4D scintillation dosimetry for the MRI-linac: proof of concept

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Abstract. A new method of time-resolved volumetric (4D) dosimetry combining transversal projected view scintillation imaging with the multi-leaf collimator (MLC) geometry information is presented and demonstrated in a magnetic resonance (MRI) guided linear accelerator (linac). The setup consisted of a time gated intensified camera and a cylindrical plastic scintillator phantom. Positioning the camera outside the 0.35 T magnetic field suppresses the interference between the MRI-linac and dosimeter camera. Transversal view images of the scintillation light were recorded at 20 Hz framerate and the light distribution along optical axis was decoded from the MLC data by Fourier algorithm. Considering scintillation light as dose surrogate, the dose volume was reconstructed with sub-millimeter resolution, and this was tested on an intensity modulated delivery of a TG119 C-shape plan. 3D gamma analysis of the recorded cumulative dose volume as compared to a Monte-Carlo simulation reported 95% pass rate at 3%/3mm criteria. By enabling the use of measurement-based 3D beam comparison metrics, the presented method may provide a comprehensive solution for volumetric end-to-end dosimetry and fast machine performance checks in this challenging environment of an MRI-linac.

1. Introduction
Novel magnetic resonance (MRI) guided radiotherapy systems offer very sharp dose gradients (<2 mm 20%-80%) and the possibility of adaptive re-planning [1,2], allowing a highly conformal beam delivery. This approach may enable not only an unprecedented tissue localization, but for certain cases also the use of hypofractionation, leading to fewer patient visits and improved clinical outcomes [2]. Unsurprisingly, the fusion of an MRI scanner with a linear accelerator has brought unique challenges to dosimetry practice [3]. The strong magnetic field impedes quality assurance (QA) efforts by inducing angular-dependent corrections in ionization chambers and semiconductor dosimeters, as well as by introducing MRI-compatibility requirement on the used equipment. The latter issue basically rules out actuator-based point scanning systems, leaving hand-actuated systems [4] and film dosimetry as the remaining options. A large effort has been invested into alternative techniques, including gel dosimeters [5] and MRI-compatible 2.5D diode arrays [4,6,7]. Non-trivial response calibration and handling of the gel dosimeters or additional calculation tools in the case of 2.5D diode array may, however, introduce other sources of significant uncertainties. Spatial resolution of the 2.5D diode array is also generally
Both end-to-end dosimetry as well as fast machine performance checks would benefit from a stand-alone dosimeter that is MRI-compatible and minimally intrusive to the scanning sequence, MRI tissue equivalent (e.g. water-based), sensitive (equivalent noise level <3 cGy), able to record or reconstruct a 3D dose distribution, and with spatial resolution high enough to resolve steep dose gradients (<1 mm³). Especially for end-to-end dosimetry, 3D imaging capability would allow analysis of dose volume histograms (DVH) in addition to standard gamma analysis [8] thereby enabling direct assessment of the clinical impact of dosimetric errors [9]. Based on earlier research of Cherenkov [10] and scintillation dosimetry [11,12] imaging, we developed a 3D dose imaging system that combines projection view scintillation imaging of a transverse dose distribution with orthogonal (cross-beam profile) geometry information. In [10] we demonstrated this technique using a portal image for decoding the 3D dose distribution. As the portal imagers are not available on all MRI-linac systems, our presented approach uses multi-leaf collimator (MLC) geometry information obtained from machine log files. In combination with “traditional” transverse projectional imaging of dose-induced visible scintillation in a cylindrical phantom, we demonstrate for the first time a 4D (3D + time) reconstruction of dose delivered by MRI-linac using a complex intensity modulated radiation therapy (IMRT) plan.

2. Materials and methods

2.1. Experimental setup

Our dosimetry technique consists of three basic components: a) a scintillating phantom, b) an intensified camera with long focal distance lens; and c) a reconstruction algorithm that used camera data stream and MRI-linac log files to reconstruct 3D dose. The initial experiment was performed on a 0.35 T MRI-linac (MRIdian, ViewRay, Mountain View, CA), and the actual setup is shown in the Fig. 1. A cylindrical plastic scintillator (Ø5” ×2.5”, Bicron BC412, Saint-Gobain Crystals, OH) was aligned such that the center of gravity coincided with the isocenter, and the base was parallel with the rotational axis of the linac gantry. An intensified complementary metal oxide semiconductor (iCMOS) camera (C-Dose, DoseOptics, Lebanon, NH) was positioned at the distal couch base, optical axis coinciding with bore axis of the linac. The intensifier of the iCMOS camera was gated to the linac pulses using a stray X-ray point detector with fast transient response time [13]. A total of 5220 scintillation image frames at 20 Hz framerate were captured upon delivery of a C-shape plan for TG119 using a 6 MV flattening-filter free beam. Linac log files recorded MLC positions, beam on-off time, fractional dose, and gantry angle, with temporal resolution of 10 μs.

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\text{Figure 1.} \ a) \text{ experimental setup showing iCMOS camera setup at bore axis and focused on target at the isocenter; b) montage of selected beamlets as imaged by the camera at 20 fps, 5220 frames total (800cGy delivery).}
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2.2. Reconstruction algorithm

The reconstruction algorithm relies on an approach developed earlier for 3D electronic portal imaging device (EPID)-Cherenkov dosimetry [14]. We establish our analysis in assumption that the scintillation image is a surrogate of the dose distribution, projected along optical axis and scaled by calibration constant

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d_{\tau} = a I_{\tau}
\]
For clarity, it is assumed that the scintillation image is free of any optical aberrations and scatter due to the phantom re-absorption.

After extracting necessary beam information from linac log files, the reconstruction begins with creating a geometrical projection of the MLC shape across the phantom volume, yielding beam volumes $m$ for each individual field. A transverse projection MLC image $m_T$, similar to the transverse scintillation image $I_T$ but lacking any information about depth dose distribution and penumbra, is created by summing the voxel intensities of the MLC shape volume in the direction of the optical axis of the camera:

$$m_T = \sum_v m(x, y, z, t)$$  \hspace{1cm} (2)

In the case of MRI-linac, the coordinate space used here ($x,y,z$ for real space, $u,v,w$ for corresponding spatial frequency space, and $t$ for time) corresponds to IEC61217 gantry coordinate system. The dissimilarity between projection MLC image $m_T$ and scintillation image $I_T$ can be expressed for each cross-beam profile across axis of the beam $z$ as a projectional correction kernel $\hat{s}_T$:

$$\mathcal{F}_x\{I_T\} = \hat{s}_T(u, z) \mathcal{F}_x\{m_T\}$$  \hspace{1cm} (3)

Where $\mathcal{F}$ represents fast Fourier transform and the subscript $x$ indicates the independent variable over which the transform is applied. We follow this notation in the rest of this manuscript. Equation 3 also indicates that the depth dose distribution is captured by the magnitude of correction kernel, and specific penumbra shape is represented by the spectral shape of $\hat{s}_T$.

Until now, we were operating exclusively with the transverse projection images. Assuming circular symmetry of the dose distribution kernel (isotropic, homogeneous phantom), we can expand the projectional correction kernel $\hat{s}_T$ to the third dimension (along optical axis of the camera, $y$-coordinate) by exploiting principles of the projection-slice theorem and Fourier-Hankel-Abel (FHA) cycle. As $\hat{s}_T$ represents a projection of a circularly symmetric 2D kernel for each slice along beam axis, we can reconstruct the 3D correction kernel from $\hat{s}_T$ by applying the FHA cycle [14]:

$$\hat{s}(u,v,z,t) = H_{x,\gamma} A^{-1}_x F^{-1}_u \left[ \frac{\mathcal{F}_x\{I_T\}}{\mathcal{F}_x\{m_T\}} \right]$$  \hspace{1cm} (4)

Where $HA^{-1}_xF^{-1}_u$ represents a series of zeroth order Hankel, inverse Abel, and inverse Fourier transforms, respectively. Again, subscripts of the transform symbols indicate independent variables over which the transforms are applied. The final dose volume is then calculated as partial convolution of the 3D correction kernel with the MLC-based beam volumes $m$:

$$d(x,y,z,t) = aF^{-1}_{u,v}\{\hat{s} m\}$$  \hspace{1cm} (5)

2.3. Image post-processing

The image post-processing and dose reconstruction was implemented in Matlab (Mathworks, Natick, MA) environment. First the image dataset was corrected for the optical scatter and divided into image sub-stacks for each beam field. Timestamps of all frames were shifted to match the MRI-linac log file data. Based on gantry angle and MLC shape information from the log files, an array of MLC-based beam volumes was created with resolution of 0.5 mm. Coordinates of MLC data and scintillation data were matched and scaled according to the preceding optical calibration, followed by the 3D analysis. Resulting 3D dose distribution movie was summed in time to get cumulative dose distribution. Finally, we performed gamma analysis-based validation of the measured data with Monte-Carlo based treatment
planning system (TPS), and the results were rendered using The Visualization Toolkit (VTK, Kitware, NY) software system.

3. Results and Discussion
The imaged cumulative dose distribution (represented by cumulative scintillation image, Fig. 2a) matches well the transverse projection view distribution calculated by the Monte-Carlo based treatment planning system (ViewRay, Mountain View, CA) (Fig. 2b). 3%/3 mm gamma test success rate for the scintillation image as compared against projected TPS image data was 96% for the pixels with intensity over 10% of the maximum intensity value. Importantly, comparing the reconstructed 3D dose to the TPS calculated dose volume, 3%/3 mm gamma test success rate was 95% for the voxels receiving over 10% of the maximum dose.

![Figure 2](image)

**Figure 2.** a) Cumulative dose distribution calculated by ViewRay TPS, summed across the 57 mm depth of the scintillator; b) measured cumulative dose distribution. Both dose distribution profiles are normalized. c) Result of gamma analysis, indicating 96% pass rate at 3%/3mm criteria and 10% threshold.

![Figure 3](image)

**Figure 3.** a) Volumetric rendering of the reconstructed cumulative 3D dose volume; b) 3D distribution of gamma values comparing reconstructed cumulative dose and TPS dose volumes.

4. Conclusion
We presented a novel 4D dosimetric system that is capable of reconstructing time-resolved dose volumes from projections of dose (scintillation images) and cross-beam profile information based on logged MLC configuration data. The innovation of this study lies in enabling fully volumetric dose imaging with high spatial (0.5 ×0.5 ×0.5 mm³) and temporal (50 ms) resolution. The system is MRI compatible, as demonstrated on a proof-of-concept measurement of a dose delivered by 0.35 T MRI-guided radiation
therapy linac. Gamma analysis revealed a good match between standard C-shape plan prepared for TG119-based commissioning. 2D and 3D 3%/3mm gamma test rates were 96% and 95%, respectively, indicating an elevated region in one of the lobes of the planning target volume. The source of discrepancy will be studied further. Following the current proof-of-concept study, an investigation of the accuracy of the presented technique is warranted. Similarly, the phantom composition will be optimized for minimal optical scatter, water equivalency, and MRI detectability. The presented system potentially offers a comprehensive solution for volumetric end-to-end dosimetry as well as a rapid machine performance assessment not only for MRI guided therapy, but also for standard C-arm and tomotherapy systems.

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6. References
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