Radiotherapy dosimetry using a commercial OSL system

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A commercial optically stimulated luminescence (OSL) system developed for radiation protection dosimetry by Landauer, Inc., the InLight™ microStar reader, was tested for dosimetry procedures in radiotherapy. The system uses carbon-doped aluminum oxide, Al_2O_3:C, as a radiation detector material. Using this OSL system, a percent depth dose curve for ^{60}Co gamma radiation was measured in solid water. Field size and SSD dependences of the detector response were also evaluated. The dose response relationship was investigated between 25 and 400 cGy. The decay of the response with time following irradiation and the energy dependence of the Al_2O_3:C OSL detectors were also measured. The results obtained using OSL dosimeters show good agreement with ionization chamber and diode measurements carried out under the same conditions. Reproducibility studies show that the response of the OSL system to repeated exposures is 2.5% (1sd), indicating a real possibility of applying the Landauer OSL commercial system for radiotherapy dosimetric procedures. © *2008 American Association of Physicists in Medicine*. [DOI: 10.1118/1.2841940]

Key words: optically stimulated luminescence, Al_2O_3:C, radiation therapy dosimetry

I. INTRODUCTION

The application of radiotherapy for cancer treatment is a complex process that includes all aspects of diagnosis, treatment planning and treatment delivery. Among the most important steps for an accurate radiation treatment are the beam dosimetry and quality assurance procedures, including *in vivo* measurements. These measurements become even more important in light of conformal treatment techniques such as intensity-modulated radiotherapy (IMRT) that requires dosimeters with high spatial resolution and high sensitivity. There are many types of dosimeters available for this purpose and new dosimeters and dosimetric systems are continually being developed with this aim.[1] Presently, the most popular techniques for dosimetric quality assurance in radiotherapy use ionization chambers, thermoluminescence dosimeters (TLDs), diodes and metal–oxide–semiconductor field effect transistors (MOSFETs). The use of TLDs is hindered by the complicated readout process of heating and annealing, which precludes instantaneous or near-instantaneous dose readout. Diodes which are widely used for *in vivo* dosimetry[2–5] give an instantaneous readout, however, their calibration factor is temperature dependent and they have nonuniform angular response to radiation. Recent MOSFET dosimeters are free of the above drawbacks, however, their lifetime is limited.

Optically stimulated luminescence (OSL) has been used extensively in radiation protection and has more recently been investigated for dosimetric applications in radiotherapy.[6–9]

Optically stimulated luminescence is similar to thermoluminescence, but utilizes light instead of heat to provoke the radiation-induced luminescence. Once exposed to ionizing radiation, the material is illuminated with a steady source of light of an appropriate wavelength and intensity from light emitting diodes (LEDs) or lasers and the luminescence stimulated from the dosimeter during this procedure is monitored as a function of the stimulation time. The integral of the luminescence emitted during the stimulation period is a measure of the dose of radiation absorbed by the material. Through an appropriate calibration of OSL signal against known values of the dosimetric quantity of interest, this quantity can be evaluated.[10] This process of stimulation does not involve annealing the sample as in the case of TLDs and therefore the signal from an OSL detector can be read multiple times. This permits the storage of the detector as a record of the delivered dose where it can be reanalyzed if necessary.

Different thermoluminescent materials show OSL properties, including carbon-doped aluminum oxide ($Al_2O_3:C$), which is the most commonly used OSL dosimetric material. As a thermoluminescent material, it is extremely sensitive: 40–60 times the TL sensitivity of LiF:Mg,Ti (Ref. 11) or higher.[12] This property permits the use of small samples of this material for dosimetric measurements, a very interesting characteristic if high spatial resolution is required, as for *in vivo* measurements involving high dose gradients or for the measurements of low dose regions of a modulated field.

Recent advances in the use of OSL in radiotherapy quality control for *in vivo* measurements use remotely placed light sources and optical fibers to simultaneously stimulate the OSL dosimeter and detect the resulting luminescence.[13–17] Huston *et al.*[18] investigated the possibility of remotely monitoring radiation doses via fiber optic cables for radiation protection using a doped glass as the detector. In this case, the radiation induces opacity changes within the glass that are monitored via fiber optic cables.

More recently, Aznar *et al.*[8] and Andersen *et al.*[9] tested a prototype OSL system that combined an $Al_2O_3:C$ detector with fiber optic cables for real time dose measurements in clinical radiotherapy settings. Their measurements showed that the $Al_2O_3:C$ was independent of energy for 6 and 18 MV beams and that the response increased linearly with dose rate. The time resolution of the system is 0.1 s and the spatial resolution <0.5 mm.[9] The detectors also showed good results when compared to a treatment planning system calculations for patient treatments, including IMRT.

Yukihara *et al.*[7] investigated the use of $Al_2O_3:C$ OSL detectors for use in quality assurance of clinical radiotherapy beams. The $Al_2O_3:C$ dosimeters were made by Landauer, Inc. (Glenwood, IL) and were circular disks measuring 7 mm in diameter. A Risø TL/OSL-DA-15 reader (Risø National Laboratory, Denmark) was used to quantify the OSL signal. The uncertainty of a single OSL measurement, estimated from the variance of a large sample of dosimeters irradiated with the same dose, was 0.7%. They found that the reproducibility of the OSL signal for multiple irradiations was on the order of 1%. They used the OSL dosimeters to measure a depth dose curve in a 6 MV photon beam and found that the OSL results agreed with commissioning data to within 1.1% for all depths from 0.5 to 15 cm. Their study concluded that OSL dosimeters were suitable for dosimetry measurements in radiotherapy beams and that they could be integrated into quality assurance programs.

A recent study by Schembri and Heijmen evaluated the response of OSL films produced by Landauer.[19] They performed a number of irradiations in order to characterize the response of the films but did not have a reader at their disposal so sent the irradiated dosimeters to Landauer, Inc. for readout. Their dosimeters, which they call films, were not encapsulated inside plastic holders, as were the detectors we used in our experiments, but were instead sealed in light-tight black envelopes. Therefore, additional uncertainty related to reproducible opening of holders inside the reader was not present in Schembri and Heijmen study.

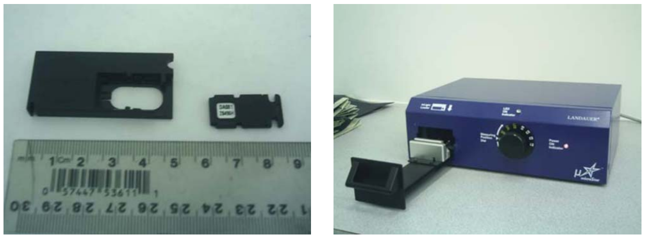
(a) (b)

FIG. 1. Landauer InLight™ microstar reader and OSL detector. The OSL detector located in a light-tight plastic case (a) is inserted into a larger holder at the time of readout. The holder is placed into the InLight™ reader (b) and the drawer is closed for readout.

In an effort to incorporate OSL dosimetry more easily into radiotherapy clinics, Landauer, Inc. has developed a simple and efficient commercial OSL system for dosimetric applications: the InLight™ system.[12] It utilizes $Al_2O_3:C$ as the radiation sensitive material. This system was designed for individual monitoring programs in radiation protection and presents the possibility to be applied for quality control in radiotherapy. Therefore, it is important to assess its dosimetric characteristics in order to investigate if the InLight™ system is suitable for this application.

In the present work the InLight™ microstar reader was evaluated for its suitability for use in clinical radiation dosimetry. A batch of Landauer dosimeters, based on $Al_2O_3:C$, were used without any modification. These dosimeters are called “dots” and they are encapsulated in special light-protective plastic holders. Using this system, a series of experiments was carried out to investigate the accuracy of the OSL system as a dosimeter for clinical measurements and to evaluate the dosimetric properties of the system. Where applicable, the results obtained with the OSL system were compared with similar measurements carried out with an ionization chamber and photon diode and with the results from Schembri and Heijmen.[19]

II. MATERIALS AND METHODS

The OSL system used was a commercial InLight™ microStar reader, manufactured by Landauer, Inc. The detectors consist of $Al_2O_3:C$ samples enveloped in plastic holders. The plastic casing is light-tight and is removed only during readout of the dosimeter. Figure 1(a) shows the plastic holders containing the small $Al_2O_3:C$ dot. During readout, these plastic holders are placed into the larger holder also shown in Fig. 1(a). This is then placed into the reader drawer, shown open in Fig. 1(b). Inside the reader, the plastic case over the detector is slid open and the $Al_2O_3:C$ chip is stimulated using an LED. It is possible to use this reader in two different modes: hardware test or standard operating mode. In hardware test mode (which is not described in the current user manual and therefore not easily available to the average user), the system outputs the raw photomultiplier counts. In standard operating mode, the system outputs the readings which are raw counts divided by the sensitivity of the detector and the calibration factor expressed in counts per mrem.

In our study we used the microStar reader exclusively in the standard operating mode. The dot readings obtained from the system were converted off-line into dose using the individual calibration factors obtained for each dot based on exposure to a known dose.

All irradiations used the detectors shown in Fig. 1(a). Most irradiations were carried out in a 30×30 cm^2 solid water phantom. A solid water slab was machined to have a small slot into which the detector would fit neatly, centered in and aligned with the surface of the block. The ^{60}Co irradiations were carried out using Theratron 780 machine. Two Siemens linear accelerators were used for irradiations with 6, 10, and 18 MV beams. When absolute dose values were of interest, the dose was monitored using an NE2571 ionization chamber in solid water simultaneously with the OSL detector irradiation. Unless otherwise indicated, all irradiations were carried out using an SSD setup (80 cm for ^{60}Co and 100 cm for the accelerator beams) and a 10×10 cm^2 field. Each set of measurements was performed using a new set of detectors and following irradiation, the detectors were stored in black envelopes to minimize exposure to ambient light.

The stability of the reader was evaluated by reading the detectors several times following a single exposure. These readings showed that there were random fluctuations in the signal which were on the order of 1% (1sd). The system itself is designed such that the detectors can be read multiple times without losing the OSL signal. The manufacturer estimates that with each reading approximately 0.2% of the signal is removed. This is below the threshold of detection based on statistical fluctuation of the readings. For all subsequent measurements, each detector was read three times following an irradiation and the average of those three readings was taken as the detector signal.

Of greater interest is the reproducibility of the detector response when exposed to identical doses multiple times. This was investigated for ^{60}Co by irradiating each detector to 50 cGy ten times and taking three readings after each exposure. After each exposure the response of the detector was calculated in terms of net reading, by subtracting the previous reading from the current one. The reproducibility for a given detector was evaluated as one standard deviation of the mean response, averaged over ten separate irradiations.

One of the advantages of OSL detectors is that a single readout does not anneal the chip and it can be stored and readout at a later date. In order to evaluate the change in signal as a function of the postirradiation time prior to readout, five detectors were irradiated to an initial dose of 50 cGy in ^{60}Co and were read out at various time intervals spanning 1 h–21 days. The change in the signal was normalized to the initial reading, taken 1 h post-irradiation.

The suitability of the InLight™ system for clinical use is also dependent on the variation in signal with beam energy. For measurements that aim to provide a dose value, the detectors would have to be calibrated in order to convert the measured signal into an absorbed dose. The energy dependence of the OSL detectors response was measured for ^{60}Co and for nominal photon beam energies of 6, 10, and 18 MV using a Siemens linear accelerator. Detectors were irradiated at the depth of maximum dose ($d_{\max}$) to a dose of 50 cGy. The absolute dose during each irradiation was simultaneously measured using an NE2571 ion chamber. A calibration factor was determined for each detector in units of cGy/rdg. Using the variable dose rate available for a 6 MV linac beam, the dose rate dependence of the OSL detectors was also evaluated for dose rates of 200, 400, and 600 cGy/min.

Since a calibration factor is determined for specific calibration conditions to a known dose, in order to be useful clinically the dose response of the detectors should follow a known relationship so that the calibration factor can be applied for a variety of delivered doses. The dose response relationship for the OSL detectors in ^{60}Co was investigated for doses ranging from 0.5 to 4 Gy. The irradiations were carried out in solid water at a depth of 0.5 cm. Each detector was irradiated a single time to a specified dose and the reading was determined. Three separate detectors were used for each dose value.

A series of dosimetry measurements were performed using the OSL detectors in comparison to ion chamber or diode measurements. These measurements evaluated the accuracy of the OSL detectors for routine dosimetry and the impact of changing field parameters on the OSL signal. Percent depth dose curves were determined between 0.0 and 5.0 cm in solid water for a ^{60}Co beam. Three OSL detectors were used at each depth and the detector response was taken as the average of three readings for the same irradiation. The results were compared with a percentage depth dose (PDD) curve measured under the same experimental conditions with a Markus ionization chamber and with a PDD curve measured with a Scanditronix photon field diode in a water tank. The percentage depth dose data were normalized at 0.5 cm depth for all detectors.

The relative output factors were also determined in a ^{60}Co beam for five different field sizes, ranging from 5×5 cm^2 to 22×22 cm^2. The detectors were irradiated in solid water at a depth of 5 cm. Three detectors were used for each field size. The OSL results were compared with relative output factor measurements from an NE2571 ionization chamber.

The response of the OSL detectors for different distances from the source was measured for SSDs from 85 to 105 cm. The field size was 10×10 cm^2 and the detectors were placed at a depth of 5 cm in solid water. Three detectors were used for each SSD. The detector response was calculated as the average of three readings for the same irradiation conditions. The OSL results were compared with ionization chamber measurements carried out at the same experimental conditions.

The ultimate goal of this project is to evaluate whether OSL dots can be used for *in vivo* dosimetry on patients. For such a purpose, buildup caps are recommended for entrance dose measurements.[20] Therefore, as a final part of this study we have evaluated custom made buildup caps of 2 mm thick

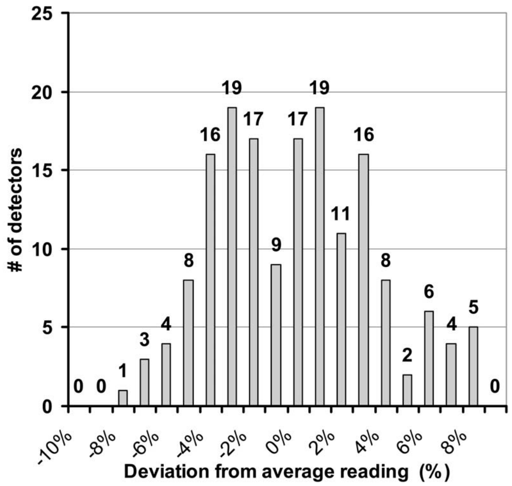

FIG. 2. Stability of OSL system for multiple detectors irradiated under identical conditions. Each detector was irradiated to 50 cGy and its reading was compared to the average reading for all 165 detectors. The figure shows the deviation from the average.

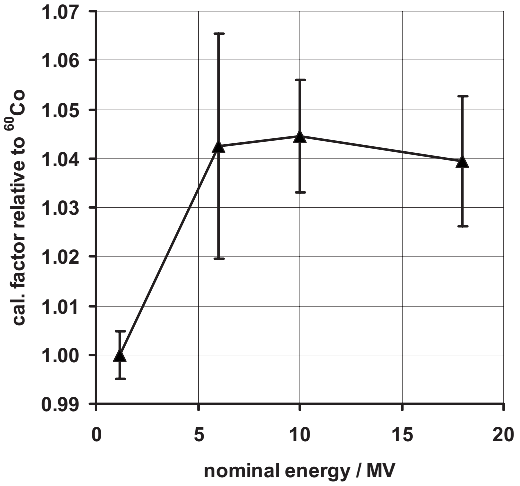

FIG. 3. Energy dependence of the response of OSL detectors. The calibration factors are determined in units of cGy/rdg and are presented here normalized to the value for ^{60}Co. Seven detectors were used at each energy and the figure presents the average over all detectors.

aluminum for cobalt-60 beam and of 2 mm thick stainless steel for 6 MV beam.

III. RESULTS AND DISCUSSION

When subjected to the identical irradiations, the response reproducibility of an individual detector was 2.5% (1sd). The reproducibility of the response of the batch of 165 detectors was 4.2% (1sd). Figure 2 compares the readings from multiple detectors, each subjected to a 50 cGy irradiation in a ^{60}Co under identical conditions. The figure shows the deviation of a given detector reading from the average value for all detectors. All detectors are within 9% of the mean and 87% of the detectors fall within 5% of the mean. This shows good stability of the system and implies that detectors from a given batch might be used with a single calibration factor depending on the level of precision required. Our results show somewhat larger spread than those of Schembri and Heijmen for dose 200 cGy.[19] However, this is consistent with their conclusion that interdetector response variation has larger spread for lower than for higher doses.[19]

The effect of the time lapse between irradiation and readout on the OSL signal was investigated by subjecting five detectors to a single irradiation of 50 cGy and then reading the response repeatedly over a period of 1 h–21 days. Within the first 6 h following irradiation, there is no noticeable change in the OSL signal. Within the first 5 days, there is about a 2% reduction in the OSL signal. Beyond 5 days, the signal is stable, at least up to 21 days post irradiation. This is consistent with the findings of Schembri and Heijmen[19] who measured a reduction in OSL signal of slightly less than 2% over 38 days. Their first reading was not taken until day 17 since the OSL films had to be sent away for reading.

The response of OSL detectors as a function of energy is shown in Fig. 3. For energies above 6 MV, there is no clear dependence on the energy. This is in agreement with the results of Aznar *et al.*[8] who found no energy dependence of $Al_2O_3:C$ detectors in the energy range 6−18 MV. Compared with the calibration factors for cobalt, there is on the order of a 4% increase in the calibration factor for the higher energies (or in other words 4% decrease in sensitivity). This implies that for detectors calibrated in ^{60}Co energy, an energy correction factor should be applied if the detectors are used for dose assessments at higher energies. However, a single calibration factor would be applicable for energies in the range 6−18 MV. Aznar *et al.*[8] and our findings support the manufacturer's claim that energy response for high energy photon and electron beams is within +/−1% for these detectors. Schembri and Heijmen[19] also investigated OSL response for photon beam energies 4–18 MV and electron beam energies 6–22 MeV. They exposed the films to the dose of 200 cGy. They found that the response of their OSL films was within +/−2% of its average value for photon beams,

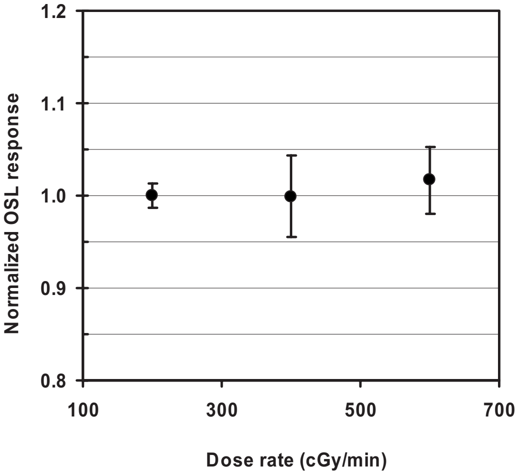

FIG. 4. Dose rate dependence of the response of OSL detectors in a 6 MV linac beam. The response is normalized to the reading at a dose rate of 200 cGy/min. Four detectors were used at each dose rate setting.

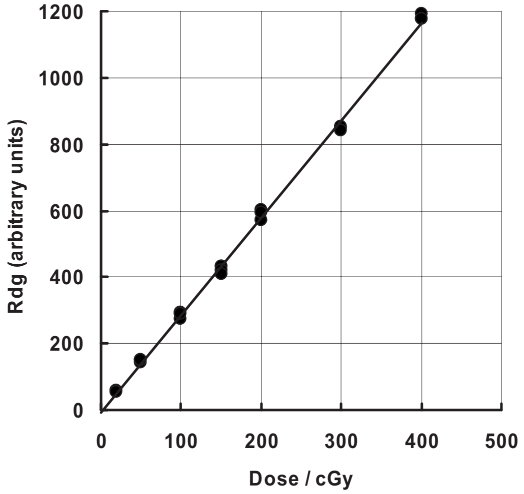

FIG. 5. Response of OSL detectors in arbitrary units vs an accumulated dose in cGy for ^{60}Co irradiations. Three different detectors were used at each dose value. For each detector, the response is the average value of three readings. The solid line shows a linear fit through all of the data points, with an intercept at zero dose of −6.4 ± 6.0.

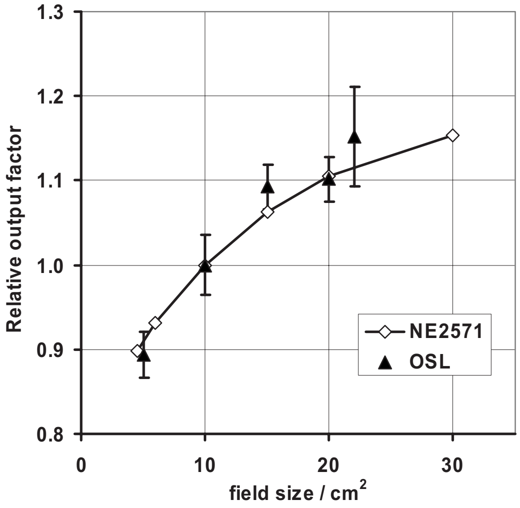

FIG. 7. Relative output factors for a ^{60}Co beam measured with OSL detectors and with an NE2571 ionization chamber. The measurements were made in a solid water phantom at a depth of 5 cm and a SSD of 80 cm.

with some trend of decrease in sensitivity (about 4% sensitivity drop between 6 and 18 MV) for higher beam energy. The sensitivity of the film in electron beams was by 3.7% lower than in photon beams. The difference between Aznar *et al.*[8] and our findings and those of Schembri and Heijmen[19] requires further investigation.

4 shows the OSL response as a function of the dose rate, normalized to the response at the nominal dose rate of 200 cGy/min. Our findings agree with those of Schembri and Heijmen which also show that there is no discernible dose rate effect for Al_2O_3:C OSL dosimeters.

5 shows the OSL response versus dose. At each dose value, there are three data points, representing the three different detectors used for each dose. The solid line in Fig. 6 is a linear fit to all of the data points and has $R^2 = 0.9983$. The standard deviation of the residuals for this fit is 4%. Extended to zero dose, the fit line shows an intercept of −6.4 ± 6.0. This suggests that the system does not show a systematic offset that may impact the readings at low doses. Figure 5 suggests that the OSL response from the Landauer detectors is linear with a dose in the range 0.5–4 Gy, meaning that a single calibration factor is applicable throughout this range of accumulated doses. Our results are in agreement with those of Schembri and Heijmen[19] who also found the OSL film signal to be linear with a dose up to about 400 cGy.

To investigate the feasibility of using this OSL system for dosimetric measurements, ^{60}Co percent depth dose curves were measured using the OSL detectors, a Markus ionization

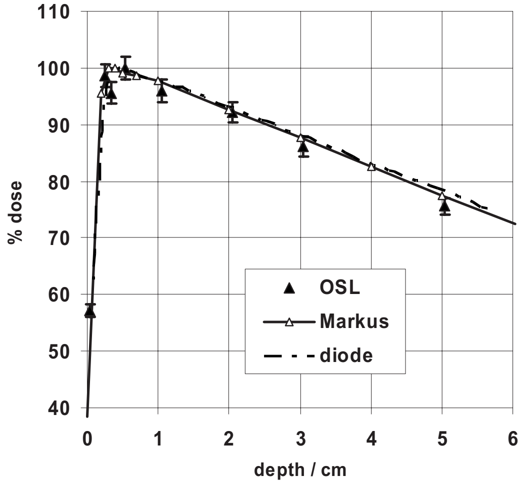

FIG. 6. PDD curves in solid water for 10 × 10 field in a ^{60}Co beam measured with OSL detectors and normalized to 0.5 cm depth. Shown for comparison are PDD curves measured in solid water using a Markus parallel-plate ionization chamber and in a water tank using a photon diode for the same irradiation energy.

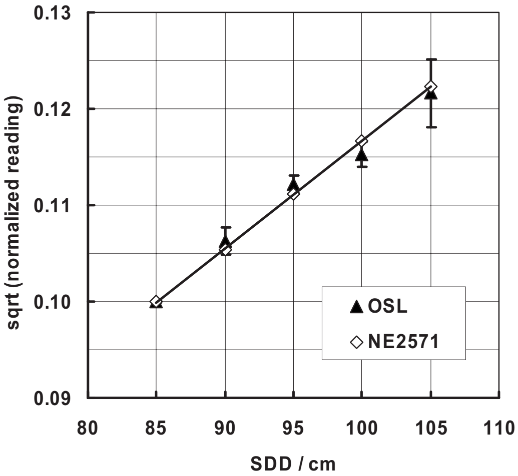

FIG. 8. SSD dependence of OSL signal for ^{60}Co irradiations compared with measurements using an NE2571 ion chamber. The data are presented as the square root of the reading, normalized to 85 cm SSD. The solid line shows a linear fit to the ion chamber measurements.

TABLE I. Dose measured with OSL detectors with buildup caps on the surface of solid water phantom and without buildup caps at d_{max}. Irradiations were carried out for a dose at d_{max} of 0.68 Gy for ^{60}Co and a dose of 0.50 Gy for 6 MV beam.

	^{60}Co dose measured with OSL (Gy) aluminum caps (2 mm thick)	6 MV dose measured with OSL (Gy) stainless steel caps (2 mm thick)
Surface with caps	0.68	0.50
d_{max} without caps	0.68	0.50

chamber, and a photon diode and are shown in Fig. 6. In the case of the OSL detectors the plotted depth takes into account the 0.42 mm thick plastic cover. The OSL curve shows good agreement with the other two curves, particularly in the buildup region and in the region immediately beyond the depth of dose maximum. This indicates that the OSL detectors can be used to accurately measure dosimetric quantities of interest and are suitable for dose measurements near the surface.

The relative output factors as a function of field size are presented in Fig. 7. The measurements with the OSL detectors and the ionization chamber show very good agreement, with a difference of less than 1% for all the field sizes considered. There is no systematic trend in the effect of field size on the OSL signal relative to the effect on the ion chamber response. This indicates that they can be used for measurement of relative output factors instead of ionization chamber. Since these OSL detectors cover a very small area, they can be very valuable for measurements of very small field sizes, such as those used in radiosurgery. Our findings are in agreement with those of Schembri and Heijmen.[19]

The SSD dependence is presented in Fig. 8. The OSL measurements agree with the ion chamber measurements to within 1% for all SSD values measured. The solid line shows a linear fit to the ion chamber data and has an R^2 value of 0.9999.

The buildup caps for OSL detectors, evaluated for ^{60}Co and 6 MV beam irradiations, showed an adequate performance as presented in Table I. Therefore, they can be used as the buildup layer in cases of entrance dose *in vivo* measurements.

IV. CONCLUSIONS

The present study shows that the Landauer InLight™ commercial OSL system is suitable for a variety of dosimetry-related measurements. It shows good agreement with both ion chamber and diode measurements. The independence of OSL response for energies greater than 6 MV, suggests that a single calibration factor could be used for exposures above this energy. The stability of the system and linear dose-response relationship make it a good candidate for *in vivo* dosimetry and the efficiency with which measurements and readings can be made make it an appealing option for dosimetry measurements in a clinical radiotherapy setting.

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