Development of a liquid scintillator-based 3D detector for range measurements of spot-scanned proton beams

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Abstract The goal of the present study is to develop a liquid scintillator-based 3D detector and investigate its performance for beam range measurements in real time. The detector design consists of a tank filled with water-equivalent liquid scintillator that scintillates in response to incident proton beams. Three scientific complementary metal-oxide semiconductor (sCMOS) cameras collect the resulting optical signals. Preliminary measurements have shown that the optical detection system can record light distribution profiles anywhere inside the tank with an average spatial resolution of 0.23 mm. By employing multiple cameras the system is capable of capturing high resolution 2D depth and beams-eye profiles of the delivered beam and can consequently localize the position of beam anywhere inside the tank. It is also shown that it is capable of accurately measuring proton beam range with no more than 0.3 mm difference from the nominal range. The detector system thus demonstrates its ability to perform fast, high-resolution, and precise beam range measurements in 3D.

1. Introduction
Ionization chambers (IC) are the prevailing standard for dosimetry on account of their accuracy and are routinely used for QA measurements of machines and treatment plans [1]. However, drawbacks such as the ability to verify only a limited number of dose points along the beam, volume-averaged readings, and time requirement for dose verification may render them obsolete especially for modern radiation therapy modalities such as intensity modulated radiation therapy (IMRT), or intensity modulated proton therapy (IMPT). Therefore, novel detectors are needed that can measure complex radiation distribution and steep dose profiles in 3D [2]. Our liquid scintillator-based detector has the capability to overcome these difficulties by generating high resolution spatially and temporally resolved 3D dose profiles in clinically relevant time-frames. In this paper, we are examining few important performance characteristics of this 3D detector with scanning proton beam systems.

2. Materials and methods

2.1 Liquid scintillator detector design
The detector consists of a liquid scintillator (LS) filled cubic tank (20 x 20 x 20 cm³) constructed out of acrylic. Clear tank surfaces allow scintillation light to be captured by the cameras while the opaque surfaces minimize influence of ambient light on the measurements. Optiphase HiSafe 3 (PerkinElmer, Waltham, MA) was our choice of LS because of its water-equivalency (0.986 g/cm³) and the ability to double up as a phantom as well as an active element thereby ensuring that the radiation field is measured without perturbation. To capture additional beam profiles, two mirrors located on the top and side
surfaces of the tank mounted at 45° angles to the tank surfaces redirect the scintillation light to the respective two cameras (Figure 1.a). Multiple cross-hair markers imprinted on the inner tank surfaces allow geometric calibration that converts pixel location in camera frame to physical location within the tank. This geometric calibration takes into account refraction, perspective, and lens distortion, in addition to rotational and translational shift of the camera relative to the tank, and effective distance between the tank and lens center [3]. Finally, the tank along with the three cameras are securely mounted on a breadboard for system portability, rigidity, and robustness.

2.2 Optical system design

We used three identical Zyla 5.5 scientific complementary metal-oxide-semiconductor (sCMOS) cameras for this work (Andor Technology Ltd., Belfast, UK). The camera sensor is equipped with 2560 x 2160 pixels with an individual pixel size of 6.5 µm. A 16-bit analog-to-digital converter digitizes the light signal. The camera is cooled and maintained at 0°C via an on-board thermoelectric cooling system. It is capable of sustaining a full frame transfer rate of 30 frames per second (fps) using USB3.0 interface. Additionally, unlike CCDs, CMOS offers the flexibility to readout smaller region of interests (ROI) that further speeds up data transfer. For example, we use 1100 x 1100 pixel area to image 20 x 20 cm tank. This smaller ROI drastically improves the frame transfer rate to 85 fps, resulting in better time-resolution for the 3D detector.

The cameras are fitted with identical 20.5 mm fixed focal length objective lenses (Schneider Optics, Van Nuys, CA). The lens selection was based on calculations of depth of field (DoF) [4] and field of view (FoV). With 750 mm working distance and f/8 aperture lens setting a 20 cm focal depth around the tank center is obtained. This allows sharp scintillation images to be captured anywhere inside the tank volume. Finally, this work requires all cameras to simultaneously acquire scintillation light from three orthogonal angles. For this study, we connected the three cameras in a daisy chain configuration with the first camera triggering the next with an insertion delay of 2 µsec. Thus, all three cameras fire in tandem capturing instantaneous beam profile from different orientations.

![Figure 1. (a) Schematic of liquid scintillator (LS) detector fitted with two mirrors. (b) Picture of actual 3D system setup at proton therapy center at MD Anderson Cancer Center at Houston, TX.](image)

All our experiments were conducted within the scanning beam gantry at the Proton Therapy Center at The University of Texas MD Anderson Cancer Center at Houston (PTCH). As seen in Figure 1.b the imaging setup was placed on the patient couch. The beam isocenter was then aligned with the center of the left tank surface. The gantry angle was set to 270°.

2.3 Spatial resolution

In order to accurately localize the beam position inside the tank volume it is critical to determine the effective image pixel size as a function of the distance from the lens. Therefore, using a checkerboard pattern we experimentally recorded the pixel size at three different distances from the lens: (i) the front
wall of the tank (650 mm), tank center (750 mm), and tank back wall (850 mm). We also measured the pixel size as a function of the distance from the image center to the lateral tank edge [5].

2.4 Range measurements of spot scanned proton beams

Proton ranges were estimated using a previously developed geometric calibration technique in our lab [3]. We measured the proton ranges at five different energies: 161.6 MeV, 144.9 MeV, 124 MeV, 100.9 MeV and 85.6 MeV using the liquid scintillator system. They were then compared with measurements from a parallel plate chamber that were collected as part of the beam commissioning, and referred henceforth as nominal measurements. A total of 5 MU dose was delivered for each energy and the images were captured with 100 msec camera integration times. For this study, the beam energies were selected such that the Bragg peaks from the scintillation light profiles fall within the tank. Higher energy beams with ranges greater than the tank dimensions can be measured by placing plastic water buildups in front of the tank side wall.

3. Results and Discussion

![Figure 2](image)

**Figure 2.** (a) Pixel size as a function of the imaging distance from camera. The solid line shows a linear fit for extracting pixel sizes for any intermediate imaging distances (b) Pixel size as a function of the lateral distance from the image center. The dotted line indicates 2% change in pixel size.

Figure 2.a represents change in pixel size (mm/pixel) as a function of the working distance from the camera. A linear fit through the three points can be used to calculate pixel size at any depth inside the tank. A sub-millimeter pixel size suggests that this system is capable of high-resolution measurements. It is therefore useful for comprehensive mapping of areas with steep-dose gradients and for generating 3D plots of dose distributions. Figure 2.b shows the change in pixel size from the image center to the periphery in the central plane of the tank. The dotted line indicates that this change is less than 2% over a distance of 100 mm.

Figure 3.a shows normalized light profiles of five energies delivered to the tank. Measuring beam range involves identifying a typical reference point on dose profile. At PTCH, beam range for scanning beams is defined at 90% of integral depth dose distal to the Bragg peak [6]. Proton studies involving liquid scintillators typically exhibit under-response near the Bragg peak known as ionization quenching. Thus, a point on dose distribution cannot be directly correlated with a point on the light distribution. In order to do so, previous study tested several reference points for minimizing the difference to the nominal range [5]. Our recent study using geometric calibration approach found that distal 90% to light peak provides the most accurate range measurements without having to correct for quenching [3]. Determining quenching is still a challenge and will need to be addressed for successful reconstruction of 3D dose. Table 1 summarizes the nominal and measured proton range showing an excellent agreement to within less than 0.3 mm for all five energies.
4. Conclusion

This preliminary study shows that generating longitudinal and lateral profiles of scanned proton beams with submillimeter resolution (0.23 mm) is now possible with this 3D detector. Post system calibration, proton range is calculated within a duration of few seconds and with a range estimation accuracy of 0.3 mm from the nominal value. The liquid scintillator-based detector can therefore be used for fast, high-resolution, and precise mapping of dose distributions in real-time and potentially for QA of scanning proton beams in 3D.

Table 1. Comparison between nominal and measured proton range for five energies.

| Beam Energy (MeV) | Nominal Range (mm) | Measured Range (mm) | Difference (ΔR) (mm) |
|-------------------|--------------------|---------------------|----------------------|
| 161.6             | 175.988            | 175.879             | -0.108               |
| 144.9             | 144.940            | 144.993             | 0.053                |
| 124               | 109.473            | 109.709             | 0.237                |
| 100.9             | 74.973             | 75.174              | 0.198                |
| 85.6              | 55.059             | 55.279              | 0.219                |

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6. References

[1] Hill R et al 2009 Med. Phys. 36 3971-81
[2] Baldock C et al 2010 Phys. Med. Biol. 55 R1-63
[3] Hui C K et al 2015 Biomed. Phys. Eng. Express 1 025204
[4] Beddar S et al 2009 Med. Phys. 36 1736-43
[5] Archambault L et al 2012 Med. Phys. 39 1239-46
[6] Gillin M T et al 2010 Med. Phys. 37 154-63