Utilizing Pencil Beam Scan Dynamics and a Scintillation Screen to produce 3D Dose Distribution of Proton Beams

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Abstract. This study describes a method for producing volumetric dose profiles of proton beams from 2D slices of optical scintillation images. The method relies on a high frame rate camera acquisition (100 frames per second), the spot scanning capabilities of current proton pencil beam scanning systems, and a water equivalent scintillation screen. The acquired slices are corrected for optical blurring and ionization quenching and stacked to produce volumetric dose distribution. The volumetric optical dose profile had a pass rate of 98.3% for 2%/2mm local gamma analysis, suggesting the method can accurately measure dose profiles. The method can potentially image all clinical proton beams for pencil beam scanning systems and can extend to imaging patient plans, with further verification.

1. Introduction
In proton therapy, validation of dose distribution is vital to verifying the irradiation system can treat patients and deliver the prescribed plan, particularly due to the significant dose deposition at the Bragg peak. There has been an interest in measuring precise three-dimensional (3D) dose profile of proton beams and scintillation and radioluminescence imaging has shown promising methods of acquiring volumetric dose distributions [1-3]. Optical imaging inherently provides two-dimensional (2D) dose profiles and require reconstruction methods to produce a 3D dose distribution. Some methods include multiple acquisitions of 2D radioluminescence slices of the dose profile [1] or imaging the scintillation projection dose profiles from orthogonal views with multiple cameras [3].

In this study, proton pencil beam scanning (PBS) dynamics produced a 3D dose profile while imaging optical emission from a scintillation screen in a water tank phantom. The imaging method required a single camera acquisition and a single dynamic treatment plan, where the beam scanned across the scintillation screen, for reconstruction of the 3D dose profile. We produced 2D profiles along the crossline of the treatment room from the frames in the acquisition. The optical properties of the scintillation screen and camera detector system (i.e. optical blurring, ionization quenching) were characterized to correct the acquired images. The post-processed dose profile was compared to Monte Carlo (MC) simulated dose profiles to determine the validity of the technique for quality assurance of proton beams.

2. Methods and Materials

2.1. Experimental Set up
We utilized the PBS system (ProBeam, Varian Medical Systems, CA, USA) for imaging dosimetry tests. The phantom, shown in Figure 1a, consisted of an acrylic tank \((15 \times 15 \times 30 \text{ cm}^3)\) with water and an organic scintillation screen \((27 \times 14.8 \times 0.1 \text{ cm}^3, \text{EJ240G screen, Eljen Technology, TX, USA})\). The camera imaged the scintillation screen through one face of the tank while the other four faces were covered by black matte plastic. The top face was not included in the phantom to prevent beam broadening or shift in range due to an obstructing slab of material [4]. The scintillation screen was aligned parallel to the imaged face and submerged in water with an edge aligned to the water surface. The phantom was aligned with the centre of the water surface at the isocentre and the scintillation screen along the crossline (XZ) plane. The intensified camera (C-Dose Research, DoseOptics LLC., NH, USA) used a 100mm lens (Canon Inc, Japan) and was positioned 3.25 m away from the isocentre with the optical axis aligned with the isocentre. The camera, capable of high frame rate imaging at 100 frames per second, acquired images of the scintillation in continuous mode with a 10 ms exposure time while a 100 MeV proton beam irradiated the phantom [5]. Cherenkov production in the water from the beam was negligible in comparison to the scintillation intensity from the screen [6,7]. The PBS steering magnet initially positioned the beam along the central axis for imaging of a 2D crossline profile. It then scanned the beam inline (+20 mm to -20 mm, 0.1 mm spot spacing, and 400 MU per spot), while the phantom was kept fixed, for acquisition of optical slices for 3D reconstruction of the dose distribution (described further in Section 2.4). Figure 1b shows example images acquired during acquisition of the scanned beam (and central axis profile).

**Figure 1.** a. Experimental set up for imaging a scintillation screen with the C-Dose camera focused on the scintillation screen along the XZ plane of the isocentre. Camera height and water surface height aligned to height of isocentre. b. Montage of example frames acquired while proton beam irradiated at the central axis (top left frame) and scanned axially across the scintillation screen, along the optical axis of the camera. c. Point spread function of the camera and scintillation screen detector system. d. Quenching correction factor data and fit (A, and kB fit values with 95% confidence) from optical images and MC dose

2.2. **Point Spread Function of the System**

The blurring of the optical profiles \((\text{Opt}_{\text{Uncorr}})\) from the images was due to the scintillation screen \((\text{PSF}_{\text{scr}}(x,z))\) and partially from the camera and lens system\(^2\) \((\text{PSF}_{\text{cam}}(x,z))\). The point spread function of the entire detection system \((\text{PSF}_{\text{sys}}(x,y))\) can be described as a convolution of each part \((\text{PSF}_{\text{sys}}(x,y) = \text{PSF}_{\text{scr}}(x,z) \otimes \text{PSF}_{\text{cam}}(x,z))\). \(\text{PSF}_{\text{sys}}(x,y)\), blurring of the entire system, was determined by exciting the scintillation screen with a UV lamp through a small pinhole and imaging the screen at the same distance from the phantom. The measured \(\text{PSF}_{\text{sys}}(x,y)\), shown in Figure 1c, was fitted to a triple gaussian function \((\text{PSF}_{\text{sys,fit}}(x,z))\). The optical intensity profiles along the central axis
and slices from the scan were deconvolved with the \( PSF_{sys, \text{fit}}(x, z) \) to remove blurring from to the detector system \((Opt_{\text{UnCorr}}(x, z) = Opt_{PSFCorr}(x, z) \otimes PSF_{sys, \text{fit}}(x, z))\).

2.3. Determine Quenching Parameters

The ionization quenching parameters were determined using the central axis 2D dose distribution profile. The quenching of the scintillations screen response, particularly near the Bragg peak, can be described using Birks’ Law [8]:

\[
Sc = \frac{A \cdot LET}{1 + kB \cdot LET} \phi \propto \frac{(A \cdot D)}{1 + kB \cdot LET}
\]

Where \( Sc \) is the scintillation response, \( D \) is dose delivered by radiation source, \( LET \) is the linear energy transfer, \( \phi \) is the particle fluence, \( A \) is scintillation efficiency, and \( kB \) is Birks’ or quenching constant.

2D equivalent of the methods described in Wang et al. were implemented to determine the quenching parameters \( A \), and \( kB \) [9]. The Monte Carlo (MC) dose (\( D_{MC} \)) and LET (\( LET_{MC} \)) profile were simulated on the Geant4 based TOPAS MC toolkit, using a validated beam model based on the commissioned beam energies of the Varian ProBeam System [10-13]. The simulations assumed the scintillation screen was water equivalent. The quenching correction factor (\( QCF \)) was the ratio of MC dose profile to the measured optical intensity profile. It was defined in a region of interest (\( 6 \times 9 \text{ mm}^2 \) ROI) around the Bragg Peak (i.e. region of greatest variation in \( LET(x, z) \)). The \( QCF_{data} \), shown in Figure 1d, was fitted to

\[
QCF_{\text{fit}} = \frac{1}{A} + \frac{kB}{A} \cdot LET_{MC}
\]

The optical image was shifted in the x and z direction to minimize the root-mean-square (rms) error of the fit and the shift with the minimum error provided the quenching parameters \( \delta z = -1.42 \text{ mm}, \delta x = +1.37 \text{ mm} \). The optical profile was corrected for ionization quenching using the fit and the following equation:

\[
Opt_{QCorr}(x, z) = Opt_{PSFCorr}(x, z) \cdot QCF_{\text{fit}}(x, z)
\]

2.4. Reconstruction of 3D Dose Profile

The frames acquired during scanning of the proton beam inline provided a temporal mean intensity profile (from an ROI of \( 2.5 \times 2.5 \text{ mm}^2 \) along the central axis at a depth of 20 mm, where LET was approximately constant). The profile was fitted to a Fourier series and the period was determined to be 98.54 ± .06 seconds (95% confidence interval). The period and the distance traveled over one scan (40 mm) provided the velocity of the scan, and the number of frames to accumulate for 1mm slices (246 frames) of the dose distribution. Intermittent delay between acquisition of consecutive frames exhibited negligible effect on the measurements [5]. The reconstructed 1mm slices were corrected for optical blurring and ionization quenching, determined in section 2.2 and 2.3, respectively.

3. Results and Discussion

The imaged central axis dose distribution (Figure 2a), after correcting for detector blurring and ionization quenching, matched well with the MC dose distribution with a 2%/2mm local gamma (10% threshold above the maximum dose) pass rate of 99.8%. The central axis depth dose curve, in Figure 2b, agreed relatively well with MC depth dose curve but showed some discrepancy at the entrance of the beam to about 30mm depth (max of 3.2% relative to maximum dose) and fall off at the distal edge of the Bragg peak. The method of scanning the beam across the scintillation screen for acquiring 3D dose distribution was stable because the corresponding frames in future cycles of the scan had a standard deviation of 2.2% (normalized to maximum intensity). The reconstructed 3D dose distribution, shown in Figure 3, agreed well with the MC dose distribution with a 2%/2mm local gamma pass rate of 98.3%.

The imaging method can potentially be improved by considering the optical and physical properties of the phantom. While the density of the scintillation screen is nearly water equivalent (1.023 g/cm³, \( 3.33 \times 10^{23} \text{ electrons/cm}^3 \)), the slight discrepancy can lead to a difference in stopping power compared to water and range shift [14]. The shift may be negligible due to the thin screen thickness of...
1mm, but a future study will incorporate the properties of the scintillation screen into the MC simulations, to quantify its impact on the proton range. While utilizing image processing methods described in Rahman et al., and correcting with the \( P_{\text{sys}}(x, y) \) addressed optical artefacts from the lens, and camera imaging system, optical distortion from the camera perspective and refraction within the phantom were not considered. At a viewing distance of 3.25m, the 100 MeV beam would result in a shift <0.5mm for the entire profile, but for higher clinically relevant proton energy (e.g. 240 MeV) there can be a shift of ~2 mm potentially requiring a correction for perspective. It is worth noting while this method can resolve volumetric profile for a pencil beam, for complex treatments its potential use would be in providing a 2D dose profile along the central axis.

**Figure 2.** Dose profiles (normalized to sum of dose above 10% threshold of max dose) of the imaged 100 MeV proton beam at the central axis. **a.** Crossline quenching corrected optically measured dose distribution and 2%2mm local gamma analysis (10% threshold above max dose) in comparison to the MC dose. **b.** Central Axis depth dose curves (MC simulated dose, and optical with no corrections, PSF correction, and quenching correction) and central axis MC LET.

**4. Conclusion**

A novel method was described here that utilized the PBS dynamics to reconstruct an accurate 3D dose distribution of clinical proton beams. A resolution of 1 mm³ was achieved and the dose distributions (both central axis and 3D) agreed with the MC dose distribution with a 2%/2mm local gamma pass rate of above 98% The central axis depth dose curve’s discrepancy at the entrance of the beam and at the distal edge of the Bragg peak need to be explored. The method assumed that the steering magnets scanning the proton beam inline over 40 mm do not significantly perturb the beam shape, which needs to be verified. The imaging technique needs to be verified for accuracy across the clinical proton beam energy range. This method provides fast acquisition of the volumetric dose profile and is dictated by the time of a single scan across the scintillation screen. The scintillation screen can potentially image central axis profiles of dynamic treatment plans and will require further validation (i.e. spread out Bragg peak).
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6. Disclosures
Brian W Pogue reports financial interest in DoseOptics LLC, a company developing cameras and software for the use of Cherenkov imaging in radiotherapy dosimetry.

7. References
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