Cavity Phantom with Film

The second phantom studied was a smaller acrylic block with 3 milled cylindrical cavities (3cm, 2cm, 1cm), that was split into two opposing blocks so that a piece of radiochromic film (EBT3) could be inserted. The film was situated in the transverse plane of the MRI bore, perpendicular to the optical axis of the camera, so that it produced an image of dose in the same plane as the optical imaging system. A single 24cm × 4.15cm beam was used to irradiate all three cavities simultaneously. The top apex of each cavity was 5cm below the acrylic surface for this phantom, compared to the 3cm depth of the cavities in Phantom A.

2.5. Camera System and Imaging Setup

As shown in Figure 2(a), a 1200×1600 pixel intensified complementary metal-oxide-semiconductor (ICMOS) camera system (C-Dose, DoseOptics LLC., Lebanon, NH, USA) with an attached Canon 135mm lens set to f/4 was aligned at the foot of the treatment couch, just behind the 5-G line of the MRI, approximately 4.25m away from treatment isocenter. The camera height was adjusted so that the optical axis was at the same height as the treatment isocenter with respect to the floor. All room lights were turned off or covered during acquisition to diminish background signal and eliminate any possible effects of glare.

C-Dose Research Software (DoseOptics LLC., Lebanon, NH, USA) was used to acquire the image data. A remote trigger device that detects the presence of radiation in the room was used to synchronize the optical image acquisition to the radiation pulses, without the need for a physical interface with the Viewray linac components, using a 6μs exposure duration per detected radiation pulse.³⁹ Background images were acquired using the same settings with an internal trigger module, without the presence of the radiation. The internal camera hardware was set to perform a temporal median filter across 5 subsequent images during readout, removing most of the stray radiation noise in the images.

The spatial resolution of the experimental setup was measured using a custom test target with known feature sizes positioned at the treatment isocenter, shown in the right half of the components in Figure 2(b). To ensure the depth of field matched the sizes of the imaged phantoms, this test target was also imaged at 5cm increments to range -15cm to +15cm. An additional test of depth of field was performed using the high frequency pattern in the left most portion of Figure 2(b). The pattern was aligned at a 45° angle with respect to the treatment couch, to span a 14cm depth along the optical axis, and analyzed for changes in light and dark contrast between line pairs to indicate level of focus in the imaging field along the optical axis.

2.6. Control Experiment Imaging

To validate the approach, Phantom A was also irradiated and imaged on a Varian TrueBeam linac (Varian Medical Systems, Palo Alto, CA, USA) with 6MV FFF beams and dose rate of 600 MU/min to mimic the MR-linac beam without the presence of the magnetic field. The same ICMOS, lens and acquisition setup described in section 2.5 was used, placed at the foot of the treatment couch approximately 4.25m away from treatment isocenter.

The 3cm, 2cm, and 1cm diameter cavities were aligned to the treatment isocenter, with the top surface of the phantom at SSD=95cm. The phantom was irradiated with 200MU of a 8.3cm × 8.3cm beam. The phantom was then aligned so that the beam was directly between the 3cm diameter and 2cm diameter holes, and irradiated and imaged with an 8.3cm × 8.3cm beam. The Eclipse TPS (Varian Medical Systems, Palo Alto, CA, USA) was used to generate dose volume distribution data for both scenarios using the Acuros XB dose calculation algorithm. The Acuros XB algorithm was chosen over the Anisotropic Analytical Algorithm (AAA) due to the established superior dosimetric performance in literature, particularly when air gaps are included in the irradiated volume.⁴⁰ Finally, the optical images were compared to the simulation results in MATLAB.

2.7. Image Post Processing

MATLAB was used for all post-processing of optical images and TPS data. Each test setup of 200MU delivery produced stacks of approximately 175 frames. A rolling minimum intensity filter, taking the minimum value between the five collocated pixels in each group of five temporally subsequent frames, was applied to remove any residual stray radiation noise manifesting as artificially hot pixels. The stack of images was then summed into one composite image, and normalized to a flat field image of the camera response under uniform illumination. An edge-preserving bilateral filter with Gaussian half-width of 11 pixels was applied to each composite image to smooth remaining noise.

For phantom A, the average intensity calculated in a 6×6 pixel region of interest at the depth of maximum dose ($d_{\max} \approx 1.4\text{cm}$) was used to determine the normalization factor for each image. The dose volume data from the TPS were imported into MATLAB, and all slices were summed into a single image along the optical axis for comparison to the optical images, to parallel the concept of the optical image being a summed projection of the dose along the optical axis.²⁷

Conversely, for comparison with film results (Phantom B), a planar TPS dose slice adjacent to the central axis slice was chosen; the actual central slice was avoided due to the added uncertainty of the physical interface of the two slabs of the phantom. An additional difference in the analyses of Phantom A and B is the selected normalization point, because the radiochromic film had to be trimmed to fit within the constraints of the metal securing pins, and subsequently film dose data begins at depths just beyond $d_{\max}$. Therefore, the geometric region at 2.5cm phantom depth and approximately 4mm to the right of the metal pin edge was chosen as the dose normalization point for all Phantom B analyses, to remain close to the central axis of the treatment beam without overlapping the obstruction.

3. ERE Imaging Results

3.1. Control Imaging of Phantom A

Both the TPS (Eclipse, Acuros algorithm) generated integrated dose images, and the experimentally measured optical images for the four control cases of Phantom A irradiation on the Varian TrueBeam linac, without magnetic field present, are shown in Figure 3. Note the TPS models the dose in air, while the experimental images are insensitive to air dose.

The projection percent depth dose curves (pPDDs) were extracted from each image by taking the intensity profiles along the central vertical axis down the height of the phantom.²⁷ In addition, the pPDD tangent to each cavity at a 5mm lateral shift from each cavity edge was plotted and compared for each case; when the area between two cavities was irradiated, this plot reflects a lateral shift 1cm from the central pPDD axis. Finally, the projection cross beam profiles (pCBPs) at the depth of maximum dose ($d_{\max}$) were obtained from each image by taking a horizontal line plot. These plots are presented in Figure 4.

Given the expected variation between the TPS and optical pPDDs at the location of the air cavity, error was predominantly studied for the fourth case (Figure 4(d)), where the phantom was irradiated in the solid area between the 3cm and 2cm diameter cavities. This error plot is presented in Figure 5, with error statistics summarized in Table 1.

3.2. Phantom A MR-Linac Results

To observe the ERE, Phantom A was irradiated on the Viewray system by a 6MV FFF beam in the presence of the 0.345-T primary magnetic field of the MR-linac. The TPS simulation projection dose images for the 1cm, 2cm and 3cm cavity cases are shown in the top row of Figure 7, and the respective experimental optical images are shown in the bottom row of Figure 7. Each image is normalized to the dose or light intensity at the depth of maximum dose. Again, the TPS models the dose in air, while the proposed imaging method, as expected, cannot detect dose within the cavity. Unfortunately, the size of the bore with respect to the size of the phantom prohibited direct, isocenter-aligned irradiation of the 0.5cm and 4cm cavities; these data sets are not shown.

Line profiles at three areas of interest were compared to evaluate the differences between the TPS and experimental data sets. First, the pCBPs for each cavity, drawn as the horizontal profile containing the center of each cavity, are shown in Figure 7 (a). Next, the pPDDs, taken as the vertical profiles through the centers of the beams and cavities, are presented in

Figure 7. Finally, the pPDD curves were captured in an area of expected dose escalation due to the ERE, 2mm to the left of each cavity; these profiles are shown in Figure 7 (c).

3.3. Phantom B MR-Linac Results

The second acrylic phantom, Phantom B, was irradiated on the MR-linac with a piece of EBT3 radiochromic film sandwiched between the two halves to produce a film reading (Figure 8(b)) in the same orientation as the optical image (Figure 8(c)). The optical image was acquired during a second irradiation with the same setup and beam, without the film in place. The results are compared to the TPS generated image of the central axis planar dose (Figure 8(a)), as well as the TPS projection dose. Each of the figures are independently normalized to the same geometric region in the phantom, at a phantom depth of 2.5cm and approximately 4mm to the right of the edge of the metal pin.

Figure 9 compares the three modalities (TPS 2D dose distribution, radiochromic film, and the proposed optical imaging technique. Figure 9 (a) shows the registration between the film and the TPS image, with drawn lines to designate the locations of the four subsequent 1-dimensional percent depth dose (PDD) and projection percent depth dose (pPDD) line plots, as well as two cross beam profile (CBP) or projection cross beam profile (pCBP) plots, Figure 9 (c–h). Note that the radiochromic film results should be compared to the planar TPS data, while the optical images are more intuitively compared to the projection TPS dose images. All image intensities were normalized to the dose reported at the same geometric small region of interest above the cavities (2.5cm phantom depth), roughly 4mm to the right of the metal pin.

A 2D comparison between the processed film dose values and the planar dose from the TPS isodose volume was conducted using gamma-index analysis with a 3%/3mm criteria, shown in Figure 9 (b) with a passing rate of 88.3%. All pixels shown in this image were included in the passing rate calculation; pixels with $\gamma \geq 1$ (blue to white in the color scale) are regarded as passing.

4. Discussion

ERE has traditionally been a difficult phenomenon to measure, because it is of most interest at solid-gas interfaces, where the electrons are permitted to move and be manipulated by the magnetic field after they are liberated from the solid medium and ejected into a gas medium. Surveying the entire area with point measurements is fundamentally challenging, because any solid-gas phantom arrangement would restrict the possibilities of physical placement of a point detector with its own inherent footprint. This has limited the ability to rigorously, quantitatively validate the simulation results modeling magnetic field effects on radiation dose deposition.

Optical imaging of scintillation and Cherenkov light to observe the ERE is the novel idea proposed here, with the attractive asset that the irradiated medium itself is the generator of the detected signal, measured remotely in real time. The optical images were compared to the TPS simulation results of the integrated dose along the optical axis. In addition, the reported dose from radiochromic film is compared to a single plane of dose near the

interface, to assess the strengths and limitations of the proposed method in comparison to the most readily available alternate dosimetry tool.

The control experiment was conducted on Phantom A to evaluate the accuracy of scintillation as a surrogate for dose in the given experimental setup without the magnetic field effects, where TPS dose simulations are inherently trusted. As expected, given the fact that the TPS is able to model dose deposited in air while the intensity of scintillation in air is much too low to be detected in the given scheme,⁴¹ the major deviation between the two results occurs within the air cavities (Figure 3 and Figure 4). This is because the air itself does not physically generate optical photons through scintillation and Cherenkov emission at the same rate as the solid acrylic (or water). While it is possible to image air scintillation, much longer exposures are necessary,⁴¹ and therefore only the optical signal from the solid acrylic can be regarded for relative dosimetry in this application. The air cavity intensity must consequently be disregarded in the optical images when comparing them to the TPS dose images.

An interesting result here is the trending drift of increased error with depth. The error plot is shown in Figure 5. While the average error is still relatively low over the entire depth (1.65%), it becomes more pronounced after depth of 6cm. In fact, over the first 6cm, the average error is only 0.48%, spanning $\pm 2\%$. One possible reasons for this depth-deviation which will be investigated further is beam hardening.

Beam hardening is a likely cause of the discrepancy at larger depths, since the phantom is coated on the outside with optically black tape and paint. Optical photon stimulation is to a degree dependent on beam energy, whereby higher energy x-rays create more optical photons than lower energy x-rays. These results are consistent with the beam hardening conclusion, given the depth-dependent errors manifest as larger in the scenarios where more acrylic is in place to filter the radiation beam (i.e. the no/small cavity cases). The error is less prevalent for the large cavity cases, where less beam energy filtering is occurring since more of the beam is passing through air.

Given the physical differences between the interfaces of the phantom at different cavity sizes, another possible source of the error at depths is the change in scatter for the x-ray photons, secondary electrons, and possibly the optical photons. Air gaps of different sizes would affect the charged particle equilibrium to different extents, with the larger cavities more severely affected. Further experiments would be required to more accurately isolate the cause.

In the presented setup, the dosimetric confidence of the proposed method in this phantom is the highest for the first 6cm. To this end, the cavities of Phantom A were placed 3cm below the surface, and end at or before the 6cm mark. Phantom B did not have this design flexibility, and the cavities are 5cm below the top surface, and therefore results from this phantom must be considered accordingly.

An additional observation from the control experiment is the low intensity ring in the immediate vicinity of the solid-air interface. This is thought to be an optical artifact originating from the inner surfaces of the holes, which were painted black and therefore

absorbing some of the generated light. This causes the holes to appear slightly larger in the pPDD plots than the known physical size of the cavity, but otherwise can be acknowledged and disregarded as a spatially restricted artifact. This, combined with the known issue of accuracy at larger depths, prompts the primary focus of this study to be the areas of escalated dose localized near the superior edges of the cavities, and not the mirrored areas of decreased dose concentrated on the inferior sides of the cavities.

A final note about the control experiment is that because it was carried out on a different machine, with a different physical beam than the MR-linac and slightly different setup, direct quantitative comparison between the data sets presented in Figure 3 and Figure 6 is not inherently valuable, as each dataset follows a relative dosimetry paradigm, rather than definitive absolute dosimetry. This is due in part to expected variations in the beam quality (i.e. energy spectra, to which Cherenkov production is dependent) of the Truebeam-produced 6MV FFF beam versus the ViewRay 6MV FFF beam. The physical beam differences would produce different responses in optical photon production, which therefore cannot be compared directly unless a beam quality correction factor were applied, which to date has not been investigated for the novel dosimetry methods proposed. In addition, the distance between the camera and the phantom varied between the two data sets to accommodate different room and treatment couch geometries, indicating a need for an imaging distance correction factor should direct comparison be required. Therefore, the two data sets are more appropriately regarded as independent relative measurements. A better control would be a comparison on the ViewRay with and without the magnetic field active, however, such a measurement was not immediately possible.

The ERE results of dose measurement in the presence of the magnetic field are qualitatively convincing for both phantoms. The differences between the optical and integrated dose TPS images for Phantom A (Figure 7) suggests that it is possible the TPS is under-reporting the extent of dose escalation due to ERE, as well as the localization of the dose escalation, and is in need of further investigation. This is particularly true for the larger cavities. The pPDDs measured 2mm to the left of each cavity (Figure 7 (d)) show that while the 1cm cavity results between the TPS and optical imaging technique largely agree, the dose escalation differs by 10% and 20% for the 2cm and 3cm cavities respectively.

Likewise, the pCBPs presented in Figure 7 (c) also reinforce this observation. The TPS dose simulations model similar magnitudes of dose escalation, however it is localized within the air cavity, which logically starts where the optical and TPS curves intersect at the two points closest to the center of the beam (0cm point). The dose most pertinent to clinical implementation, however, is the dose deposited within the solid. The optical images suggest that this dose is instead localized in the solid rather than the air, raising concern that the TPS may be misrepresenting the actual dose to tissue in clinical cases.

Data from Phantoms B includes reference to film data to more completely establish the differences between the TPS and experimentally measured intensities to assess their validity. However, the film data is not without its own weaknesses. There are pronounced artifacts from the metal pins used to hold the two acrylic blocks sandwiched together; future experiments would benefit from acrylic or nylon pins instead. Great care should be taken to

ensure the film is tightly situated between the two acrylic pieces without air gaps, and sits close enough to the phantom surface to include the depth of maximum dose in the image. The material interfaces (acrylic-film-acrylic), especially along the central axis of the beam path as in this scenario, can lead to disruptions of the ionized electron paths and therefore changes to the dose profile. Therefore, there is an expected effect of the presence of the film on the electron fluence in and near the cavities, meaning even film dosimetry for this application should be regarded with some reservations. This is a benefit of the optical method over the film, since the optical method allows for an intact phantom without additional heterogeneities to be measured.

There are demonstrated effects of magnetic fields on Gafchromic-film response.²⁰ Reynoso *et al.*⁴² used scanning electron microscopy to assess changes in the crystal orientation and polymerization within the active layer of radiochromic film after exposure to a 0.35T magnetic field of a similar MR-IGRT system. After comparing films irradiated with and without the magnetic field, the results of their work concluded there was a dose-dependent under response of the radiochromic film. However, at the doses used for film irradiation in this experiment, the dose discrepancy should be limited to only ~2%, meaning there should be limited influence of this effect for the given dataset.

Finally, the film was also intact within the air cavities, causing hot and cold spots within the air cavity on the film readout. This makes it a nontrivial task to isolate the true boundaries of the air cavity on the film. Future film experiments should implement additional fiducials coincident with the film and the phantom to properly validate image registration.

Despite these challenges with radiochromic film dosimetry, consideration of the results from the Phantom B experiment show similar degrees of agreement between the film and planar TPS dose, compared to the optical image and projection TPS dose, when considering the solid sections of the phantom (i.e. Figure 9 (c), and areas around the air cavities). This reinforces the validity of the optical results obtained for Phantom A. The fact that the optical image is an aggregate representation of the ERE along the width of the phantom down the optical axis provides more information than can be feasibly captured with film dosimetry. Overall, optical imaging of air cavity phantoms is a useful tool in the further investigation of the accuracy of TPS dose calculations in the presence of magnetic fields.

Validation of this approach with an externally validated Monte Carlo simulation model is highly desirable, but not currently available.

5. Conclusion

In this work, the electron return effect (ERE) was imaged for the first time in an MR-linac using a novel optical method imaging system, from a combination of scintillation and Cherenkov light in clear acrylic phantoms. This method allows for relative 2-D longitudinal measurement of dose distributions along the MR bore without the need to perturb the beam and solid-air interface with any detector. There are some observed artifacts that can be accounted for, such as the inability of the optical method to measure the dose within the air cavities. Regardless, the method is promising to provide dosimetric value to quantifying the

spatial heterogeneities in dose from the ERE, and suggests that the dose escalation due to ERE at interfaces may be underestimated by a currently implemented treatment planning system. Further validation of the ViewRay TPS along with this experimental method are warranted through future Monte Carlo simulation studies.

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Figure 1.
a) Acrylic phantom A: 5 cylindrical cavities in a solid block; b) Acrylic phantom B: 3 cylindrical cavities and film in the center of the two part block.

Figure 2.

a) Experimental setup with ICMOS at foot of treatment couch, looking down the MR-linac bore to the phantom; b) cropped and scaled resolution and depth of field test targets.

Figure 3.

Phantom A control experiment, without magnetic field. Top row: Eclipse TPS integrated dose images (Acuros algorithm); Bottom row: optical images for each test case. All images are independently normalized to the dose at $d_{\max}$.

Figure 4.

Control experiment line plots for the same sized beams impinging the: a) 1cm cavity, b) 2cm cavity, c) 3cm cavity, and d) the solid phantom area. Column 1 presents the pPDDs through the center of each beam. Column 2 shows the pPDDs taken 5mm to the left of each cavity in a)-c), and 1cm to the left of the central axis in d). Column 3 contains the pCBPs at the d_{max} .

Figure 5.
Percent error between the optical and TPS pPPDs in Figure 4 (d) for depths 3–90mm.

Figure 6.

Top Row: Projection view images of TPS dose predictions of ERE resulting from 1cm, 2cm, and 3cm cavities in Phantom A, normalized to the dose at $d_{\max}$. Bottom Row: associated optical images for each respective cavity irradiation in the presence of the magnetic field, normalized to the dose at $d_{\max}$.

Figure 7.

a) pCBPs through the center of each cavity comparing TPS and optical results for each cavity size; b) pPDDs through the central axis of each cavity, containing $d_{\max}$ comparing TPS and optical results; c) cavity-tangent pPDDs, or the vertical dose profiles at a 2mm lateral shift from the left cavity boundary, comparing TPS and optical results.

Figure 8.

Phantom B with magnetic field present. a) TPS planar dose simulation; b) readout of radiochromic film irradiated while within phantom; c) optical dose measurement (film was not inside the phantom at the time of optical acquisition). Dark artifacts in the optical image are the result of the tape (bottom corners) and metal pins used to secure the phantom pieces together.

Figure 9.

Projection (optical) and planar (film) PDD and CBP plots comparing the results of the three modalities, optical imaging, TPS dose prediction, and film measurement. a) Registered film (magenta/gray) and TPS (green) planar dose images, with colored line plots designating the physical locations of the subsequent plots (c-h); b) gamma index image between the registered film dose image and a TPS dose plane adjacent to the film; c) projection (optical) and planar (film) CBPs at 2.5cm phantom depth (green horizontal line); d) projection (optical) and planar (film) CBPs intersecting all three cavities at 5.5cm phantom depth (blue horizontal line); e) projection (optical) and planar (film) PDDs directly to the right of the metal phantom alignment pin (white line); f) projection (optical) and planar (film) PDDs 2mm to the left of the 3cm cavity (yellow line); g) projection (optical) and planar (film) PDDs 2mm to the left of the 2cm cavity, intersecting the metal pin (magenta line); f) projection (optical) and planar (film) PDDs 2mm to the left of the 1cm cavity (cyan line).

Table 1.

Statistical summary of errors plotted in Figure 5.

Depth Span	Average	Error (%)		
		Maximum	Minimum	Standard Deviation
3–60mm	0.5	1.9	–1.7	0.8
3–90mm	1.7	4.6	–1.7	–1.8