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Purpose: Optically stimulated luminescent detectors (OSLDs) have a number of advantages in radiation dosimetry making them excellent dosimeters for quality assurance and patient dose verification. Although the dosimeters have been investigated in several modalities, relatively little work has been done in examining the dosimeters for use in clinical proton beams. This study examined a number of characteristics of the response of the dosimeters in the spread-out Bragg peak (SOBP) region of clinical proton beams.
Methods: Optically stimulated luminescence (OSL) dosimeters from Landauer, Inc., specifically the nanoDot dosimeter, were investigated. These dosimeters were placed in a special phantom with a recess to fit the dosimeters without an air gap. Beams with nominal energies of 160, 200, and 250 MeV were used in the passively-scattered proton beam at the MD Anderson Cancer Center Proton Therapy Center. Dosimetric properties including linearity, field size dependence, energy dependence, residual signal as a function of cumulative dose, and postirradiation fading were investigated by taking measurements at the center of SOBPs.
Results: The dosimeters showed 1% supralinearity at 200 cGy and 5% supralinearity at 1000 cGy. No noticeable field size dependence of the detector was found for field sizes from 2×2 cm^2 to 18×18 cm^2. Residual signal as a function of cumulative dose showed a small increase for measurements up to 1000 cGy. Readout signal depletion of the dosimeters after consecutive readings showed a slightly larger depletion in protons for doses up to 500 cGy but not by a clinically significant amount. Within the center of various SOBP widths and proton energies the variation in response was less than 2%. An average beam quality factor of 1.089 with experimental standard deviation of 0.007 was determined and applied to the data such that the results were within 1.2% of ion chamber data.
Conclusions: The nanoDot OSL dosimeter characteristics were studied in the SOBP region of clinical proton beams. To achieve accurate dosimetric readings, corrections to the dosimeter response were applied. Corrections tended to be minimal or broadly consistent. The nanoDot OSLD was found to be an acceptable dosimeter for measurement in the SOBP region for a range of clinical proton beams.
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Key words: optically stimulated luminescence, proton therapy, dosimetry

I. INTRODUCTION

Optically stimulated luminescence (OSL) is an area of development that has been and can be very useful for radiation dosimetry and measurement in numerous applications.[1–6] Optically stimulated luminescent detectors (OSLDs) present a number of advantages over thermoluminescent detectors (TLDs), which have been widely used for in vivo and quality assurance measurements. Aluminum oxide with carbon doping (Al_2O_3:C), the most common OSL material in use, has little energy dependence in the MV range, less postirradiation fading than TLDs, and 40–60× the sensitivity, allowing measurement of very small or large doses.[7–9] Despite these advantages, there remain some drawbacks to OSL and Al_2O_3:C. The dosimeter must always be enclosed in some form of light-proof packaging due to the nature of the phenomenon. Additionally, the effective atomic number is 11.28, which presents challenges of tissue equivalence and enhanced response at low photon energies.[10–12] As with TLDs, OSLDs exhibit supralinearity with dose as well as a transient signal after irradiation, although this is significantly shorter than TLDs.[13,14]

The substance is made by growing aluminum oxide with carbon impurities, causing lattice defects. The defect of interest is the F-center, where an oxygen atom is replaced by two captured electrons. Ionizing radiation can give an electron enough energy to jump from the valence band to the conduction band. The defects in the crystal cause localized energy traps that can capture free electrons. The locally-trapped electrons can be freed with a broad spectrum of light, typically peaking at 475 nm.[15]

The resulting emission spectrum has two peaks. The first and most prominent is the F-center emission, centered at 420 nm, and a much smaller ultraviolet (UV) emission at 330 nm. The F-center emission changes little with time postirradiation, but the UV emission shows a strong increase in signal with time postirradiation.[16] Thus, for readout systems using continuous-wave (CW) OSL or filters that allow UV emission to pass, both timing of the readouts and knowledge of the filters used are important considerations for accurate dosimetry.

Al_2O_3:C has proven itself to be a reliable dosimeter in photon and electron dosimetry.[9,14,17] If the advantages of the material are similar in protons it would increase the possibility and popularity of using them as dosimeters of various functions within the modality. There have been a number of studies describing the response of Al_2O_3:C in proton and heavy ion irradiations with emphasis given to the optical emission efficiency as a function of linear energy transfer (LET), particularly in reference or research conditions such as in space dosimetry and heavy charged particle (HCP) research, with more recent works approaching clinical situations in proton beams.[5,6,14,18–21] However, there remains a number of questions to be answered regarding the characteristics of Al_2O_3:C in the context of clinical proton dosimetry. Because of the complex microscopic dose distribution of protons and HCPs, the assumptions and correction factors applied in low-LET situations cannot be applied without modification or at least verification.

The concentrated dose distribution around the core of a proton or HCP track can cause a saturation of the local energy traps, which will affect the overall response of the detector.[2] The dosimeter response is a function of the radial dose distribution of the incoming particle, which will depend on its charge and energy as well as absorbing material.[5,20] The response will also depend on the readout technique and filters used.[22,23] The difference in dose deposition in this modality could also change aspects not studied previously, such as postirradiation fading, readout depletion, and field size factors. Thus, this study characterized basic clinical factors for a commercial OSLD and OSL reader using Al_2O_3:C in proton beams. In this way, the accuracy of dosimetry in these beams could be quantified and increased, allowing improved protocols to be developed and introduced to clinics using OSLDs.

II. METHODS AND MATERIALS

II.A. Dosimeters and Readout System

The OSL dosimeters used in this study were InLight nanoDots from Landauer, Inc. (Glenwood, IL). The sensitive material, Al_2O_3:C, is a 5 mm diameter disk, 0.2 mm thick. The disc has binding foils on top and bottom, such that in total it is approximately 0.3 mm thick. It is encased in a plastic light-tight casing, measuring $10 \times 10 \times 2$ mm^3, with a density of 1.03 g/cm^3. The disk can slide out of the casing, done so during reading or optical bleaching. Before any irradiation, the dosimeters were bleached for 24 h under fluorescent lights combined with a UV filter (FG-408UV, Bergen Industries, Las Vegas, NV).

To read the OSLDs, two microStar OSL readers, also from Landauer, were used. These readers incorporate the use of CW-OSL for short periods. The readers are specially modified for the Radiological Physics Center (RPC) to illuminate and read the dosimeter for 7 s, as opposed to the more common 1 s. The readers are identical in model, design, and filter pack. There is a stimulation power difference between them, i.e. different count/dose ratios, but Al_2O_3:C has been shown to be linear with stimulation power.[24] Difference in the response between readers is accounted for with control dosimeters (see Sec. II B). The stimulation photons are produced by light emitting diodes and filtered by an OG-515 high band-pass filter to eliminate photons in the same range as the dosimeter luminescence. Luminescent light is detected with a photomultiplier tube (PMT) with Schott BG-12 and Hoya B-370 band-pass filters in front to discriminate between stimulation and luminescence photons. The combined transmittance of the filters along with the emission spectra of Al_2O_3:C is shown in Fig. 1. Characterization of the reader with these types of filters is important considering the UV emission and its time dependence.

II.B. Absolute dose determination

Dose (D) to water using OSLDs was determined using a proposed formalism,[2] similar to that of the AAPM TG-51 and IAEA TRS-398 protocols[25,26]

$$D_{w,Q} = M_Q^{\mathrm{OSL}} * N_{D,w,Q_0}^{\mathrm{OSL}} * k_{Q,Q_0}^{\mathrm{OSL}}. \tag{1}$$

Here, $D_{w,Q}$ is the dose absorbed in water, w, irradiated with a beam of quality Q. M_Q^{OSL} is the reading of the dosimeter after the application of dosimeter-specific correction factors. $N_{D,w,Q_0}^{\mathrm{OSL}}$ is the calibration coefficient in terms of absorbed dose to water in a beam of known quality. k_{Q,Q_0}^{OSL} is the beam quality factor between the experimental beam and the beam of known quality.

The M_Q^{OSL} term consists of several correction factors related to the irradiation conditions and readout process and is expanded in Eq. (2).

$$M_Q^{\mathrm{OSL}} = R * k_F * k_L * k_D * k_S. \tag{2}$$

Here, R is the net number of counts of the PMT in the microStar OSL reader. The net number is taken as the raw value after irradiation minus the value read before irradiation, or the residual signal. Before reading any dosimeters, the microStar system has an internal procedure to eliminate the effects of dark current and background signal from the displayed readings. The remaining k terms, F, L, D, and S, are

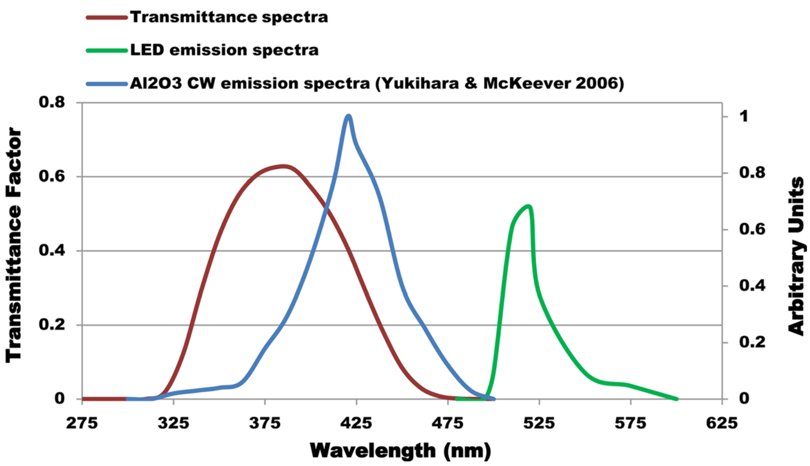

FIG. 1. Spectra related to the microStar reader. The LEDs and high band-pass filter peak transmission is near 530 nm, while the combined transmission of the two filters peak at 385 nm. Both spectra are plotted according to the left y-axis as a transmission factor. Total Al_2O_3:C emission spectra is reproduced from Yukihara and McKeever, 2006 (Ref. 16) and is plotted according to the right y-axis.

corrections for fading of signal postirradiation, absorbed dose linearity, readout depletion, and individual sensitivity, respectively. These terms do not represent a comprehensive list of all possible correction factors, but those thought to have the greatest effect on the results.

To determine a reader calibration factor for our experiments we followed the method adopted by the RPC, which is to irradiate a designated set of nanoDot dosimeters to 100 cGy dose to muscle under reference conditions in cobalt-60. This dose was then converted to dose to water, as this is more universal and to make the data comparison consistent with the IAEA TRS-398 absorbed dose protocol used at UT MD Anderson Cancer Center Proton Therapy Center in Houston (PTCH).[26,27] At the time of readout the calibration dosimeters are read out just before the experimental ones, giving cGy/count (the reader calibration coefficient).

The correction factors applied to the raw signal readings in Eqs. (1) and (2) are relevant for nonreference irradiations. All of these correction terms for proton irradiations were the subject of investigation in this study. Thus, these terms were not initially applied for those irradiations, but applied retrospectively where applicable.

The individual sensitivity term k_S is used to increase the accuracy of the results. In our experiments, the sensitivity of the dosimeters was derived by irradiating the group of OSLDs with no prior irradiation in cobalt-60 to 25 cGy. After irradiation, the dosimeters were read out in one session; the response of each dosimeter was then divided by the average response of all the dosimeters. Most factors ranged within 5% from 1.00. This method has been used previously in TL and OSL dosimetry.[28,29]

The calibration coefficient $N^{\mathrm{OSL}}_{D,w,Q_0}$ is session-dependent. Reference dosimeters are irradiated to a known dose to water in the reference beam and read out in the same session as the experimental. The PMT counts of the reference dosimeters give a cGy/count coefficient to be applied to the experimental dosimeters.

Finally, the k^{OSL}_{Q,Q_0} factor in select proton beams was also under investigation in this study; thus, it was not initially included in the dose calculation. The factor can be described with two components for this situation, the absorbed dose and relative luminescence efficiency[2,30]

$$k^{\mathrm{OSL}}_{Q,Q_0} = 1\Big/\eta^{w}_{Q,Q_o} = 1\Big/\eta^{Al2O3}_{Q,Q_0} * \left[D_Q/D_{Q_o}\right]^{w}_{Al2O3} = \frac{D_{w,Q}}{M^{\mathrm{OSL}}_{Q}}\Bigg/\frac{D_{w,Q_0}}{M^{\mathrm{OSL}}_{Q_o}}, \tag{3}$$

where η is the relative luminescence efficiency, which can be defined for a reference medium as in the second expression, or for the dosimeter material, as in the third expression. Efficiency defined in the reference medium is common for clinical situations and comparison against other materials. Efficiency defined in the dosimeter material is of interest for proton and HCP irradiations as a study of microscopic volume irradiation.[3,5] Comparison of relative luminescence efficiency between materials can be done by accounting for the difference in absorbed dose between materials. Response of the detector varies widely with the transmission filters and readout method, whether the initial signal observed in the first few seconds or the integrated signal of several minutes.[22,31] Since the k^{OSL}_{Q,Q_0} of the dosimeter in proton beams is a function of the readout method and transmittance filters it will be specific to each reader or reader model.

In summary, dosimeters are read after an experimental irradiation and the signal corrected for the context of the irradiation and readout procedure, then converted from a raw signal to absolute dose using a calibration term, and corrected for a beam quality difference if applicable.

II.C. Proton Beam Irradiation Conditions

The proton beam measurements were taken at the PTCH. All proton beam irradiations used the passively-scattered beam produced from the Hitachi ProBeat system. The beams were calibrated using the IAEA TRS-398 absorbed dose to water protocol.[26,27] Absorbed doses reported here refer to physical doses to the respective material; i.e., no biological effectiveness factor is applied. A water equivalent phantom was used that held the dosimeters in place and would allow centering along a beam. The phantom had dimensions of $20 \times 20 \times 0.5$ cm^3 ($L \times W \times D$). A volume was milled that allowed up to eight nanoDot dosimeters to fit flush with the top of the phantom. All irradiations at depth were done with

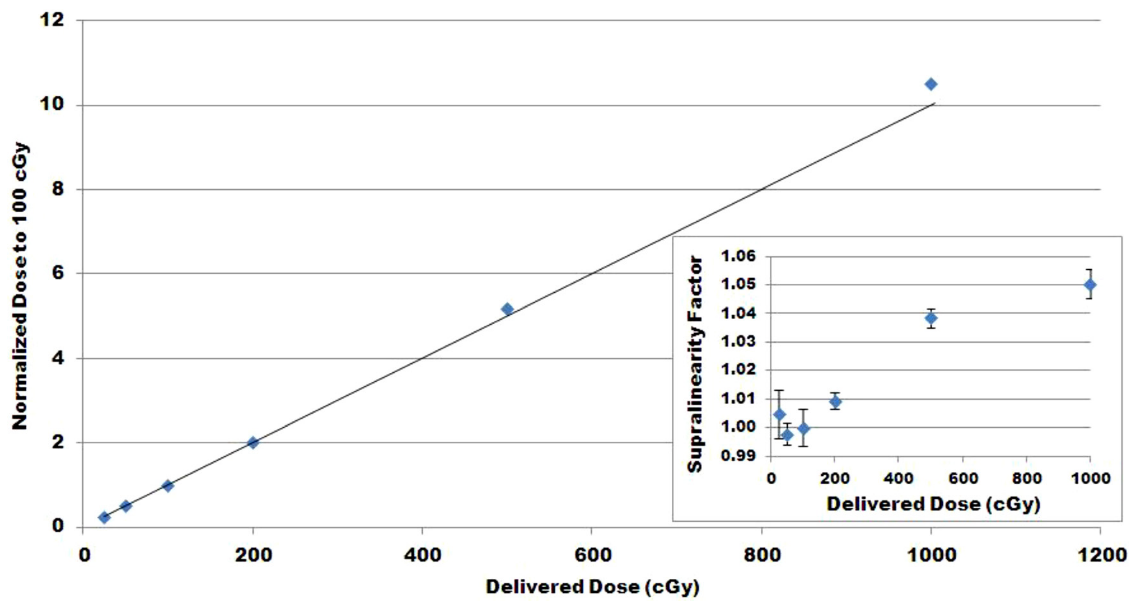

FIG. 2. Supralinearity of OSLD response to varying absorbed dose in a 250 MeV proton beam, centered in a 10 cm SOBP with a linear response line drawn for comparison. The inset shows the response normalized to a linear response, i.e., the supralinearity factor.

this phantom and with the dosimeters perpendicular to the incoming proton beam. The dosimeters were centered within the spread-out Bragg peak (SOBP) and along the central axis (CAX). Plastic Water (CIRS, Norfolk, VA) was used for beam buildup, as is used for other quality assurance measurements at the PTCH. For irradiations involving absolute dose, ion chamber measurements were taken immediately prior to the OSLD irradiations to determine the output of the beam following TRS-398 formalism using a Markus type 23343 parallel-plate ion chamber calibrated in water by an Accredited Dosimetry Calibration Laboratory.

The following OSLD characteristics were evaluated.

First, OSLD dose linearity was evaluated. Three dosimeters were irradiated at each of the following dose levels: 25, 50, 100, 200, 500, and 1000 cGy. The dosimeters were placed at the center of a 10 cm wide SOBP in the 250 MeV proton beam with a 10×10 cm^2 field size.

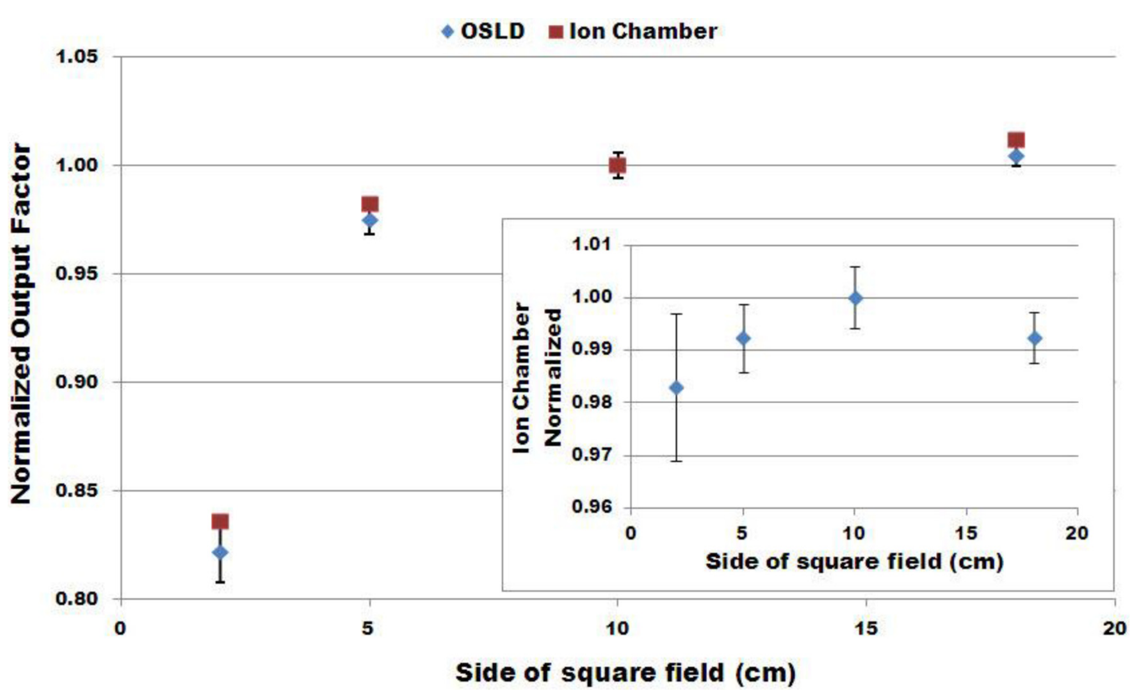

FIG. 3. OSLD response as a function of field size. Measurements were taken in a 250 MeV proton beam, centered in a 10 cm SOBP. Error bars represent one standard deviation of the measured data. The inset shows the output factor of various field sizes normalized to the PTCH commissioning data at the same location.

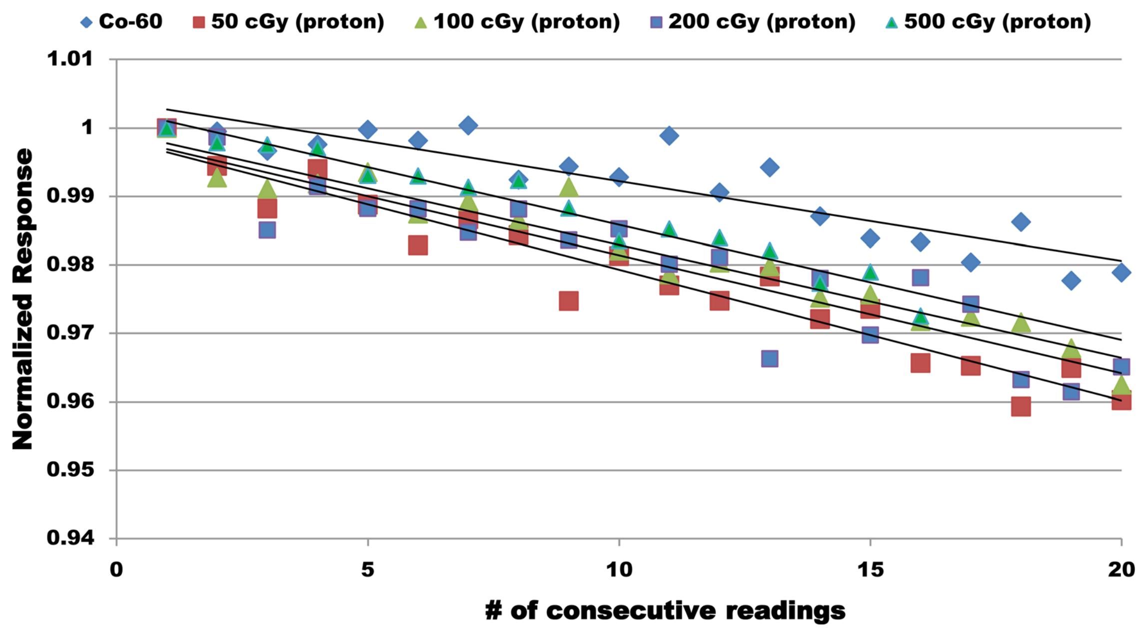

FIG. 4. Readout signal depletion for dosimeters irradiated to varying doses in a 250 MeV proton beam. Readings were shown as raw PMT counts as well as compared to typical dosimeter response when irradiated to 100 cGy in cobalt-60.

Second, the response of the OSLDs to different field sizes was examined. This consisted of irradiating four dosimeters in a 10 cm wide SOBP in the 250 MeV proton beam. Four field sizes were chosen: 2×2, 5×5, 10×10, and 18×18 cm^2. In each case, 100 monitor units were delivered and the response compared to output factors determined at the time of commissioning with an ion chamber under the same conditions.

Third, the depletion of the OSLD signal was evaluated. The OSLD signal is reduced with each reading; the amount of depletion has been found to be in the range of 0.05%–0.20% in photon beams, but has not been quantified for proton beams.[13,32] In this study, a dosimeter irradiated to doses ranging from 50 to 500 cGy were readout 20 consecutive times to determine both whether proton irradiation made a difference in depletion and if this was a function of dose level. For comparison, a dosimeter irradiated to 100 cGy in cobalt-60 was also read out.

Fourth, OSLD response to nominal energy and SOBP widths were examined using four dosimeters per irradiation at energies of 160, 200, and 250 MeV in SOBP widths of 4, 6, and 10 cm at each of the three energies. The dosimeters were located at isocenter and in the middle of the SOBP for all situations. All irradiations had a geometric field size of 10×10 cm^2 at isocenter. Ion chamber measurements were taken under the same conditions to determine dose to water at all investigated points.

Fifth, the extent of influence of cumulative dose on the residual signal of the dosimeters was examined. Considering the difference in local dose deposition associated with different LET, OSLDs irradiated in proton beams may present different characteristics of the residual signal than in photon beams. After a given experiment, the OSLDs were read out and then put under fluorescent lights with a UV filter for 24 h for optical bleaching, then read again to determine the

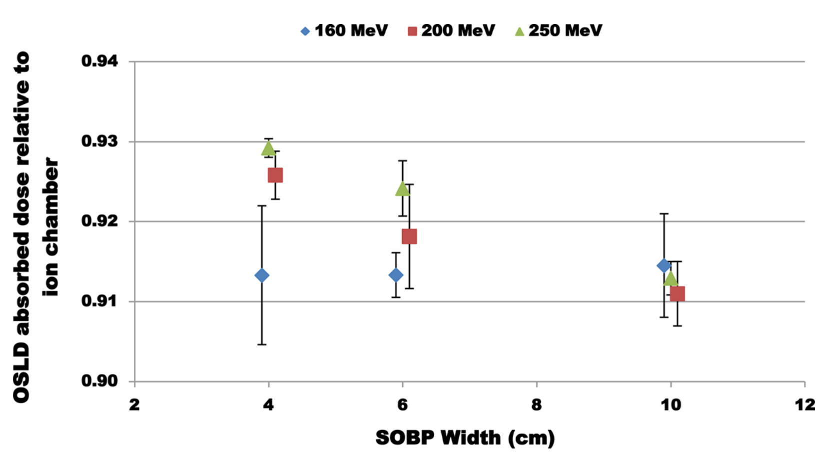

FIG. 5. OSLD response when irradiated with 160, 200, and 250 MeV proton beams. For each energy the response in a 4, 6, and 10 cm SOBP was examined. For every irradiation, the dosimeters were centered along the CAX and in the middle of the SOBP. Data are offset around each measurement point for visualization.

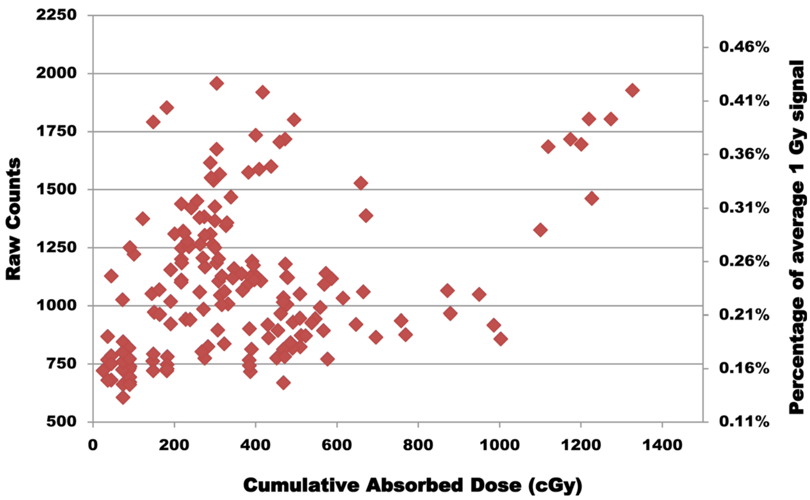

FIG. 6. Residual signal measured as a function of cumulative absorbed dose. The readings were taken after optically bleaching the dosimeters for 24 h under fluorescent light with a UV filter.

residual signal. Bleaching has been shown to effectively remove most signal but will not completely remove all trapped charges in the Al_2O_3:C.[13] Thus, a residual signal is left and accumulated on the dosimeter throughout its lifetime. The amount of residual signal was tracked on the dosimeters used in this experiment and plotted as a function of cumulative absorbed dose to determine the extent of this effect. All of the results were corrected for the sensitivity of the individual dosimeter as described earlier.

Sixth, fading of the OSLD signal postirradiation was examined. This was done by irradiating four dosimeters to three different dose levels: 25, 100, and 500 cGy. All irradiations were at the center of a 10 cm SOBP in the 250 MeV beam. The dosimeters were read at various times postirradiation to determine a fading curve. The fading correction factor accounts for the time between irradiation of the dosimeter to the time of readout. In photons, a transient signal has been observed that falls off sharply within 10–15 min of irradiation, stabilizing to within 2% in 24 h.[13,14]

III. RESULTS

Results of the dosimeter linearity are shown in Fig. 2. All results are normalized to the response at 100 cGy. There was an observed supralinearity of 1% at 200 cGy that increased to 5% over response at 1000 cGy. A linear response is shown for comparison, drawn by extrapolating the line from the origin to the 100 cGy response. The readings normalized to a linear response are shown in the inset of Fig. 2; this is essentially the inverse of k_L, the linearity factor.

Field size dependence measurements are shown in Fig. 3, normalized to the 10×10 cm^2 field size response. The inset of Fig. 3 shows the OSLD response relative to each field's ion chamber readings, again normalized to the 10×10 cm^2 response. The 2×2 cm^2 field shows the worst agreement with a difference of 1.8% compared to the ion chamber. The other field sizes show agreement within 1%.

Readout signal depletion was examined for various proton doses, as well as 100 cGy in cobalt-60 for comparison. Results for each dosimeter were normalized to the first reading of each, and are shown in Fig. 4. Depletion rate for the proton doses was constant across the dose range. On average, the signal depletion for all proton dose levels was 0.18% per reading. This is compared to the dosimeter irradiated in cobalt-60, which had a depletion value of 0.12% per reading. Regression analysis using an F-test statistic shows that none of the proton dose depletion rates were significantly different than that of cobalt-60. The standard error of the responses ranged from 0.26% for the 100 cGy proton results to 0.45% for that of 200 cGy; the cobalt-60 standard error was 0.34%.

Response of OSLDs as a function of SOBP width and nominal proton energies is shown in Fig. 5. The results of the OSLDs are shown as a ratio to the measured dose to water of the ion chamber. On average, the response across

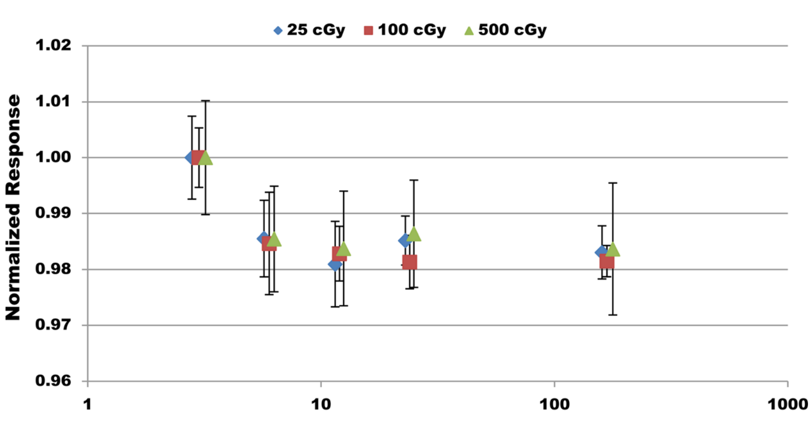

FIG. 7. OSLD response as a function of time postirradiation. Three levels of dose were examined; all response levels were normalized to the 3 h postirradiation time point. Data are slightly offset around each time point for visualization.

the SOBPs and energies is consistent, showing a result of 91.8 cGy with an experimental standard deviation of 0.7 cGy when irradiated to 100 cGy to water. The results of the 10 cm SOBP are the most stable, with very little response difference amongst the energies; the 4 cm SOBP irradiations show the most variation, with the average response being 1% greater than that with the 10 cm. The difference in response is largely a result of the differences observed in the 250 and 200 MeV beams that varied almost 2% across SOBPs. However, the standard deviations of the 160 MeV results were the largest.

OSLD residual signal data were collected throughout the study and are shown in Fig. 6. Because readers and bleaching protocols can vary with each clinic's needs, the values are not meant to be absolute but are indicative of the percent of residual signal that may be encountered with a similar protocol. Figure 6 shows the residual signal as a function of previously absorbed dose in protons in raw counts given by the OSL reader. As a means of comparison for different readers, the right axis shows the data as a percentage relative to the average reading of an OSLD irradiated to 100 cGy in cobalt-60. Response for cumulative doses between no previously absorbed dose and 1400 cGy varied between 0.12% and 0.43%. Overall, the residual signal is significantly linked to the previously absorbed dose ($p < 0.01$), although the slope of this effect is small, with a relatively large coefficient of variation of the data at 3.6%. Error introduced from not correcting the residual signal for accumulated dose is less than 0.3% for doses of 1400 cGy or less.

To examine fading, a number of dosimeters irradiated to different doses were read at different times postirradiation. Results of the fading dependence of the three different dose levels are shown in Fig. 7, with readings at 3, 6, 12, and 24 h postirradiation, normalized to the 3 h response. The response quickly dropped approximately 1.5% between the 3 and 6 h readings, but stayed constant after that. Furthermore, the fading rate was found not to change with dose.

IV. DISCUSSION

Al_2O_3:C OSLDs have been shown to have response that is supralinear with dose in photon beams, and the same effect appears present in protons. Figure 2 demonstrates linear response of the nanoDot dosimeter from 25 to 100 cGy within experimental uncertainty. Normalized to the 100 cGy response, a 1% supralinearity effect is seen at 200 cGy and this increases to approximately 4% and 5% at 500 and 1000 cGy, respectively. Other published data for photon beams have shown a supralinear response starting at 200,[14] 300,[13] 400,[17,33] and 500 cGy,[9] although the readers and readout methods varied in some cases. Data from other institutions irradiating OSLDs in clinical proton beams in similar conditions appeared to show supralinearity at near the same dose level as this study and were observed to have slightly less dose supralinearity than MV photon results.[14] Thus, the amount of supralinearity with dose of OSLDs in proton beams appears to be similar if not slightly less than that in photon beams. Both the magnitude and start of observed supralinearity will likely depend on the readout method employed.

Field size dependence of OSLDs in proton beams was found to be small. Measurements of this dependence matched within 2% of the ion chamber measured output factors (Fig. 3). The OSLD reading for the 2×2 cm^2 field size had the worst agreement of all field sizes. However, with such a small field size, small geometrical setup errors of either the ion chamber or OSLDs could affect the value. Additionally, while for field sizes above 5×5 cm^2 the flatness of the center 80% field width is always <3%, for the 2×2 cm^2 field the dose profile starts to approximate a Gaussian distribution.[27] The Al_2O_3:C disk within the nanoDot dosimeter is slightly offset from center. Profile data of the 2×2 cm^2 field analyzed from the PTCH commissioning shows that the difference in absorbed dose if the disk was offset by 1 mm from the CAX is approximately 1.5%, which would account for most of the difference observed. Ultimately though, as the OSLD output factor values agreed with ion chamber measured values within 2%, this dosimeter was deemed suitable for irradiations at these field sizes. Similar conclusions were drawn in previous studies for using OSLDs to measure output factors in MV photon beams.[9,17]

Using the microStar reader, OSLDs can be read several times to reduce the uncertainty of the readings. Readout depletion (Fig. 4) showed that overall, protons have a slightly sharper depletion rate than that of 100 cGy in cobalt-60 and there was no correlation of depletion rate with dose. Previous work showed that the drop off of luminescence of Al_2O_3:C when studying the entire decay curve was sharper with increasing LET, at least for heavy charged particles.[22] It was shown elsewhere that even for beta irradiation of different doses, there was signal dependence as a function of time using CW-OSL.[34] However, investigation with a microStar reader at dose levels of 100 and 400 cGy in cobalt-60 showed no dose depletion rate dependence.[35] Thus, for photon and proton OSLD measurements using a microStar reader, there seems to be no effect on the depletion according to the dose absorbed by the dosimeter.

Typical clinical practice involves reading the dosimeter at most only a few times. This will minimize the effect of variation in depletion. Considering a protocol using a depletion correction factor based only on photon data, errors introduced to proton measurements from depletion differences are on average within 0.2% for up to three readings per dosimeter and within 0.3% for five readings with an absorbed dose up to 500 cGy.

Figure 5 shows the results of irradiating OSLDs in SOBPs of different widths and with different nominal energies. The results show that the response across the SOBPs and nominal energies is fairly consistent, giving an average of 91.8 cGy and standard deviation of 0.7 cGy when irradiated to 100 cGy in water. The results with a 10 cm wide SOBP are the most stable, with very little response difference amongst the energies; the 4 cm SOBP irradiations show the most variation, with the average response being 1% greater than that with the 10 cm. Similar studies have shown relatively little variation in response to varying therapeutic proton energies within the SOBP.[6,14] It should be noted that this is all that is needed to answer the question of the k_{Q,Q_0}^{OSL} value: dose to water in the reference beam and proton beam are known; the observed signal is then entered into Eq. (3). In reality, two possible effects have been combined to produce this result.

If the signal per unit absorbed dose to the detector changes between the reference beam and experimental beam it is said to be intrinsically energy dependent. This information can be determined by examining the relative luminescence efficiency of Al_2O_3:C. The link between the relative luminescence efficiency of Al_2O_3:C and that of water is the absorbed dose difference between them. Ideally, the intrinsic energy dependence is unity and only the absorbed dose is different. To determine the absorbed dose to the dosimeter, information about the energy and LET of the proton beam as well as cobalt-60 must be determined.

Within SOBPs, the proton energies will range from nearly zero for particles at the end of their path length to those with the full residual energy to reach the end of the SOBP. Monte Carlo calculations based on the PTCH beam line of dose-averaged LET in proton beams with energies similar to those studied in this work have been done.[21] For a 140 MeV beam at the center of a 4 cm SOBP the $\overline{\mathrm{LET}}$ in water was calculated as 2.22 keV/μm, and 1.84 keV/μm at the center of a 10 cm SOBP in a 250 MeV beam. This represents the range of $\overline{\mathrm{LET}}$ observed at the central portion of the SOBP for these beams. We expect then that the $\overline{\mathrm{LET}}$ range of our experiments will be similar to this relatively narrow range. Based on these $\overline{\mathrm{LET}}$ values, we can approximate the effective proton energy experienced at the dosimeter location. Data from the NIST PSTAR database show that monoenergetic protons in the range of 25–30 MeV (2.18–1.88 keV/μm, respectively) have LET similar to the $\overline{\mathrm{LET}}$ experienced in the central SOBP region.[36] Carbon impurities are in the range of 100–5000 ppm and thus contribute negligibly to the stopping power ratio.[8] Note that these data are only to determine absorbed dose to the Al_2O_3:C, not to invoke an LET-to-luminescence relationship, as the $\overline{\mathrm{LET}}$ within the SOBP is by definition a combination of LETs. Within this 25–30 MeV energy range the absorbed dose to Al_2O_3:C and to water can be determined. The ratio of absorbed dose in protons to water to that of Al_2O_3:C is 1.263 at 25 MeV and 1.258 at 30 MeV. For all intents and purposes, these values are close enough that the average can be used without adding undue error. A similar determination of the absorbed dose ratio in cobalt-60 was done also using NIST data.[37] For cobalt-60, the ratio of absorbed dose to water to that of Al_2O_3:C is 1.131. This data can be combined with the observed SOBP luminescence data at the two LET extremes: 160 MeV, 4 cm SOBP and 250 MeV, 10 cm SOBP. These data actually have the same response despite the average differences between SOBP widths. The relative luminescence efficiency of Al_2O_3:C was found to be 1.017, indicating that the dosimeter may not be perfectly ideal in protons, having a small intrinsic energy dependence.

Results from other investigations show variation in the relative luminescence efficiency as defined in water, depending largely on the transmittance filters used. Collective results of irradiations of Landauer's Luxel dosimeters by He beams with LET of ≈2.2 keV/μm showed an average luminescence efficiency of 1.024 and 0.906 for filters collecting UV and F-center and F-center only emissions, respectively.[3,5] Other studies found an efficiency of approximately 1.06 in proton beams compared to 6 MV photons, and 1.04 in He beams of 2.2 keV/μm, but the filter spectra was not specified in either study.[14,18] Results from our study are closer to that of the F-center only filter used in the collective results; however, a comparison of filters show that our filter collects far more F-center emissions.[23] Thus, in proton and HCP irradiations where the UV emissions start to influence the response, how much of the F-center emission is collected affects the response. Our experimental observations determined an average k_{Q,Q_0}^{OSL} of 1.089 with an experimental standard deviation of 0.007, for the energies and SOBPs studied here, such that all the results were then within 1.2% of the absorbed dose to water as measured by the ion chamber.

From Fig. 6, the residual signal as a function of accumulated dose from 0 to 1400 cGy ranged, on average, from 0.12 to 0.43% of a typical response of 100 cGy in cobalt-60. Although the results showed significant dependence on the accumulated dose, the clinical significance is likely negligible as the change was less than 1%. This indicates that one residual (or background) signal correction factor is sufficient for cumulative doses below 10 Gy. A change in the residual signal could indicate a change in the sensitivity of the dosimeter, although little to no observable difference has been found for doses less than 10 Gy in photons.[13,35] However, because of the overall low signal of the residual counts, no inference is made here regarding the change in sensitivity. If the sensitivity was found to change with dose, this would have to be accounted for either by making the k_S term a function of dose or introducing another correction factor.

The fading results of Fig. 7 indicate that the nanoDot dosimeter loses signal during the first hours after irradiation. When normalized to the 3 h reading, the signal stabilized to within 0.5% for times after 6 h postirradiation. These data agree within experimental uncertainty with similar data presented for photons using another Landauer system.[38] Measurements done with the microStar reader irradiated by a ^{90}Sr/^{90}Y source over a longer period of time show a smooth decay of signal of 2% between 10 min and 24 h after irradiation, and around 3% between 24 h and 3 months postirradiation.[6] Due to physical and time limitations we were not able to properly analyze the dosimeters before 3 h. This effect should be investigated before using the dosimeters so soon after irradiation. The small variation in results from differing dose levels shows that fading was not affected by the dose received by the dosimeter up to 500 cGy and up to 1 week postirradiation for the 10 cm SOBP and 250 MeV beam.

Several caveats must be understood from this study. We used a microStar reader to gather data, which is less accurate than other methods presented but more likely to be used in clinical practice given its commercially-packaged form.[39] The characteristics of the filters in the microStar will change from reader to reader due to the production method of filters and the reader. This can change the transmittance characteristics and thus the signal, especially the F-center to UV signal ratio. Thus, a calibration of relevant parameters is strongly encouraged for each reader used. However, we believe some of the results shown will be consistent for similar irradiation and readout protocols. Specifically, the field size and residual signal aspects should be similar as they rely more on irradiation, dosimeter, and bleaching conditions. Linearity will be affected based on the filters used and whether they transmit the UV-center emission spectrum or just the F-center as well as how much of the F-center emissions get collected. The readout depletion and overall energy dependence will change depending on the stimulation power and filters used in the individual OSL reader and can be markedly different for readers even of the same manufacturer and model; however, we could recommend initially using the same readout depletion correction for that of photons if reading the dosimeter 5 or fewer times. If irradiating in proton or HCP beams the relative luminescence efficiency should be determined before clinical use. While the luminescence efficiency did not change much between energies and SOBPs, this will not hold true for HCP irradiations where the luminescence efficiency can change dramatically. If the filters allow UV transmission, the fading curve could give different results given the characteristics of that emission band. Signal fading at less than 3 h was not examined. Until further examination shows otherwise, we recommend waiting 6 h or more before reading dosimeters irradiated in protons to reduce the effects of fading if accuracy greater than 1.5% is required.

While a number of clinical factors were examined, several still remain unanswered or were not addressed in this study. Neither the effect of sensitivity on the dosimeter with accumulated dose, the beam perturbation caused by the OSLD, nor the effect of dose rate was examined. It is worth noting that angular dependence of the same OSLD model in a clinical proton beam was studied earlier with no observable effect on the response.[40]

V. CONCLUSIONS

This study aimed at examining many of the factors that influence the response of Al_2O_3:C OSLDs irradiated in therapeutic proton beams. When irradiated in the middle of an SOBP, the dosimeters show linear dose response up to 100 cGy after which a supralinearity of 1% at 200 cGy and 5% at 1000 cGy was observed. No clinically significant field size effect was found. Signal depletion was slightly stronger for protons than for photons but was not significant nor should it affect protocols that only use a handful of consecutive readings. A proton beam quality dependence of 8.9% was found, consisting of absorbed dose difference and luminescence efficiency of the dosimeter material relative to the reference beam. The OSLD response varied slightly with SOBP width, agreeing within 0.5% for 10 cm and within 2% for 4 cm. The residual signal left after optical bleaching was found to be between 0.1% and 0.4% of a typical 100 cGy response for doses below 1400 cGy, and although the signal significantly increased with accumulated dose, this was a minor effect. Signal postirradiation fell approximately 1.5% between 3 and 6 h, after which no change was observed up to 1 week. Neither was the fading found to be dose dependent up to 500 cGy when irradiated in the SOBP. Al_2O_3:C OSLDs, specifically the nanoDot, show characteristics in clinical proton beams that can be described and corrected for where necessary and can therefore be used for proton therapy dose verification in common situations.

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