

Real-time volumetric scintillation dosimetry

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Abstract. The goal of this brief review is to review the current status of real-time 3D scintillation dosimetry and what has been done so far in this area. The basic concept is to use a large volume of a scintillator material (liquid or solid) to measure or image the dose distributions from external radiation therapy (RT) beams in three dimensions. In this configuration, the scintillator material fulfills the dual role of being the detector and the phantom material in which the measurements are being performed. In this case, dose perturbations caused by the introduction of a detector within a phantom will not be at issue. All the detector configurations that have been conceived to date used a Charge-Coupled Device (CCD) camera to measure the light produced within the scintillator. In order to accurately measure the scintillation light, one must correct for various optical artefacts that arise as the light propagates from the scintillating centers through the optical chain to the CCD chip. Quenching, defined in its simplest form as a nonlinear response to high-linear energy transfer (LET) charged particles, is one of the disadvantages when such systems are used to measure the absorbed dose from high-LET particles such as protons. However, correction methods that restore the linear dose response through the whole proton range have been proven to be effective for both liquid and plastic scintillators. Volumetric scintillation dosimetry has the potential to provide fast, high-resolution and accurate 3D imaging of RT dose distributions. Further research is warranted to optimize the necessary image reconstruction methods and optical corrections needed to achieve its full potential.

1. Introduction

Volumetric scintillation is a promising new area of study aiming at making fast, high resolution and accurate measurements of absorbed dose distributions by imaging the light emitted by a scintillating medium in three dimensions. It was initially proposed by Kirov *et al* 2005 [1] for the dosimetry of brachytherapy Ru-106 eye plaque applicators using a liquid scintillator (LS) and then by Fukushima *et al* 2006 [2] for proton range measurement using a plastic scintillator block.

Our research group has been engaged over the last five years in the conception and development of real-time volumetric scintillation dosimetry using large-volume liquid or solid scintillators with the goal of rapidly measuring or imaging 3D dose distributions from photon or proton beams. Beddar *et al* 2009 [3] have investigated the feasibility of using a 3D LS detector system for the verification and characterization of proton beams in real time for intensity and energy-modulated proton therapy. They irradiated a plastic tank filled with LS with pristine proton Bragg peaks. Scintillation light produced during the irradiation was measured with a CCD camera. Acquisition rates of 20 and 10 frames per second (fps) were used to image consecutive frame sequences. These measurements were then compared to ion chamber measurements and Monte Carlo simulations. The light distribution measured from the images acquired at rates of 20 and 10 fps had standard deviations of 1.1% and



0.7%, respectively, in the plateau region of the Bragg curve. Differences were seen between the raw scintillation signal and the ion chamber due to the quenching effects of the scintillator and due to the optical properties of the imaging system. Beddar *et al* 2009 [3] have demonstrated the feasibility of using such detector system to measure the dose distribution produced by a proton beam in 2D and verified that the system is capable of acquiring multiple images within a single proton pulse with good signal-to-noise ratio and submillimeter image resolution. Although the size of the LS detector was limited ($7 \times 7 \times 14 \text{ cm}^3$), they were able to conclude that a larger detector volume could be used.

Poenisch *et al* 2009 [4] investigated a larger LS system for application for high-energy photon beam dosimetry. They were able to acquire series of light signal images when irradiated with a 6 MV photon beam. The scintillation light data measured with the CCD camera were filtered to correct for scattering of the optical light inside the liquid scintillator. Depth-dose and lateral profiles as well as two-dimensional (2D) dose distributions were found to agree with results from the treatment planning system. Further, the corrected light output was found to be linear with dose and independent of dose rate. They concluded that the short time needed for image acquisition (10 s) and processing could make this system ideal for fast verification of the beam characteristics of the treatment machine and this new detector system shows potential usefulness for 2D Quality Assurance for 3D conformal RT or Intensity Modulated Radiation Therapy (IMRT) treatment verification. The limitation of this system is that it can only acquire images in two dimensions because there is no information as to the distance between the light source and the single camera. The addition of a second camera with an orthogonal view will lead to the eventual goal of measuring or imaging dose distributions in 3D.

Archambault *et al* 2012 [5] developed a similar LS detector system ($20 \times 20 \times 20 \text{ cm}^3$) viewed by a CCD camera with a field view of $\sim (26 \times 20 \text{ cm}^2)$ and a pixel size of 0.4 mm to measure the spatial location, intensity, and depth of penetration (energy) of the proton beamlets in near real-time. Pencil beams with nominal ranges in water between 9.5 and 17.6 cm were scanned to irradiate an area of $10 \times 10 \text{ cm}^2$. Image frames were acquired at 50 ms per frame. Proton range measured from the light distribution produced in the LS was accurate to within 0.3 mm on average. The largest deviation seen between the nominal and measured range was 0.6 mm. The lateral position of the measured pencil beam was accurate to within 0.4 mm on average (see Section 3.3). The intensity of single proton spots was measured with precision ranging from 3% for the smallest spot intensity (0.005 MU) to 0.5% for the largest spot (0.04 MU).

Robertson *et al* 2013 [6] undertook the challenging task to study the quenching problem exhibited by scintillating materials when subjected to proton beams to be able to accurately convert the light signals to absorbed dose. The goal of their study was to develop an ionization quenching correction method to restore the linear dose response of scintillators when irradiated by proton beams. The quenching correction method they used was based on an empirical model that predicts quenching on the basis of the linear energy transfer (LET) of the proton beam developed by Birks 1951 [7,8]. The results of this study (see Section 5) demonstrated the effectiveness of a correction method that restores the linear dose response of a liquid scintillator throughout the proton beam range. The correction method was applied to a large-volume scintillation detector, addressing a major obstacle to fast 3D dosimetry of proton beams. This new correction method allows for closer agreement than a prior study by Wang *et al* 2012 [9] to correct for quenching in plastic scintillator detectors which showed an agreement within 5% between scintillator and ionization chamber measurements of proton beams.

Recently, a systematic study was performed by Robertson *et al* 2014 [10] to characterize the optical artefacts affecting measurement accuracy of such detector systems and to develop methods to correct for these artefacts. The optical artefacts addressed were photon scattering, refraction, camera perspective, vignetting, lens distortion, the lens point spread function, stray radiation, and noise in the camera. These artefacts were evaluated by theoretical and experimental means. The correction methods effectively mitigated the artefacts, increasing the average gamma analysis pass rate from 66% to 98% for gamma criteria of 2% dose difference and 2 mm distance to agreement.

Goulet *et al* 2014 [11] have recently developed a $10 \times 10 \times 10 \text{ cm}^3$ plastic scintillator embedded inside an acrylic phantom and imaged using a plenoptic camera. The 3D light distribution produced by

the scintillator was reconstructed at a 2 mm resolution in all dimensions by back-projecting the light collected by each pixel of the light-field camera using an iterative reconstruction algorithm constrained by a beam's eye view projection of the incident dose acquired using the electronic portal imager integrated with the linac. The absolute dose difference between the reconstructed 3D dose and the expected dose calculated by the treatment planning software Pinnacle³ was on average below 1.5 % of the maximum dose for both integrated IMRT and Volumetric Modulated Arc Therapy deliveries, and below 3% for each individual IMRT beam for a seven step-and-shoot IMRT plan of a brain tumor.

2. Materials and detector designs

All scintillation detectors that have been described in the literature have used either a liquid scintillator or a block of plastic scintillator, contained or embedded inside of an enclosure (tank or cell) and have at least one transparent side through which the light can be transmitted to a light imaging device. The scintillation light from the detector is collected and transmitted to a CCD or Complementary Metal Oxide Semiconductor (CMOS) chip using an appropriate optical lens that would capture the entire field of view onto the chip.

2.1. Material properties

Scintillators are designed to be transparent to their own light emissions. However, the scintillation photons produced within their own matrix material will experience Rayleigh scattering and additional scattering due to absorption and re-emission. The physical characteristics of the most widely used liquid scintillator (BC-531) [3,4,5,6,10] is compared to the most commonly known and used plastic scintillator (BC-400) [6] manufactured by Saint-Gobain Ceramics & Plastics, Valley Forge, PA and the recently used plastic scintillator EJ-260 [11] manufactured by Eljen Technology, Sweetwater, TX. These scintillating materials that do not contain any high Z materials are compared to polystyrene and water to ascertain there water-equivalence.

Table 1. Physical characteristics of the liquid scintillator BC-531 compared to the BC-400 plastic scintillator, EJ-260 plastic scintillator, polystyrene, and water.

	BC-531	BC-400	EJ-260	Polystyrene	Water
Emission (% of anthracene)	59	65	60	N/A	N/A
Peak wavelength (nm)	425	423	490	N/A	N/A
Electron density (10^{23} e/g)	2.930	3.272	3.350	3.238	3.343
Specific gravity (g/cm ³)	0.870	1.032	1.023	1.060	1
Composition	1:11.98	1:8.420	1:8.460	1:7.740	1:11.19
[Z: fraction by weight (%)]	6:88.02	6:91.58	6:91.54	6:92.26	8:88.81

Another organic liquid scintillator [12], OptiPhase HiSafe 3 (PerkinElmer, Waltham, MA) is being evaluated as a replacement for future scintillator detectors because its density is nearer to water than that of BC-531 (0.986 vs. 0.869 g/cm³) has not been included in the above table.

2.2. Detector design and construction

When designing any kind of detector system, one will always consider the primary clinical applications being sought, the type of ionizing radiation being detected and the required performance (dosimetric characteristics, spatial resolution, speed etc...). There will always be trade-offs in the selection and construction of all the components of such systems to make them useful for as many applications as possible, the most challenging one being clinically practical and cost effective.

2.2.1. Proton therapy beams. Figure 1 shows the LS detector systems used by Archambault *et al* 2012

[5] and Robertson *et al* 2013, 2014 [6,10]. The CCD camera used in both studies is a Luca^{EM} S 658M (Andor Technologies, Belfast, Northern Ireland). This is a 14 bits CCD with a resolution of 658 x 496 pixels and a physical pixel size of 10 μm , capable of measuring 37 full frames per second.

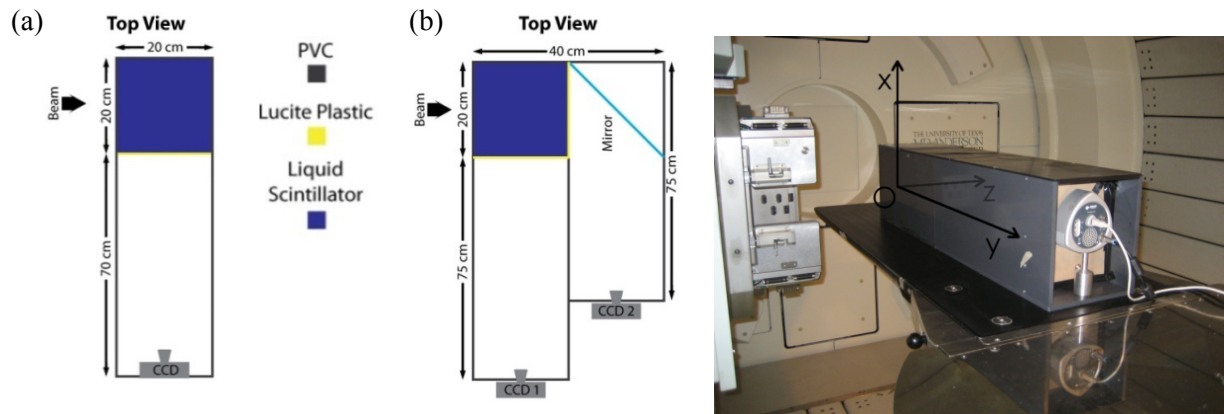


Figure 1. (a) Schematic of the LS detector system with one CCD camera (b) the configuration with two CCD cameras to resolve the 3rd dimension and (c) the actual LS detector system illustrated in Figure 1 (a) being set-up on a robotic couch at the scanning beam gantry at the Proton Therapy Center of the University of Texas MD Anderson Cancer Center in Houston TX.

2.2.2. Photon therapy beams. Although it is not evident to justify the need for developing additional 3D QA or *in vivo* dose verification of photon RT treatments, studies have looked at using volumetric scintillation dosimetry for these purposes and have proven to be an attractive and an efficient option.

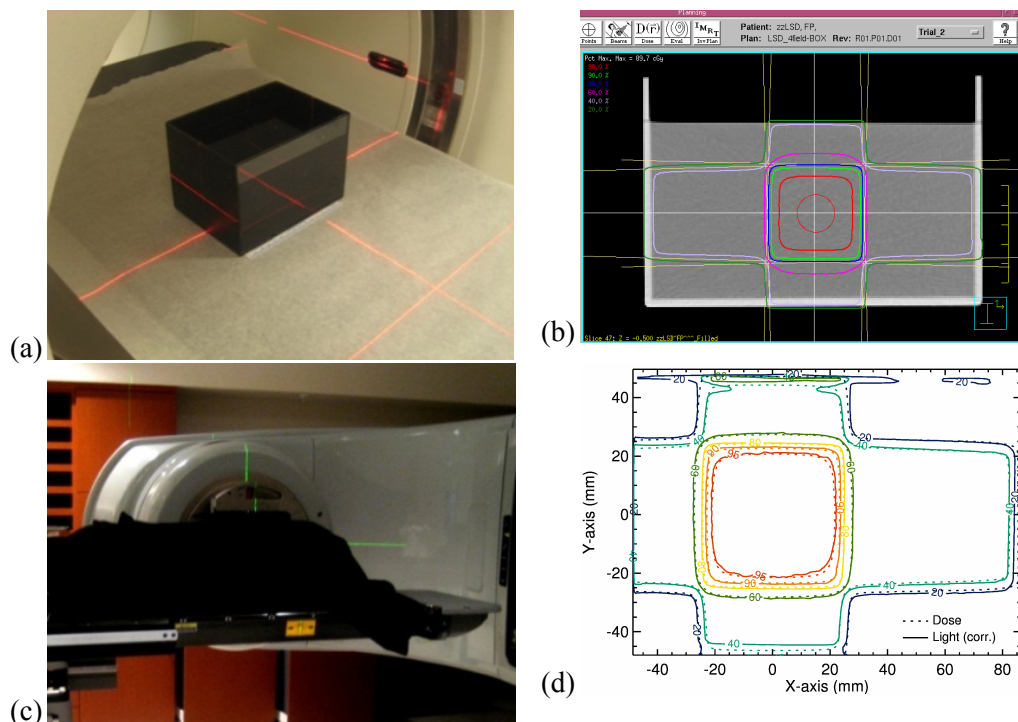


Figure 2. Experimental validation of the LS system for the QA of photon RT: a) CT simulation, b) treatment planning, c) treatment delivery, d) and image acquisition/processing of the light distribution produced within the LS detector compared to the expected dose distribution.

This approach would have to compete with EPID dosimetry, the ArcCHECK system, the ScandiDos Delta 4 system, the MatriXX system or the MapCheck system to establish itself in the marketplace. Despite these uncertainties, initial studies have been performed investigating the capabilities of these systems for photon RT. Fig. 2 shows an extension of the system that was described by Beddar *et al* [3] for IMPT and the steps undertaken by Poenisch *et al* 2009 [4] to verify the dose distribution of a simple four-field-box technique. Excellent agreement was found between the corrected scintillation light and the dose distribution calculated with the treatment planning system throughout both the high and low dose regions

The system used by Goulet *et al* 2014 [11] (*in press*) is shown in Fig. 3. They have developed and validate a novel type of high-resolution 3D dosimeter based on the real-time light acquisition of a plastic scintillator volume using a plenoptic camera. Using this, they were able to perform millimeter resolution, water-equivalent dosimetry of an IMRT and VMAT plan over a whole 3D volume.

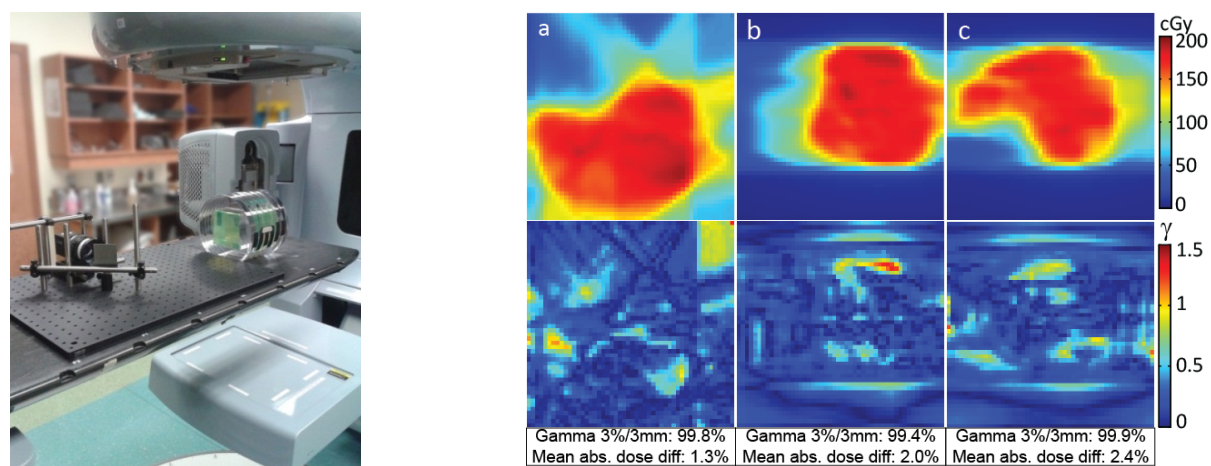


Figure 3. (Left): A plastic block scintillator system using a plenoptic camera in conjunction with an EPID for 3D photon QA. (Right) Axial (a), sagittal (b), and coronal (c) planes from a 3D reconstruction of an IMRT brain treatment measured with the plenoptic camera and the portal imaging device. The 2D gamma maps for the criteria (3% dose, 3 mm distance) are also shown.

3. Basic properties

3.1. Linearity with absorbed dose

An ideal detector system should respond linearly to the absorbed dose deposited within the detector. Poenisch *et al* [4] measured the light signal of the LS using a 6 MV photon beam, keeping the background noise constant by using a constant acquisition time of 25 s, a dose rate of 600 MU/min and a field size of 4x4 cm².

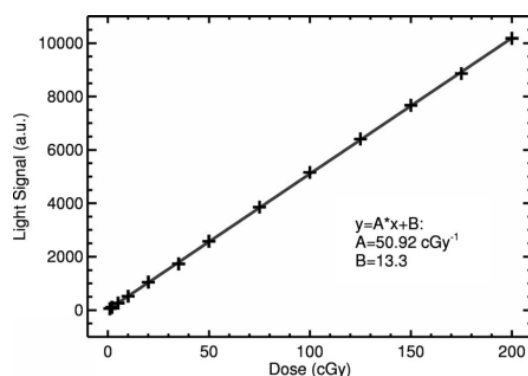


Figure 4. Measured light-dose response shown by symbols. The solid line depicts a linear fit with a correlation coefficient of $R=0.99995$.

3.2. Spatial and temporal resolution

In general the spatial and temporal resolution is dictated by the physical characteristics of the CCD camera (pixel size and image acquisition time). However, these two properties should be experimentally determined for the particular LS system and application. Furthermore, the spatial resolution is determined by the blurring of the image due to light diffusion and the discrete nature of the pixels. The system described by Beddar *et al* [5] had a spatial resolution of 0.38 mm. In general all the systems that have been described here are capable of providing submillimeter spatial resolution.

The temporal resolution is limited by the maximum frame rate of the camera but also will be limited by the signal to noise ratio needed to make accurate measurements. This is one of the reasons that we selected cooled CCD cameras in our work. Another important consideration is the readout time of the camera, which determines the dead time of the detector between image frames.

3.3. Determination of particle range and lateral position for spot scanning

Fig. 5 (a) shows the difference between the nominal proton range in water and the range measured in the LS detector system as reported by Archambault *et al* 2012. They also determined the lateral position of single proton spots with great precision with an average difference between the measured and expected spot position of 0.11 ± 0.34 mm (1σ) and a maximum difference of 0.6 mm.

Fig. 5 (b) shows a sample of a proton spot measured with a similar LS detector system illustrating the intensity in color scale by Robertson (unpublished).

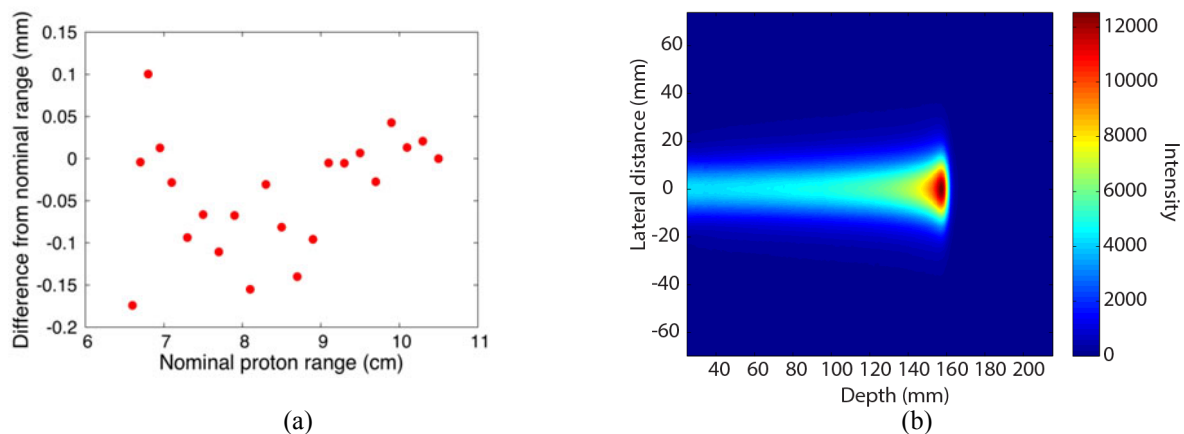


Figure 5. (a) Comparison of measured proton beam range in the LS detector system with expected range [5]. (b) A 144.9 MeV proton pencil beam measured with a liquid scintillator.

4. Optical artefact corrections

A number of artefacts need to be corrected in order to accurately measure the scintillation distribution produced inside the LS system. Table 2 lists their sources, physical phenomena and their effects [10].

Table 2. Optical artefacts in a volumetric scintillation dosimetry system.

Artefact source	Physical phenomenon	Effect
Light propagation in the scintillator and tank	Photon scattering	Blurring of light signal
	Refraction	Changes in effective pixel size and intensity
	Perspective	Changes in effective pixel size with depth
Optical train	Vignetting	Decreased brightness at image periphery
	Lens distortion	Radial variation in pixel size and location
	Lens point spread function	Blurring of light signal
CCD chip	Stray radiation	Hot pixels and streaks
	Background noise	Measurement uncertainty and pixel value offset

Robertson *et al* [10] developed correction methods for these artefacts and compared corrected light signals to calculated light distributions from a validated Monte Carlo model. The calculated and corrected light distributions were compared using gamma analysis. Table 3 gives the results, showing marked improvement in passing rates after applying the corrections to images of proton pencil beams.

Table 3. Gamma analysis pass rates for proton pencil beam light distributions as compared to projected dose distributions calculated using Monte Carlo methods.

Gamma Criteria	85.6 MeV		100.9 MeV		144.9 MeV		161.6 MeV	
	Orig	Corr	Orig	Corr	Orig	Corr	Orig	Corr
3%/3mm	80.7%	99.1%	81.9%	99.7%	84.5%	100.0%	95.3%	100.0%
3%/2mm	61.2%	95.3%	63.5%	98.4%	65.3%	99.9%	76.2%	99.9%
2%/2mm	60.6%	94.9%	62.9%	98.3%	64.6%	99.9%	75.5%	99.8%

5. Quenching correction

The LET-dependent response of organic scintillators, called ionization quenching, is problematic for proton beam dosimetry because of the steep increase in LET in the proton Bragg peak. The most common and simple model for ionization quenching of scintillators is the Birks model, which assumes that the degree of quenching is linearly proportional to the LET [7,8]. Using this model, Robertson *et al* [6] developed a voxel-by-voxel quenching correction method for use in volumetric scintillation dosimetry of proton beams. This model was tested on proton beams of four different energies. The corrected central-axis light signal matched the Monte Carlo-calculated dose within $\pm 5\%$ (Fig. 6a). Fig. 6b shows a comparison of the uncorrected light signal measured with the scintillator detector, the calculated dose, and the corrected light signal, showing a good qualitative agreement between the calculated dose and corrected light distributions.

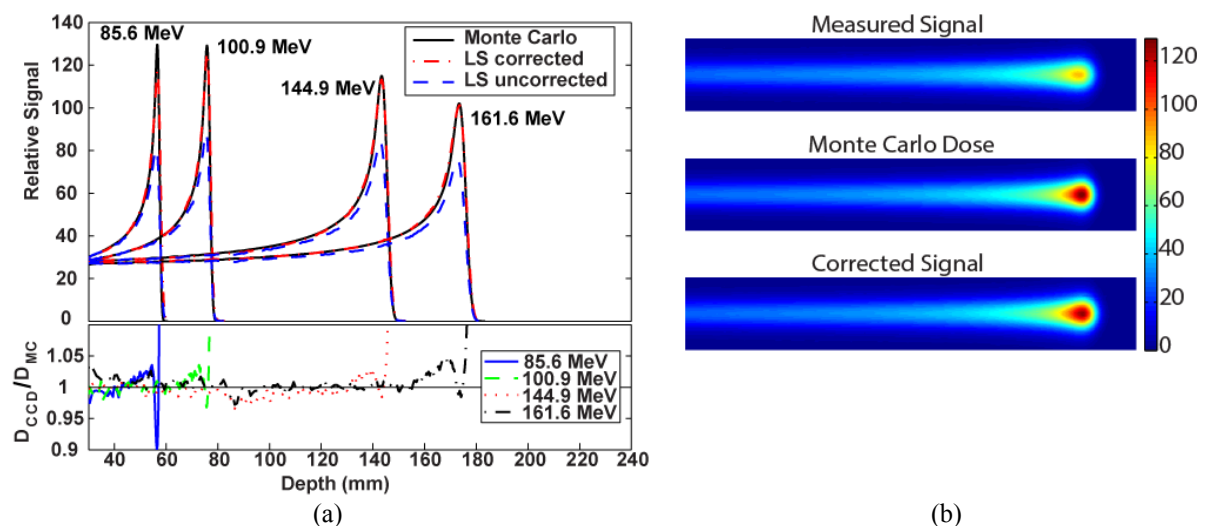


Figure 6. (a) The central axis depth-dose profiles for proton pencil beams (top): the dose calculated by the validated Monte Carlo model is shown in black, the uncorrected scintillation signal is shown in blue and the corrected scintillation signal is shown in red, and (below) the ratio of the corrected scintillation signal to the Monte Carlo dose. (b) The measured scintillation signal (top), the dose calculated using Monte Carlo methods (centre) and the corrected scintillation signal (bottom) for a 100.9 MeV proton pencil beam [6].

6. Conclusion and future directions

All the work developing liquid or plastic scintillator systems and the related theoretical studies that

have been done so far have demonstrated the feasibility and utility of using such detector systems for 2D and 3D dosimetry for photon and proton RT. These systems have been shown to be capable of submillimeter spatial resolution and the ability to perform real-time measurements.

Studies have shown that optical artefacts introduce non negligible deviations into the scintillation light distribution measured by these volumetric detectors and indicated that it is essential to correct for them to accurately measure the intensity and the spatial distribution of the scintillation light emission.

When these systems are used in proton beams, one must also correct for scintillation quenching. Methods have been developed to do so with sufficient accuracy to fulfil dosimetric quality assurance and verification purposes. These methods require prior knowledge of the LET distribution of the beam and the Birks model parameters for the scintillator. Further research is still warranted in this area to get a better understanding of this phenomenon and the theory behind it and to improve the accuracy of quenching correction techniques.

Nevertheless sufficient work has been done over the past five years in real-time volumetric scintillation dosimetry that it is ready for technology transfer from the laboratory setting to commercial research and development (R&D) settings.

7. References

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