Mrčela I., Bokulić T., Izewska J., Budanec M., Fröbe A., Kusić Z. (2011) ***Optically stimulated luminescence in vivo dosimetry for radiotherapy: physical characterization and clinical measurements in (60)Co beams.*** **Physics in Medicine and Biology, 56 (18). pp. 6065-82. ISSN 0031-9155**

http://iopscience.iop.org/0031-9155

http://dx.doi.org/10.1088/0031-9155/56/18/018

http://medlib.mef.hr/1440

Optically stimulated luminescence in-vivo dosimetry for radiotherapy: physical characterization and clinical measurements in ^{60}Co beams

I Mrčela[1], T Bokulić[1], J Izewska[2], M Budanec[1], A Fröbe[1] and Z Kusić[1]

[1] Department of Oncology and Nuclear Medicine, University Hospital "Sestre milosrdnice", Zagreb, Croatia
[2] Dosimetry Medical Radiation Physics section, International Atomic Energy Agency, Vienna, Austria

E-mail: iva.mrcela@kbcsm.hr

Abstract
A commercial optically stimulated luminescence (OSL) dosimetry system was investigated for in-vivo dosimetry in radiation therapy. Dosimetric characteristics of InLight dot dosimeters and a microStar reader (Landauer Inc.) were tested in ^{60}Co beams.
Reading uncertainty of a single dosimeter was 0.6%. The reproducibility of a set of dosimeters after a single irradiation was 1.6%, while in repeated irradiations of the same dosimeters it was found to be 3.5%. When OSL dosimeters were optically bleached between exposures, the reproducibility of repeated measurements improved to 1.0%. Dosimeters were calibrated for the entrance dose measurements and a full set of correction factors was determined. A pilot patient study that followed the phantom validation testing included more than 100 measured fields with a mean relative difference of the measured entrance dose from the expected dose of 0.8% and the standard deviation of 2.5%. In conclusion, these results demonstrate that OSL dot dosimeters represent a valid alternative to already established in-vivo dosimetry systems.

1. Introduction

In-vivo dosimetry in radiation therapy is a well established and recommended procedure (AAPM 1994, Essers and Mijnheer 1999, IAEA 2008) for the estimation of the dose delivered to a patient during the radiation treatment. It provides an independent check of the treatment procedure that aims at detection of possible errors in calculation, patient setup and data transfer. In vivo dosimetry is recognized as a part of the quality assurance programme in radiotherapy. It became even more important with the emerging use of new and more complex radiotherapy techniques such as intensity modulated (IMRT) or image guided radiation therapy (IGRT). Today, the most common dosimeters for patients' dose verification through in-vivo measurements are semiconductor diodes, thermoluminescent dosimeters (TLDs), or electronic portal imaging devices (EPID). There is also an increased application of metal-oxide-semiconductor field effect transistors (MOSFET) for clinical radiotherapy dosimetry. Diode based in-vivo dosimetry has been widely explored and reported in literature (Millwater *et al* 1998, Alecu *et al* 1999, Jornet *et al* 2000, Fiorino *et al* 2000, Jursinic 2001). Similar investigations have been reported with TLDs (Loncol *et al* 1996, Swinnen *et al* 2004) and with MOSFET dosimeters (Ramaseshan *et al* 2004, Scarantino *et al* 2005, Cherpak *et al* 2009). Comprehensive guidelines for the implementation of in-vivo dosimetry with diodes are published by ESTRO (2001) and AAPM (2005). Generally, in vivo dosimeters are available as passive detectors with delayed readout or as real time detectors with immediate readout. Although real time techniques have obvious advantage over passive dosimeters, latter are still widely used in routine work for several reasons. Usually there is larger number of dosimeters provided for in-vivo measurements which allows for

simultaneous dose determination at several positions on the patient. No cable connection between dosimeters and the reader, allows their usage on different treatment machines in the department. Moreover, it is not necessary for the hospital to be equipped with the reading unit; instead readout can be arranged in an outside institution.

Optically stimulated luminescence dosimeters (OSLDs) are well known as suitable dosimeters for various dosimetric applications from retrospective dosimetry i.e. luminescence dating, to personal, environmental and space dosimetry (McKeever 2001). Until recently, TLDs were the most common choice of passive dosimeters for medical purposes. However, OSLDs have an advantage compared with TLDs (McKeever and Moscovitch 2003), since no heating is required for the dose measurement, which makes the reading equipment cheaper, and easier to handle and maintain.

OSLDs potential in medical dosimetry (Akselrod *et al* 2006) was recognized due to their small size suitable for point measurements, high sensitivity, non-destructive readout and reanalysis and the use of relatively simple readers with high degree of automation. Technical advances in production of OSL materials and availability of the reading devices make them a very promising dosimetric tool for medical purposes. Current developments in this field are given in the review by Yukihara and McKeever (2008) and specific applications of OSLDs in diagnostic radiology and radiation therapy are reconsidered in the review by Yukihara *et al* (2010).

The physical principle of OSL is analogous to TL, making use of light instead of heat as a source of stimulation energy (Bøtter Jensen *et al* 2003). When exposed to ionizing radiation, the energy is stored in the crystal lattice of the detector material in a form of the trapped charge. In a stimulated relaxation process, electrons and holes recombine and the luminescence light is emitted. The rate of relaxation is related to the amount of trapped charge and consequently to the luminescence intensity. Thus, an integral of the luminescence intensity over the stimulation period is a function of the initial absorbed dose. Many materials, like carbon doped aluminium oxide Al_2O_3:C, exhibit both TL and OSL, depending on the stimulation. The dosimetric properties of Al_2O_3:C as OSLD in a single crystal or powder form, have been extensively investigated over past years for different applications and types of radiation (Yukihara *et al* 2004, McKeever *et al* 2004,). Preferable characteristics, such as very high sensitivity, all-optical nature of measurement and a possibility of optical bleaching make it now a common material for OSLD. Still, Al_2O_3:C has a relatively high effective atomic number (Z=11.2) compared to water that is responsible for the energy dependence especially in the low energy region (Reft 2009). Andersen *et al* (2008) have shown that OSL response from Al_2O_3:C is dependent on the temperature during both irradiation and stimulation process. Sensitivity changes with accumulated dose were also reported by several authors (Yukihara *et al* 2004, Yukihara and McKeever 2008, Jursinic 2009).

Although several studies have investigated different types of Al_2O_3:C based OSLDs for radiotherapy dosimetry (Gaza *et al* 2004, Yukihara *et al* 2005, Schembri and Heijmen 2007, Miller and Murphy 2007, Jursinic 2007, Viamonte *et al* 2008, Yukihara *et al* 2008, Reft 2009) there are only few published results on actual clinical in-vivo measurements with OSL dosimeters. Aznar *et al* (2004) tested a real time OSL fiber dosimeter for direct measurement of the dose delivered to one patient in IMRT treatment of head and neck. Luxel (Landauer) OSL dosimeters were used by Meeks *et al* (2002) for in-vivo measurements of extra target doses in tomotherapy for nine patients.

In this work we have investigated the performance of a commercial OSLD system based on Al_2O_3:C for entrance dose measurements in external ^{60}Co photon beam radiotherapy and performed in-vivo studies that involved extensive measurements on patients. After physical characterization and validation of the entire procedure on an anthropomorphic phantom, about 100 in-vivo patient measurements were made. Investigation of OSLDs, as possible in-vivo dosimeters in radiotherapy, was a part of the research project supported by the International Atomic Energy Agency (IAEA). The same generic protocol was proposed for the implementation of in-vivo dosimetry in clinical practice with the recently introduced MOSFET dosimeters and OSLDs, as for the already established in-vivo dosimeters such as diodes and TLDs. Characterization and comparison of these solid-state dosimeters are presented in an IAEA report (IAEA 2011).

2. Materials and methods

2.1 Equipment

All irradiations in this study were done with a ^{60}Co unit (Cirus, Cis BioInternational, France). The ionisation chamber used was a Farmer type 0.6 cm^3, model 30002, connected to the Unidos electrometer, both PTW, Freiburg, Germany. Measurements were performed in a white polystyrene phantom (RW3, PTW) with a special adapter for the chamber. Computer treatment planning system (TPS) was Theraplan Plus 1000 (MDS Nordion, Canada).

The OSL system was a commercial InLight^{TM} OSL system (Landauer, Inc. Glenwood, Illinois, USA) that included OSL dot dosimeters, a microStar reader and an external PC with dosimetry software. The dot dosimeter consists of an OSL active element (Al_2O_3:C), 7 mm in diameter and 0.3 mm thick, placed in a 2 mm x 12 mm x 24 mm lightproof plastic housing. The housing opens automatically during the reading process, and it could be opened manually for optical bleaching purposes, i.e. for exposing the OSL element to light (figure 1). Specially made, 2 mm thick, aluminium build-up caps of 9 mm diameter, which fit to the active OSL element, were taped on dots in all irradiations. The microStar reader uses a green light emitting diode (LED) array, with the wavelength centered at 530 nm, for the optical stimulation. The reader operates in a so-called continuous wave (CW) stimulation mode with a reading time of approximately one second. During the stimulation, only a small fraction of the trapped charge is released so that dosimeters can be re-read multiple times.

Figure 1. a) Landauer's OSL dot dosimeters with build-up caps and a plastic adapter for the reader, b) microStar reader.

2.2 OSL reading process

The microStar reader can operate in two modes with different light power stimulation depending on the dose received by the dosimeter. Every dosimeter is first stimulated with a low power light beam and the initial response is monitored. If the response is large enough (high doses) the reading process continues with a low power light beam, otherwise a high power beam is employed to get a signal large enough to measure small doses. These modes of operation are activated automatically. In a typical radiotherapy dose range, used in this study, the reader operates in a low power LED stimulation mode.

During the readout process, the dots are placed in a plastic adapter and then inserted into the reader's loader. The reader is equipped with a rotary knob that is turned into the reading position for opening the housing and exposing the dot to light. Turning back to home position closes the housing and allows withdrawing the loader. This operation needs to be done slowly and gently. Otherwise, dosimeters were often only partially pulled from the adapter or the housing was not fully closed after the reading, which caused inconsistent readouts, and consequently poor reproducibility.

The stability of the reader was tested daily, by the intrinsic system check: the measurement of the dark current values, the readout from the irradiation with a built-in small ^{14}C source, and the readout with the

light beam on, to indicate the stability of the beam intensity. The dark current values were always well below the manufacturer's recommended value of 30 counts. The variation between the results of the other two daily system checks were around 3%, which is inside the tolerance levels stated at ±10%, indicating therefore an acceptable stability of the reader.
For all measurements, dosimeters were read five times, every time opening the loader and repositioning the dot. The average of five repeated readings was taken as a dosimeter signal.

2.3. Basic characterization of the OSL system
Initially, OSLDs were tested for a perturbation of the radiation field beneath the dosimeter, multiple reading depletion correction, optical bleaching techniques, the system stability and reproducibility, and the fading of OSL signal with time. Attenuation of the radiation field beneath the OSLD equipped with a build-up cap was measured by placing a radiographic film between the plastic phantom plates at 5 and 10 cm depths. The optical density of a developed film was determined by a point densitometer (DensiX, PTW, Freiburg) and compared to the profiles measured in a water phantom with the ionization chamber.

When performing multiple readings of the same dosimeter, each stimulation process empties only a fraction of a trapped charge. To estimate the depletion fraction and its possible dependence on the absorbed dose, two dot dosimeters were exposed to two different doses, i.e. one dot to 100 cGy and the other to 400 cGy. Both were read a hundred times in a sequence.

Two approaches for reusing the dots were examined; accumulation of the dose in repeated irradiations and optical bleaching before another exposure. In the first case, the dose was estimated by subtracting a previously measured dose from the current one. Two sets of three dosimeters were taken to compare the reproducibility of the system for these two techniques. All dosimeters were irradiated eight times with 100 cGy.

As mentioned above, OSLDs can be easily bleached by exposing them to the light. Several light sources and different illumination times were used to investigate the bleaching process. Optical bleaching at room temperature with a standard tungsten 100 W light bulb was compared to a 75 W reflector halogen lamp with UV filter (OSRAM, Halopar 30, model 64841 FL). According to the manufacturer the filtering is provided with doped quartz glass that entirely blocks high energy UV radiation and reduces low energy UV to a half of its initial value. Luminous intensity of this lamp was 2200 cd with colour temperature of 2900 K. The illumination distance was about 50 cm to avoid overheating of dosimeters, which could damage the plastic housing. At this distance the maximal temperature at dosimeters level was around 35^{o}C. In further characterization, OSLDs were bleached with a halogen lamp, when needed.

The reproducibility of the OSL system, including both the reader and dosimeters, was evaluated by irradiating a large number (N=206) of dosimeters under identical conditions. Each dosimeter was read five times following a single exposure with 50 cGy. The mean reading R_i and the experimental standard deviation $SD(R_i)$ were calculated. The standard deviation (SD) reported throughout this work is an experimental SD and not the SD of the mean (GUM 2008). The mean value of relative SDs for all dots was stated as the overall reading uncertainty:

$$\text{reading uncertainty} = \frac{1}{n}\sum_{i}^{n}\frac{SD(R_i)}{R_i}\cdot 100\% \tag{1}$$

Here the number of dosimeters in a set is denoted with n. The reproducibility of a system was determined as a relative SD of all mean readings:

$$\text{system reproducibility} = \frac{n}{\sqrt{n-1}}\frac{\sqrt{\sum_{i}^{n}\left(R_i-\frac{1}{n}\sum_{i}^{n}R_i\right)^2}}{\sum_{i}^{n}R_i}\cdot 100\% \tag{2}$$

To investigate a potential change of the system reproducibility with the dose, a different set of 20 dots was irradiated with 100 cGy.

The decrease of the OSL signal with the time after the irradiation was measured with four dosimeters. Readings started as early as 40 s after the irradiation, then after 2, 3, 5 and 6 min, later every 10 minutes for the first hour, and finally every 30 min during three hours. For the evaluation of a long-term fading, the readings were taken every day for the first week and then less frequently. The last measurement was taken 2 months after the irradiation.

The change of the temperature after the detector contact with patient skin was measured using Fluke 561 IR thermometer.

2.4. Dose response and calibration

The dose response of the OSL system was tested in a range of doses from 20 cGy to 1000 cGy. OSL signal was compared to the ionization chamber measurements. A nonlinear response at higher doses (Yukihara et al 2004, Edmund et al 2006, Jursinic 2009) was the reason to study the dose dependence and the system sensitivity in more detail. Supralinearity effect can be corrected by the dose response non-linearity correction factors (IAEA 2011), or it can be incorporated in the calibration itself, which is an option employed here. Individual dots sensitivity factors along with a dose dependent calibration curve for the set of dosimeters were used to convert the OSLDs readings into the dose.

For the determination of the sensitivity factors, OSLDs were irradiated on the top of the phantom with the same 50 cGy dose. The source to surface distance (SSD) of 80 cm and the square field size of 10 x 10 cm^2 were arranged. Sensitivity factors are defined as:

$$s_i = \frac{nR_i}{\sum_i^n R_i} \tag{3}$$

A set of 20 dosimeters was used for the dose response study using the same field arrangement. OSLDs were optically bleached as described above and then irradiated with different doses ranging from 20 to 1000 cGy, defined at the depth of the maximum dose (d_{max} = 0.5 cm). Two OSLDs with build-up caps were placed at the same time on the phantom surface as illustrated in figure 2. The ionization chamber was placed inside the plastic phantom at a reference depth of 5 cm. The chamber readings were later converted to the dose in water at d_{max} employing the percent depth doses (PDDs). As PDDs were previously measured in water and employed in TPS calculations, a plastic to water conversion factor defined as a ratio of ionisation chamber readings in water and in phantom (IAEA 2011) under same irradiation conditions at d_{max}, was determined.

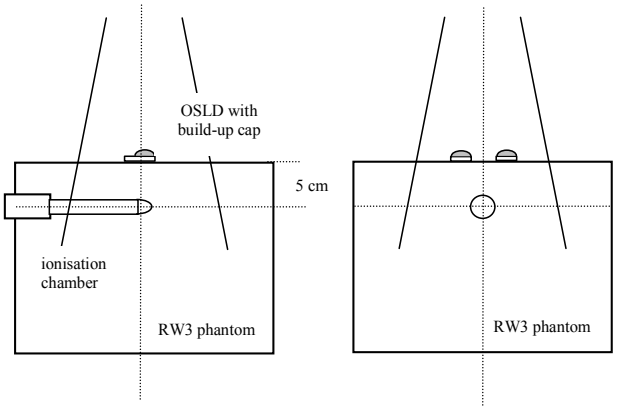

Figure 2. The calibration setup.

The dose response curve that relates the ratio M of OSL reading R and respective sensitivity factor s, to the dose D delivered to the dosimeter set, was determined as a second degree polynomial fit to the measured data:

$$M = a_0 + a_1 D + a_2 D^2 \tag{4}$$

The dose D was determined from the ionisation chamber readings and represents the dose in water at d_{max}. An unknown OSL dose D_{OSL} can then be calculated as a positive solution of the quadratic equation (4).

$$D_{OSL} = \left(-a_1 + \sqrt{a_1^2 - 4a_2 \left(a_0 - M \right)} \right) / 2a_2 \tag{5}$$

2.5. Correction factors for clinical conditions

All correction factors were measured according to the IAEA guidelines (IAEA 2011) with OSLDs that were previously irradiated with 50 cGy. They are generally defined as a ratio of dosimeter response in clinical and reference conditions. The dose of 50 cGy was delivered in all measurements except for wedge factors measurements where 100 cGy was used. For angular correction test a single dosimeter with a build up cap was placed on the phantom surface with an active element at the isocentre. The gantry was rotated by 15° from 0° to 75°, in both directions. The purpose was to simulate clinical setups where, due to the patient anatomy irregularities, is not possible to place the dosimeter perpendicularly to the beam axis. The field size corrections were investigated by varying the square field size from 5 cm × 5 cm to 25 cm × 25 cm. SSD correction factors were measured in the range of 70 cm to 100 cm. Wedge correction factors were determined for a set of nine physical wedges. OSLDs were carefully placed with an active element at the field centre and with a longer axis of the OSLD housing oriented along the unwedged direction. Block and tray factors were measured by placing triangularly shaped blocks on a tray at the field corners. Two square field sizes, 10 cm and 15 cm, with varying block dimensions were considered. The entrance dose D_E for a given treatment arrangement is then determined according to IAEA (2011):

$$D_E = D_{OSL} \cdot \left(\frac{SSD - d_s}{SSD + d_{\max}} \right)^2 \prod_i k_i \tag{6}$$

where the term in the brackets is the distance correction with d_s representing the distance between the patient skin and the active OSL element (d_s = 0.1 cm) and k_i are correction factors relevant to the particular beam arrangement. The uncertainty related to the D_E, was expressed as a combined standard uncertainty $u_c(D_E)$ (GUM 2008). The formula includes variance terms for the OSL reading R, sensitivity factor s, correction factor k_{angle}, fit parameters $a_{0,1,2}$ defined in equations (3-5) and covariance terms between fit parameters.

2.6. Phantom and patient measurements

First dosimeter irradiations aiming at a quick check of a whole measurement chain were performed on the plastic RW3 phantom with a set of relevant setup parameters. Final validation of the procedure was done using an anthropomorphic Rando Alderson phantom. Simulation of usual radiotherapy treatments were made in a pelvic, head and neck and breast region. A percentage difference of the measured entrance dose D_E from the expected dose D_{TPS}, calculated with the TPS, was obtained as:

$$\Delta = \frac{D_E - D_{TPS}}{D_{TPS}} \cdot 100\% \tag{7}$$

Patient measurements were taken within the first three fractions of the radiation treatment. All patients received their treatment at the ^{60}Co unit. OSLDs were taped to the patient skin and precisely positioned at a treatment field centre. Dosimeters were read one day after the irradiation. More than a hundred entrance dose measurements for different treatment set-ups were done. Expected doses varied in a range from 37 cGy to 270 cGy. In this final part of the study, we used dosimeters that were bleached before patient measurements because all available OSLDs had already been irradiated for previous investigations.

3. Results and discussion

3.1 Basic OSLDs characteristics

Attenuation of the radiation field beneath the OSL dot with a build up cap, at 5 cm and 10 cm depth in the RW3 phantom, was found to be 3%. These values are similar to the results reported with diode detectors and suggest that in-vivo measurements do not significantly affect the dose delivered to the patient if performed in a few treatment fractions only (Essers and Mijnheer 1999).

The depletion rate of the dosimeter signal due to repeated readings for dosimeters exposed to 100 cGy and 400 cGy was nearly the same (figure 3). Our results showed that each reading decreases the OSL signal by approximately 0.04% which is much smaller than 0.2% given by the manufacturer. Similar result was reported by Jursinic (2007), who found 0.05% signal depletion per readout. The OSL signal is proportional to the concentration of the filled traps at the time of stimulation and to the light beam power. Since the microStar reader uses a low beam power in the high dose region, this might be a reason why the measured depletion rate was much smaller then that by the manufacturer. The depletion correction can be neglected when making repeated readings of the same dot, in dosimetric conditions typical for radiotherapy.

Optical bleaching at the room temperature with an ordinary halogen lamp is a convenient way to erase previous doses given to OSLD. It turned out that illumination of about four hours was sufficient to decrease the signal to the level below 0.1% of the initial reading. Although it is known (Yukihara *et al* 2004) that OSL materials cannot be entirely cleared of the radiation-induced effects without heating, due to the filled deep traps, the residual dose can be neglected or subtracted when measuring relatively high doses as in radiotherapy applications.

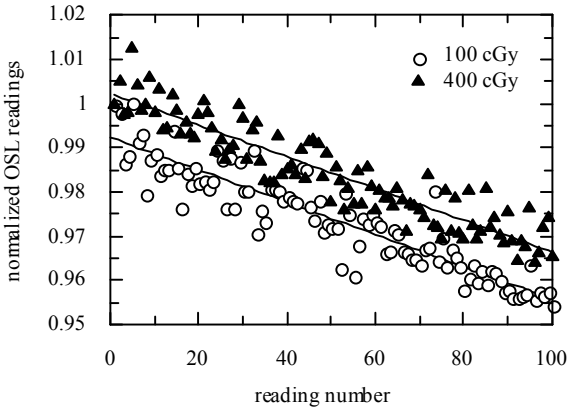

Figure 3. Depletion of an OSL signal in repeated readings for two dosimeters exposed to 100 cGy and 400 cGy. Data are normalized to the first reading. The solid lines represent a linear fit to the measured data.

Single irradiation of a large number of OSLDs, followed by multiple readings, gave the information on the system reproducibility. The uncertainty of the dosimeter reading was (0.6 ± 0.3)%. This value is

comparable to other published results. For a similar dosimetry system, Jursinic (2007) also reported 0.6% reading uncertainty while Viamonte *et al* (2008) had 1%. Miller and Murphy (2007) obtained 0.5% uncertainty when studying the same OSL material but with the different reading equipment. Our first measurements with the OSL system were far from promising with the reading uncertainty (2.9 ± 0.9)% measured on the set of 70 dosimeters, due to the faster rotation of the microStar reading knob.
The reproducibility of the OSL set consisting of 206 dosimeters was 1.6% for irradiations with 50 cGy and 1.3% for 20 dosimeters irradiated with 100 cGy. Raw reading data without sensitivity correction were used. The result compares favourably with the results by other authors, i.e. 4.2% found by Viamonte *et al* (2008) in a batch of 165 dots, and even 7% obtained by Miller and Murphy (2007). Schembri and Heijmen (2007) who have investigated OSL film dosimeters found an inter film response variation in a range of 1% to 3.2%. Better results have been reported by Yukihara *et al* (2005) in a reproducibility test with ten packages, each containing five OSL dosimeters, irradiated in a 6MV photon beam. They found that the maximum difference between the mean package readings from the overall mean was 0.7%.
In-phantom reproducibility for a small group of 8 dosimeters was 1.8%, which implies that attaching and positioning a build-up cap to each dot, did not contribute to the uncertainty in measurements significantly.

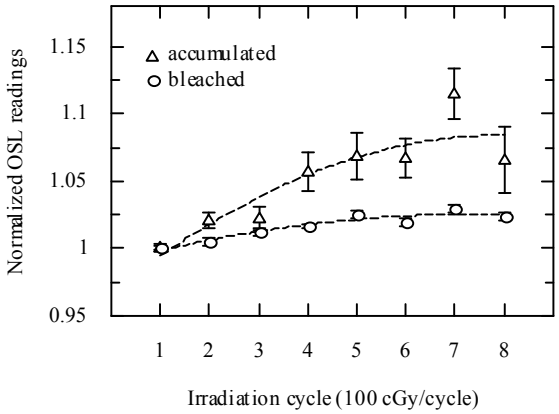

Figure 4. OSLDs reproducibility in repeated measurements. Triangular symbols represent the group of dosimeters that accumulated the dose at each exposure, while circles denote the group that was bleached before next irradiation. Each data point represents the weighted mean of normalized readings (individual sensitivities included) for three dosimeters that were irradiated in one fraction. Error bars represent the weighted mean uncertainties. Dashed lines are second degree polynomial fits to the measured data.

Table 1. Summary of results from testing the uncertainty and reproducibility of the OSL dosimetry system.

Test (delivered dose)	*Description*		*Number of dosimeters*	*Value %*
Reading uncertainty (50 cGy)	Average value of relative SDs calculated from five consecutive readings for each dot (one irradiation of each dot)		206	0.6
Reproducibility (50 cGy)	Relative SD of all averaged readings over five consecutive readings for each dot		206	1.6
Reproducibility (100 cGy)			20	1.3
Reproducibility of a single dosimeter in multiple irradiations (8 x 100 cGy)	Relative SD of weighted averages for a group in an irradiation fraction	accumulated	3	3.5
		bleached	3	1.0

The response of two groups of dosimeters to a series of repeated irradiations is shown in figure 4. One OSLD group was bleached after each subsequent irradiation and the other one was allowed to accumulate

the dose without bleaching. The total delivered dose for both groups was 800 cGy in eight fractions. OSLDs that were bleached between exposures showed a small dose response increase about 2.5%, whereas the signal from dosimeters that accumulated the dose demonstrated significant over response that depended on the previous irradiation history (figure 4). This confirms the results presented by Jursinic (2009) who reported increased supralinearity effect with accumulated dose. The effect is small, but still noticeable also in bleached dosimeters, which can be explained through complex competing processes between the deep hole and electron traps. This test demonstrates that dot dosimeters can be conveniently reused a number of times with optical bleaching between the irradiation sessions, but taking care of the sensitivity changes with repeated irradiations. Results of OSLD response tests are summarized in table 1.

Fading investigation proved OSLDs strong over-response if read shortly (< 10 minutes) after the irradiation (figure 5), which is consistent with the results published by Jursinic (2007). Measured data were normalized to the mean of three readings taken one hour after the irradiation. Since the dosimeters were read 50 times in total, the depletion correction for repeated measurements was applied. There is an obvious transient signal that decays with time and fades out after a few minutes. To extract the parameters of this decaying curve, an exponential plus linear function was fitted to the depletion corrected data:

$$y(t) = A + Bt + C\exp\left(-\frac{\ln 2 \cdot t}{T_{1/2}}\right) \tag{8}$$

Here *y(t)* stands for the OSLD signal at time *t* after the irradiation, and *A, B, C* and $T_{1/2}$ are fit parameters.

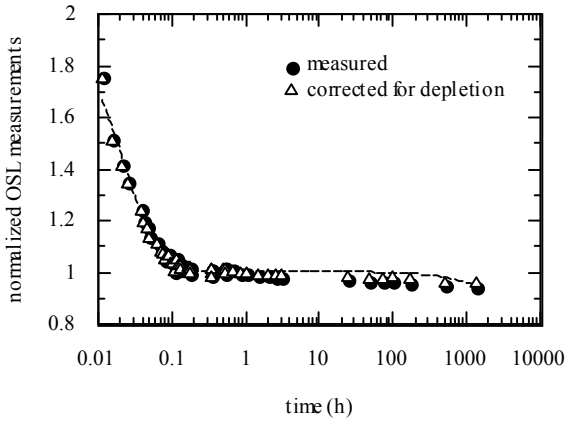

Figure 5. Fading of the OSL signal after the irradiation. Measured data are presented with circles and depletion corrected with triangles. Early measurements include results from four different dots, while data points after 1h are an average of the four dosimeters. A decay curve (8) fitted to the corrected data is denoted as the dashed line.

The half-life time $T_{1/2}$ was (1.03 ± 0.05) minutes consistent with the published results of Jursinic (2007) of 0.8 to 1.4 minutes for different sets of detectors. The stated uncertainty here is a standard uncertainty of the $T_{1/2}$ fit parameter. Stability of the OSL signal should be reached after ten half-lives or about 10 minutes. A small decrease of the signal can also be noticed for data points obtained from readings taken with a larger delay. For example, readings performed after 58 days were, on average, 4% lower than readings taken one hour after the irradiation. The similar results were found by Yukihara *et al* (2010). They address this issue to be a reader related and connected with a very short stimulation time. Over response immediately after the irradiation is explained by the charge leakage from shallow dosimetric

traps (Jursinic 2007).
When compared to diodes or MOSFETs (Ciocca *et al* 2006) with immediate readout or new real-time techniques such as fibre optic based OSL dosimeters (Aznar *et al* 2004, Polf *et al* 2003), dot dosimeters fall short. On the other hand, their advantages are simple handling and a permanent dose record that can be re-evaluated if needed, taking care of the decrease in OSL signal caused by the long term fading effect. This can be easily overcome by employing the fading correction factor.

Measurement of the temperature change for the dot OSLDs taped to the patient skin showed that after typical set up and irradiation time of 5-10 minutes the temperature increase was less than 2°. When dosimeters were placed at the patient table or at the thermoplastic mask used for patient immobilization, the temperature change was negligible. According to the work of Andersen *et al* (2008) temperature coefficients for OSL response are around 0.2%/K. Based on these results, temperature correction due to the contact of dosimeter with patient skin is not required for the clinical in-vivo dosimetry with this type of OSLDs, since it is expected to be well below 1%.

3.2. Calibration and correction factors

For the system calibration, individual dosimeter sensitivities (figure 6) were determined according to equation (3). Sensitivity factors were inside ±5%, with the largest one at 4.1%. Relative standard uncertainties of individual factors were in the range from 0.1% to 2.3%, while spread of values among all sensitivity factors, stated as a relative standard deviation, was 1.6%.

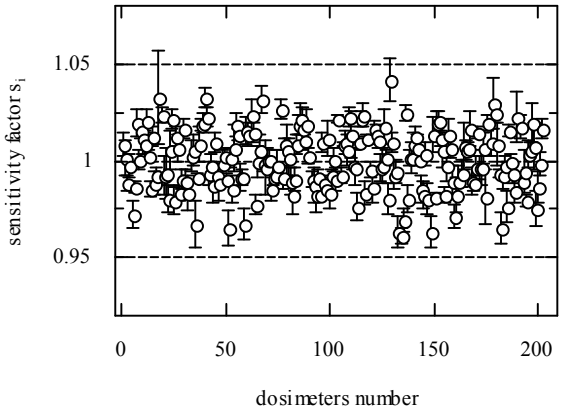

Figure 6. Distribution of sensitivity factors for 206 dosimeters irradiated with 50 cGy. Sensitivity factors were expressed as a ratio of the mean dosimeter reading over the average of all dosimeter's readings in a set. Error bars represent standard uncertainties of individual sensitivity factors.

The OSLD dose response in the range of doses from 20 cGy to 1000 cGy is shown in figure 7. A supralinearity at higher doses was observed and a second degree polynomial was used to model the experimental results. Supralinear response is in agreement with similar studies reported by Schembri and Heijman (2007), Jursinic (2007, 2009) and Reft (2008). They observed a linear response up to around 200 cGy followed by an increase in sensitivity. Many authors have explored OSL dose response of Al_2O_3:C samples experimentally and with numerical analysis of the theoretical energy level models. Yukihara *et al* (2004) investigated a wider dose range and reported linear behaviour for low doses followed by supralinearity and decrease to plateau for very high doses. Quadratic dose dependence was reported by Chen and Leung (2000) as a result of numerical simulation for a simple model with a single trapping state and only one type of a recombination centre, assuming short stimulation time of 1s. In a later work by Chen *et al* (2006) a more complex model with two trapping centres and two types of recombination centres showed that dose response of integral OSL is a nonmonotonic peak shaped curve with plateau at high doses similar to the results reported by Yukihara *et al* 2004. Phenomenological description of the

OSL dose dependence for Al_2O_3 is rather complex but it is generally explained by a competition between deep electron and hole traps during the irradiation or stimulation process, which enhances the OSL signal at higher doses. It is also known (Yukihara and McKeever 2008) that linearity of the dose response varies for different types of OSLDs and readers and also depends on the irradiation history of the detector.
The quadratic dose response curve in figure 7 was used for calibration according to (4) and (5). Parameters were extracted from the fit: a_0 = (-3.7 ± 1.9), a_1 = (1.150 ± 0.009) cGy^{-1} and a_2 = (6.5 ± 0.8)x10^{-5} cGy^{-2}. It turned out that this calibration procedure was quite practical because ionisation chamber measurements were required only for a small set of OSLDs as opposed to an individual calibration where every dot has to be calibrated against the dose determined from the chamber readings. In addition, calibration based on the quadratic relation (4) is valid for the whole range of measured doses and for the whole set, or a batch of dosimeters. Each dosimeter is related to the set through its sensitivity factor. That is particularly useful in the determination of correction factors. They are generally assessed with smaller groups of dosimeters, and then assumed to be valid for the whole set. On the other hand, in the individual calibration, the relation between single dosimeter and a set is not determined.

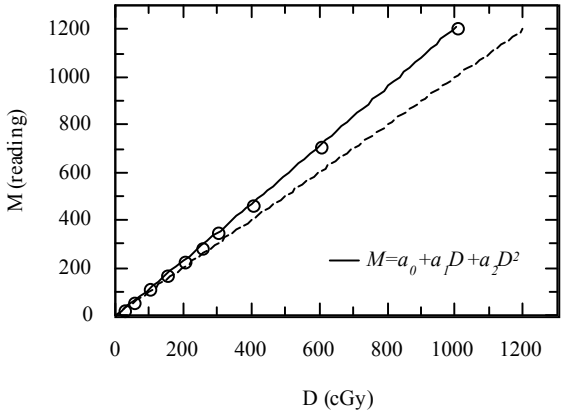

Figure 7. OSLD dose response. The ratio *M=R/s* was averaged over two dots used per each dose. Solid line represents a quadratic polynomial fit and the dashed line represents an ideal linear dose response.

Figure 8 shows angular correction factors averaged for the same angles in both directions of the gantry rotation. The response of the dosimeters increased with the gantry angle, resulting in a correction factors up to (0,904 ± 0,009) at 75°. Although several authors (Aznar et.al. 2004, Jursinic 2007) have reported that an in-phantom OSL response for different OSL dosimeters was independent of the beam incidence angle, the situation is fairly different in the presence of build-up caps. Aluminium build-up caps used in this study were not hemispherical but flattened top, causing different beam path lengths depending on the incidence angle. A hemispherical build-up caps needs to be thicker to appropriately cover the whole detector area, so although it would provide more uniform angular response, additional attenuation on the beam axis would become too large (IAEA 2011). It needs to be pointed out that angular correction is also related to the setup which mimics the clinical situation and it is not an intrinsic characteristic of the dosimeter itself.

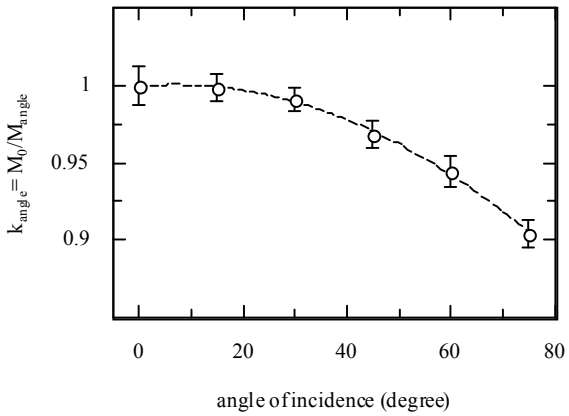

Figure 8. Correction factors for angle of beam incidence to OSLDs with build-up caps.

Variation of the field size and the SSD showed no significant change, less than 1%, in the OSL response. These results are in agreement with Yukihara *et al* (2008) and Viamonte *et al* (2007) who reported no change in the OSL response when varying the SSD. However, Schembri and Heijmen (2007) found deviations of the overall mean response of OSL films within 2.5% when comparing different field sizes, at various depths in the phantom. No dependence of the OSL response with the dose rate was also found by other investigators (Jursinic 2007 and Schembri and Heijmen 2007). Accessories correction factors, including wedge, and block and tray factors were also within 1%. The only exception was the largest wedge (60^o/10 cm width in a wedged direction) with the factor of (1.022 ± 0.009). It is a common practice to apply only correction factors that are larger than 1%, because smaller values fall within the measurement uncertainty. Therefore, all correction factors, except the ones for the different angle of incidence (k_{angle}), were neglected in later in-vivo entrance dose measurements. Appropriate k_{angle} was estimated by linear interpolation.

3.3. Phantom and patient measurements

Relative combined standard uncertainties in the estimation of the entrance dose D_E, defined as $u_c(D_E)/D_E \times 100\%$, were on average 1.7% in phantom, and 1.0% in patient measurements. Possible explanation of a considerable difference in uncertainty values lays in the fact that dosimeters used in patient measurements were optically bleached, while dots used in the phantom study were not. Uncertainties connected with the ionization chamber measurements and treatment planning system calculations were not included in the uncertainty budget.

Table 2. Rando Alderson phantom OSL study results.

Anatomical site	*N (number of measurements)*	*Mean Δ (%)*	*SD (%)*
All	32	0.1	1.4
Head & neck	18	0.6	1.3
Pelvis	10	-0.4	0.9
Breast	4	-1.6	1.5

Initial testing of the complete procedure performed on RW3 phantom for eight field arrangements with different irradiation parameters gave the mean relative difference Δ of the measured from the expected dose and respective experimental standard deviation of (-0.8 ± 1.2)%. Verification of the dosimetric procedure was concluded with Rando Alderson phantom measurements for anatomical sites that are typically treated in the department. Results summarized in table 2 confirmed the suitability of this

OSL dosimetric system for in-vivo dosimetry. Differences from the expected dose were generally below ±3%, except for the one case of two opposed fields where it was 3.8%. This could be attributed to the rigidity of the dots plastic housing that made it difficult to attach properly the dosimeter to the phantom, especially in the neck region.

The patient study comprised 103 entrance dose measurements. Data analysis was done according to the treatment site (table 3). Head and neck region treatments were arranged with two laterally opposed isocentric fields, with or without a thermoplastic mask. A box technique with four fields for gynaecology patients and three fields for rectal carcinoma patients were included in the pelvis site group. Conventional tangential fields, most often with wedges, were used for breast treatments. The spread of the results was the same in these three groups, with SD = 2.4%. In abdomen/thorax group most of the treatments were done with only one field in a fixed SSD technique. All other patient measurements were sorted to the "rest" group. The explanation for a few relatively high values of Δ, although lesser than tolerance levels, measured in the pelvis or thorax region could be attributed to a wrong estimation of the beam incidence angle. Since k_{angle} factors are quite large, the angles need to be determined very precisely, which might be a problem in some situations.

In a routine in-vivo dosimetry with diode dosimeters (Scanditronix EDE-5) that was previously established in our department, more than 700 treatment field measurements were made at the same cobalt treatment machine. Based on that experience the tolerance levels for Δ were set at ±5%. In the OSL in-vivo measurements only 6% of all measured fields exceeded the tolerance level, which is a similar result compared to the 5%, result that we had with diodes. Differences Δ between 5 and 10% were observed in 4 measured fields, while in 2 cases Δ was larger than 10%. These two large differences were discovered in one patient treated with tangential breast fields. Later inspection of a treatment plan showed the wrong fractionation of the total prescribed dose, which resulted in a lower dose per fraction. The results of repeated measurements following the correction of the treatment plan were satisfactory. Other causes of differences exceeding tolerance levels can be attributed to a bad positioning of the patient or an incorrect SSD.

Table 3. Patient data analysis according to the treatment anatomical site: the total number of measured fields, the mean difference between the measured and the expected dose Δ, SD of Δ, a number of measurements with Δ between ± (5-10) %, and a number of measurements with Δ larger than ±10%. Numbers in the brackets refer to the values without two major detected discrepancies.

Anatomical site	*N*	*Mean Δ (%)*	*SD (%)*	*5% <Δ <10%*	*Δ >10%*
All	103 (101)	0.4 (0.8)	3.7 (2.5)	4	2
Head & neck	17	1.1	2.4	1	0
Pelvis	34	2.2	2.4	2	0
Breast	21 (19)	-2.1 (-0.3)	6.0 (2.4)	0	2
Abdomen & thorax	21	-0.6	1.5	0	0
Rest	10	0.3	2.7	1	0

Differences from the expected entrance doses Δ, for all OSL in-vivo measurements are presented as a frequency distribution in figure 9. When all data were included, the mean Δ was (0.4 ± 3.7) %, whereas with the exclusion of two major detected differences it changed to (0.8 ± 2.5) %. Since there is a lack of published patient measurements with OSL as passive dosimeters, clinical results can be compared to other in-vivo studies with different dosimeters. We had got very similar results from 727 diode patient measurements; the mean relative difference of the measured entrance dose was 0.5% with the standard deviation of 3.2% (IAEA 2011). Fiorino *et al* (2000) in a large study of 2700 patients reported the mean difference of 0.4% (3% SD) with diodes. Vordeckers *et al* (1998) described the implementation of diode in-vivo dosimetry in a small department. They found -1.5% mean difference with 4% SD in 650 measured doses. Similar results were also obtained in several other diode studies (Millwater *et al* 1998, Jursinic

2001, Shakeshaft *et al* 1999). Patient study with TLDs was reported by Loncol *et al* (1996) where the mean difference of 1.3% and 4.1% SD were obtained.

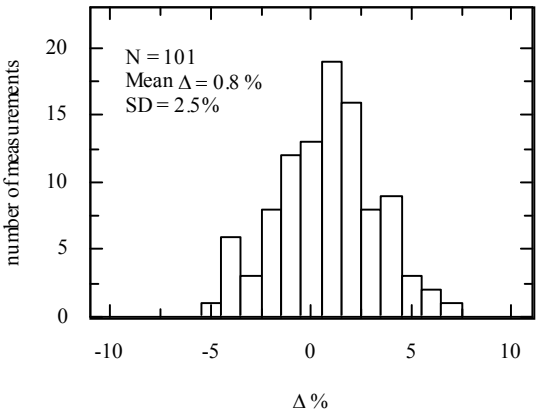

Figure 9. Frequency distribution of percentage differences between the measured and expected dose for in-vivo patient measurements.

4. Conclusion

The results presented in this study show that OSL dot dosimeters are suitable for in-vivo dosimetry in radiotherapy. They demonstrated sufficient accuracy and acceptable reproducibility in measuring the entrance doses in ^{60}Co photon beams. The OSL response was found to be independent of various radiation parameters relevant for clinical practice. Significant angular correction is not inherent to the dosimeters, but it is related to the setup parameters and also to the use and design of the build-up caps. Investigated dosimetric system with sensitive, small and lightweight dot OSLDs along with portable microStar reader, proves to be convenient and easy to use in hospital environment. These characteristics make it appropriate for clinical in-vivo applications with the potential to substitute or complement TLDs or diodes, especially with financial considerations taken into account. On the other hand, issues like dependence of the dosimeters response with accumulated dose and a small long term fading effect require a careful consideration in practical application. More extensive investigations that will include different radiation qualities that are necessary for the in-vivo dosimetry in photon beams other then ^{60}Co, with larger number of patient measurements, would certainly contribute to the enhanced utilization of OSLD in radiotherapy.

Acknowledgments

This research project was supported by the International Atomic Energy Agency (IAEA Research Contract 13115 RBF) within the CRP E2.40.14 project “Development of procedures for in-vivo dosimetry in radiotherapy”. The authors would like to thank Landauer Inc. for providing the OSL Microstar reader and supplying the dosimeters. We are grateful to Dr. Cliff Yahnke for his practical suggestions to this work and to Dr. Eduardo Yukihara for his valuable comments during the project implementation.

References

AAPM 1994 American Association of Physicists in Medicine, Task Group 40 of the Radiation Therapy Committee Comprehensive QA for Radiation Oncology *(AAPM Report No. 46) Med. Phys.* **21** 581-618

AAPM 2005 American Association of Physicists in Medicine, Task Group 62 of the Radiation Therapy Committee *Diode In Vivo dosimetry for Patients Receiving External Beam Radiation Therapy (AAPM Report No. 87)* (Madison WI, Medical Physics Publishing)

Akselrod M S, Bøtter Jensen L and McKeever S W S 2006 Optically stimulated luminescence and its use in medical dosimetry Radiat. Meas. **41** S78-S99

Alecu R, Loomis T, Alecu J and Ochran T 1999 Guidelines on the implementation of diode in-vivo dosimetry programs for photon and electron external beam therapy *Med. Dosim.* **24** 5-12

Optically stimulated luminescence in-vivo dosimetry for radiotherapy

Andersen C E, Edmund J M, Damkjær S M S and Greilich S 2008 Temperature coefficients for in vivo RL and OSL dosimetry using Al2O3:C *Radiat. Meas.* **43** 948-953

Aznar M C, Andersen C E, Bøtter Jensen L, Bäck S A J, Mattsson S, Kjær-Kristoffersen F and Medin J 2004 Real-time optical-fibre luminescence dosimetry for radiotherapy: physical characteristics and applications in photon beams *Phys. Med. Biol.* **49** 1655-1669

Bøtter Jensen L G, McKeever S W S and Wintle A G 2003 *Optically stimulated luminescence dosimetry* (Elsevier, Amsterdam)

Chen R and Leung P R 2000 Nonlinear dose dependence and dose rate dependence of optically stimulated luminescence and thermoluminescence *Radiat. Meas*. **33** 475-481

Chen R, Pagonis V and Lawless J.L. 2006 The nonmonotonic dose dependence of optically stimulated luminescence in Al_2O_3:C: Analytical and numerical simulation results *J. Appl. Phys*. **99** 033511 1-6

Cherpak A, Ding W, Hallil A, Cygler J E 2009 Evaluation of a novel 4D in vivo dosimetry system *Med. Phys*. **36** 1672-1679

Ciocca M et al 2006 Real time in vivo dosimetry using micro-MOSFET detectors during intraoperative electron beam radiation therapy in early-stage breast cancer *Radiother. Oncol.* **78** 213-216

Edmund J M, Andersen C E, Marckmann C J, Aznar M C, Akselrod M S and Bøtter Jensen L 2006 CW-OSL protocols using optical fibre Al_2O_3:C dosemeters *Rad. Prot. Dosim.* **119** 368-374

Essers M and Mijnheer B J 1999 In vivo dosimetry during external photon beam radiotherapy *Int. J. Radiation Oncology Biol. Phys.* **43** 245-259

Fiorino C *et al.* 2000 Quality assurance by systematic in vivo dosimetry: results on large cohort of patients *Radiother. Oncol.* **56** 85-95

Gaza R, McKeever S W S, Akselrod M S, Akselrod A, Underwood T, Yoder C, Andersen C E, Aznar M C, Marckmann C J and Bøtter Jensen L 2004 A fiber-dosimetry method based on OSL from Al_2O_3:C for radiotherapy applications *Radiat. Meas.* **38** 809-812

GUM 2008 *Evaluation of measurement uncertainty – Guide to the expression of uncertainties in measurement*, *GUM 1995 with minor corrections* Joint Committee for Guides in Metrology

Huyskens D, Bogaerts R,Verstraete J, Lööf M, Nystrom H, Fiorino C, Broggi S, Jornet N, Ribas M and Thwaites D I *Practical Guidance for the Implementation of In Vivo Dosimetry with Diodes in External Radiotherapy with Photon Beams (Entrance Dose)* (*Physics for Clinical Radiotherapy ESTRO Booklet No.5)* (Brussels: European Society for Therapeutic Radiology and Oncology (ESTRO))

IAEA 2008 *Setting up a Radiotherapy Programme: Clinical, Medical Physics, Radiation Protection and Safety Aspects* (Vienna: International Atomic Energy Agency)

IAEA 2011 *Development of procedures for in vivo dosimetry in radiotherapy* (Vienna: International Atomic Energy Agency), at press

Jornet N, Ribas M and Eudaldo T 2000 In vivo dosimetry: Intercomparison between p-type based and n-type based diodes for the 16-25 MV energy range *Med. Phys.* **27** 1287-1293

Jursinic P A 2001 Implementation of an in vivo diode dosimetry program and changes in diode characteristics over a 4-year clinical history *Med. Phys.* **28** 1718-1726

Jursinic P A 2007 Characterization of optically stimulated luminescent dosimeters, OSLDs, for clinical dosimetric measurements *Med. Phys.* **34** 4594-4604

Jursinic P A 2009 Changes in optically stimulated luminescent dosimeter (OSLD) dosimetric characteristics with accumulated dose *Med. Phys.* **37** 132-140

Loncol T, Greffe J L, Vynckier S and Scalliet P 1996 Entrance and exit dose measurements with semiconductors and thermoluminescent dosemeters: a comparison of methods and in vivo results *Radiother. Oncol.* **41** 179-187

McKeever S W S 2001 Optically stimulated luminescence dosimetry *Nucl. Instrum. Methods Phys. Res. B* **184** 29-54

McKeever S W S and Moscovitch M 2003 On the advantages and disadvantages of optically stimulated luminescence dosimetry and thermoluminescence dosimetry *Radiat. Prot. Dosim.* **104** 263-270

McKeever S W S, Blair M W, Bulur E, Gaza R, Gaza R, Kalchgruber R, Klein D M and Yukihara E G 2004 recent advances in dosimetry using the optically stimulated luminescence of Al_2O_3:C *Rad. Prot. Dosim.* **109** 269-276

Meeks S L, Paulino A C, Pennington E C, Simon J H, Skwarchuk M W and Buatti J M 2002 In vivo determination of extra target doses received from serial tomotheraphy *Radiother. Oncol.* **63** 217-222

Miller S D and Murphy M K 2007 Technical performance of the Luxel Al_2O_3:C optically stimulated luminescence dosemeter element at radiation oncology and nuclear accident dose levels *Radiat. Prot. Dosim.* **123** 435-442

Millwater C J, MacLeod A S and Thwaites D I 1998 In vivo semiconductor dosimetry as part of routine quality assurance *Br. J. Radiol*. **71** 661-668

Polf J C, Yukihara E G, Alselrod M S, McKeever S W S 2003 Real time luminescence from Al_2O_3 fiber dosimeters *Radiat. Meas*. **38** 227-240

Ramaseshan R, Kohli K S, Zhang T J, Lam T, Norlinger B, Hallil A and Islam M 2004 Performance characteristics of a micro MOSFET as an in vivo dosimeter in radiation therapy *Phys. Med. Biol.* **49** 4031-4048

Reft C S 2009 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.* **36** 1690-1699

Scarantino C W, Rini C J, Aquino M, Carrea T B, Orintz R D, Anscher M S and Black R D 2005 Initial clinical results of an in vivo dosimeter during external beam radiation therapy *Int. J. Radiat. Oncol. Biol. Phys.* **62** 606-613

Schembri V and Heijmen B J M 2007 Optically stimulated luminescence (OSL) of carbon-doped aluminium oxide (Al_2O_3:C) for film dosimetry in radiotherapy *Med. Phys.* **34** 2113-2118

Shakeshaft J T, Morgan H M and Simpson P D 1999 In-vivo dosimetry using diodes as a quality control tool-experience of 2 years and 2000 patients *Brit. J. Radiol* **72** 891-895

Swinnen A, Verstraete J and Huyskens D P 2004 Feasibility study of entrance in vivo dosimetry dose measurements with mailed thermoluminescence dosimeters *Radiother. Oncol.* **73** 89-96

Viamonte A, da Rosa L A R, Buckley L A, Cherpak A and Cygler J E 2008 Radiotherapy dosimetry using a commercial OSL system *Med. Phys.* **35** 1261-1266

Vordeckers M, Goosens H, Rutten J and Van den Bogaert W 1998 The implementation of in vivo dosimetry in a small radiotherapy department. *Radiother. Oncol.* **47** 45-48

Yukihara E G and McKeever S W S 2008 Optically stimulated luminescence (OSL) dosimetry in medicine *Phys. Med. Biol.* **53** R351-R379

Yukihara E G, Gasparian P B R, Sawakuchi G O, Ruan C, Ahmad S, Kalavagunta C, Clouse W J, Sahoo N and Titt U 2010 Medical applications of optically stimulated luminescence dosimeters (OSLDs) *Rad. Measur.* **45** 658-662

Yukihara E G, Whitley V H, McKeever S W S Akselrod A E and Akselrod M S 2004 Effect of high-dose irradiation on the optically stimulated luminescence of Al_2O_3:C *Radiat. Meas*. **38** 317-330

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

Yukihara E, Mardirossian G, Mirzasadeghi M, Guduru S and Ahmad S 2008 Evaluation of Al_2O_3:C optically stimulated luminescence (OSL) dosimeters for passive dosimetry of high energy photon and electron beams in radiotherapy *Med. Phys.* **35** 260-269