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Performance characterization of a 3D liquid scintillation detector for discrete spot scanning proton beam systems

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Abstract

Existing systems for proton beam dosimetry are limited in their ability to provide a complete, accurate, and detailed account of volumetric dose distribution. In this work, we describe the design and development of a portable, fast, and reusable liquid scintillator-based three-dimensional (3D) optical detection system for use in proton therapy. Our long-term goal is to use this system clinically for beam characterization, dosimetry, and quality assurance studies of discrete spot scanning proton beam systems. The system used a $20 \times 20 \times 20\text{-cm}^3$ liquid scintillator volume. Three mutually orthogonal cameras surrounding this volume captured scintillation photons emitted in response to the proton beams. The cameras exhibited a mean spatial resolution of 0.21 mm over the complete detection volume and a temporal resolution of 11 msec. The system is shown to be capable of capturing all 94 beam energies delivered by a synchrotron and performing rapid beam range measurements with a mean accuracy of 0.073 ± 0.030 mm over all energies. The range measurement uncertainty for doses less than 1 cGy was found to be ± 0.355 mm, indicating high precision for low dose detection. Finally, we demonstrated that using multiple cameras allowed for the precise locations of the delivered beams to be tracked in 3D. We conclude that this detector is capable of real-time and accurate tracking of dynamic spot beam deliveries in 3D. The high-resolution light profiles it generates will be useful for future 3D construction of dose maps.

1. Introduction

Even as photons continue to be used for conventional external-beam radiation therapy, the use of proton therapy continues to grow. The steep dose gradients around the Bragg peak for proton beams can be used to develop treatment plans that efficiently target tumors while reducing the exposure to surrounding healthy tissues. In particular, the intensity-modulated proton therapy technique facilitates implementation of three-dimensional (3D) conformal therapy (Kooy and Grassberger 2015). It uses complex radiation fields that vary dynamically in space and time. Spot-scanned or pencil-beam proton therapy is an attractive approach because it can efficiently paint complex tumors in three dimensions, avoid critical organs, and lower the harmful neutron background (Xu *et al* 2008).

However, the developmental pace of clinical dosimeters has lagged behind that of the significant recent technological advances in proton delivery systems. Thus, reliable monitoring of complex beams is challenging with the currently available clinical dosimeters.

The gold-standard for dosimetry, the ionization chambers, in principle, can generate a 3D dose distribution by scanning them through a water phantom. However, in practice, these single-point measurements are time consuming and are therefore poorly suited for mapping complex and dynamic treatment plans. They also suffer from volume averaging, which further impedes their use in high-resolution dose mapping. A two-dimensional (2D) array of ionization chambers (Poppe *et al* 2006) or diodes (Létourneau *et al* 2004) is a good alternative to a single ionization chamber for mapping large-field dose variations and symmetry. However, 2D arrays suffer from relatively poor sampling frequency because of their size, large inter-chamber separations, and are subject to field disturbances caused by their electrical connections (Archambault and Rilling 2016). High water-equivalency for plastic scintillators, in addition to their energy independence, and flexibility in measuring both x-ray and electron radiation fields make them a good fit for precision dosimetry applications (Beddar 1992). Thin plastic scintillation sheets have been used for performing high resolution and rapid beam measurements for 2D photon (Petric *et al* 2006, Collomb-Patton *et al* 2009) and proton dosimetry (Torrissi 2000).

3D chemical dosimeters that undergo chemical composition changes upon irradiation, such as Fricke and polymer gels, have been extensively studied (Lepage and Jordan 2010). However, practical difficulties such as preparation time, sensitivity to gel preparation, and loss of spatial information from ion diffusion (in Fricke gels) have limited their use in clinical practice (Schreiner 2015). Commercially available 3D radiochromic gels exhibit little to no diffusion of active species and have been used for commissioning studies of small-field systems (Jordan 2010). However, these dose-integrating dosimeters cannot be used for dynamic measurement studies. In addition, they all require a separate dose readout system. Light generation in scintillator depends on the linear energy transfer (LET) of the protons. This LET-dependence causes the detector to under-respond around the Bragg peak and is known as ionization quenching. Quenching must therefore be corrected to obtain accurate dose calculations. While quenching can be precisely corrected for individual proton beams, estimating quenching corrections for 3D chemical dosimeters with intensity integrated signal delivered by multiple beams with variable LETs is not always straight forward (Doran *et al* 2015). Small-volume plastic scintillators are a promising alternative to chemical dosimeters for 3D dosimetry (Kroll *et al* 2013), but need further development to be useful for clinical applications.

Therefore, volumetric or 3D detectors that can measure both large and small fields and overcome the dosimetric shortcomings of existing 3D dosimeters are urgently needed. Early work with volumetric liquid scintillation detectors for 3D dosimetry showed that these detectors have several promising features (Beddar *et al* 2009). First, they are tissue- or water-equivalent, enabling dose detection with minimal field perturbation (Ingram *et al* 2015). Second, they have high spatial resolution (Archambault *et al* 2012). Third, they exhibit good temporal resolution, with nanosecond scintillation-response time and detection time in the range of milliseconds. This high temporal resolution allows quenching correction

for individual scanning beamlets. Thus, accurate determination of the total dose delivered by a treatment plan is now possible. Fourth, they are flexible for applications involving photons, protons, and other heavy-ion particles.

Earlier work by the author's group has focused on the development of a large volume of liquid scintillator coupled to optical camera detectors for 2D dosimetry of photons (Ponisch *et al* 2009) and proton beams (Beddar *et al* 2009, Archambault *et al* 2012, Hui C *et al* 2015). In the work described here, we describe the design and development of a new 3D detection system based upon liquid scintillation. We then discuss the characterization methods used for validating the detector's performance for proton therapy.

2. Methods and Materials

2.1. Design of the 3D detection system

Figure 1(a) depicts the 3D liquid scintillator detector system and setup. The scintillation volume consisted of a $20 \times 20 \times 20\text{-cm}^3$ liquid scintillator-filled tank. This scintillation volume was constructed from a combination of clear and opaque acrylic (density 1.18 g/cm^3) with a wall thickness of 11.6 mm. The entire volume was securely sealed to prevent leakage of the scintillator liquid. Figure 1(b) shows typical integrated scintillation light profiles of a proton beam captured by the three cameras.

The reference coordinate system was oriented as shown in figure 2(a). Cameras X, Y, and Z image the x-, y-, and z- projection images from the scintillation volume, respectively. As illustrated in figure 2, camera X was positioned such that it pointed directly into the front surface of the tank perpendicular to the proton beam direction (the z-axis). For this work, we considered the center of the scintillation volume as the origin point. The beam was incident on the left surface of the scintillation volume and traveled through the origin point along the z-direction. Additional beam profiles were captured using two mirrors oriented at an angle of 45° to the top and right surfaces of the scintillation volume. These mirrors redirected the scintillation photons to cameras Y and Z. Thus, cameras X and Y captured depth profiles of the beam while camera Z captured a beam's-eye view without directly exposing the camera sensor to ionizing radiation. Thus, three mutually orthogonal perspectives of the scintillation light were captured.

The liquid scintillator-filled volume and the two mirrors were housed in an opaque acrylic housing to reduce light leakage and for providing mechanical stability to the structure. Finally, a base-plate was used to securely mount the scintillator volume and the mirror assembly on an optical board.

Eight cross-hair markers with known geometries and relative positions were equidistantly placed on the inner sides of each tank surface. These built-in calibration markers were a big improvement over the earlier prototype, which required laborious acquisition of multiple images of a ruler inside the volume. A previously developed geometric calibration technique (Hui C *et al* 2015), made use of these cross-hair markers to determine the optical length from the surface of the scintillation volume to the camera sensor plane, angular shift, and translational shift of the volume relative to the camera. The optical length and angular shift

were determined by minimizing the square difference between the measured and calculated distances for multiple pairs of cross-hair markers. Likewise, pixel locations of the markers from the front and back surfaces of the volume along with the angular shift were used for calculating the translational shift of the camera center with respect to the center of the volume. Based on the calibration process, identical optical lengths were set for all the cameras (650 mm).

OptiPhase HiSafe 3 (PerkinElmer, Waltham, MA) was used as the liquid scintillator for this study. This liquid scintillator, with a density of 0.963 g/cm^3 , serves as both a phantom and a detector medium (Ingram *et al* 2015). The peak emission wavelength of the scintillator is around 420 nm with a decay time of 1.4 ns and quantum yield of 0.83.

Finally, the system was mounted on an optical board ($1200 \times 600 \times 58 \text{ mm}$) that secured the relative orientation of the components, making the system robust and portable. The rigid optical setup also ensured that the optical calibration would not change accidentally while transporting the system or during daily operation.

2.2. Optical system design

Three identical Zyla 5.5 scientific complementary metal oxide semiconductor (sCMOS) cameras were used in this work (Andor Technology Ltd., Belfast, UK). sCMOS offers relatively lower read noise ($< 2.9 \text{ e}^-$ root mean square at the fastest pixel readout speed, 560 MHz), better resolution, and wider dynamic range over interline charge-coupled device technology that was previously used for earlier detector prototypes. The sCMOS cameras were each equipped with a 5.5-megapixel sensor (2560×2160 pixels) with an individual pixel size of $6.5 \text{ }\mu\text{m}$. They supported a full-frame transfer speed of 30 frames per second (fps) using a 16-bit data range. The imaging sensors were thermoelectrically cooled to 0° C to suppress dark current. The cameras were fitted with identical fixed focal length objective lenses.

2.3. Synchronizing image acquisitions with proton beam deliveries

The proton beam delivery system at our institution does not produce a camera-compatible trigger signal that can be supplied to the cameras for synchronizing them to the beam delivery. Therefore, in this work, we developed a microcontroller-based transistor-transistor logic pulse-generating circuit that produced a trigger signal upon receiving a beam initiation signal from the synchrotron. This method ensured that the camera acquired the images only during the duration of the beam delivery.

The synchrotron at The University of Texas MD Anderson Cancer Center Proton Therapy Center generates up to 94 different beam energies. Each energy is delivered in a different “spill” of the beam, such that only one proton energy is available per spill. A spill, in turn, can consist of multiple beam spots, with a typical spot duration of 4 ms (Gillin *et al* 2010). For this work, we selected two synchrotron-generated signals, labelled as START and STOP in figure 3, that indicate the beginning and end of the beam delivery. An Arduino-based microcontroller was used to condition the START/STOP signals and to output a trigger signal to the cameras. The extraction start signal was generated about $500 \text{ }\mu\text{s}$ prior to the beam delivery, the extraction stop signal is produced towards the end of the beam delivery. It

was observed that a small dose was delivered following the STOP signal. The microcontroller was therefore programmed to capture this dose by delaying the next trigger by 100 μ s. Because the cameras were active only during the beam delivery times, their readout times did not overlap with the dose-delivery times. As a result, the cameras were able to capture all the delivered beams.

To improve the frame capture rates, the cameras were operated in an overlap mode that allowed simultaneous exposure of the next-frame while the present frame was being readout. Figure 3 shows image frames captured during the previous cycle (labelled “S#” for signal frame) being read out while, simultaneously, the camera was acquiring the next frame. The camera also read out a reference frame from the sensor (labelled “R#” for reference frame) to account for residual pixel charge. “Beam-pause” interlocks that are occasionally generated by the beam delivery system prolong the camera exposure times in the absence of proton delivery and thus contribute noise. The selected camera mode upon receiving a beam-pause interlock, terminated the current signal frame and started another frame (labelled as “blank”) that lasted for the duration of the interlock. Once the interlock was cleared, a new signal frame coincident with the beam delivery was then started. The blank image frames were identified and discarded in post-processing.

2.4. Optical artefact corrections

Before the three cameras can be used for scintillation detection, optical artefacts associated with each camera were individually tested and corrected (Robertson *et al* 2014). This involved setting similar optical settings, including lens focus and aperture, and identical optical lengths. Corrections for refraction at the air-liquid scintillator interface and changes in image perspective with distance were implemented. Lens distortion was corrected using the Camera Calibration Toolbox for MATLAB (Bouguet 2010). The three lenses demonstrated a mean distortion of 0.22 pixels along the radial distance, and their focal lengths matched within 0.2%.

The background signal contributed by ambient light, dark current, and camera read noise was also taken into account. To ensure that the cameras had similar dark current and read noise levels, image acquisitions commenced after the cameras reached a stable operating temperature of 0 °C. The background signals for the three cameras were within 1% of each other for camera integration times varying from 10 ms to 1 s. A spatial median filter was applied to the 3 × 3 region surrounding each pixel for eliminating the influence of stray radiations on the sensors.

Because all three cameras demonstrated matching characteristics, only the performance characteristics of the camera X are described here. All experiments were performed using the discrete spot scanning proton beam gantry at the MD Anderson Proton Therapy Center. As shown in figure 1, the 3D detection system was placed on the patient couch such that the beam isocenter aligned with the left surface of the scintillation volume. The gantry angle was set to 270°. All image computations and analyses were performed using MATLAB (version R2014b; MathWorks, Natick, MA).

2.5. Spatial resolution

One of the advantages of using volumetric detectors is their ability to capture dose distributions with high spatial resolution. The resolution limit depends on several aspects of the optical system design, including sensor size, individual pixel size, choice of objective lens, distance of the imaging plane from sensor, and optical artefacts. To produce a sharp image, objective lenses for volumetric imaging must be able to maintain good focus over the entire depth of the scintillation volume. The distance over which the lens can retain reasonably good focus is referred to as the depth of field. Therefore, the lens selection was based on calculations of depth of field (Beddar *et al* 2009). Using a 20.5 mm fixed focal length objective lens (Schneider Optics, Van Nuys, CA) with an f/8 aperture and a working distance of 650 mm, a 205 mm depth of field was obtained. These settings resulted in a field of view that was larger than that required to image the 20 cm² area of the scintillation volume face. We therefore chose a smaller region of interest (1100 × 1100 pixels). Furthermore, spatial resolution changes as a function of the object's distance from the lens. It is important to accurately quantify this change in spatial resolution through the entire scintillation volume depth, especially for 3D dose reconstruction.

2.6. Camera linearity

For quantitative imaging, the camera is expected to demonstrate a linear response so that a simple yet accurate conversion of the detected light to the deposited dose is possible. To verify the camera's linearity, the intensity counts at the peak of the light profile for six doses ranging from an average of 0.1 Gy to 12 Gy were recorded. These doses were delivered three times for each of five proton beam energies: 161.6 MeV, 144.9 MeV, 124 MeV, 100.9 MeV, and 85.6 MeV. We also calculated a non-linearity factor describing the deviation in camera counts at each dose level from the corresponding linear regression fit. Additionally, the signal-to-noise ratio (SNR) was calculated for individual pixels over an area of 21 × 21 pixels around the peak light intensity profile, where the majority of the proton dose is

deposited. The SNR was calculated using the equation, $20 \times \log\left(\frac{\text{signal} - \text{background}}{\sigma_{\text{background}}}\right)$, where σ is the standard deviation in the background signal.

2.7. Camera stability

Another important performance characteristic for the detector is the ability to maintain a stable light output for a given dose input over a period of time. To assess the camera stability, a proton beam with an energy of 161.6 MeV and a dose of 1.16 Gy was delivered to the detector 10 times over a single imaging session. The experiment was then repeated

over three sessions spanning a week. Stability is defined using the equation $\frac{\sigma_{\text{signal}}}{\overline{\text{signal}}}$, where σ_{signal} indicates the standard deviation for an individual pixel intensity while $\overline{\text{signal}}$ is the average intensity for that pixel over three sets of measurements. An area of 21 × 21 pixels around the maximum signal intensity was then selected and stability was assessed for all the individual pixels in that area. The reported stability for the system is the mean stability over the entire area.

2.8. Proton range measurement accuracy

The detector's proton range measurement accuracy was assessed for all 94 proton energies generated by the synchrotron, ranging from 72.5 MeV to 221.8 MeV. The measurements were divided in two batches: low-energy batch consisted of energies ranging from 72.5 MeV to 141.6 MeV with a mean proton range less than 20 cm; high-energy batch delivered energies from 129.2 MeV to 221.8 MeV with a mean proton range exceeding 20 cm. To ensure that the Bragg peaks from the high-energy batch were restricted to the scintillator medium, a plastic-water buildup with a water-equivalent thickness of 105 mm was placed in the beam path. An average dose of 0.25 Gy was delivered on the central beam axis for each beam energy and all the energy deliveries were repeated nine times. To determine the beam range, pixels of the captured scintillation image were vertically summed to generate a one-dimensional (1D) integrated light profile along the z-axis. At the MD Anderson Proton Therapy Center, the proton range is defined as 90% integral depth dose distal to the Bragg peak (Gillin *et al* 2010). Based on a recent study, we found that the distal 90% of the light peak provides the most accurate range measurements without correcting for quenching (Hui *et al* 2015).

Since this study involved range measurements in a liquid scintillator; therefore, the range measured in the scintillator medium was first converted to the water-equivalent range by accounting for the ratio of the stopping power of water to that of the liquid scintillator. This stopping-power ratio was experimentally found to be 0.995. Ionization chamber measurements collected as part of the beam commissioning studies served as the reference measurements.

For clinical applications, the detector must be able to detect the smallest dose fractions delivered in treatment plans. At these small dose fractions, the influence of noise on range measurement accuracy is expected to increase. The minimum deliverable spot MUs from the synchrotron at The University of Texas MD Anderson Cancer Center Proton Therapy Center are 0.005. Considering the dose at the Bragg peak to be about 0.25 Gy/MU (for 72.5 MeV) to 1 Gy/MU (for 221.8 MeV), the minimum dose a spot can deliver at the Bragg peak is in the range of 0.13 cGy to 0.5 cGy. Therefore, to determine the impact of noise on range-measurement accuracy at very low delivered doses, the beam ranges were measured as a function of beam dose for three separate energies (161.6 MeV, 144.9 MeV, and 124 MeV), with the beam intensities varying over average doses of 0.87 cGy to 1.16 Gy.

2.9. Lateral beam profiles

Lateral beam profiles can be used for determining the size of individual beam spots that vary with the proton energy. In this study, we used the beam images captured by camera X to plot lateral 1D integrated light profiles perpendicular to the beam axis. For the 161.6-MeV proton-beam energy, these 1D lateral profiles were plotted at two different depths within the scintillation volume: $z = 20$ mm and $z = 166$ mm. The normalized integrated 1D light profiles were compared with the normalized dose profile of the proton beam as determined using a Monte Carlo radiation transport code MCNPX, version 2.7d. The full width half maximum (FWHM) for these 1D profiles was selected as the quantification parameter. Lateral light profiles for off-axis beams that were displaced ± 4 cm and ± 8 cm along the y-

axis, were plotted at $z = 20$ mm and compared with the lateral profile of the beam on the central axis.

2.10. Determining 3D beam locations

The unique benefit of employing multiple imaging cameras is that combining 2D images from different cameras produces accurate location of the 3D proton beam anywhere inside the tank. Here, we used the cameras Z and X to locate the beam position in the z and x projection images, respectively. A beam with an energy of 161.6-MeV was selected. The beam was initially positioned on the central beam axis and then displaced 4 cm off-axis along both directions of the x -axis. Similar beam translations were then repeated along both directions of the y -axis. A dose of 0.28 Gy was delivered three times at each location.

3. Results

Figure 4 shows 1D integrated light profiles generated for five beam energies. Depending on the dose magnitude, each proton beam was delivered and captured for durations ranging from 11 ms to 4 s. Figure 4(a) shows the normalized depth light profiles for five beam energies imaged using camera X. The higher-energy profiles show wider Bragg peaks caused by greater energy straggling. Similarly, figure 4(b) shows the normalized lateral profiles of the proton beams at the same five energies obtained using camera X. As expected, proton beams with lower energies had a wider lateral profile.

3.1. Spatial resolution

As shown in figure 5, the detector demonstrated good spatial resolution that varied linearly from the front (65 cm from the sensor plane) to the back (85 cm from the sensor) of the scintillation volume, with a mean resolution of 0.208 mm. This sub-millimeter resolution can be used for generating detailed light distribution profiles and is a clear improvement over that of the currently used detectors for clinical proton beam dosimetry.

3.2. Camera linearity

The recorded light intensity from camera X increased as a function of increasing dose, as shown in figure 6(a) for the 161.6-MeV beam energy. A non-linearity parameter calculated for the camera was 0.39% over five separate energies, indicating that the camera had an excellent linear response over a three-order-of-magnitude change in delivered dose. The SNR varied linearly from 20 dB to 85.5 dB over a dose range of 5.3 mGy to 10.6 Gy. The lowest dose that can be delivered by our synchrotron at 161.6 MeV is 2.9 mGy; therefore, the detector was able to detect proton beams just above this lowest dose (figure 6(b)).

3.3. Camera stability

The average stability of the detector was 1.08% over 3 weeks, with week-to-week variation of 0.04%. These results indicated that the camera had an excellent, stable response to the delivered dose.

3.4. Proton range measurement accuracy

Figure 7(a) shows the proton-beam range-measurement accuracy plotted as a function of the beam locations inside the scintillation volume measured using camera X. The range-measurement accuracy is calculated as the difference between the measured and reference beam ranges, which is shown for all 94 synchrotron-generated energies. After linear fitting, the average measurement accuracy over 94 energies was -0.073 ± 0.030 mm.

Relatively larger errors (still within 1 mm) in range measurements were observed for the proton beams with Bragg peaks located towards the right and left edges of the scintillation volume. We attributed this effect to beam reflections from the inner surfaces of the scintillation volume. To confirm that this was the case, five discrete energies from 211.6 MeV to 221.8 MeV with Bragg peaks located near the scintillation volume edge were delivered. The same energy beam deliveries were then repeated while attenuating their ranges using a water-equivalent phantom in the proton path, so that the Bragg peaks were shifted away from the volume edge. The range measurement errors were 0.78 mm when the Bragg peaks were located near the edges and 0.14 mm away from the edges. This confirmed that internal beam reflections are a source of error in range verification studies. Therefore, Bragg peaks of beams must be located away from the edges of the scintillator volume during range measurements.

We also assessed the impact of low-dose deliveries on proton-beam range measurement accuracy over three different energies using six dose levels. Whereas the difference in measured range was within ± 0.2 mm of the reference range, reducing dose levels increased the uncertainty in the range measurements. As shown in figure 7(b), for a beam energy of 161.6 MeV, the uncertainty increased from ± 0.006 mm to ± 0.701 mm when the delivered dose was reduced from 1.16 Gy to 0.87 cGy. Despite this increase, the overall uncertainty was less than 1 mm. The average increase in uncertainty over all three energies for the highest and lowest doses were ± 0.01 at 1.16 Gy and ± 0.355 mm at 0.87 cGy. These results illustrated that this detector is capable of accurate beam-range measurements even for extremely low doses.

3.5. Lateral beam profiles

The FWHM of lateral beam profiles was used for calculating the beam spot sizes. Figure 8(a) shows the normalized integrated 1D lateral light profile at a depth of $z = 20$ mm inside the scintillation volume along with a profile obtained from Monte Carlo simulation at the same depth. The FWHM of the lateral profile at $z = 20$ mm depth along the beam axis was 1.6 mm (9.5%) wider than that of the Monte Carlo simulation. The FWHM of the lateral profile at depth $z = 166$ mm (not shown) was 2.2 mm (10.6%) greater than that of the Monte Carlo simulations. The photon travel paths for both profiles from the location of scintillation generation within the tank volume to camera X were approximately the same; therefore, we observed a similar widening of the measured profiles.

Lateral 1D light profiles measured at $y = \pm 4$ cm and $y = \pm 8$ cm from the central beam axis at a depth of $z = 20$ mm are shown in figure 8(b). The lateral light profile along the central beam axis ($y = 0$ cm) was used as the reference. The FWHMs of the off-axis spots were

2.6% (at -4 cm), 4.6% (at -8 cm), 2.6% (at 4 cm), 2.5% (at 8 cm) wider than that of the single energy spots located along the central axis. This broadening of the off-axis beam profiles can be accounted for in the future by correcting them for the vertical angle they subtend with respect to the central beam axis. Thus, practically identical lateral light profiles can be obtained away from the central beam axis for a given depth.

3.6. 3D beam locations

The location of a beam in three dimensions can be determined by combining the 2D views of a proton spot from multiple cameras. We used the cameras X and Z to record the 2D beam profile in the x- and z-projections images, respectively. The geometric calibration algorithm then accurately mapped the physical location of these spots from their recorded pixel locations. The nominal beam locations programmed into the beam delivery system and the measured 3D locations are compared in figure 9. Figures 9(a) and 9(b) show the beam location accuracy in the z- and x-projections, respectively. We observed a location accuracy of 0.3 mm in the x-projection but an accuracy of 2 mm along the lateral z-projection. Lower accuracy in the z-projection could be attributed to a combination of errors such as beam positioning by the steering magnets and imperfect detector calibrations. Further, in this work, the location accuracy was assessed by comparing the measured beam locations to the nominal or programmed beam locations. Greater location accuracy is expected by comparing the measured locations to the actual delivery locations as recorded by the spot position monitor.

4. Discussion

This study demonstrated that a liquid scintillator-based detector can be used with multiple orthogonally placed cameras to sample large-area (20 cm²) proton-therapy fields in three dimensions. Because the liquid scintillator is water-equivalent, measurements can be performed under pristine beam conditions, with minimal alteration to the proton fluence. The detector also generated high-resolution (0.2 mm) light distribution profiles.

The detector is uniquely suited for dynamic imaging of discrete proton spots because it can perform rapid imaging of the proton beams with high temporal resolution (11 ms). Moreover, the temporal resolution could be further improved for use in quality assurance studies where the placement of the beam path is predetermined, for example, in beam range validations. Using a smaller region of interest tightly encompassing the beam would produce faster image acquisition rates for such studies.

We found that all three cameras used for 2D detection of the proton beams displayed good stability and dose linearity. Good camera stability indicates that no correction or additional compensation factors were necessary to correct for systemic drift or camera output change over time when measuring a given beam condition. We also tested the cameras for their ability to linearly capture scintillation light for doses up to 12 Gy. Camera response linearity allows for meaningful interpretation of captured light to dose and is critical for dosimetry. For camera Z that captured the beam's-eye view, camera saturation was reached earlier because of light integration in the beam direction. Thus, most hypo-fractionated treatments (< 8 Gy) can be imaged with these cameras. One approach to avoiding camera saturation is

by programming the beam deliveries such that the intended large dose is split into multiple synchrotron spills. Refer to figure 3 for timing chart on proton delivery by synchrotron. This way the camera can collect signal from each spill, read out the integrated signal, and be ready for the next signal acquisition from the succeeding spill.

Figure 4(a) shows depth scintillation light profiles with tails towards their distal ends. Such tails were not present in the depth dose profiles measured using ionization chamber. Further, lateral integrated light profiles (refer figure 8(a)) were relatively wider compared to those obtained through Monte Carlo simulations. These differences could be attributed to image blurring from photon scatter inside the scintillation liquid as well as from imperfections in the lens optics. Robertson *et al* (2014) has shown that both these artefacts can be accounted for by determining and later deconvolving the acquired proton beam images with point spread function for the system. The next step therefore will be to experimentally determine the total point spread function of the optical system, which includes contributions from both photon scatter and lens optics, at all locations within the scintillation volume.

Unlike our previous detector, this detector is not enclosed inside a light-tight box [Hui *et al* 2015]. Therefore, the proton-imaging workflow with this detector requires that the room lights be turned off. Future designs will incorporate camera-mounted band-pass filters that selectively allow scintillation light (~ 420 nm) to reach the sensor while rejecting ambient light wavelengths. This approach will improve the image SNR and enable beam detection studies to be performed with the room lights on.

One area of concern is the risk of exposing cameras to harmful radiation, such as that emitted within the proton beam gantry by secondary neutrons with high linear energy transfer. Secondary radiation is likely to damage electronics, particularly CMOS devices (Shoga and Binder 1986). The effect of such radiations on the cameras can range from temporary loss of function to permanent damage. Shielding electronics with borated polyethylene has produced only limited benefits (Kry *et al* 2011). Although our unshielded cameras have been exposed to doses of more than 100 MU, we have yet to experience any loss of camera functionality. This could be because the present work concerned a spot scanning system, in which the neutron fluence generated is considerably less than that of a passive scatter system. In addition, the cameras were located away from the primary beam incidence. Nonetheless, a risk of radiation-induced electronics failure is always present when using cameras for proton-beam imaging.

The next goal of our work will be to use the 2D scintillation light images from three mutually orthogonal cameras to produce 3D reconstructions of light distribution. A simulation study with this detector configuration performed by our group demonstrated the feasibility of this setup for 3D reconstruction (Hui *et al* 2014). In addition, ionization quenching correction models that will be required for gathering correct dose information from the scintillation light map are currently being investigated.

5. Conclusion

In this work, we designed and developed a 3D liquid scintillator-based detector to rapidly acquire and accurately assess beam characteristics for discrete spot scanning proton beam systems. Highlights of the system capabilities include its high spatial (0.21 mm) and temporal (11 ms) resolutions and its large 3D volume that can cover most treatment plans (20 cm³). It requires virtually no preparation time after a onetime optical calibration, can be quickly reused, is portable and durable, and provides real-time or near real-time dosimetry data.

The key design features of this detector include (a) its three-camera setup, which will support future 3D dose verification methods; (b) its single-board setup with fixed optical equipment to preserve optical calibration and make the system more rugged and portable; (c) its use of embedded imaging markers for rapid calibration; and (d) its ability to synchronize camera acquisition with beam delivery.

In this study, we identified and analyzed several performance criteria for the 3D detector. First good camera performance parameters such as spatial and temporal resolution, stability (1%), SNR (20 dB at 0.5 cGy), and dose response linearity (99.6%) were found. Second, the detector was found to be suitable for rapid and accurate beam range verifications for several proton beam energies with an average error of -0.073 ± 0.030 mm. Third, the detector enabled accurate determination of spot beam sizes for all energies. Fourth, the detector facilitated 3D location of the beam position within the detector with good accuracy.

Having validated the performance of this detection system, our next steps are to develop 3D reconstructions of the measured light distribution and to improve ionization quenching correction models for converting the measured light distribution into corresponding dose values.

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Figure 1.

(a) Photograph of the 3D liquid scintillator detector setup placed on a patient couch in a discrete spot scanning proton beam gantry. (b) Integrated light profiles of a proton beam captured using the three-camera setup.

Figure 2.

Schematic of the 3D liquid scintillation detector. (a) 3D model of the system. (b) Right-side view of the detector system with cameras X and Y imaging the scintillation volume. A 45° mirror mounted on the top of the volume redirected scintillation light to camera Y. Camera Z and side mirror are not shown in the 2D schematic for simplicity.

Figure 3.

Timing diagram depicting synchronization of sCMOS camera triggering with synchrotron-generated beam pulses.

Figure 4.

(a) Normalized depth scintillation light profiles for five beam energies measured with camera X. (b) Normalized lateral scintillation light profiles of the same beam energies. On both figures, the x-axis shows the location of the Bragg peak with respect to the center of the tank volume.

Figure 5.

Spatial resolution of the detection system as a function of working distance from the camera sensor plane to different imaging planes within the scintillation volume. The change in resolution was modeled using a linear fit.

Figure 6.

(a) Linear response of the camera to increasing dose for 161.6 MeV beam energy. (b)

Signal-to-noise of camera X as a function of the delivered dose (Gy).

Figure 7.

(a) Proton-beam range-measurement accuracy plotted for all 94 energies delivered by the synchrotron. The difference between the measured (R_{measured}) and reference ($R_{\text{reference}}$) beam ranges is plotted as a function of their range inside the tank volume. The energies were delivered in two batches: low-energy (72.5-141.6 MeV) and high-energy (129.2-221.8 MeV). (b) Uncertainty in range measurements for low-dose deliveries. Uncertainty in measurements is represented by the error bars in both the figures.

Figure 8.

(a) Lateral 1D integrated light profile recorded by camera X is compared to Monte Carlo simulation of lateral dose profile at depth $z = 20$ mm from the entrance of the scintillation volume. (b) Lateral 1D integrated light profiles of spots of single beam energies incident at different locations from the central axis ($y_{\text{shift}} = 0$ cm) along the y-axis ($y_{\text{shift}} = \pm 4$ cm and ± 8 cm) are shown.

Figure 9.

Spot location accuracy calculated as the difference between measured and programmed beam locations inside the scintillation volume is plotted for a beam energy of 161.6-MeV (nominal location of 60.85 mm within the scintillation volume) and for the beam locations, (0,0), (± 4 cm,0), (0, ± 4 cm), in the (a) z-projection, and (b) x-projection.