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Technical Note

Study of dosimetric characteristics of a commercial optically stimulated luminescence system

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Abstract

Background: Optically stimulated luminescence dosimeters (OSLDs) have a number of advantages in radiation dosimetry making them an excellent dosimeter for in vivo dosimetry. The study aimed to study the dosimetric characteristics of a commercial optically stimulated luminescence (OSL) system by Landauer Inc., before using it for routine clinical practice for in vivo dosimetry in radiotherapy. Further, this study also aimed to investigate the cause of variability found in the literature in a few dosimetric parameters of carbon-doped aluminium oxide (Al_2O_3:C).

Materials and methods: The commercial OSLD system uses Al_2O_3:C nanoDot^{TM} as an active radiation detector and InLight^{TM} microStar® as a readout assembly. Inter-detector response, energy, dose rate, field size and depth dependency of the detector response were evaluated for all available clinical range of photon beam energies in radiotherapy.

Results: Inter-detector variation in OSLD response was found within 3·44%. After single light exposure for the OSL readout, detector reading decreased by 0·29% per reading. The dose linearity was investigated between dose range 50–400 cGy. The dose response curve was found to be linear until 250 cGy, after this dose, the dose response curve was found to be supra-linear in nature. OSLD response was found to be energy independent for Co^{60} to 10 MV photon energies.

Conclusions: The cause of variability found in the literature for some dosimetric characteristics of Al_2O_3:C is due to the difference in general geometry, construction of dosimeter, geometric condition of irradiation, phantom material and geometry, beam energy. In addition, the irradiation history of detector used and difference in readout methodologies had varying degree of uncertainties in measurements. However, the large surface area of the detector placed in the phantom with sufficient build-up and backscatter irradiated perpendicularly to incident radiation in Co^{60} beam is a good method of choice for the calibration of a dosimeter. Understanding the OSLD response with all dosimetric parameters may help us in estimation of accurate dose delivered to patient during radiotherapy treatment.

Keywords: aluminium oxide; ionisation chamber; optically stimulated luminescence dosimeter; radiation dosimetry

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INTRODUCTION

Cancer treatment is a complex process which includes all aspects of diagnosis, treatment planning and treatment delivery. Accurate treatment delivery can be verified using in vivo measurements.[1] There are some dosimeters available commercially like thermoluminescent dosimeters (TLD), PN junction-type diodes, or metal–oxide–semiconductor field-effect transistor (MOSFET) detectors and optically stimulated luminescence dosimeter (OSLD) for this purpose.

For a number of decades, TLD have been the passive detector of choice, able to perform in vivo dosimetry as well as remote quality assurance checks of radiation therapy delivery systems. Because of its long history, TLDs have been well characterised, making them reliable with relatively small margins of error. Their major disadvantage, which is the time required for readout, can be considerably decreased by a good choice of the TLD reader and a good methodology.[2,3] Semiconductor PN junction-type diodes have some advantages over TLD for in vivo measurements such as instant readout and reproducibility of signal. The physical and dosimetric properties of modern diodes have been studied and found satisfactory for use in the clinic.[4–7] However, they require handling of long cabling from treatment area to measurement station and they have cumbersome calibration with many corrections. MOSFET dosimeters provide for fast readout of the dose with permanent storage of information in the MOSFET.[8,9] Their small size makes them very useful for measurements in high-dose gradient regions, typically, for example, in brachytherapy. Their main disadvantage is their limited lifetime.[10]

In recent years, new materials and methods have been proposed to improve passive radiation dosimetry in clinical applications and in radiation protection. One of these includes optically stimulated luminescence (OSL) to overcome a number of drawbacks inherent to the TLD and MOSFET. The OSL technique became a successful tool in personal and environmental radiation dosimetry, geological and archaeological dating, retrospective/accident dosimetry, and medical applications of radiation in diagnostic imaging as well as in radiotherapy in last two decades. The use of OSL for radiation dosimetry was first suggested in the 1950s and 1960s.[12–14]. However, the use of OSL for various dosimetric applications started in mid-1990s. The main obstacle at that time was the non availability of sensitive OSL phosphors.[11–15] Pulsed optically stimulated luminescence (POSL) technique for radiation dosimetry using anion-deficient carbon-doped aluminium oxide (Al_2O_3:C) has been developed[19] and is commercially implemented in LUXEL^{TM} technology (Landauer Inc., Glenwood, IL, USA). In 1998, the first commercial OSL dosimetry service based on Al_2O_3:C and the POSL technique (Luxel^{TM}) was introduced by Landauer Inc.[16–22] There are several other OSL materials developed in recent decades but Al_2O_3:C is the only OSL material and is the most popular commercially available for various dosimetric applications in radiotherapy.

The dosimeter is generally characterised by some of the dosimetric quantities such as precision, accuracy, dose linearity, energy, dose rate and angular dependence. However, the response of dosimeter depends upon its physical form or dimension, construction and incidence angle of radiation beam. Al_2O_3:C-based OSL systems appear to possess the properties near to an ideal dosimeter, such as high sensitivity, high spatial resolution, availability in different shapes and sizes, no or few dependencies on beam parameters, capability to measure absorbed dose in real time for both photon and electron beams, and temperature independence for the ease of calibration and use.[23] However, Al_2O_3:C-based OSL systems possess some disadvantages in that non-tissue equivalent material having an effective atomic number[24] (Z_{eff}) of 11·28, requires a light protective environment to read and only single vendor of Al_2O_3:C material is commercially available for OSL-based dosimetry.

OSL utilises materials and electronic processes similar to thermoluminescence but interrogation of the detector is performed by light (ultraviolet, visible or infrared) instead of heat and emits a light signal; the wavelength of the emitted light is a characteristic of OSL material and the intensity of emitted light signal is proportional to the irradiation dose.[25–29] High sensitivity, precise delivery of light, fast readout times, simpler readers and easier automation are the main advantages of OSL in comparison with TLD. OSL allows for re-reads of the detector multiple times while maintaining the precision, and yet it still can be used as an erasable measurement technique.

A number of papers have been published in past few years that describe the use of OSLDs for clinical measurements in radiation oncology.[23,30–41] Jursinic[31] published a very comprehensive study discussing the dosimetric properties of a commercially available OSL detector encapsulated in a light-tight plastic holder that is readout with a simple and efficient system Landauer Inc. Jursinic found their response to be independent of energy for megavoltage photons 6 and 15 MV. The dose sensitivity coefficient of variation of OSLDs from a batch of detectors was found to be 0·9%. The dose response was linear with absorbed dose over a test range of 1–300 cGy; >300 cGy a small supra-linear behaviour occurs. He demonstrated that the OSL signal stabilised after 8-minute post-irradiation.

Viamonte et al.[23] also investigated some dosimetric characteristics of the same OSLD and readout system Landauer Inc. They found the detector response to repeated exposures to be within 2·5%, no energy dependence for 6, 10 and 18 MV photons, but ~4% lower response for Co^{60} γ rays. They demonstrated that the dose response was linear with a dose range 50–400 cGy. Their measurements with OSL detector and the ionisation chamber showed a very good agreement of <1% for relative output factor measurement. Yukihara and McKeever[22] published a comprehensive review article on the fundamental and practical properties of OSLD in medicine. Pradhan et al.[42] also published a comprehensive review article on the fundamentals, materials available, practical dosimetric properties and radiation dosimetry by OSLDs.

OSL detectors are relatively new to medical dosimetry compared with TLDs. However, much characterisation has been done in recent few years on OSLD nanoDot™ (Landauer Inc., Glenwood, IL, USA) but it lacks the comprehensive characterisation of many mainstream TLD materials. The purpose of this work is to characterise commercially available OSL detector for use in certain clinical and dosimetric situations relevant to radiation dosimetry in radiotherapy. In the present work, a series of experiments were carried out using this OSL system to obtain the dosimetric quantities and the results obtained were compared with similar measurements carried out with an ionisation chamber.

MATERIALS AND METHODS

OSLD and readout system

The OSL system used in our study was commercially available OSL detection system, manufactured by Landauer Inc. The system consists of an OSL nanoDot™ dosimeter[43] as detector and InLight™ microStar® (Landauer Inc., Glenwood, IL, USA) reader[44] as readout assembly. The detectors consist of Al_2O_3:C (Landauer Crystal Growth Facility, Stillwater, OK, USA) encapsulated in a light-tight plastic holder with dimensions measuring $10 \times 10 \times 2\,mm^3$ as shown in Figure 1a. During initial manual preparation to readout OSLDs, these plastic holders are placed into larger holder and then placed into the reader sliding drawer, shown open in Figure 1b. During readout, inside the reader, the plastic case over the detector is slid open and the detector's active part Al_2O_3:C chip is optically simulated using light-emitting diodes (LEDs). The microStar reader uses an array of 36 LEDs for stimulation of nanoDot™. One of two exposure levels is automatically selected depending on the dose level determined by the software. In 'low-dose' mode, all 36 are illuminated, whereas in 'high-dose' mode only six LEDs are illuminated.[40] In present study, all nanoDots™ were exposed to levels deemed 'high' by the software and were therefore readout in high-dose mode with a stimulation time of about 1 second. The microStar OSL reader operates in the continuous wave optically stimulated luminescence (CW-OSL) mode. The CW-OSL mode of stimulation considered to be the most sensitive mode of OSL, because the intensity of the emitted signal is the highest during a stimulation and there is no restriction on the duration of a stimulation or recording for an optimisation of the signal-to-noise ratio as compared with other stimulation modes, that is, linearly modulated OSL, POSL.[45] When green light from LED is incident on Al_2O_3:C chip, the trapped electrons get exited (the number of

(a)

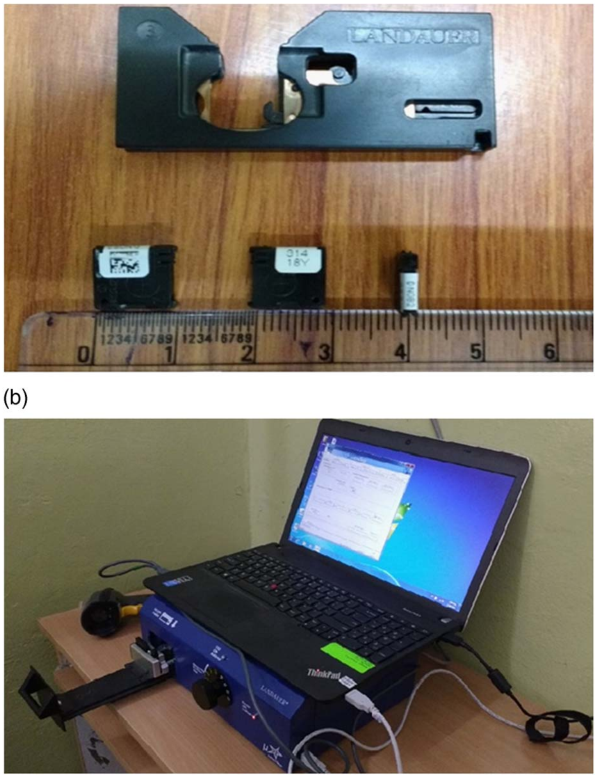

Figure 1. (a) Photograph of InLight nanoDot™ optically stimulated luminescence dosimeter form Landauer Inc., with large holder (above) to place the dosimeter for readout. (b) Photograph of the Inlight microStar® reader system from Landauer Inc.
Notes: (a) Three nanoDots placed in different orientation showing (left) back, (middle) front and (right) side on profile of closed dosimeter. (b) Closed dosimeter dimensions are $10 \times 10 \times 2\,mm^3$. The complete reader system consists of a barcode scanner to facilitate record keeping and data entry, a loader to load dosimeter in reader for readout and a laptop to show readout result and record keeping of data.

trapped electrons proportional to the exposed dose), and emits blue luminescent light as a result of de-excitation of electrons. The photomultiplier tube that is installed in reader counts the number of blue luminescent light photons, which is proportional to the exposed dose in the reader. All the readouts were performed after a time delay of 10 minutes following irradiation of dosimeters to avoid transient signal in all of the OSLDs as was suggested by Jursinic.[31]

Therapy units and phantom

In the present study, Bhabhatron-II TAW (Panacea Medical Technologies Pvt. Ltd, Bengaluru, India) unit cobalt-60 telecobalt radiation was used for most of the measurements carried out. Varian Clinac-iX linear accelerator (Varian Medical Systems, Palo Alto, CA, USA) was also used for 6 and 10 MV energy X-ray photon beams to study dose rate and energy dependence. All irradiations performed in a $30 \times 30\,cm^2$ custom made acrylic solid slab phantom, which is placed on the couch of the machine. The solid water phantom slab having a slot for accommodating parallel plate chamber is filled with a little amount of tissue equivalent material, that is, paraffin at level of phantom surface and a slot was fabricated and filled with paraffin in a size comparable with size of nanoDot™, for accommodating the OSL nanoDot™ in the phantom, placing the detectors in a solid slab phantom without incurring air gaps. The use of this method avoided the need for the slabs of solid water (Standard Imaging, Madison, WI, USA) to be machined to fit the various detectors. The total thickness of the slab phantom was kept at 10 cm beyond the point of measurement, which was sufficient to provide adequate backscattering for the photon beam used in this study.

Experimental set up condition

Unless otherwise mentioned, most of irradiations were carried out using an source-to-surface distance (SSD) set up (80 cm for Co^{60} and 100 cm for linear accelerator beams) with a $10 \times 10\,cm^2$ field size and the irradiation platform is perpendicular to the incident radiation beam. Before each irradiation session, the SSD was carefully monitored and verified with the optical distance indicator. The dosimeter in the phantom was then positioned at the phantom surface and aligned with the central axis of the teletherapy machine. OSL detector is placed at either 0·5 or 5 cm depth in phantom with an appropriate build-up of acrylic solid slab phantom as suggested by Werner et al.[46] in a flat homogeneous phantom for Co^{60} and high-energy beams. In our experiments, a high level of precision for high-energy beam irradiation was achieved by placing nanoDot™ with appropriate build-up thickness and with full scatter conditions for full phantom geometry. The dosimeters are then irradiated to a known dose. After each irradiation, the dosimeter group was read in one session to reduce statistical uncertainty associated with the reader. Each dosimeter was read three times consecutively to reduce the measurement uncertainty. The apparent reading of each individual dosimeter was taken as the average of the three readings performed on it consecutively. When absolute doses were of interest, absolute dose was measured using a TM30013 Farmer type [Physikalisch Technische Werkstätten (PTW), Freiberg, Germany] cylindrical ionisation chamber in acrylic solid slab phantom.

Response studies for Al_2O_3:C

OSL detectors were exposed to identical doses of 50 cGy irradiated in Co^{60} telegamma beam. Eight detectors were used to evaluate the inter-detector response variation. Absolute dose was also measured using a TM30013 Farmer type PTW ionisation chamber in acrylic solid water slab phantom. Average OSLD response was compared with practically measurable 'gold standard' ionisation chamber measurements.

One of the important characteristics of a dosimetric system is energy response due to the fact that the energy absorbed by the material (detector) is usually proportional to the dose absorbed. Dosimeter system may exhibit energy dependence due to higher atomic number (increase in detector response at low-energy beams), different energy beams have different scattering properties which can result in slight change in detector response indirectly during measurement. Detectors calibrated in only a particular radiation field generally Co^{60} as reference beam, any deviation from calibrated reference beam may result in significant change in detector response. It is difficult to develop an ideal dosimeter that is tissue equivalent and energy dependent over the entire clinical energy range used in radiotherapy. Thus, the suitability of this commercial OSL system for clinical use in radiotherapy may also be dependent on the variation in the response with beam energy, needs to be investigated. Further, there is variability found in the literature for energy dependency of OSLD which needs to be analysed. In this present study, energy dependence of OSL was investigated for Co^{60}, 6 and 10 MV beams, delivering an identical dose of 50 cGy each at dose maximum depth (d_{max}) in phantom with a 10×10 cm^2 field size at source-to-axis distance (SAD) set up for all three energies. Further, the dose rate dependence of OSL detector is evaluated for dose rates of 200, 400, 600 MU/minute in linear accelerator by delivering a dose of 50 cGy at SAD at d_{max} in with 6 MV energy X-ray photon beam. Energy and dose rate dependency measurements aimed to provide a corrected dose value, after calculating a calibration factor by comparing dose received in a particular radiation field to that of Co^{60} for reference.

OSL response with given dose is investigated for doses ranging from 50 to 400 cGy. Irradiations of dosimeters were carried out to known doses in reference Co^{60} beam at 0·5-cm depth in phantom. The response of a dosimeter should be independent of the field size and depth. However, it is recommended to check the dosimeter response with these dependences before using it for dosimetry. To analyse OSL response with the field size and depth, nanoDots^{TM} were irradiated with fields of 4×4, 10×10 and 30×30 cm^2 and depths of 1·5, 5, 10 and 14 cm in Co^{60} beam. Before a measurement session, for each field size/depth combination, the irradiation time was determined such that the delivered dose would be as close as possible to 25 cGy, using the established baseline phantom percentage depth dose (PDD) curves traceable to British Journal of Radiology Supplement 25 data.[47] OSL nanoDots^{TM} were then irradiated with these calculated irradiation time.

Angular response is an important dosimetric parameter of the dosimeter and must be analysed before use in in vivo dosimetry and patient quality assurances for multi-field treatments in radiotherapy. The response of the dosimeter varies with radiation incidence angle depending on their various physical parameters such as construction, physical size, shape and energy of incident radiation. In our study, because the construction of OSLDs from Landauer consists of a thin disk of Al_2O_3:C-coated material encased in plastic with a small air gap, the irregular geometry mandates that angular dependence is an important characteristic to determine. The magnitude of incidence beam angularity effect was evaluated by measuring dose on central axis of the beam at the depth of maximum dose with along with investigating build-up thickness effect in rectangular geometry phantom. The dosimeter angular response of OSLD was studied in solid slab phantom at various gantry angles ranging from 90 to 270° relative to the axis of gantry rotation, at an interval of 30°, with two different experimental set up having a 0·5 cm build-up of solid slab phantom placed over detector and without build-up, respectively. The responses of OSLD at any angle normalised to response of OSLD at 0° gantry angle, where the radiation incidence was perpendicular to detector.

During multiple readouts of OSL for dose were reanalysed the subsequent readout results in a decrease in signal. In present study, Landauer InLight microStar® reader OSL system was used in standard operating mode. After single irradiation to a known dose subsequent readouts were performed of nanoDots™. Long-term fading of the signal is an important parameter of the dosimeter for the use in dosimetric audits, permanent dose record and for the utility of dosimeter in periodic dose assessment applications. For this purpose, OSLDs were irradiated to a dose of 50 cGy. The first reading of the OSL nanoDots™ was performed, taken as reference and thereafter, the readouts were performed weekly and monthly.

A set of radiation dosimetry measurements were carried out using OSL detectors and the results were compared with ionisation chamber measurements. These measurements were aimed to check the accuracy of OSLDs for routine relative dosimetry in radiotherapy. For this purpose, PDD curves were measured with OSLDs at a depth ranging from 0·0 to 14 cm in solid slab phantom for Co^{60} beam. The relative output factors were also measured with six different field sizes, ranging from $4 \times 4\ cm^2$ to $22 \times 22\ cm^2$. Ionisation chamber measurements were performed with TM04102 PTW Markus parallel plate ionisation chamber (PTW, Freiburg, Germany) for PPD measurement and TM30013 PTW Farmer type cylindrical ionisation chamber is used for output factor measurements, under the same experimental condition.

RESULTS AND DISCUSSION

Inter-detector response

This study investigated dosimetric characteristics of OSLDs in megavolt energy photon beam and some relative dosimetric quantities were also measured. The inter-detector variation of OSL detector was found to be within 3·44% SD, with coefficient of variation of 0·035. Mrcela et al.[37] investigated inter-detector variation for irradiating at various identical doses, that is, 1·6% for 50 cGy and 1·3% for 100 cGy. However, the reproducibility of OSL InLight™ Dot irradiated eight times × 100 cGy with accumulating dose from each irradiation was found to be 3·5% and it was observed only 1% for OSL Dot optically bleached (illuminated to light to remove some of electron trap for optical resetting of dosimeter) before each irradiation. Another author Jursinic showed for six InLight™/OSL Dot dosimeters subjected to identical dose of 100 cGy had a coefficient of variation of 0·93%. In addition to that, an OSLD was repeatedly exposed, read and then optically annealed to six times provided, an analysis of six data points and had a coefficient of variation of 0·63%.[31] However, it was later shown for 17 individual nanoDots™ that were new and never irradiated, that they had unique sensitivity (coefficient of variation 5·1%) to low dose and unique supra-linearity (coefficient of variation 28%). In addition, these characteristics were shown to change with accumulated dose.[35] Apart from high accumulated doses, which lead to a drop in OSL sensitivity, the size of the dosimeter, reduced in case of nanoDots™ than in Dot dosimeters, was considered as a contributing factor in the large deviation in coefficient of variation in both published results. It has been reported by Jursinic, for a sample of 78 new nanoDots™, the range of their relative intrinsic sensitivity was found within 0·92–1·09 due to the inhomogeneous composition of the OSLD disc.[41] Viamonte et al. showed 4·2% (1 SD) inter-detector variation of a batch of 165 OSL Dot detectors exposed to 50 cGy irradiation suggesting good stability of the system and implied that detectors from a given batch might be used with a single calibration factor depending on the level of precision required. Our results are consistent with the findings of Mrcela et al.[37] and Schembri and Heijmen.[48] Schembri and Heijmen who investigated inter-film variations in 228 OSL films for a fixed dose of 200 cGy in six measurement sessions and found a variation in the range of 1–3·2% (1 SD). They exposed 125 OSL films to doses ranging from 5 up to 202 cGy irradiation in a 6 MV photon beam under identical conditions of irradiation and suggested that on average, the spread in readings for low doses <30 cGy was larger than for higher doses and showed a decrease in SD of 0·3%/100 cGy through linear regression of data points. There was variability found in the literature as far as delivered doses are concerned ranging from 25 to 200 cGy. This can also be considered as a minor contributing factor variation found in the coefficient of variation in the literature. However, the present study utilised 50 cGy identical doses for performing the experiments.

In conclusion, based on our experimental results and comparing them with other studies, we recommend minimum threshold of delivered dose of 50 cGy for calibration studies of OSL dosimeters, because inaccuracy in measurements may be significant and the dosimeter is not going to serve the purpose. More than 50 cGy delivered dose for the OSL calibration studies will increase accumulated dose to OSL dosimeter without any significant improvement in accuracy of measurements.

Further, an average response of the OSLDs was evaluated 0·982 relative to the IAEA TRS-398[49] calculated absolute dose using an ionisation chamber. The maximum percentage deviation in an OSLD was found to be 4·5%, which is low relative to the ionisation chamber measurement. A possible explanation of an under response of the OSLD observed, could be the result of inherent scatter conditions within the detectors due to the ~0·85 mm of air gap on each side, between the aluminium oxide (as an active detector material) and plastic casing.

Energy response

There was a difference observed in response to Co^{60} compared with 10 MV photon beams of ~3% as given in Table 1; however, it was found within experimental uncertainty. Our results for energy dependency suggested that there is an energy independent response of OSLD in Co^{60} to 10 MV photon beam. Dunn et al.[40] performed an energy dependency test for a similar type of OSL nanoDot, this showed little dependence on energy. The largest variation was in response to 6 MV for photons and for electrons was attributed to 1·2% for Co^{60} and 1·6% for 20 MeV beams, respectively. Two different Monte Carlo studies performed by Mobit et al.[50] and Chen et al.[51] on Al_2O_3:C found a decrease in relative response of 2% in 15 MV and 6–24 MV photons, was determined to be 1%, respectively. Both studies concluded that the independent response of OSLD is a function of energy. However, Schembri and Heijmen[48] found a difference of ~4% between 6 and 18 MV photon beams in OSL films of Al_2O_3:C. Previous authors[23,31] performed experiments with the OSL InLight Dot dosimeters, unlike this present study that used OSL nanoDot™ dosimeters. However, the general geometry, construction, and casing material are similar to the InLight nanoDot™. The Dot measures $24 \times 12 \times 2\ mm^3$ with the aluminium oxide disk having a 7-mm diameter, whereas nanoDot™ measures $10 \times 10 \times 2\ mm^3$ with the oxide disk having a 5-mm diameter.[52] However, Jursinic found no energy dependence between 6 and 15 MV photon beams within experimental uncertainty, which is consistence with our results. However, Viamonte et al.[23] found similar results for 6 and 18 MV beams suggesting a single calibration factor in high megavolt beams; however, there was a clear difference in response to Co^{60} compared with megavolt beams of ~4% requiring an energy correction factor for dose assessment at higher energies for detectors calibrated in Co^{60} energy. In contrary to that for high-energy photon beams, Yukihara et al.[32] reported OSLD response for 18 MV was $(0{\cdot}51 \pm 0{\cdot}48)$% of the response for the 6 MV photon beam. However, their results for the response of OSLD in a range of 6–20 MeV electron beams, for 'uncorrected data' OSLD response is on average 1·9% higher than the response to the 6 MV photon beam,

Table 1. The relative energy response of optically stimulated luminescence (OSL) detectors as a function of incident photon energy for Co^{60}, 6 and 10 MV beam

Energy	Relative response of OSLD
Co^{60}	1
6 MV	0·995
10 MV	0·97

Note: The response is normalised to the reading for Co^{60}. Three detectors were used at each energy setting.
Abbreviation: OSLD, optically stimulated luminescence dosimeter.

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demonstrated need of a fixed correction factor. Both the studies by Viamonte et al.[23] and Yukihara et al.[32] normalise the data to response of OSLD for Co^{60} and 6 MV photon beam, respectively. Yukihara et al. eliminated the machine output variation uncertainty in measurements by dividing OSLD response by the machine calibration factor. However, Yukihara et al. investigations were made by circular discs measuring 7 mm in diameter with a Risø TL/OSL-DA-15 reader (Risø National Laboratory, Denmark) was used to read OSL signal. Energy dependence of Al_2O_3:C has been studied several times in the past but the results are varied, the reasons for the differences reported in the literature is unclear. However, in our study the OSLD was calibrated in to a reference Co^{60} beam, any deviation in beam quality for each measurement was expected and was found within experimental uncertainty, suggesting response of the OSLD is energy independent.

Dose rate response

The response of OSLD with dose rate was observed in a clinical treatment range from 200 to 600 MU/minute in 6 MV beam and the response of OSLD was found to be independent of dose rate. Jursinic[31] performed measurements by varying dose-per-pulse ranging from 53·4 Gy/second to 3,208 Gy/minute achieved by changing in source to detector distance delivering a dose of 100 cGy at SAD at d_{max} in 6 MV beam and showed no change in OSL InLight™ Dot dosimeter response for a large 388-fold change in dose-per-pulse. Schembri and Heijmen[48] also tested the dose rate dependency of OSL films by delivering an identical dose of 200 cGy at SAD at d_{max} in 6 MV beam and found that the deviations from the mean OSL response for all films remain within ±1%. Viamonte et al.[23] and Yukihara et al.[32] also showed the response of OSLD independent of dose rate up to 400 cGy/minute in Co^{60} beam and 600 MU/minute in 6 MV photon beam in their respective studies. However, Yukihara et al. also investigated dose rate check with 1,000 MU/minute in a 9 MeV electron beam and the overall OSL response was found within ±1%. Sharma et al.[39] reported no dose rate effect on OSLD in high-dose rate brachytherapy measurements ranging from a change in dose rate of 3·5–0·14 cGy/second at 2 and 10 cm, respectively. Our findings were inconsistent with all the above studies for dose rate dependency suggesting that there was no effect of dose rate on response of OSLD.

Dose response

Our experimental results for dose linearity showed that OSL dosimetry provides a good linearity until ~250 cGy of dose with a coefficient of determination (R^2) value 0·997 but after that OSL dose shows supra-linear response at higher doses as shown in Figure 2. One other similar study, performed by Jursinic[31] reported a linear response with absorbed dose over a test range of 1–300 cGy; >300 cGy a small supra-linear behaviour occurs. Our supra-linear response is in agreement with a few similar studies reported by Reft,[34] Jursinic,[35] Mrcela et al.[37] and Schembri and Heijmen.[48] They observed dose linearity up to around 200 cGy followed by an increase in OSL sensitivity. A large number of charge transfer reactions take place during irradiation and stimulation of an OSL material, which is a complex phenomenon. However, an increase in OSL sensitivity at higher doses is explained by an increase in deep and intermediate electron trap concentrations in a competitive manner with accumulated doses. The shift in the concentrations of deep and intermediate electron traps impacts on the magnitude of supra-linearity at higher doses.

Field size and depth dependency

OSLDs were irradiated with a Co^{60} beam using three different field sizes of 4×4, 10×10 and $30 \times 30\ cm^2$, and four depths of 1·5, 5, 10 and 14 cm. OSL responses (counts/delivered dose) relative to the overall mean response for all fields and depths were observed within ±3·5%, which was within experimental uncertainty. Mrcela et al.[37] reported variation in OSL sensitivity with field size <1%. Yukihara et al.[32] investigated the field size dependency and the maximum variation was ±1% as compared with ionisation chamber measurement for four different field sizes ranging from $5 \times 5\ cm^2$ to $30 \times 30\ cm^2$ irradiated with 200 MU at a depth of 10 cm in water 6 MV beam. Our results are in agreement with Schembri and Heijmen[48] who found deviations in the overall mean response of OSL

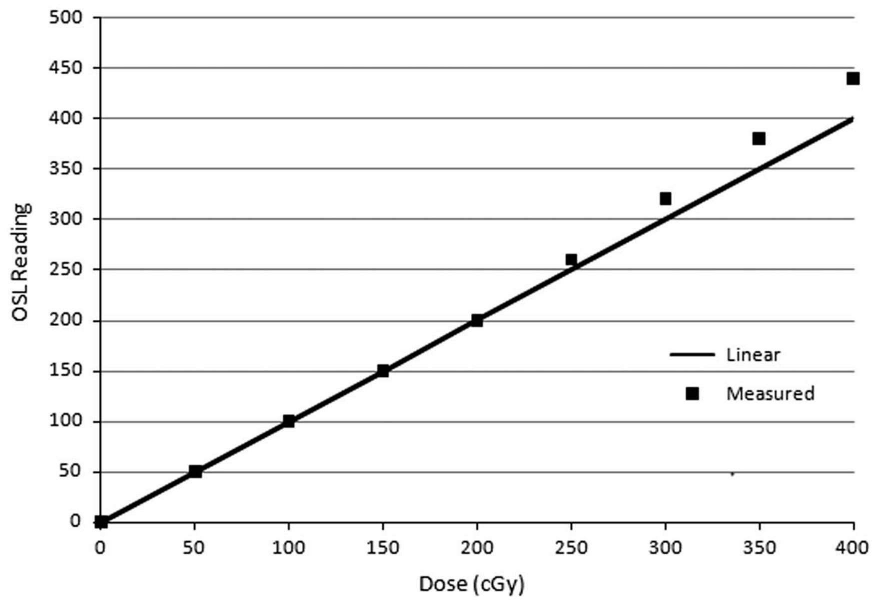

Figure 2. The dose linearity curve showing measured response of optically stimulated luminescence dosimeter (OSLD) as a function of delivered absorbed dose.
Note: The solid line shows linear dependency on dose based on the response of OSLD data upto 400 cGy.
Abbreviation: OSL, optically stimulated luminescence.

films within 2·5% when comparing different field sizes, at various depths in the phantom.

Angular response

The angular response was studied for OSLD with 0·5 cm build-up of solid slab phantom placed over detector and without using build-up, respectively, in a Co^{60} beam and the results of angular correction factors are shown in Figure 3. The maximum deviation of OSLD was found to be, 8% relative to the response of OSLD at gantry angle 0° with build-up, the maximum deviation observed was an under response of OSLD by 38% without build-up set up, respectively. The results of the test of normal incidence of radiation on detector for without build-up observed an under response of measured dose due to lack of build-up. This supports the findings of Jursinic and Yahnke[36] in that the use of the detector without a build-up cap is the appropriate way to measure dose on the surface of the patient but it is an unsuitable method to measure dose at other depths in the patient. However, placing the detector on the surface of a patient will perturb dose delivered by the attenuation and scatter of beam. The results of the OSLD showed an under response of the dosimeter at angles approaching 90–270° in both with and without build-up set up studies. In our experiment, a drop in signal at the 90–270° points is due to the limitations in phantom and dosimeter geometry at these extreme angles. This experimental method has the limitation of using a rectangular geometry phantom in which dose to a point in the rectangular phantom changes with the incident angle of radiation. At more oblique angles, the dose decreases due to attenuation of the beam passing through the edge of the phantom.

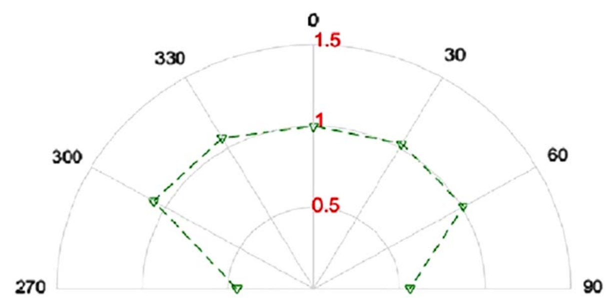

Figure 3. Angular response curve for optically stimulated luminescence dosimeter (OSLD). All dosimeters were irradiated in a 10 × 10 cm² field, at a depth of 0·5 cm.
Notes: The angular response of OSLD measured at gantry ranging from 90 to 270° relative to the axis of gantry rotation, at an interval of 30°. The responses of OSLD at any angle normalised to response of OSLD at 0° gantry angle.

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Thus, the angular dependence noticed at the extreme angles can simply be ignored due to experimental uncertainty. Cylindrical phantoms were best suited for angular dependency check with no angular dependency and eliminating above-mentioned experimental uncertainty. Jursinic[41] performed a study in which nanoDots™ were irradiated in cylindrical, cubical or rectangular phantoms in a 6 MV beam and showed a maximum angular dependence of 1% or less at an incidence angle of 90°. In our study, the results for angular dependency of nanoDots™ in a Co^{60} beam, disagree with Jursinic, the possible reason for discrepancy is the use of a cylindrical phantom with relatively small dimensions and in the 6 MV beam the gantry was stationary and the phantom was rotated for the measurements. However, our measurements were performed in a rectangular phantom in the Co^{60} beam and the gantry was rotated, whereas the phantom was stationary. The findings of our data were consistent with two Monte Carlo simulation studies performed on OSL nanoDots™ by Lehmann et al.[53] and Kerns et al.[52] and suggested a small angular dependency of OSLD due to the variation observed in the response of OSLD when irradiated with the incident photon beams parallel to the plane of the dosimeter (0°) to the response of OSLD when irradiated with the incident beam at any other angle to the plane of the dosimeter. Measurements performed by Lehmann et al. were also made in a rectangular phantom and they reported a small angular dependence of ~2%, which needs to be considered for measurements involving other than normal incident beam angles. Kim et al.[54] showed a 70% angular dependence when irradiation is done in a highly asymmetric field that occurs on the surface of a phantom. Kerns et al.[52] reported a drop in nanoDot™ response at 90° versus 0° incident angle of the radiation beam for 6 and 18 MV, this was found to be 4 and 3%, respectively. Their Monte Carlo simulations at 6 MV showed similar results to their experimental values. He suggested that the larger size of Dot dosimeter contributes more to the increase in angular dependency than the small size OSL nanoDot™ dosimeter.

Signal depletion

The re-readout property of OSL dosimeter is an invaluable advantage. This allows multiple readings for better statistical results and a permanent record that can be read again much later on if need be. However, each readout of the dosimeter depletes the amount of trapped charge by a small fraction. For a comprehensive approach, we have evaluated the depletion in signal per readout. The decrease in signal was found to be 0·29% per readout for OSL nanoDot™. A similar study using OSL Dot dosimeters was performed by Mrcela et al.[37] and demonstrated a reading uncertainty of a single dosimeter was found to be 0·6% delivering 50 cGy with one irradiation of each dot. Jursinic[31] showed that the luminescence signal is reduced by ~0·2%.

Long-term fading

The long-term fading of signal of the dosimeter was studied and was found to be 1·4% in 1 week. The signal fading data of three dosimeters were recorded and shown in Figure 4, which shows that the fading behaviour was very much consistent between dosimeters. In a long-term study, the response showed a signal fading of ~1·8% after 37 days compared with 1 day post-irradiation. Dunn et al.[40] showed that the nanoDots™ lost ~2·5% over 30 days if the readout time was normalised to 2 days following irradiation. Mrcela et al.[37] reported a 4% lower response post-58 days when first readout was taken 1 hour after the irradiation. Viamonte et al.[23] showed a

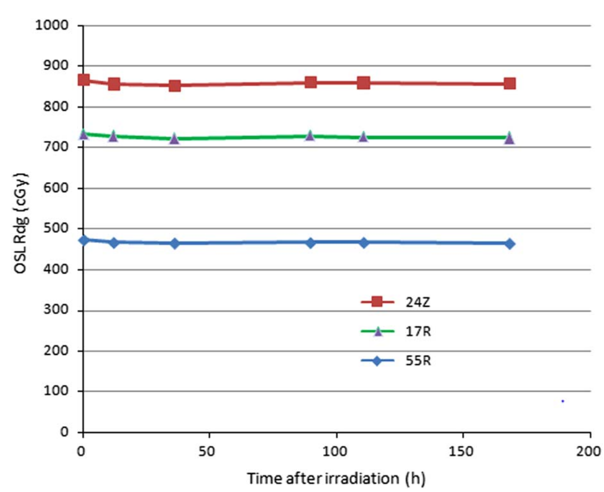

Figure 4. Post-irradiation fading response of optically stimulated luminescence (OSL) dosimeter in a week period.
Note: Three dosimeters were used identified as 24Z, 17R, 55R, the three digits of unique alphanumeric code of dosimeter.

drop in signal of about 2% within the first 5 days after irradiation. Beyond 5 days, the signal was stable up to 21 days post-irradiation. One of the other studies of long-term fading between 3 weeks by Schembri and Heijmen[48] measured fading of OSL films, was <1·8%. However, their first reading for normalisation was not taken until day 17, as the OSL films had to be sent away for reading. Our results are in agreement with all the above-mentioned studies. However, small variation were observed due to different types of OSLDs and time elapsed between taking first readout after irradiation because the signal decays very fast in an exponential manner for the first few minutes after the irradiation. The long-term fading property study showed that the dosimeter is a suitable for keeping a permanent dose record. The results of this present study supports the statement of Dunn et al.[40] that the fast read-out, accuracy and reusability of nanoDot™ dosimeters make the dosimeter is a viable replacement for TLD in large-scale dosimetry operations for dosimetry audits. However, in the present scenario, these commercial dosimeters are more expensive than the TLDs.

PDD and relative output factor measurement

The PDD curves for Co^{60} beam measured using OSL detectors and a Markus ionisation chamber are shown in Figure 5. There was an inherent build-up of 0·42-mm thick plastic cover in the case of OSLD and 0·2 mm in the case of the ionisation chamber measurements, respectively. The OSL curve shows a good agreement with the ionisation chamber measurements, at greater depths, the OSL measurements were within ±1% relative to ionisation chamber measurements. Our results are in agreement with Yukihara et al.[32] who reported that maximum relative errors were measured until a depth of 15 cm and, were found to be 1·7% for 6 MV photon beam ($d = 13$ cm) and 0·7% for 18 MV photon beam ($d = 2$ cm) as compared with commissioning data acquired with Scanditronix/Wellhofer CC13 (Radiation Products Design Inc., Albertsville, MN, USA) ionisation chambers measurements. Further, the relative output factors were also measured using OSL detectors and Farmer type ionisation chamber, which are

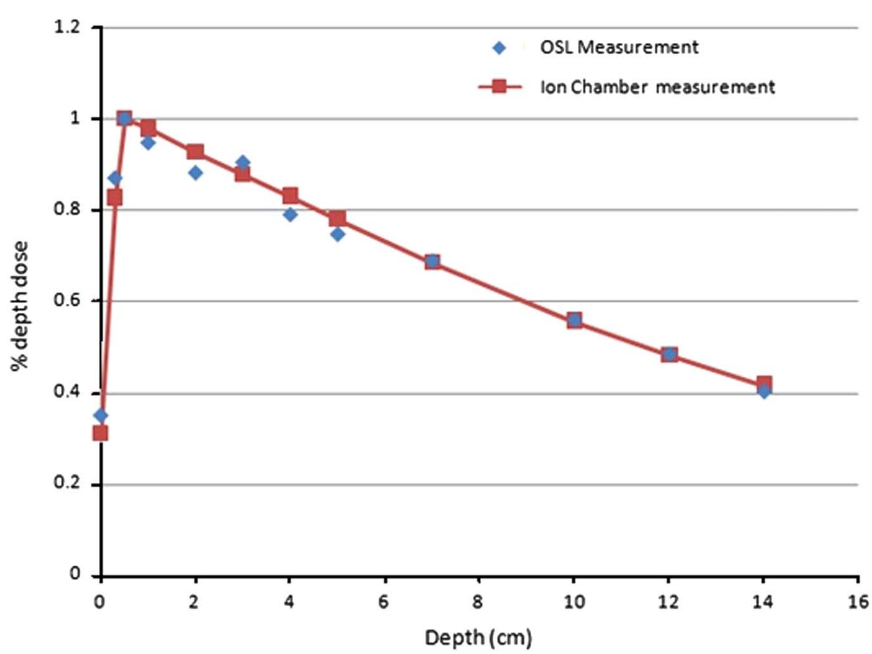

Figure 5. Showing comparison of percentage depth dose (PDD) curves in solid water for 10 × 10 cm² field at source-to-surface distance 80 cm in a Co^{60} beam upto 14·0-cm depth.
Note: The solid line shows PDD curve measured using a Markus parallel plate ionisation chamber for the irradiation conditions and energy.
Abbreviation: OSL, optically stimulated luminescence.

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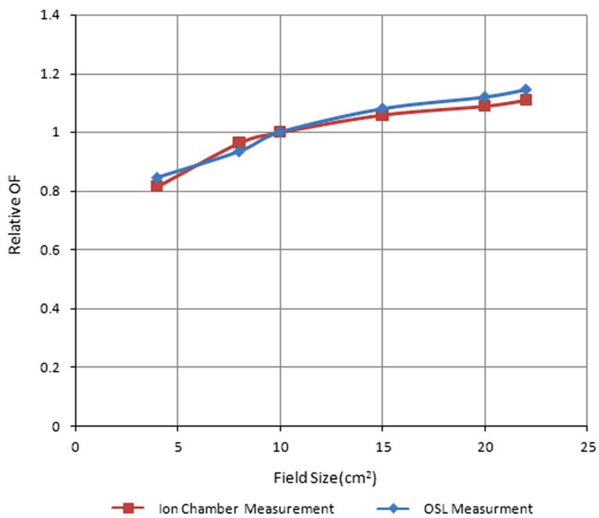

Figure 6. Showing comparison of relative output factor curves measured using optically stimulated luminescence (OSL) dosimeters and a Farmer type cylindrical ionisation chamber, in solid water for field size ranging from $4 \times 4\,cm^2$ to $22 \times 22\,cm^2$ in a Co^{60} beam.

presented in Figure 6. The variation in OSL measurements for all field sizes considered were found to be within ±3% relative to ionisation chamber measurements. Our findings are in agreement with Viamonte et al.[23] who demonstrated a difference of <1% in relative output factors measurement for Co^{60} beam with an NE2571 (PTW, Freiburg, Germany) ionisation chamber placed at a depth of 5 cm in a solid water phantom.

The literature reports several studies on Al_2O_3:C OSL material. Most of them were performed on single crystal of Al_2O_3:C,[15] OSL film strips[48] and in house-developed powder and discs of Al_2O_3:C, OSL Luxel™ dosimeters[18]/ OSL InLight™ Dot[23,31,32] dosimeters from Landauer Inc. Schembri and Heijmen[48] used Al_2O_3:C-based OSL film strips and Viamonte et al.,[23] Mrcela et al.[37] used OSL Dots with the microStar reader system in their studies, respectively. In contrast, Yukihara et al.,[32] used an automated Risø TL/OSL-DA-15 reader to carry out experiments and before irradiation, the dosimeters were optically illuminated. However, to date the OSL InLight nanoDot™ from Landauer Inc.,[36–41,52–54] is relatively new for use in in vivo radiation dosimetry and was used in the present study with a InLight™ microStar® reader as OSL readout assembly. Despite of the fact that the OSL Al_2O_3:C material remains the same as well as the general geometry, construction, outer covering material, there is variability found in literature for experimental set up condition. This variability found in terms of irradiation dose, beam energy and phantom geometry, etc., irradiation history of detector, that is, the use of new/reuse of dosimeters with accumulated dose for experiments/various optical bleaching methods applied before experimental irradiation of OSLD to eliminate any background signal and difference in readout methodologies with different types of readout assembly used for readout may result in alteration in some of the characteristics of OSL dosimeter. This is the possible reason for the variability found in literature with respect to present study for few characteristics, for example, angular and energy dependency of the Al_2O_3: C-based OSLD.

CONCLUSION

The dosimetric characteristics of the commercial OSL system were studied. The results demonstrate good accuracy and precision as compared with the ionisation chamber. It was observed that the OSL response is energy and dose rate is independent. Our experimental results for dose linearity show a linear OSLD response until ~250 cGy, which was just above the normal clinically relevant dose range in radiotherapy but above this dose, a supra-linear behaviour of OSLD was observed. Thus, dose evaluation by OSLD at high doses requires a non-linear calibration factors or a high-order polynomial fits need to be applied in evaluation of delivered dose. This study suggests the need to use appropriate build-up during surface dose measurements in multi-field angular beam delivery in radiotherapy for accurate dose estimation, as insufficient build-up may result in inaccurate dosimeter readings. Based on the results of this study, the linear dose response in clinically relevant dose range and the high sensitivity of OSL dosimeter and the ease of use and stability of OSL system, make it a good choice of dosimeter for in vivo dosimetry in radiotherapy. The variation in the response of OSLD for measuring PDD and relative output factors were measured and compared with ionisation chamber absolute measurements, which were found very close to ionisation chamber measurements. This shows that OSLD is a good relative dosimetric tool which can be used as a relative radiotherapy dosimeter in the future.

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Conflicts of Interest

The authors declare that they have no conflicts of interest.

References

1. Kutcher G J, Coia L, Gillin M et al. Comprehensive QA for radiation oncology. Report of Task Group No. 40 of the Radiation Therapy Committee of the American Association of Physicists in Medicine. AAPM Report No. 46. Med Phys 1994; 21 (4): 581–618.
2. Kirby T H, Hanson W F, Gastorf R J, Chu C H, Shalek R J. Mailable TLD system for photon and electron therapy beams. Int J Radiat Oncol 1986; 12 (2): 261–265.
3. Dam J V, Marinello G. Methods for In Vivo Dosimetry in External Radiotherapy ESTRO Physics Booklet No. 1, 2nd edition. Brussels: ESTRO, 2006.
4. Kumar A S, Sharma S D, Ravindran B P. Characteristics of mobile MOSFET dosimetry system for megavoltage photon beams. J Med Phys 2014; 39 (3): 142–149.
5. Jursinic P A, Nelms B E. A 2-D diode array and analysis software for verification of intensity modulated radiation therapy delivery. Med Phys 2003; 30: 870–879.
6. Saini A S, Zhu T C. Dose rate and SSD dependence of commercially available diode detectors. Med Phys 2004; 31: 914–924.
7. Yorke E, Alecu R, Ding L et al Diode in vivo dosimetry for patients receiving external beam radiation therapy. Report of Task Group No. 62 of the Radiation Therapy Committee of the American Association of Physicists in Medicine. AAPM Report No. 87. https://www.aapm.org/pubs/reports/RPT_87.pdf. Accessed on 27th August 2016.
8. Soubra M, Cygler J, MacKay G. Evaluation of a dual bias dual metal–oxide–silicon semiconductor field effect transistor detector as radiation detector. Med Phys 1994; 2: 567–572.
9. Lambert J, Nakano T, Law S, Elsey J, McKanzie D R, Suchowerska N. *In vivo* dosimeters for HDR brachytherapy: a comparison of a diamond detectors, MOSFET, TLD, and scintillation detector. Med Phys 2007; 34: 1759–1765.
10. International Atomic Energy Agency. Development of procedures for in vivo dosimetry in radiotherapy. IAEA Human Health Reports No. 8. Vienna: IAEA, 2013. http://www-pub.iaea.org/MTCD/Publications/PDF/Pub1606_web.pdf. Accessed on 27th August 2016.
11. Bhatt B C. Thermoluminescence, optically stimulated luminescence and radiophotoluminesence dosimetry: an overall perspective. Radiat Prot Environ 2011; 34 (1): 6–16.
12. Antonov-Romanovskii W, Keirum-Marcus I F, Poroshina M S, Trapeznikova Z A. IR stimulable phosphors. Conference of the Academy o f Sciences o f the USSR on the Peaceful Uses o f Atomic Energy. USAEC Report AEC-tr-2435, Moscow, 1955, Session of the Division of Physical and Mathematical Sciences. Moscow: USAEC, 1956: 239–250.
13. Braeunlich P, Schaefer D, Scharmann A. A simple model for thermoluminescence and thermally stimulated conductivity of inorganic photoconducting phosphors and experiments pertaining to infrared-stimulated luminescence. Proceedings o f the First International Conference on Luminescence Dosimetry, Stanford, June 1965. Stanford, CA: USAEC, 1967: 57–73.
14. Sanborn E N, Beard E L. Sulfides of strontium, calcium, and magnesium in infrared stimulated luminescence dosimetry. Proceedings of the First International Conference on Luminescence Dosimetry, Stanford, June 1965. Stanford, CA: USAEC, 1967: 183–191.
15. Akselrod M S, Kortov V S, Kravetsky D J, Gotlib V I. Highly sensitive thermoluminescent anion-defective-Al2O3:C single crystal detectors. Radiat Prot Dosimetry 1990; 32: 15–20.
16. McKeever S W S, Akselrod M S, Markey B G. Pulsed optically stimulated luminescence dosimetry using Al2O3:C. Radiat Prot Dosimetry 1996; 65: 267–272.
17. Akselrod M S, Lucas A C, Polf J C, McKeever S W S. Optically stimulated luminescence of Al2O3. Radiat Meas 1998; 29 (3–4): 391–399.
18. McKeever S W S, Akselrod M S. Radiation dosimetry using pulsed optically stimulated luminescence of Al2O3:C. Radiat Prot Dosimetry 1999; 84: 317–320.
19. Akselrod M S, McKeever S W S. A radiation dosimetry method using pulsed optically stimulated luminescence. Radiat Prot Dosimetry 1999; 81 (3): 167–176.
20. Bøtter Jensen L G, McKeever S W S, Wintle A G. Optically Stimulated Luminescence Dosimetry. Amsterdam: Elsevier, 2003.

21. Akselrod M S, Bøtter-Jensen L, McKeever S W S. Optically stimulated luminescence and its use in medical dosimetry. Radiat Meas 2007; 41: S78–S99.

22. Yukihara E G, McKeever S W S. Optically stimulated luminescence (OSL) dosimetry in medicine. Phys Med Biol 2008; 53 (20): R351–R379.

23. Viamonte A, Da Rosa L A R, Buckley L A, Cherpak A, Cygler J E. Radiotherapy dosimetry using a commercial OSL system. Med Phys 2008; 35 (4): 1261–1266.

24. Bos A J J. High sensitivity thermoluminescence dosimetry. Nucl Instrum Methods Phys Res B 2001; 184: 3–28.

25. Edmund J M, Andersen C E. Temperature dependence of Al_2O_3:C response in medical luminescence dosimetry. Radiat Meas 2007; 42: 177–189.

26. Aznar M C, Andersen C E, Bøtter-Jensen L et al. Real-time optical-fiber luminescence dosimetry for radiotherapy: physical characteristics and application to photon beams. Phys Med Biol 2004; 49: 1655–1669.

27. Gaza R, McKeever S W S, Akselrod M S et al. A fiber dosimetry method based on OSL from Al_2O_3:C: for radiotherapy applications. Radiat Meas 2004; 34: 809–812.

28. Polf J C, Yukihara E G, Akselrod M S, McKeever S W S. Real-time luminescence from Al_2O_3:C fiber dosimeters. Radiat Meas 2004; 34: 227–240.

29. Yukihara E G, Yosimura E M, Lindstrom T D, Ahmad S, Taylor K K, Mardirossian G. High-precision dosimetry for radiotherapy using optically stimulated luminescence technique and thin Al_2O_3:C dosimeters. Phys Med Biol 2005; 50: 5619–5628.

30. Takegami K, Hayashi H, Okino H et al. Practical calibration curve of small-type optically stimulated luminescence (OSL) dosimeter for evaluation of entrance skin dose in the diagnostic X-ray region. Jpn J Radiol Technol 2015; 8 (2): 286–294.

31. Jursinic P A. Characterization of optically stimulated luminescent dosimeters, OSLDs, for clinical dosimetric measurements. Med Phys 2007; 34 (12): 4594–4604.

32. Yukihara E G, Mardirossian G, Mirzasadeghi M, Guduru S, Ahmad S. Evaluation of Al2O3:C optically stimulated luminescence (OSL) dosimeters for passive dosimetry of high-energy photon and electron beams in radiotherapy. Med Phys 2008; 35 (1): 260–269.

33. Sawakuchi G O, Yukihara E, McKeever S W S et al. Relative optically stimulated luminescence and thermoluminescence efficiencies of Al2O3:C dosimeters to heavy charged particles with energies relevant to space and radiotherapy dosimetry. J Appl Phys 2008; 104: 124903.

34. Reft C S. The energy dependence and dose response of a commercial optically stimulated luminescent detector for kilovoltage photon, megavoltage photon, and electron, proton, and carbon beams. Med Phys 2009; 36: 1690–1699.

35. Jursinic P A. Changes in optically stimulated luminescent dosimeter (OSLD) dosimetric characteristics with accumulated dose. Med Phys 2010; 37 (1): 132–140.

36. Jursinic P A, Yahnke C J. In vivo dosimetry with optically stimulated luminescent dosimeters, OSLDs, compared to diodes; the effects of buildup cap thickness and fabrication material. Med Phys 2011; 38 (10): 5432–5440.

37. Mrčela I, Bokulić T, Izewska J, Budanec M, Fröbe A, Kusić Z. Optically stimulated luminescence in vivo dosimetry for radiotherapy: physical characterization and clinical measurements in ^{60}Co beams. Phys Med Biol 2011; 56 (18): 6065–6082.

38. Scarboro S B, Followill D S, Kerns J R, White R A, Kry S F. Energy response of optically stimulated luminescent dosimeters for non-reference measurement locations in a 6 MV photon beam. Phys Med Biol 2011; 57 (18): 2505–2515.

39. Sharma R, Jursinic P A. In vivo measurements for high dose rate brachytherapy with optically stimulated luminescent dosimeters. Med Phys 2013; 40 (7): 071730-1–12.

40. Dunn L, Lye J, Kenny J, Lehmann J, Williams J, Kron T. Commissioning of optically stimulated luminescence dosimeters for use in radiotherapy. Radiat Meas 2013; 51–52: 31–39.

41. Jursinic P A. Angular dependence of dose sensitivity of nanoDot optically stimulated luminescent dosimeters in different radiation geometries. Med Phys 2015; 42 (10): 5633–5641.

42. Pradhan A S, Lee J I, Kim J L. Recent developments of optically stimulated luminescence materials and techniques for radiation dosimetry and clinical applications. Med Phys 2008; 33 (3): 85–99.

43. Landauer. InLight nanoDot™ Dosimeter. Glenwood, IL: Landauer, 2012.

44. Landauer. InLight microStar® Reader. Glenwood, IL: Landauer, 2010.

45. Palan C B. Developments in OSL dosimetry. GJSET Publishing 2013; 14: 1–17.

46. Werner B L, Das I J, Khan F M, Meigooni A S. Dose perturbations at interfaces in photon beams. Med Phys 1987; 14: 585–595.

47. Aird E G A, Burns J E, Day M J et al. Central axis depth dose data for use in radiotherapy. London: British Institute of Radiology, 1996; 46–61.

48. Schembri V, Heijmen B J M. Optically stimulated luminescence (OSL) of carbon-doped aluminum oxide (Al_2O_3:C) for film dosimetry in radiotherapy. Med Phys 2007; 34 (6): 2113–2118.

49. Andreo P, Burns D T, Hohlfeld K et al. Absorbed Dose Determination in External Beam Radiotherapy: An International Code of Practice for Dosimetry Based on Standards of Absorbed Dose to Water. Vienna: International Atomic Energy Agency, 2000.

50. Mobit P, Agyingi E, Sandison G. Comparison of the energy-response factor of LiF and Al_2O_3 in

https://doi.org/10.1017/S1460396917000346 Published online by Cambridge University Press

radiotherapy beams. Radiat Prot Dosimetry 2006; 119 (1–4): 497–499.

51. Chen S W, Wang X T, Chen L X, Tang Q, Liu X W. Monte Carlo evaluations of the absorbed dose and quality dependence of Al_2O_3:C in radiotherapy photon beams. Med Phys 2009; 36: 4421–4424.
52. Kerns J R, Kry S F, Sahoo N, Followill D S, Ibbott G S. Angular dependence of the nanoDot OSL dosimeter. Med Phys 2011; 38 (7): 3955–3962.
53. Lehman J, Dunn L, Lye J E et al. Angular dependence of the response of nanoDot OSLD system for measurements at depth in clinical megavoltage beams. Med Phys 2014; 41 (6): 061712-1–9.
54. Kim D W, Chung W K, Shin D O et al. Dose response of commercially available optically stimulated luminescent detector, Al_2O_3:C for megavoltage photons and electrons. Radiat Prot Dosimetry 2012; 149 (2): 101–108.