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Characteristics of an OSLD in the diagnostic energy range

Rani M. Al-Senan[a)] and Mustapha R. Hatab
Department of Radiology, University of Texas Health Science Center at San Antonio, San Antonio, Texas 78229

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Purpose: Optically stimulated luminescence (OSL) dosimetry has been recently introduced in radiation therapy as a potential alternative to the thermoluminescent dosimeter (TLD) system. The aim of this study was to investigate the feasibility of using OSL point dosimeters in the energy range used in diagnostic imaging.
Methods: NanoDot OSL dosimeters (OSLDs) were used in this study, which started with testing the homogeneity of a new packet of nanoDots. Reproducibility and the effect of optical treatment (bleaching) were then examined, followed by an investigation of the effect of accumulated dose on the OSLD indicated doses. OSLD linearity, angular dependence, and energy dependence were also studied. Furthermore, comparison with LiF:Mg,Ti TLD chips using standard CT dose phantoms at 80 and 120 kVp settings was performed.
Results: Batch homogeneity showed a coefficient of variation of <5%. Single-irradiation measurements with bleaching after each OSL readout was found to be associated with a 3.3% reproducibility (one standard deviation measured with a 8 mGy test dose), and no systematic change in OSLDs sensitivity could be noted from measurement to measurement. In contrast, the multiple-irradiation readout without bleaching in between measurements was found to be associated with an uncertainty (using a 6 mGy test dose) that systematically increased with accumulated dose, reaching 42% at 82 mGy. Good linearity was shown by nanoDots under general x-ray, CT, and mammography units with an $R^2 > 0.99$. The angular dependence test showed a drop of approximately 70% in the OSLD response at 90° in mammography (25 kVp). With the general radiography unit, the maximum drop was 40% at 80 kVp and 20% at 120 kVp, and it was only 10% with CT at both 80 and 120 kVp. The energy dependence study showed a range of ion chamber-to-OSLDs ratios between 0.81 and 1.56, at the energies investigated (29–62 keV). A paired *t*-test for comparing the OSLDs and TLDs showed no significant variation ($p > 0.1$).
Conclusions: OSLDs exhibited good batch homogeneity (<5%) and reproducibility (3.3%), as well as a linear response. In addition, they showed no statistically significant difference with TLDs in CT measurements ($p > 0.1$). However, high uncertainty (42%) in the dose estimate was found as a result of relatively high accumulated dose. Furthermore, nanoDots showed high angular dependence (up to 70%) in low kVp techniques. Energy dependence of about 60% was found, and correction factors were suggested for the range of energies investigated. Therefore, if angular and energy dependences are taken into consideration and the uncertainty associated with accumulated dose is avoided, OSLDs (nanoDots) can be suitable for use as point dosimeters in diagnostic settings.
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Key words: OSL dosimetry, x-ray, CT, mammography

I. INTRODUCTION

Optically stimulated luminescence dosimeters (OSLDs) were recently introduced into medical and environmental dosimetry as a potential alternative to thermoluminescent dosimeters (TLDs).[1–5] Optically stimulated luminescence (OSL) dosimetry was first introduced by Huntley *et al*.[6] as a new technique for dating of sediments. Afterward, research shifted to studying the feasibility of using OSL materials in environmental and medical dosimetry.[7–9] Carbon-doped aluminum oxide (Al_2O_3:C), the OSL material being used for both radiation protection and clinical dosimetry measurements, was originally utilized as a TLD due to its high thermoluminescence (TL) output.[10,11] Later work showed that Al_2O_3:C was a more effective dosimeter when read by pulsed optical stimulation.[8,12–14] The theory of OSL phenomenon is well understood and is described thoroughly in various papers.[2,3,5,7]

Although some earlier reports cast doubt over replacing TLDs with OSLDs in clinical dosimetry,[15] the recent adoption of OSL dosimetry by the Radiological Physics Center (RPC) for remote verification of therapy units output, thus ending the era of TLD audit system,[16] is a sign of acceptance of the OSL systems. The employment of these novel dosimeters did not rapidly migrate into the diagnostic arena. This imbalance in utilization has led to a preponderance of studies characterizing OSLDs in radiation therapy, but a sparse corresponding research in the diagnostic energy range. One reason could be that with the exception of mammography and interventional fluoroscopy, the need for *in vivo* dosimetry is not as prevalent as it is in radiation therapy departments. Another reason could be the relatively high effective atomic number of the Al_2O_3 ($Z_{eff} = 11.28$),[17] which causes the material to exhibit an over-response to low energy photons as a result of high contribution from the photoelectric effect.[5,10,18–20]

One of the first reported uses of OSL dosimetry in diagnostic radiology came from Aznar[21] where the dosimeters were used to measure entrance and exit *in vivo* doses in film/screen mammography. Their results indicate that the presence of the OSLDs did not interfere with the diagnostic quality of the mammograms and, more importantly, that the measurements showed good reproducibility (3%) and linearity, but did demonstrate an energy dependence. Peakheart *et al*.[22] evaluated the use of small OSLDs in CT quality assurance with an acrylic CT body phantom. The results they obtained showed good correlation with the ion chamber for 120 and 140 kVp. Yukihara *et al*.[19] employed OSLDs in measuring dose profiles in CT as well as measuring CTDI with them.[20] The authors recognized that at typical tube potentials used for CT, the energy dependence of the OSLDs would confound the measurements and thus had to determine field-specific energy correction factors for their experiments. These examples, even though at opposite ends of the diagnostic energy spectrum, demonstrate the need for proper characterization of the dosimetric properties of OSLDs at radiologic energies.

In this work, we present an evaluation of commercial OSLDs as point dosimeters in the diagnostic energy range. We investigated the feasibility of implementing these detectors to measure patient doses from radiological procedures and under various exposure conditions. The examined characteristics included batch homogeneity, reproducibility and effect of optical treatment (i.e., erasure of remaining signals), linearity response, and angular and energy dependences. Correction factors for a range of beam quality used in diagnostic radiography are presented. In addition, response comparison with TLDs was also studied.

II. MATERIALS AND METHODS

The OSLDs used for this work were aluminum oxide-based (Al_2O_3:C) (nanoDots, Landauer Inc., Glenwood, IL). They consisted of a 0.2 mm thick disk-shaped detector with a diameter of approximately 5 mm encased in a light-tight $10 \times 10 \times 2$ mm^3 plastic carrier which has a mass density of 1.03 g/cm^3 (Ref. 2). Figure 1 demonstrates the structure of a nanoDot.

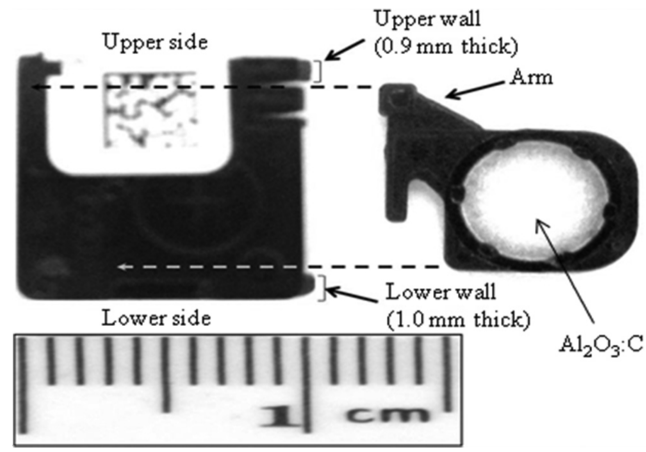

Fig. 1. NanoDot dosimeter and its plastic case ($10 \times 10 \times 2$ mm^3). The attaching arm allows the Al_2O_3:C detector to protrude from its case for readout or bleaching. Upper and lower sides of the plastic are sealed with 0.9 and 1.0 mm thick walls, respectively.

Irradiations were carried out using a general radiography unit (Digital Diagnost, Philips Healthcare, Andover, MA), mammography (Selenia, Hologic, Bedford, MA), and a 64-slice CT scanner (Brilliance, Phillips Healthcare, Andover, MA).

Statistical analysis was performed using GraphPad Prism 5 for Windows (version 5.01, 2007, La Jolla, CA). A p value of less than 0.05 was considered statistically significant.

II.A. OSLDs readout

OSLDs were read using a microStar Reader (Landauer Inc., Glenwood, IL) which has an array of 36 green light emitting diodes (LEDs) as a high intensity stimulating source. In all measurements, reading was carried out between 0.5 and 24 h after irradiation. Each OSLD was read at least twice and only the average reading was utilized for the study. Variations in the repeated readings were found to be in the order of 1%. As part of the microStar Reader's QC procedure, the variations in the reader's sensitivity were checked daily by measuring: background signal (DRK), photomultiplier tube (PMT) counts from the ^{14}C source (CAL), and counts from PMT with the shutter open and the LEDs on to indicate beam intensity (LED). The procedure was repeated three times, and the average counts were recorded to ensure that the variations are within the recommended limits: DRK < 30 and CAL and LED $= \pm 10\%$ of the reader's established average. The reader was calibrated using five preirradiated nanoDots provided by the manufacturer, which had been exposed to known amounts of absorbed dose to air ranging from zero (unexposed) to 1 Gy using a 80 kVp beam with a HVL of 2.9 mm Al (Ref. 23) (which approximately corresponds to 33 keV effective energy). MicroStar Reader employs two calibrations: low- and high-dose, in which the LED beam operates in high power mode (for low-dose) or in low power mode (for high-dose). Readouts in this work were performed in the low-dose mode. All OSL counts reported here indicate the PMT counts as displayed by the reader.

II.B. Batch homogeneity

This test was undertaken to look at the variation among different dosimeters with 47 new nonscreened nanoDots (i.e., having $\pm 5\%$ variation of the labeled sensitivity values, as opposed to screened nanoDots that have only $\pm 2\%$ variation). Batch homogeneity studies were performed using 25, 80, and 120 kVp. Variation of x-ray exposure within the irradiated field was checked using a 15 cc parallel-plate ion chamber, model 96035B, (35050AT TRIAD TnT Dosimeter, FLUKE, Everett, WA), which had a recent NIST-traceable calibration with accuracy of $\pm 2\%$.

II.C. Reproducibility, effect of optical treatment, and signal depletion

Three unexposed new nanoDots were used for this test in order to examine their reproducibility and the effect of optical treatment to clear all the dosimetric traps. NanoDots in this measurement were placed over a $10 \times 10 \times 1$-in.[3] PMMA sheet. Irradiation was performed with 120 kVp and 80 mAs, which yielded a dose of 8 mGy as recorded by the ion chamber. Following the reading, OSLDs were optically bleached by placing them under a 75-W fluorescent bulb for approximately 8 h. NanoDots were then read to ensure that they had been adequately bleached (i.e., less than 200 counts). In the low-dose mode, the calibration constant was found to be about 2600 counts/mGy, and thus, the bleached nanoDots' dose was <80 μGy. The same cycle of irradiation, readout, and bleaching was repeated 13 times. Reproducibility of each dosimeter was measured as the coefficient of variation (CoV) of the 13 readout values. To estimate the signal depletion from readouts, ten OSLDs with average counts of 23 000 were read for a total of ten sequential readings per OSLD.

II.D. Effect of accumulated dose on the net counts

Using the general radiography unit, 12 nanoDots, which had been previously bleached, were irradiated together to a dose of 6 mGy. Irradiation followed by reading was repeated eight times using the same technique. No optical treatment was performed between measurements. After a cumulative dose of 42 mGy, a single dose of 40 mGy was added to the dosimeters making the total accumulated dose 82 mGy. NanoDots were then irradiated with the 6 mGy dose and read again. Net counts (i.e., previous counts subtracted from the new counts) from every 6-mGy irradiation were obtained, and the uncertainty (one standard deviation (1SD) from the 12 OSLD readings) was calculated.

II.E. Linearity

The purpose of this particular experiment was to investigate the linearity response of the OSLDs when exposed to doses from planar (general x-ray and mammography) or tomographic (CT) x-ray units.

In general radiography, the test was carried out at two kVp settings (80 and 120 kVp), and in mammography only the 25 kVp was used. Different mAs stations were utilized. The parallel-plate ion chamber was used to estimate the irradiated dose. In all three measurements, nanoDots and the chamber were placed over the PMMA sheet.

The tomographic linearity was investigated using the CT scanner. A piece of tape was extended from the scanner bed along the z-axis and taped onto a table placed at the back of the scanner (Fig. 2). The height of the bed and table was adjusted so that the tape intersected the scan isocenter. A standard axial scan was selected with a collimation of 40 mm (16×2.5 mm). NanoDots were exposed at the isocenter using 80 and 120 kVp with different mAs stations. A 3.2 cc pencil ion chamber, model 500–100 Victoreen® (ELIMPEX, Austria), which has a calibration traceable to NIST, was used to obtain the exposure.

FIG. 2. Tomographic linearity measurement. Two nanoDots (arrow) were placed at the scanner isocenter.

II.F. Angular dependence

The variability of nanoDots response to the incident x-ray beams from various angles was investigated. Eight different angles were used with the general x-ray and mammography units. A 45° wedge-shaped piece of hard paper was made, on which the dosimeters were placed to give the angles 45°, 135°, 225°, and 315°. The angular dependence was tested using two techniques in general x-ray: 80 kVp/200 mAs and 120 kVp/200 mAs, and only at 25 kVp/100 mAs in mammography. Tests were carried out with and without the presence of a backscattering material (the $10 \times 10 \times 1$-in.[3] PMMA sheet). In each measurement, the parallel-plate ion chamber was placed at the edge of the 15×15 cm^2 field to ensure the stability of the tube output.

With the 64-slice CT scanner, angular dependence was investigated using four angles. NanoDots were placed at the scanner isocenter over the tape in the same way the tomographic linearity test was performed. The angulation was done along the x/y-axis and the z-axis. A standard single-slice axial scan was used with 80 kVp/500 mAs and 120 kVp/400 mAs. The CT tube reproducibility was tested prior to performing the test using the pencil ion chamber, which was also placed on the same tape.

II.G. Energy dependence

With the calibrated parallel-plate ion chamber placed on the radiographic table, measurements were taken at the following tube potentials: 50, 60, 80, and 120 kVp. Aluminum plates were used to alter the effective photon energy. Measurements were then repeated using three nanoDots per exposure in place of the ion chamber. HVL at every measurement was calculated, and the corresponding effective photon energy (keV) was estimated using data of the mass attenuation coefficient for aluminum.[24] Doses from the ion chamber were corrected for each beam quality, using the conversion factors available in the chamber's user manual. Ratios of

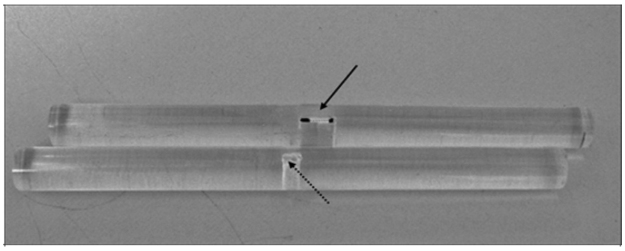

FIG. 3. Acrylic rods used for the nanoDot and TLDs. Solid arrow shows a nanoDot inserted in the slit-shaped hole. The dotted arrow shows the hole used for the TLDs.

chamber-to-nanoDot reading were used as the energy correction factors for nanoDots.

II.H. Comparison with TLDs using CT phantoms

The study was carried out on the 64-slice CT scanner. LiF:Mg,Ti TLDs (Harshaw TLD-100) with dimensions of $3 \times 3 \times 1\ \mathrm{mm}^3$ were used along with nanoDots for this investigation. The TLDs were taken from a batch with homogeneity of <3%.

The standard CTDI phantoms were used for this experiment: body (32 cm) and head (16 cm) phantoms. Drilling was carried out in two acrylic rods to accommodate the point dosimeters (nanoDot and TLDs), as shown in Fig. 3.

Starting with the head phantom, measurements were obtained at 80 and 120 kVp settings. A standard axial head scan with a collimation of 40 mm was used. Holes at the following depths in the phantom were utilized for dosimeter measurements: 0 cm (surface), 1, 4, and 8 cm. With the body phantom, in addition to the phantom surface, four holes were utilized at the following depths from surface: 1, 9, 12, and 16 cm. A standard axial body scan was selected with 40 mm collimation. Measurements were performed at 80 and 120 kVp tube potentials.

The scanner HVL was measured at the two tube potentials. TLDs were then calibrated at these tube potentials using the general x-ray unit, with 8.5 mm added aluminum to match the x-ray beam characteristics of the CT. The measured HVLs also helped to determine what correction factors to be used for nanoDot doses. TLDs were read 24 h after irradiation using a 2800M Victoreen reader. Unexposed TLDs were read to estimate the background signal to be subtracted. A paired t-test was used to determine whether the differences between dosimeter readings in different depths of the phantom were significant.

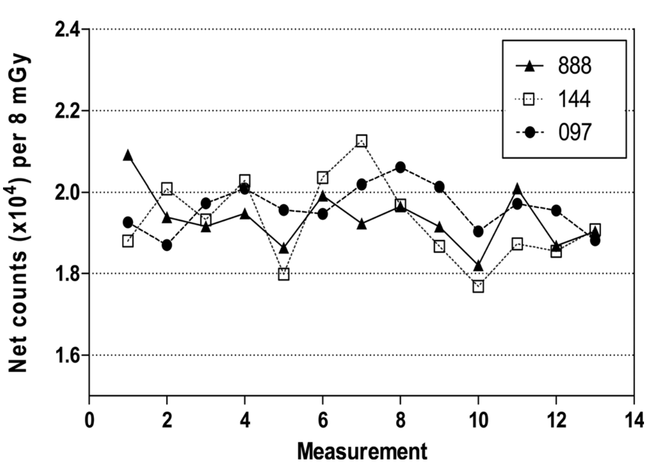

FIG. 4. Plot of the nanoDots reproducibility results. Numbers in the legend represent the last three digits of each nanoDot's serial number.

TABLE I. Mean, standard deviation (SD) and coefficient of variation (CoV) for the 13 repeated measurements of the three dosimeters.

NanoDot #	Mean (counts)	SD	CoV (%)
097	19 604	562	2.9
144	18 845	682	3.6
888	19 349	702	3.6

III. RESULTS

III.A. NanoDots batch homogeneity

The maximum variation in the beam output within the irradiated field, as measured by the ion chamber, was found to be 2.3% (1.76 R at the cathode side and 1.72 R at the anode side). The relative standard deviation of the 47 nanoDot readings was 4.3%, 4.8%, and 4.4% at kVp settings of 120, 80, and 25, respectively.

III.B. Reproducibility, effect of optical treatment, and signal depletion

Figure 4 illustrates the plot of the 13 repeated measurements. Dosimeter's reproducibility was found to be between 2.9% and 3.6%. Table I shows the mean, SD, and CoV. No noticeable trend was observed to suggest a change in sensitivity as a result of the optical treatment. Measurements reproducibility from the ion chamber readings were 1.5%. A plot of the ten repeated readouts is shown in Fig. 5. Average decrease of signal per readout was found to be 0.995, which implies that the depletion per readout is 0.5%.

III.C. Effect of accumulated dose on the added doses

The lowest SD based on reading of 12 nanoDots was 571 counts (3.8%), and was obtained from the first measurement.

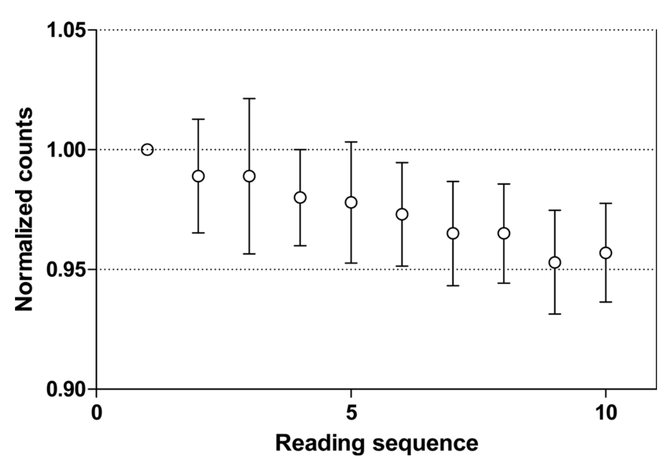

FIG. 5. OSL signal depletion from repeated readouts. Every OSLD was read ten times and the readout values were normalized to the first reading. Each point represents the mean of the ten measured OSLDs. The average of signal depletion per readout was found to be 0.5%. Error bars represent 1SD from the ten OSLDs.

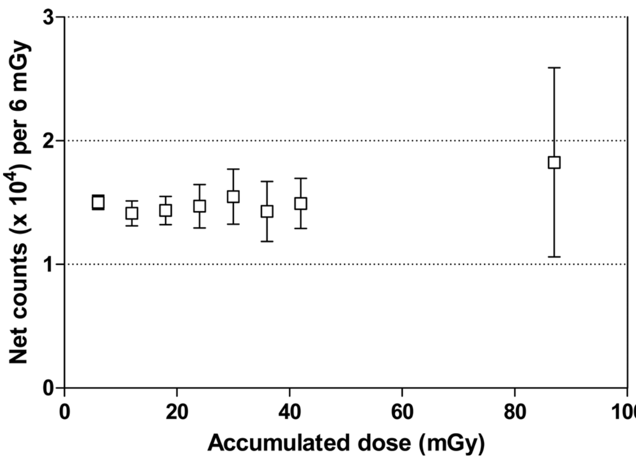

FIG. 6. Effect of the accumulated dose on the net counts (i.e., new counts - old counts) per 6 mGy irradiation. Error bars represent 1 SD from the 12 nanoDots.

However, as illustrated in Fig. 6, as the accumulated dose increased, so did the SD of the net counts for a 6-mGy irradiation. When the accumulated dose was 82 mGy, the added 6 mGy resulted in a SD of the net counts of 42%.

III.D. Linearity

As shown in Figs. 7(a) and 7(b), good linearity between nanoDot response and ion chamber doses was obtained for all five radiation qualities, with R^2 values found to be greater than 0.99 ($p < 0.05$). Moreover, the residuals were plotted [Fig. 7(c)] and a subjective evaluation of the random distribution of the points above and below the fitted curve did also indicate a linear relationship in the five measurements.

III.E. Angular dependence

Doses from angled nanoDots were normalized to the 0° dose, in which the detector's serial number was facing the beam. The 180° indicates when the opposite side (barcode) was facing the beam; 90° and 270° are when the x-ray beam was facing the upper side and lower side (Fig. 1), respectively. The most obvious deviation was found with the 90° in the mammography setup, in which the dose decreased by 60% and 70% on PMMA and in air, respectively [Fig. 8(a)]. Another noticeable drop was also seen in mammography with 270° (33%–40%). For the general x-ray unit, the same pattern of variation was seen but to a lesser degree. At 80 kVp the reduction was 42% at 90° with PMMA, and 18% at

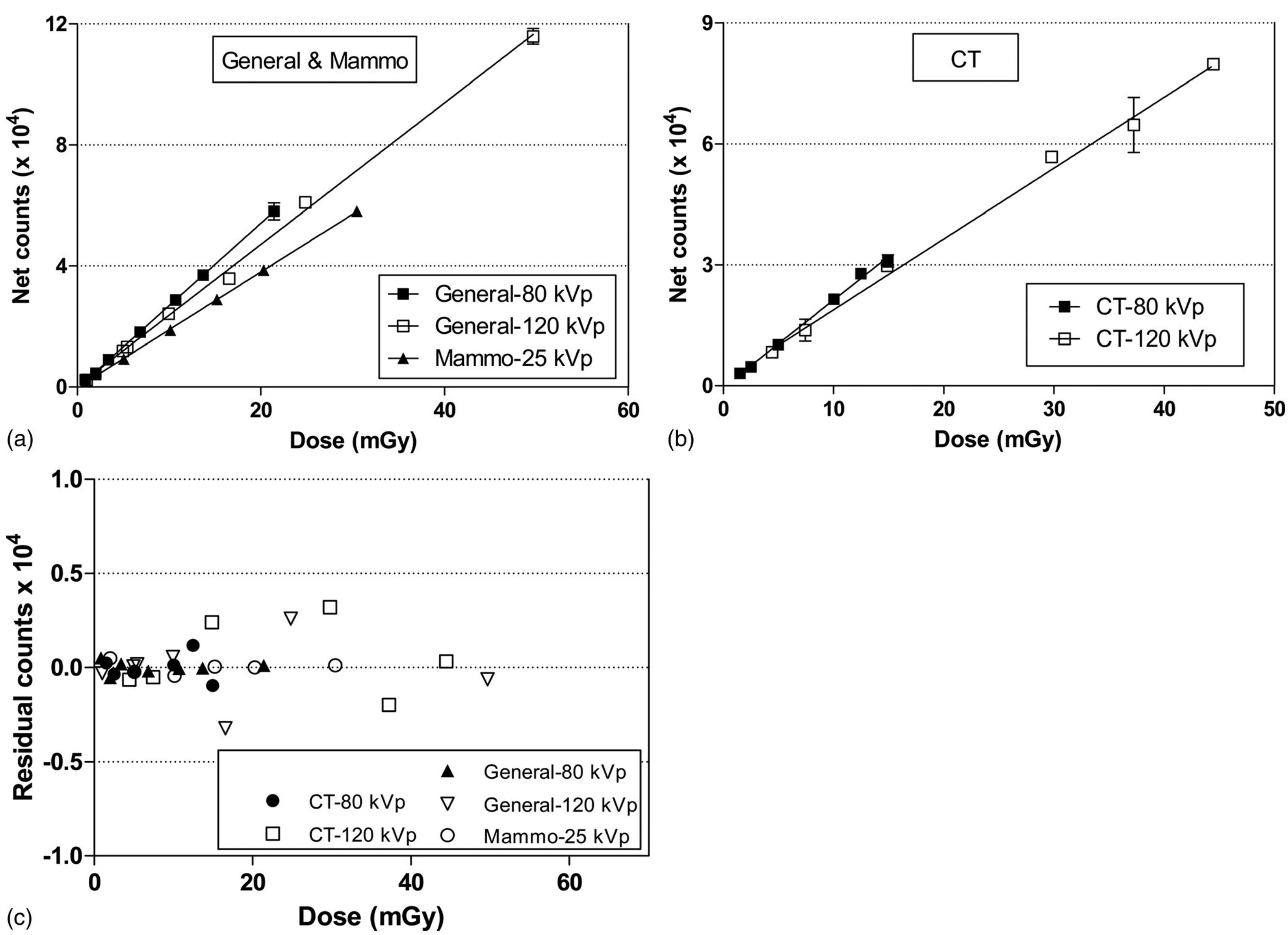

FIG. 7. Linear regression of linearity tests with (a) general radiography and mammography and (b) CT. Error bars represent 1SD from three nanoDots per mAs station in general and mammography and two nanoDots per mAs for CT. (c) A plot of the residual counts (difference between points and the fitted curve) against dose (mGy) for all five measurements to test the linearity. It can be seen that the data are randomly distributed above and under the curve.

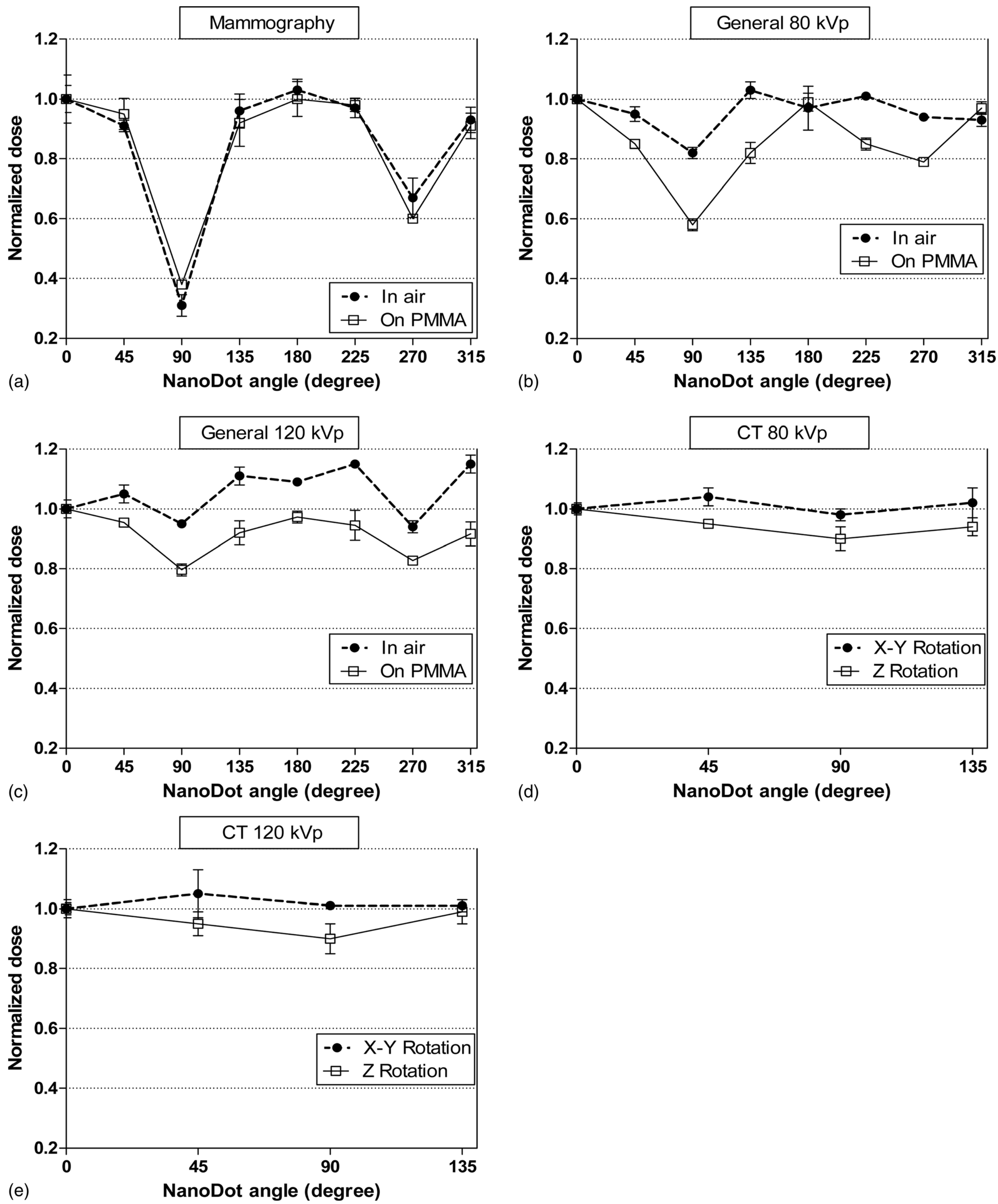

FIG. 8. Results for the angular dependence evaluation with mammography (a), general diagnostic at 80 kVp (b) and 120 kVp (c), and CT (d) and (e). Doses are normalized to the dose at 0°. Error bars represent 1 SD from three nanoDots used per angle.

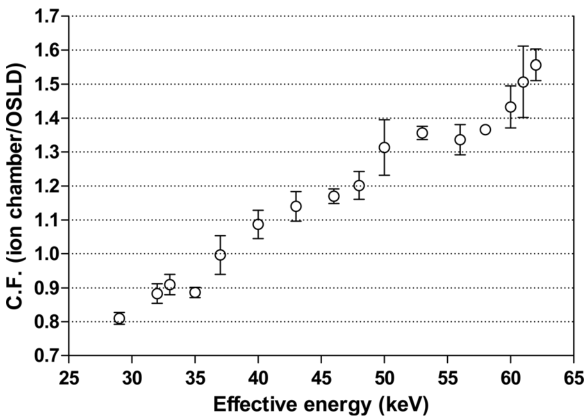

FIG. 9. OSLD correction factors as a function of the effective photon energies. Error bars represent 1 SD from three nanoDots.

90° in air [Fig. 8(b)]. With the 120 kVp technique, a decrease by 20% at 90° on PMMA was seen [Fig. 8(c)]. In air, angled nanoDots showed remarkably less variation with a fluctuation ranging between 0.95 and 1.15, the dose at 0°. CT results showed the least apparent variation in either 80 or 120 kVp with a maximum drop of 10% in the z-axis direction [Figs. 8(d) and 8(e)]. Tube output reproducibility of the general radiographic unit and CT was below 0.5%, and was 2% for the mammographic unit.

III.F. Energy dependence

Figure 9 demonstrates the obtained correction factors (CFs) for nanoDots at the different measured effective energies from 29 up to 62 keV, which correspond to HVLs of 2.0 mm Al to 9.8 mm Al. A range of CFs from 0.81 (at 29 keV) to 1.56 (at 62 keV) was found. The list of the correction factors with their 1SD uncertainties is shown in Table II.

TABLE II. Calculated HVLs with their corresponding effective energy (keV), estimated correction factor for nanoDots, and the standard deviation (SD).

HVL (mm Al)	keV	CF	SD
2.0	29	0.81	0.02
2.6	32	0.88	0.03
2.9	33	0.91	0.03
3.4	35	0.89	0.02
3.9	37	1.00	0.06
4.5	40	1.08	0.04
5.3	43	1.14	0.04
6.0	46	1.17	0.02
6.4	48	1.20	0.04
6.9	50	1.31	0.08
7.6	53	1.36	0.02
8.4	56	1.33	0.05
8.8	58	1.37	0.01
9.2	60	1.43	0.07
9.5	61	1.50	0.11
9.8	62	1.56	0.05

III.G. Comparison with TLDs using CT phantoms

The measured HVL for 120 kVp was found to be 8.9 mm Al, and it was 6.5 mm Al for 80 kVp. From the energy dependence CF results, these HVLs were found to correspond to CFs of 1.4 and 1.23 for 120 kVp and 80 kVp, respectively. TLDs' calibration factors and background subtraction were applied to TLD readout values. Figures 10(a) through 10(d) illustrate the results normalized to OSLD dose at 1 cm. In all experiments, nanoDots and TLDs showed comparable results.

The paired t-tests did not show a significant variation in any of the four experiments (the lowest p value was 0.38), as shown in Table III.

IV. DISCUSSION

The stability of OSLDs sensitivity was examined first by exposing batches of nanoDots to low and high tube potentials. The resultant batch homogeneity was between 4% and 5% which, given that the irradiating beam had a variation of 2%, can be described as excellent. Next, by exposing three nanoDots to a certain technique and repeating it 13 times, the reproducibility was between 2.8% and 3.8% (average: 3.3%). This is higher than the <1.0% reproducibility values reported by Jursinic[3] at a beam energy of 6 MeV, but it is close to the results of Viamonte *et al.*[4] who found the reproducibility of the individual OSLD to be 2.5%, and 4.2% for the batch. Another observed result was that there was no trend seen which would indicate any changes in dosimeter sensitivity as a result of optical treatment.

One of the advantages of OSLD over TLD system is that in the reading process, OSLDs are stimulated for a very short time, which would allow the dosimeter to retain its dose record. It was found that only 0.05% of the dosimeter signal was depleted when the high-dose mode was used.[3] In the low-dose (high LED beam power) mode, however, the signal depletion was found to be ten times higher (0.5%). Nevertheless, when considering the uncertainties associated with statistical fluctuations, this amount of depletion may still be considered small.

Doses in diagnostic imaging are generally low compared to those in radiation therapy, and therefore, adding a few mGy to high accumulated doses may result in the newly added dose being within the uncertainty range of the accumulated dose. Based on the data shown in Fig. 6, it is noted that when the added dose was less than 10% of the accumulated dose, the uncertainty of the net counts (and hence the dose) did increase to 42%, which is rather high in a dose estimate. Therefore, we suggest that in such cases, optical treatment be applied or alternatively, using fresh nanoDots to reduce the estimation error.

As expected, the linearity test for the nanoDots showed an overall good linear response at typical mammographic, radiographic, and CT energies, with $R^2 > 0.99$. This is in good agreement with the published results of the manufacturer[25] and those obtained by Jursinic[3] and Danzer *et al.*[26] in therapy settings. As for angular dependence, a prior study[3] concluded that when OSLDs are exposed to 6 MeV x-ray

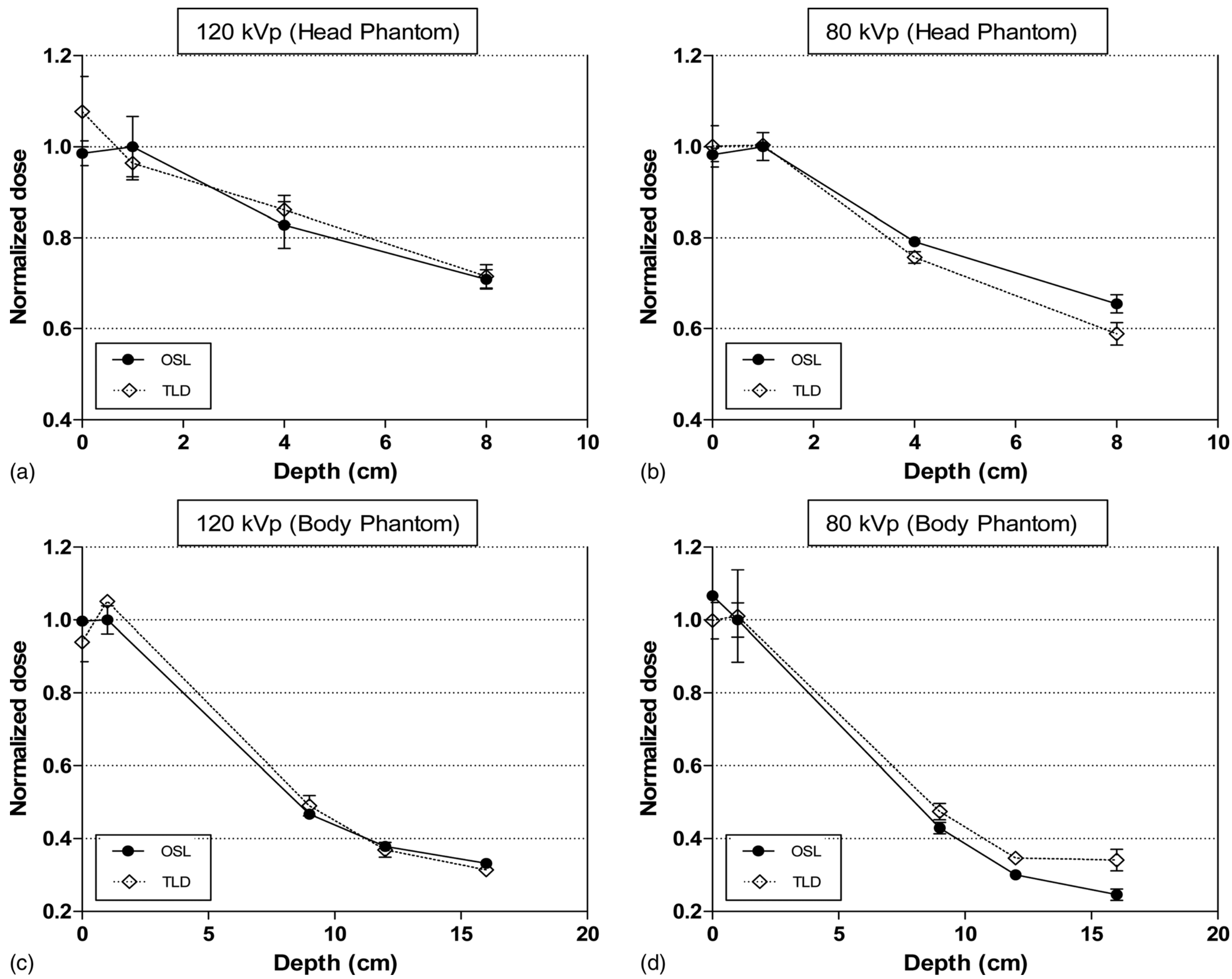

FIG. 10. Comparison between different dosimeters in CT using two phantom sizes and different techniques: the head phantom at (a) 120 kVp and (b) 80 kVp and the body phantom at (c) 120 kVp and (d) 80 kVp. Doses are normalized to the OSLD dose at 1 cm. Error bars represent 1SD from three readings.

beam, their response is independent of the incident angle. In a different study using beta radiation,[27] nanoDots response showed a decrease of 50% when irradiated at 90°. A report by the nanoDots manufacturer on the angular dependence of their product in diagnostic, CT, and mammography showed the maximum variation to be 10% at a 60° angle.[27] In this study, variations as high as 70% were observed in mammography, but tapered off as the photon energy increased. In the general radiographic range, angular dependence peaked at 40% for 80 kVp (HVL = 3.3 mm Al) and at 20% for 120 kVp (HVL = 5.3 mm Al). Even at typical CT energies, there was close to 10% variation seen for a changing beam incidence angle as exhibited in Figs. 8(d) and 8(e). This implies that the angular dependence of nanoDot dosimeters is energy dependent. This may be explained by the higher attenuation, the low energy photons undergo when entering the angled dosimeter, in which the thickness of the plastic casing that photons will encounter before reaching the detector is higher, and thus more soft x-ray beam will be filtered out. The thickness of front and back faces of the plastic casing is in the order of 0.4 mm, whereas the upper and lower side walls (Fig. 1) are about 1 mm thick. The OSLD is held by a plastic frame that has a nonuniform thickness around the detector as shown in Fig. 1. Therefore, in the case of 90° and 270° angles, this frame adds to the 1 mm thick sides. Although these thicknesses sound small, knowing that the casing material has a mass density close to that of water, low energy photons of x-ray spectra can be significantly attenuated by such thicknesses. Another influencing factor that may contribute to the angular dependency could be due to hardening of the beam by the sensitive volume of the OSLD itself. When the OSLD is oriented at 90° and 270°, the beam traverses 5 mm of Al_2O_3:C, whereas at 0° and 180°, the beam only traverses 0.2 mm. There was, in addition, a remarkable difference between the 90° and 270° doses. Possible reasons for this is that in the case of 90° the plastic arm that attaches the OSLD to its case happens to be in the upper side (Fig. 1), which may further attenuate the soft beam especially if the nanoDot was not placed precisely on the central axis of the beam. Also, because the sensitive volume

TABLE III. Results of the p values from the paired t-test to compare OSLD and TLD doses from measurements using head and body phantoms in CT. A p value of <0.05 was considered statistically significant.

	p value	
Phantom	120 kVp	80 kVp
Head	0.44	0.38
Body	0.92	0.39

of the detector is not placed in the center of the plastic casing, the beam will traverse different thickness of plastic for the 90° and 270°. However, further investigation may be necessary to confirm these hypotheses. These results of angular dependence are of great consequence when OSLDs are used in interventional fluoroscopy procedures to evaluate patient peak skin dose. If the orientation of the C-arm changes during such procedures, then the angular dependence of the OSLDs should be kept in mind when evaluating patient doses.

It has been found that the Al_2O_3:C OSLDs have no energy dependence in the therapeutic range (>6 MV).[3,4,28–30] Nevertheless, due to their relatively high effective atomic number (11.28), it has been shown that OSLDs over-respond to low energy x-ray.[10,17] This can be attributed to the high photoelectric effect in Al_2O_3:C at low photon energies,[31] which raises its mass energy absorption coefficients relative to water.[32] In this study, it was observed that even a small increment in the photon energy may result in a remarkably different response by the OSLDs. Correction factors for energies from 29 keV to 62 keV (relative to the 80 kVp calibration; see Sec. II A) were found to range from 0.81 to 1.56, with uncertainty of <7% (1SD). These factors represent the ratios of ion chamber doses to their corresponding nanoDot doses from the microStar Reader. It is thus recommended that for any study utilizing OSLDs, the beam energy (or HVL) of the imaging unit be accurately known before applying any correction and the value of the correction factor should be validated. When looking at the CFs provided by the manufacturer's calibration report,[24] we find CFs of 1.0 and 1.19 are given to the beams with HVLs of 2.9 and 8.4 mm Al, respectively, whereas the corresponding CFs in our study were, respectively, 0.91 ± 0.03 and 1.33 ± 0.05. The differences may have arisen from the fact that, as stated in the manufacturer's report, they used a backscattering material (PMMA) during exposure. In our measurements, no backscattering material was used in the energy dependence test. Furthermore, the manufacturer's report does not provide details about the measurements methodology or whether any corrections were applied to the reference dosimeter.

In comparison with other dosimeters using CT phantoms, nanoDots demonstrated very comparable response to TLDs, as shown in Figs. 10(a) through 10(d). Also, the relatively high p values using a paired t-test support the hypothesis that a good agreement between TLDs and OSLDs would be obtained since they were calibrated under similar conditions.

It should be noted that, in all OSLD results presented here, the effect of OSL signal fading may have a considerable role in the results' uncertainties. Despite the fact that readout was carried out at least 30 min after irradiations to achieve a good detector's signal stability, it has been shown by Yukihara *et al.*[33] that fading of OSL signals continues in a slower rate even after a period of 70 days post irradiation, and the decrease of signal in a period of time from 0.5 to 24 h (the time interval in which OSLDs were read in our study) was found to be in the order of 4%.

V. CONCLUSIONS

OSLDs (nanoDots) exhibited good homogeneity, reproducibility, and linearity in the diagnostic energy range. They also compared well with TLDs. However, a few factors were found to introduce considerable errors into the measurements: adding small doses to an OSLD which already has high accumulated dose, angular dependence, and energy dependence. Based on the results of these measurements, the following recommendations are made:

1. To reduce the high counting errors caused by low doses in diagnostic imaging, either optical bleaching or new OSLDs should be used.
2. When using low kVp techniques, care should be taken of the nanoDot orientation. Angled nanoDots may significantly affect the accuracy of dose estimate in such techniques.
3. It is important to determine the beam energy (or HVL) and to find the appropriate correction factor that matches it to reduce possible significant errors in dose estimation.

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[a]Author to whom correspondence should be addressed. Electronic mail: alsenan@uthscsa.edu

[1]E. G. Yukihara, G. Mardirossian, M. Mirzasadeghi, S. Guduru, and S. Ahmad, "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 (2008).
[2]P. A. Jursinic, "Changes in optically stimulated luminescent dosimeter (OSLD) dosimetric characteristics with accumulated dose," Med. Phys. **37**, 132–140 (2010).
[3]P. A. Jursinic, "Characterization of optically stimulated luminescent dosimeters, OSLDs, for clinical dosimetric measurements," Med. Phys. **34**, 4594–4604 (2007).
[4]A. Viamonte, L. A. R. da Rosa, L. A. Buckley, A. Cherpak, and J. E. Cygler, "Radiotherapy dosimetry using a commercial OSL system," Med. Phys. **35**, 1261–1266 (2008).
[5]E. G. Yukihara and S. W. S. McKeever, "Optically stimulated luminescence (OSL) dosimetry in medicine," Phys. Med. Biol. **53**, R351–R379 (2008).
[6]D. J. Huntley, D. I. Godfrey-Smith, and M. L. Thewalt, "Optical dating of sediments," Nature (London) **313**, 105–107 (1985).
[7]L. Bøtter-Jensen and S. W. S. McKeever, "Optically stimulated luminescence dosimetry using natural and synthetic materials," Radiat. Prot. Dosim. **65**, 273–280 (1996).
[8]B. G. Markey, L. E. Colyott, and S. W. S. McKeever, "Time-resolved optically stimulated luminescence from α-Al_2O_3:C," Radiat. Meas. **24**, 457–463 (1995).
[9]H. Nanto, K. Murayama, T. Usuda, S. Taniguchi, and N. Takeuchi, "Optically stimulated luminescence in KCl:Eu single crystals," Radiat. Prot. Dosim. **47**, 281–284 (1993).

[10]M. S. Akselrod, V. S. Kortov, D. J. Kravetsky, and V. I. Gotlib, "Highly sensitive thermoluminescent anion-defective $A1_2O_3$:C single crystal detector," Radiat. Prot. Dosim. **32**, 15–29 (1990).
[11]M. S. Akselrod, V. S. Kortov, and E. A. Gorelova, "Preparation and properties of alpha-Al_2O_3:C," Radiat. Prot. Dosim. **47**, 159–164 (1993).
[12]S. W. S. McKeever and M. S. Akselrod, "Radiation dosimetry using pulsed optically stimulated luminescence of Al_2O_3:C," Radiat. Prot. Dosim. **84**, 317–320 (1999).
[13]L. Bøtter-Jensen, N. Agersnap Larsen, B. G. Markey, and S. W. S. McKeever, "Al_2O_3:C as a sensitive OSL dosemeter for rapid assessment of environmental photon dose rates," Radiat. Meas. **27**, 295–298 (1997).
[14]S. W. S. McKeever, M. S. Akselrod, and B. G. Markey, "Pulsed optically stimulated luminescence dosimetry using alpha-Al_2O_3:C," Radiat. Prot. Dosim, **65**, 267–272 (1996).
[15]S. W. S. McKeever and M. Moscovitch, "Topics under debate - on the advantages and disadvantages of optically stimulated luminescence dosimetry and thermoluminescence dosimetry," Radiat. Prot. Dosim. **104**, 263–270 (2003).
[16]Radiological Physics Center (RPC), "The RPC has changed to optically-stimulated luminescence dosimetry," RPC Newsl. June, (2010).
[17]A. J. J. Bos, "High sensitivity thermoluminescence dosimetry," Nucl. Instrum. Methods Phys. Res. B **184**, 3–28 (2001).
[18]L. Duggan *et al.*, "Suitability of LiF:Mg, Cu, P and Al_2O_3:C for low dose measurements in medical imaging," Radiat. Prot. Dosim. **85**, 425–428 (1999).
[19]E. G. Yukihara *et al.*, "An optically stimulated luminescence system to measure dose profiles in x-ray computed tomography," Phys. Med. Biol. **54**, 6337–6352 (2009).
[20]C. Ruan, E. G. Yukihara, W. J. Clouse, P. B. R. Gasparian, and S. Ahmad, "Determination of multislice computed tomography dose index (CTDI) using optically stimulated luminescence technology," Med. Phys. **37**, 3560–3568 (2010).
[21]M. C. Aznar, "In vivo absorbed dose measurements in mammography using a new real-time luminescence technique," Br. J. Radiol. **78**, 328–334 (2005).
[22]D. Peakheart *et al.*, "Evaluation of an OSL dosimetry system for CT quality assurance and dose optimization," *Presented at the 48th Annual Meeting of the AAPM*, (Orlando, FL, July 30 to August 3, 2006).
[23]C. J. Yahnke, *Calibrating the MicroStar* (Landauer Inc., Glenwood, IL, 2010), URL: http://www.landauer.com/uploadedFiles/Resource_Center/microStar%20Calibration%20and%20Usage%20Instructions%2010-Jun-09.pdf
[24]J. H. Hubbell and S. M. Seltzer, "Tables of X-ray mass attenuation coefficients and mass energy-absorption coefficients from 1 keV to 20 MeV for elements $Z=1$ to 92 and 48 additional substances of dosimetric interest," NISTIR 5632 (National Institute of Standards and Technology, Gaithersburg, MD, 1995).
[25]C. Perks, C. Yahnke, and M. Million, *Medical Dosimetry Using Optically Stimulated Luminescence Dots and Microstar^{TM} Readers* (Landauer Europe, UK, 2008), URL: www.landauer.co.uk/documents/LANDAUER_EUROPE_Poster_microstar_in_medical_dosimetry_october08.pdf
[26]J. Danzer *et al.*, "Optically stimulated luminescence of aluminum oxide detectors for radiation therapy quality assurance," *Presented at the 49th Annual Meeting of the AAPM*, (Minneapolis, MN, July 22–26, 2007).
[27]T. C. Pinto and L. V. Caldas, "Characterization of OSL commercial dosimeters using a hand phantom, in standard beta radiation beams," *Presented at the International Nuclear Atlantic Conference - INAC 2009*, (Rio de Janeiro, RJ, Brazil, 2009).
[28]P. Mobit, "Comparison of the energy-response factor of LiF and Al_2O_3 in radiotherapy beams," Radiat. Prot. Dosim. **119**, 497–499 (2006).
[29]S. W. Chen, X. T. Wang, L. X. Chen, Q. Tang, and X. W. Liu, "Monte Carlo evaluations of the absorbed dose and quality dependence of Al_2O_3 in radiotherapy photon beams," Med. Phys. **36**, 4421–4424 (2009).
[30]M. C. Aznar *et al.*, "Real-time optical-fibre luminescence dosimetry for radiotherapy: Physical characteristics and applications in photon beams," Phys. Med. Biol. **49**, 1655–1669 (2004).
[31]C.S. Reft, "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 (2009).
[32]F. H. Attix, *Introduction to Radiological Physics and Radiation Dosimetry* (Wiley-VCH, Weinheim, 2004).
[33]E. G. Yukihara *et al.*, "Medical applications of optically stimulated luminescence dosimeters (OSLDs)," Radiat. Meas. **45**, 658–662 (2010).