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Flexible Optically Stimulated Luminescence Band for 1-D *in vivo* Radiation Dosimetry

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Abstract

In vivo dosimetry helps ensure the accuracy of radiation treatments. However, standard techniques are only capable of point sampling, making it difficult to accurately measure dose variation along curved surfaces in a continuous manner. The purpose of this work is to introduce a flexible dosimeter band and validate its performance using pre-clinical and clinical X-ray sources. Dosimeter bands were fabricated by uniformly mixing BaFBr:Eu storage phosphor powders into a silicone based elastomer. An optical readout device with dual-wavelength excitation was designed and built to correct for non-uniform phosphor density and extract accurate dose information. Results demonstrated significant correction of the non-uniform readout signal and excellent dose linearity up to 8 Gy irradiation using a pre-clinical 320 kV X-ray system. Beam profile measurements were demonstrated over a long distance of ~30 cm by placing multiple dosimeters in a single line and stitching the results. The performance of the dosimeters was also tested using a clinical linear accelerator (6 MV) and compared to radiochromic film. Once bias corrected, the bands displayed a linear dose response over the 1.02 – 9.36 Gy range ($R^2 > 0.99$). The proposed system can be further improved by reducing the size of the readout beam and by more uniformly mixing the phosphor powder with the elastomer. We expect this technique to find application for large-field treatments such as total-skin irradiation and total-body irradiation.

Keywords

in vivo dosimetry; wearable device; radiation therapy; optically stimulated luminescence

1. INTRODUCTION

Radiation therapy (RT) is a common treatment for cancer. The wide variety of available techniques mirrors the unique geometrical and dosimetric requirement of different tumor sites. For some treatments, *in vivo* dosimetry is required because treatment planning alone is insufficient to ensure accurate coverage of the target. For instance, total skin irradiation (TSI)^{1,2} is effective only if the radiation dose is uniformly delivered to the patient, and therefore *in vivo* dosimetry is used routinely in clinical practice to ensure the uniformity of the delivered treatment. Treatment of superficial targets that abut the patient's surface can also use *in vivo* dosimetry to verify surface dose. In addition, large-field treatments such as

total body irradiation (TBI)³ can use surface dose measurement to estimate dose to deeper tissues.

The most prominently used methods for patient-specific *in vivo* dosimetry are thermoluminescent dosimetry (TLD)^{4–6} and optically stimulated luminescence dosimetry (OSLD)^{3,7,8}. A specific number of TLDs or OSLDs (typically 10) are strategically placed over the patient's skin to sample the radiation field and verify dose uniformity. After irradiation, the stored dose information can be retrieved as an optical signal by exposing the material to heat or light, respectively. However, the number of measurements is limited since the dosimeters measure dose at single points, resulting in coarse sampling of the radiation field. Moreover, for fractionated treatments, it is difficult to track the integrated dose distribution. As a result, the completed treatment may incur cold spots or under-treated areas that potentially require additional boosts to prevent recurrence, and hot spots or over-treated areas that may lead to unintended patient toxicity.

Recently, spatially resolved *in vivo* dosimetry techniques were developed to overcome such limitations. Examples include local or whole-body dosimetry using scintillating sheets⁹, Cherenkov luminescence^{10–12}, and glass beads with TLD properties^{13,14}. While these methods provide high-resolution dosimetry images, they require the patient stage to be modified (e.g., removing the electron scatter plate for electron skin irradiation), equipment (e.g., camera) to be rigorously shielded to reduce radiation damage to the device, and images to be spatially calibrated due to image distortion and warping effects from the irregular patient surface. Other spatially resolving dosimeters were also explored using OSLD by embedding phosphor powders into plastic sheets or strips^{15,16}. While this is an effective method since OSLD strips can be collected and measured at later time points, the dosimeter may have limited flexibility and the powder density and grain size may vary spatially, potentially affecting the dose response when scanning the dosimeter.

Here we introduce a flexible dosimeter for 1-D dosimetry, and an optical readout platform that can automatically correct the dosimeter for phosphor density and efficiency variations. As proof of concept, OSLD material was embedded in a silicone elastomer and cut into thin bands. Various shapes can be achieved depending on the mold used. A dose readout and correction system that uses dual-wavelength excitation was designed and evaluated using a pre-clinical X-ray system and a clinical linear accelerator (linac). Results demonstrated a linear dose response to both kV and MV X-ray irradiation. Moreover, extended 1-D dose information was acquired by placing multiple dosimeters in series and by stitching the linear dose response profiles. Based on the results, it is expected that surface dose maps could be reconstructed by stitching multiple 1-D dose profiles wrapped around curved objects, enabling non-invasive large-field dosimetry on patients receiving superficial or large-field radiation treatments.

2. MATERIALS AND METHODS

2.A. Fabrication of flexible dosimeter bands

Flexible dosimeter bands were fabricated by embedding BaFBr:Eu storage-phosphor powders (Radiation Monitoring Devices, Inc.), into polydimethylsiloxane (PDMS) silicone

elastomers. While BaFBr:Eu is not ideal due to its high effective atomic number¹⁷, long fading time¹⁸, and afterglow (phosphorescence)¹⁹, BaFBr:Eu was selected because it is commercially available and its physical mechanism^{20,21} is similar to that of other OSLD materials such as Al₂O₃:C^{8,22}. Optically stimulated luminescence (OSL) is a phenomenon used to measure absorbed dose optically. When a storage phosphor is irradiated with ionizing radiation beam, the dopant (e.g. europium) loses an electron. The free electron is then trapped in a vacancy within the bromine lattice, between the valence band and conduction band. These electrons can relax back to their ground state when they are exposed to a suitable wavelength of light, typically 633 nm. These storage phosphor bands can therefore be readout after treatment to estimate the 1D dose profile along a curve surface.

PDMS (Sylgard 184, Dow Corning, Co.) was selected as the encapsulating material since it is optically clear, moldable, and extremely flexible. BaFBr:Eu powder was encapsulated in a PDMS base solution at a concentration of 30% w/v, and solidified by adding the corresponding PDMS curing agent concentration at a concentration of 10% v/v (Fig. 1). Once the curing agent was sufficiently mixed, residual air bubbles were removed by placing the PDMS mixture in a vacuum desiccator for 10 – 15 minutes. The mixture was then poured onto a 50.8 mm × 76.2 mm (width × length) glass plate, which had been previously pre-treated with a hydrophobic coating (Novec 1720, 3M) to prevent the solidified PDMS from bonding to the glass substrate. The PDMS was placed on a hot plate and cured at 75°C for approximately 1 hour. Once the PDMS was fully cured, the solidified sheet was peeled off and cut into 5 mm × 100 mm × 1 mm strips.

2.B. Development of dose readout system

A prototype dose readout system was developed to measure the radiation dose recorded by the flexible dosimeter band (Fig. 2a). The readout system comprises a linear translation stage, two excitation light sources, and an optical detector (Fig. 2b). The two light sources are designed to excite two different types of luminescence, namely OSL and photoluminescence (PL). In practice, this design could be adapted to a more compact rotating spool to read arbitrarily long dosimeter bands.

PL, defined as a spontaneous light emission following excitation^{21,23}, is a distinct phenomenon from OSL. Through this process, the BaFBr:Eu phosphor can be excited at a wavelength of ~300 nm to emit in the 350 – 420 nm range. Since PL is similar to fluorescence, the intensity of the light emitted in response to UV excitation is proportional to the density of the luminescent material. Hence, the PL signal from the dosimeter band can be utilized to measure the density of BaFBr:Eu that is encapsulated within the PDMS elastomer, and in this way normalize the response of the system.

The optical readout system can measure both OSL and PL signals from the dosimeter band. Since both OSL and PL have similar emission spectra, the two signals are differentiated by first measuring the OSL signal using a red excitation source, then measuring the PL signal using a UV excitation source. In this way, the OSL signal can be normalized or scaled by the PL signal. This allows self-correction of signal variations due to non-uniform density of the encapsulated storage phosphor powders. A 635 nm rectangular laser was used as the OSL excitation source (Thorlabs, CPS635), and a laser light sheet was achieved by placing an $f=$

50 mm cylindrical lens (Thorlabs, LJ1695RM) in front of the laser source. To reduce the intensity of the laser source, neutral density filters (Thorlabs, NE05A-A and NE10A-A) were attached in front of the cylindrical lens. For the PL excitation source, a 300 nm LED light source (Thorlabs, M300L4) was used. An $f=35$ mm UV fused silica bi-convex lens (Thorlabs, LB4879-UV) was placed in the beam path to collimate the LED source, followed by an $f=10$ mm cylindrical lens (Thorlabs, LJ4918) to achieve a thin UV light sheet. A diaphragm aperture (Thorlabs, SM1D12D) was placed between the spherical lens and the light source to control the thickness of the light sheet. The emitted luminescence was detected using an amplified silicon photodetector (Thorlabs, PDF10A) fitted with an $f=35$ mm lens and UV emission filters for signal clean-up (Thorlabs, FB400-40 and Semrock, FF01-440/SP).

The dosimeter bands were loaded on a linear translational stage that can travel ~ 100 mm (Velmex, XSlide XN10-0040-M01-71). The translational stage and the light sources were both controlled using a data acquisition system (National Instruments, NI-DAQ 6002). In the readout process, the OSL emission is measured by moving the translational stage in one direction with the 635 nm laser sheet turned on. The stage direction is then reversed and the excitation source is switched to 300 nm UV light sheet to measure the PL emission. The raw fluence signal for the OSL and PL signal was sampled at a rate of 11 Hz and was recorded into a separate file for further analysis. Since the bandwidth of the detector was sufficiently low (20 Hz), no low-pass filter was applied to the output signal.

2.C. Dosimeter evaluation using pre-clinical kV X-ray

The dose response of the dosimeter was evaluated using a pre-clinical X-ray system (X-RAD 320, Precision X-Ray). The system was set to deliver a 320 kV / 12.5 mA beam filtered by 2 mm thick aluminum (half-value layer ≈ 1.0 mm copper²⁴). A single dosimeter band was placed at a source-to-surface distance (SSD) of 45 cm, and radiation exposures of 2, 4, 6, and 8 Gy were delivered to 10 separate dosimeter bands. After each exposure, the dosimeters were measured using the prototype readout device. Since the storage phosphors are sensitive to light, experiments were performed with room lights turned off. Each scan measured OSL and PL signals along the dosimeter band to extract the dose information and the phosphor density, respectively. After each reading, the dosimeters were erased by exposing them to fluorescent lights for > 5 minutes. Using the calibrated dosimeter bands, the beam profile of the X-ray system was observed. Three dosimeter bands were placed end-to-end to cover a total lateral distance of ± 15 cm from the beam center, extending to the edges of the specimen stage, and were then exposed with 8 Gy.

To evaluate the spatial resolution of the combined dosimeter band and readout system, a single dosimeter band was placed at the center of the X-ray field, under a stainless-steel caliper for X-ray collimation. The opening of the caliper jaws was set to create an adjustable slit ranging from 0.24 mm to 6 mm in width. For each slit size, the band was exposed with 8 Gy and the dose information was erased following each readout.

2.D. Dose evaluation using a linac

Finally, the dose response of the dosimeter bands was evaluated using a clinical linac to assess its potential use in a clinical environment. Three dosimeter bands were placed in series, with 2 cm of overlap between them to fully cover a 24 cm wide treatment field. The dosimeters were protected from light by thick black tape, which was removed for readout. They were then placed under 5 cm-thick solid water and exposed to 6 MV X-ray at 95 cm SSD. Dose was modulated by sequentially opening leaves from the multileaf collimator (MLC), yielding a 12-step staircase dose distribution ranging from 1.02 Gy to 9.36 Gy. To validate the dosimetric performance of the dosimeter bands, a single Gafchromic film (Ashland) was placed under the dosimeters and read out using a film scanner (Epson Expression 10000XL, 300 dpi, 48 bits per pixel). The scanned images were evaluated with ImageJ using the red channel only for larger dynamic range²⁵.

3. RESULTS AND DISCUSSIONS

3.A. Calibration for translational stage travel speed

Prior to X-ray irradiation, the speed of the translation stage was measured by placing a ruler on the stage and detecting the reflected UV excitation light. The stage showed a highly reproducible behavior with a measured traveling speed of 3.75 mm/s. This calibration value was applied to all the following dosimeter readout data.

3.B. Dose response under X-Ray irradiation

The dosimeters were exposed to 0 – 8 Gy X-ray dose (320 kVp, 12.5 mA), in increments of 2 Gy, and the OSL signal was measured by scanning the band with the red laser light sheet (Fig. 3 (a)). The OSL signal was then normalized by the PL signal acquired using the UV excitation source (Fig. 3(b)), to yield the corrected dose (Fig. 3 (c)). Spatial uniformity of the signal was improved two-fold, as measured by the coefficient of variation ($COV(\%) = \text{standard deviation of the spatial distribution} / \text{mean dose} * 100\%$), which decreased from 4.6 % to 2.4%.

The signals measured ~ 5 mm from the dosimeter edges were intentionally truncated to avoid artifacts in the readout signal. This is because the transparent dosimeter bands conduct light similarly to optical fibers and the signal is elevated at the edges due to light piping. The edge effects may be reduced by applying dark coating on the sides of the dosimeters. It should be noted that the PL signal of the dosimeter is invariant, even after the dosimeter is irradiated. We observed the average difference in PL between the same irradiated and non-irradiated dosimeter to be < 0.5 %.

To confirm that this trend is reproducible with other dosimeter bands, a total of 10 different dosimeters were investigated. As shown in Fig. 4, the regression line demonstrates excellent linearity up to 6 Gy ($R^2 > 0.99$) and follows the dimensionless equation,

$\left(\frac{OSL}{PL}\right) = 0.15 \cdot \text{Dose [Gy]} + 0.02$, demonstrating that the PL signal can indeed be used as a reference value to correct the dosimeter reading for density variation across each dosimeter. The error bars in the figure represent an average COV of $\pm 4.3\%$ for the measured OSL/PL signal. The slope of the response curve for the individual dosimeters has a COV of $\sim 3.5\%$.

One thing to note in Fig. 4 is that the linear regression line was estimated between 0 – 6 Gy. This is because the OSL signal started to saturate around 8 Gy. Another point to note is that the y-intercept line has a finite bias signal of ~ 0.1 Gy of radiation dose. This bias may be due to small non-linearities in the response of the dosimeter.

The lateral profile of the 320 kV X-ray beam was also measured using multiple dosimeter bands. Three dosimeter bands were lined up in a row and exposed to 8 Gy. Results show that the dose profile of the three dosimeter bands can be stitched together to provide a single dose profile along a given direction (Fig. 5). In the figure, the dose profile is relatively uniform near the center of the field (± 5 cm). One thing to note in the figure is that the dose profile is not symmetric: the edge of the field is clearly visible on the right band but not on the left one. This suggests that the center of the platform-ray cabinet does not match the beam isocenter, and we indeed found the X-ray tube to be slightly tilted. We conclude from the figure that multiple dosimeter bands can be connected to characterize the 1-D dose profile inside a large radiation field.

Since the current dosimeter and readout platform is a proof-of-concept design, further steps must be performed to substantially enhance accuracy and precision. In comparison to clinical dosimeters, with typical COV of $< 1.0\%$ ²⁶, the average COV of the dosimeter band is significantly high ($\sim 3.5\%$). There are several factors that potentially affect the precision of the current system. One issue may be due to how the excitation beams are oriented in the reader device. The OSL excitation beam and the PL excitation beams are oriented at an angle ($\sim 45^\circ$) from each other, with the detector positioned in the middle. This may cause a discrepancy in the illuminated volume when using PL to correct the OSL signal. Future iterations of the design will include dichroic mirrors to match the optical path between the two excitation beams and emission readout.

Another phenomenon that may have increased the uncertainty in the dose response curve is the fading effect. Storage phosphors are known to have signal fading, or partial loss of stored information, even when kept in the dark²⁰. A fading experiment was performed by scanning a dosimeter with the red excitation light source 2 to 20 minutes after irradiating it with 8 Gy. The readout signal was reduced up to 7.9 % in the first 5 minutes (Fig. 6a), indicating that dose uncertainty could be significant if a varying amount of time elapses between irradiation and readout. To minimize the fading effects, readout in the current experiment was performed within 1 minute of X-ray irradiation. Uncertainty due to fading can be further reduced by measuring the dosimeter after the signal has stabilized (e.g., 10 minutes²⁷).

Other phenomena that may adversely affect accuracy include variations in delivered X-ray dose and afterglow effects. Based on 34 X-ray exposures, we observed a variation of 0.6 % (COV) in the dose delivered by the irradiator, as measured by the internal ion chamber. Afterglow was also observed in the phosphor bands. Previous studies have shown that BaFBr:Eu exhibits afterglow due to electrons trapped at the shallow traps¹⁹. We found that afterglow increased background signal by up to 1.1% relative to the OSL signal (Fig. 6b). Delaying the readout by at least 10 minutes would further reduce uncertainty.

An additional experiment was performed to measure the time constant of afterglow using a preclinical optical imager (Precision X-Ray) positioned inside the cabinet X-ray. A single band was irradiated with 2 Gy and luminescence images were captured immediately after irradiation (f -stop 0.95, 8×8 binning, 10 s exposure each frame, 5 min image time). Fitting the intensity profile with a two-exponential curve yielded two time constants of approximately 20 s and 225 s. This demonstrates that the afterglow may affect the results long after the irradiation and will depend on when the dosimeters are analyzed. While it is possible to wait for >18 minutes (5 time constants) after each irradiation to minimize such effects, the another solution to this problem is to use a material other than BaFBr:Eu. For instance, $\text{Al}_2\text{O}_3\text{:C}$, a popular material currently used as the gold standard in OSL dosimetry, has an afterglow time constant of only a few seconds^{28,29}. However, due to fading, the $\text{Al}_2\text{O}_3\text{:C}$ signal still takes ~ 8 minutes to stabilize³⁰, and the most appropriate approach to minimize uncertainty is to measure the dosimeters at consistent time points post-irradiation.

3.C. Spatial Resolution of the Dosimeter Band and the Readout System

The spatial resolution of the combined dosimeter band and readout system was evaluated by exposing the dosimeter to a collimated X-ray beam of increasing width. The beam size was controlled by placing the dosimeter under a stainless-steel caliper and varying the opening of the jaws. The jaws of the caliper, made of 3 mm-thick stainless steel, were sufficient to attenuate the 320 kV X-ray beam. The resulting dose profiles were normalized to their peak value to assess the full-width half maximum (FWHM) as a function of the slit opening (Fig. 7a). To quantify the resolution of the system, we applied the following approximation for the FWHM as a function of the collimator opening result, based on a quadrature model (Fig. 7b):

$$\text{FWHM} = \sqrt{(\text{Collimator opening})^2 + C^2}, \quad (\text{Eq. 1})$$

where the constant c represents the spatial resolution of the readout system. Based on this model, it is estimated that the resolution of the system c is $1.09 \text{ mm} \pm 4.5 \%$ (95% confidence interval). This is to be expected since the light sheet (red laser) used to read out the luminescence signal is $>1 \text{ mm}$ thick. The spatial resolution of the readout system can be improved by applying beam-reducing optics to decrease the thickness of the exciting light sheet.

3.D. Beam Characterization of Clinical Linear Accelerator

Since the goal of this project is to perform dosimetry on curved surfaces under clinical conditions, the performance of the dosimeter bands was evaluated using a clinical linear accelerator (Clinac 21EX, Varian). Three dosimeter bands were placed end-to-end and exposed to a 12-step staircase field, linearly ranging from 1.02 Gy to 9.36 Gy (Fig. 8a). While slightly varying biases were observed for different bands, potentially due to afterglow and fading effects, results demonstrate a strong linear step dose function following a simple bias-correction. Gafchromic film was used in the same experiment as a reference. The film dosimeter required a non-linearity correction prior to comparing the dose information because of the signal saturation above $\sim 5 \text{ Gy}$.

To compare the dosimeter bands and the Gafchromic film, the dose response curves of the two dosimeters were matched using the known dose delivered by the linac (Fig. 8b). Although both calibrated curves show excellent linearity, it should be noted that the film dosimeter requires a non-linear correction whereas the dosimeter band only requires a linear correction. This demonstrates that the dosimeter band can cover a larger dynamic range without correcting for non-linearity. Moreover, since the signal increase between dose steps for the Gafchromic film is smaller at higher doses, this also implies that the dosimeter band will retain higher accuracy at larger doses.

One thing to note is that the normalized emission signal (OSL/PL) from the dosimeter bands is energy dependent. This is due to the high effective atomic number of the storage-phosphor material used in this work ($Z_{eff} \approx 49$)¹⁷. The normalized intensity values of the dosimeters exposed to 6 MV X-ray beam (Fig. 8) were less than half of those exposed to the 320 kV beam, for the same dose (Fig. 4). Although the changes in MV beam energy over the surface of the patient is small, energy dependence will also need to be accounted for potential clinical applications such as those that require out-of-field measurements.

4. CONCLUSIONS

A flexible dosimeter and readout device capable of dual illumination for density correction is presented as proof of concept device for measuring radiation dose *in vivo*. The dose response and spatial resolution of the combined dosimeter and readout device were characterized using a preclinical kV X-ray irradiator. The dose profile was then measured with a clinical linac and compared with film dosimetry. The method shows that relative dosimetry can be performed using the flexible OSL dosimeter bands. Since the encapsulating polymer is highly flexible, they can be taped around the patient to acquire high-resolution 1-D dose information. Preliminary results demonstrate that the dosimeters have excellent linear response to dose over a large dynamic dose range with no linearity correction required. The uncertainty of the dose response curve may be reduced if the BaFBr:Eu is replaced with a different storage phosphor with substantially shorter afterglow and more stable OSL signal (e.g., $Al_2O_3:C$). By addressing such issues, it is expected that dosimetry can be achieved with any number of dosimeter bands since they can be self-corrected using the dual-wavelength readout device. Further improvements will be implemented in the future using micro-sieves to reduce powder size variation and industrial grade powder mixers. The optical reader could also be miniaturized for use in practical applications.

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Figure 1. Schematic diagram of (a) the fabrication steps for the flexible dosimeter bands and (b) photographs of the dosimeter bands placed on a flat surface or rolled into a spool.

Figure 2.

(a) Photograph and (b) schematic diagram of the dosimeter readout device. The dosimeter is initially scanned with the red laser sheet for OSL signal, then with the UV source for PL signal.

Figure 3.

Example of intensity profiles measured using a single dosimeter (8 Gy exposure). (a) OSL signal excited with red laser sheet; (b) PL signal excited with UV light sheet, and (c) calibrated readout obtained by computing the ratio of OSL to PL signal. The x-axis is the spatial coordinate from the start to the end of the dosimeter band.

Figure 4.

Dose response graph of dosimeters exposed to 0 – 8 Gy of X-ray irradiation. The data from 8 Gy is excluded from the linear regression line since slight saturation of the signal was observed. Error bars represent ± 1 standard deviation, from 10 dosimeter readings.

Figure 5.

Graph representing beam profile of a kV cabinet X-ray. Three dosimeters, each 10 cm long, were placed in series to cover ~30 cm.

Figure 6. Factors affecting uncertainty in the dose response curve includes (a) fading and (b) afterglow. The uncertainty may be reduced if dose readout is performed consistently after irradiation and after an extended period of time (e.g., 10 minutes).

Figure 7.

Graph representing (a) the response of the dosimeter system to a narrow X-ray beam of varying width, and (b) the measured FWHM as a function of the collimator opening size. The adjustable collimator size is achieved by opening the jaws of a stainless-steel caliper in the X-ray beam path.

Figure 8.

(a) Raw data of the dosimeter bands and Gafchromic film after exposure to 6 MV linac beam (staircase field, 1.02 Gy to 9.36 Gy), and (b) the corresponding dose response curve corrected for bias and non-linearity. Data from the three dosimeters were bias-corrected and stitched to form a single 1-D line. Corrected dose curves of both the Gafchromic film and the dosimeter bands demonstrate good linearity ($R^2 > 0.99$).