ELSEVIER

Brachytherapy ■ (2017) ■

BRACHYTHERAPY

Validation of plastic scintillation detectors for applications in low-dose-rate brachytherapy

François Therriault-Proulx[1], Luc Beaulieu[2,3], Sam Beddar[1,4,*]

[1]Department of Radiation Physics, The University of Texas MD Anderson Cancer Center, Houston, TX

[2]Département de physique, de génie physique et d'optique et Centre de recherche sur le cancer, Université Laval, Québec, QC, Canada

[3]Département de radio-oncologie et CRCHU de Québec, CHU de Québec, QC, Canada

[4]The University of Texas Graduate School of Biomedical Sciences at Houston, Houston, TX

ABSTRACT

PURPOSE: To develop a plastic scintillation detector (PSD) capable of accurately measuring dose around a low-dose-rate (LDR) iodine-125 (^{125}I) radioactive seed as a first step toward in vivo dosimetry for prostate LDR brachytherapy.

METHODS: Using a GEANT4 based Monte Carlo code, photon energy distribution at any position around a realistic ^{125}I source model was obtained. This energy distribution was convolved with the expected energy response from a plastic scintillator and dosimetry accuracy was evaluated. A PSD was constructed and validated in a water phantom for the entire range of clinically relevant positions around an ^{125}I radioactive seed.

RESULTS: The effect of energy dependence on dosimetry accuracy was shown to be limited, with a maximum relative difference of 1.2% from the calibration condition. A sophisticated approach to account for the energy dependence of PSDs is, therefore, not required if the detector is calibrated using the same model of radioactive seed or a geometrically similar one. The measurements were in good agreement with theoretical models for the entire clinical range.

CONCLUSIONS: This study shows that PSDs can be used for accurate dosimetry in real time around a single ^{125}I seed used in LDR prostate brachytherapy and is promising for clinical applications.

Keywords: Brachytherapy; Scintillation; *In vivo* dosimetry; LDR; Fiber

Introduction

Plastic scintillation detectors (PSDs) have several practical advantages for use as radiation dosimeters. Their flexibility, small size, fast response, linearity to dose, energy independence (for energy above 100 kV), and water equivalence make them ideal candidates for *in vivo* dosimetry (1, 2). Several research groups have shown that PSDs have valid applications in both external photon and electron beams in the megavoltage energy range and in iridium-192 high-dose-rate brachytherapy (1−6). There is also interest in using PSDs in lower energy treatment and diagnostic modalities, but at the energy ranges used in these modalities, PSDs cannot be assumed to be energy independent (7−11). Indeed, for photon energies under 100 keV, the mass energy−absorption coefficient of the scintillating materials (polystyrene or polyvinyl toluene) varies strongly with energy. Furthermore, plastic scintillators have also been shown to have an intrinsic sensitivity that varies with energy, often referred to as quenching (10−12).

One potential lower energy application of PSDs is real-time *in vivo* dosimetry during low-dose-rate (LDR) brachytherapy. In LDR brachytherapy for prostate cancer, the detector could be inserted into the urethra or rectum to measure the dose delivered to a specific point during needle insertion and at the end of treatment. Such a technique would be helpful for both verifying and

Received 1 February 2017; received in revised form 29 March 2017; accepted 4 April 2017.

Financial disclosure: F. Therriault-Proulx was supported in part by the Odyssey Program at The University of Texas MD Anderson Cancer Center and by the Natural Sciences and Engineering Research Council of Canada. S. Beddar and the research reported in this publication were supported in part by the National Cancer Institute of the National Institutes of Health under Award Number R01CA182450.

Conflict of interest: The authors have no relevant conflicts of interest to disclose.

* Corresponding author. Department of Radiation Physics, Unit 1420, The University of Texas MD Anderson Cancer Center, Houston, TX 77030. Tel.: (713) 563-2609; fax: (713) 563-2545.

E-mail address: abeddar@mdanderson.org (S. Beddar).

1538-4721/$ - see front matter
http://dx.doi.org/10.1016/j.brachy.2017.04.002

documenting the dose delivered to the tumor and surrounding organs at risk. The temperature dependence of the PSD would have to be accounted for as there would likely be a difference between calibration (e.g., room temperature) and *in vivo* temperatures (13−16). However, the problem of energy dependence and other challenges still have to be overcome. In particular, the steep dose gradients associated with LDR brachytherapy complicate the development of accurate detectors. Measuring a dose rate on the order of 0.1 μGy/s at 1 cm from an iodine-125 (^{125}I) seed—and a dose rate of about 68 times lower at 5 cm—necessitates the development of sensitive and optimized detectors that are versatile enough to be inserted into the tumor or surrounding healthy organs. The goal of our study was to develop such a detector and validate its dose measurement accuracy at clinically relevant positions around a single ^{125}I radioactive source. Our main assumption was that a detector capable of single-seed dosimetry should also be successful in clinical applications where multiple seeds are implanted needle by needle (with each needle generally loaded with two to four seeds). Our main goal in this study was to evaluate the effect of the energy dependence of PSDs on dose measurement accuracy using Monte Carlo simulations (17). In-phantom measurements were performed and compared with expected doses from the American Association of Physicists in Medicine Task Group no. 43 (TG-43) for the same type of radioactive source (18−20).

Methods and materials

Monte Carlo study of energy distribution

To obtain a model of the energy spectral distribution around an ^{125}I radioactive seed, Monte Carlo simulations were performed using the GEometry ANd Tracking 4.9.6 (GEANT4.9.6) toolkit (21). Our code was derived from the ALgorithm for heterogeneous dosimetry based on GEANT4 for BRAchytherapy and for which characteristics were detailed by Afsharpour *et al.* (17). The simulations were performed on the Lonestar computer cluster at the Texas Advanced Computing Center (The University of Texas, Austin, TX) (22).

An ^{125}I source model (Oncoseed 6711; Amersham Health, Atlanta, GA) was placed at the center of a water sphere of 30 cm in diameter consisting of 1-mm voxels. Dose-weighted photon distributions were calculated from generation of more than 100 million events for distances up to 6 cm from the source in the radial and longitudinal directions. Absorbed dose to the different positions was estimated using the linear track length estimator (23). To evaluate the effect of changes in the spectral distribution as a function of detector-to-source position, the obtained dose distributions (S) were convolved with a typical response spectrum of plastic scintillators with energy (R), derived from Peralta and Rego (10) for a BCF-10 (Saint-Gobain Crystals, Hiram, OH) scintillating fiber and a 0.5-mm Al filtration. This gives an estimate of the light output (LO) as a function of radial (r) and longitudinal (z) distances from the radioactive source as detailed in the following equations:

$$\mathrm{L.O.}(\mathrm{r,z}) = \sum_{E=5\ \mathrm{keV}}^{40\ \mathrm{keV}} \left[\mathrm{S}(E)|_{\mathrm{r,z}} * \mathrm{R}(E)\right] \quad (1)$$

$$R(E)|_{\mathrm{r,z}} = R_{\mathrm{PSD}}(E) * \left[\frac{(\mu_{en}/\rho)_{\mathrm{PS}}}{(\mu_{en}/\rho)_{\mathrm{water}}}\right](E) \quad (2)$$

where $R_{\mathrm{PSD}}(E)$ is the intrinsic response of the PSD with energy and $(\mu_{en}/\rho)_{mat}$ the mass energy−absorption coefficient for a given material (*mat*).

The PSD calibration point was set at 1 cm from the ^{125}I source along its radial axis (r = 1 cm, z = 0 cm), in accordance with the TG-43 formalism (18−20). The effective point of scintillation was accounted to be at the center of the scintillator throughout this study. The effect of energy dependence of PSDs on dose measurement accuracy was calculated as the relative difference in the responses between a given position and the calibration position after normalization to deposited dose (Eq. (3)).

$$\Delta\left(\frac{\mathrm{L.O.}}{D}\right)(\mathrm{r,z}) = \frac{\frac{\mathrm{L.O.(r,z)}}{D(\mathrm{r,z})} - \frac{\mathrm{L.O.(1cm,0cm)}}{D(\mathrm{1cm,0cm})}}{\frac{\mathrm{L.O.(1cm,0cm)}}{D(\mathrm{1cm,0cm})}}. \quad (3)$$

We also examined the effects of the materials of PSDs on the photon energy spectrum. Four Monte Carlo simulations were performed, during which the photon energy spectrum was acquired at r = 1 cm and z = 0 cm. A polystyrene scintillator (1 mm diameter and 5 mm long), a polymethyl−methacrylate transmission optical fiber (1 mm diameter), and a polyethylene jacket (2.2 mm outside diameter) were sequentially added to the simulation in water, and the spectra were compared. The construction and orientation of PSDs with respect to the ^{125}I radioactive seed are presented in Fig. 1.

In-phantom measurements

A 5-mm long × 1-mm diameter blue scintillating fiber (BCF-12; Saint-Gobain Crystals, Hiram, OH) was coupled to a 4-m long optical fiber with a numerical aperture of 0.6 to increase light collection (24). A Mylar reflector was placed at the tip of the scintillating element to enhance light reflectivity. A photon-counting photomultiplier tube (PMT) (H7828; Hamamatsu, Hamamatsu City, Japan) was used as the photodetector and placed inside a temperature-controlled enclosure (6.0 ± 0.2°C). A schematic of PSD and experimental setup are shown in Fig. 1.

To minimize the uncertainty related to the water equivalence and energy distribution that would occur if a solid phantom was used at the LDR brachytherapy energy range, we used a water tank. The water tank was customized in-