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Development and implementation of a remote audit tool for high dose rate (HDR) Ir-192 brachytherapy using optically stimulated luminescence dosimetry

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Purpose: The aim of this work was to create a mailable phantom with measurement accuracy suitable for Radiological Physics Center (RPC) audits of high dose-rate (HDR) brachytherapy sources at institutions participating in National Cancer Institute-funded cooperative clinical trials. Optically stimulated luminescence dosimeters (OSLDs) were chosen as the dosimeter to be used with the phantom.
Methods: The authors designed and built an $8 \times 8 \times 10$ cm^3 prototype phantom that had two slots capable of holding Al_2O_3:C OSLDs (nanoDots; Landauer, Glenwood, IL) and a single channel capable of accepting all ^{192}Ir HDR brachytherapy sources in current clinical use in the United States. The authors irradiated the phantom with Nucletron and Varian ^{192}Ir HDR sources in order to determine correction factors for linearity with dose and the combined effects of irradiation energy and phantom characteristics. The phantom was then sent to eight institutions which volunteered to perform trial remote audits.
Results: The linearity correction factor was $k_L = (-9.43 \times 10^{-5} \times \text{dose}) + 1.009$, where *dose* is in cGy, which differed from that determined by the RPC for the same batch of dosimeters using ^{60}Co irradiation. Separate block correction factors were determined for current versions of both Nucletron and Varian ^{192}Ir HDR sources and these vendor-specific correction factors differed by almost 2.6%. For the Nucletron source, the correction factor was 1.026 [95% confidence interval (CI) = 1.023–1.028], and for the Varian source, it was 1.000 (95% CI = 0.995–1.005). Variations in lateral source positioning up to 0.8 mm and distal/proximal source positioning up to 10 mm had minimal effect on dose measurement accuracy. The overall dose measurement uncertainty of the system was estimated to be 2.4% and 2.5% for the Nucletron and Varian sources, respectively (95% CI). This uncertainty was sufficient to establish a ±5% acceptance criterion for source strength audits under a formal RPC audit program. Trial audits of four Nucletron sources and four Varian sources revealed an average RPC-to-institution dose ratio of 1.000 (standard deviation = 0.011).
Conclusions: The authors have created an OSLD-based ^{192}Ir HDR brachytherapy source remote audit tool which offers sufficient dose measurement accuracy to allow the RPC to establish a remote audit program with a ±5% acceptance criterion. The feasibility of the system has been demonstrated with eight trial audits to date. © *2013 American Association of Physicists in Medicine*. [http://dx.doi.org/10.1118/1.4824915]

Key words: HDR, brachytherapy, OSL, audit, RPC

1. INTRODUCTION

The Radiological Physics Center (RPC) was established in 1968 with the mission to ensure that institutions participating in National Cancer Institute-funded clinical trials deliver clinically comparable and consistent radiation doses. One of the major efforts the RPC has developed in pursuit of this mission is the mailable optically stimulated luminescence

dosimeter (OSLD) program for external beam machine output.[1] Through this program, OSLDs are mailed to participating institutions, irradiated, and returned to the RPC for analysis. The OSLD program currently monitors nearly 15 000 megavoltage external beams but it is not used to monitor brachytherapy treatment delivery at this time. Current high dose rate (HDR) brachytherapy audit activities performed by the RPC consist mainly of questionnaires, patient plan checks, and benchmark treatment plans, with no remote measurement capabilities analogous to the external beam monitoring program. Given the recent growth in the use of HDR brachytherapy for certain diseases such as gynecological[2] and breast[3] cancer and the development of cooperative clinical trials utilizing HDR brachytherapy,[4–6] a new remote audit tool capable of verifying HDR source strength was needed to supplement the RPC's existing HDR audit capabilities.

The aim of this work was to develop and validate a new phantom suitable for remote RPC audits of HDR source strengths. The phantom was required to be durable enough to be mailed and simple enough to be easily understood and used accurately at participating institutions. Most importantly, it had to be capable of measuring dose with an accuracy that would allow the RPC to establish a ±5% acceptance criteria for HDR source strength audits, which matches the RPC's existing criteria for external beam machine output audits.[7] The use of nanoDots, a type of small planar OSL dosimeter, in the phantom allowed for the precise measurement of dose in the regions of steep dose gradients characteristic of brachytherapy sources.

2. METHODS AND MATERIALS

2.A. Dosimeter

The dosimeter selected for use in the proposed HDR brachytherapy audit tool was the nanoDot OSLD (Landauer, Inc., Glenwood, IL). Each nanoDot consists of a disk 5 mm in diameter and 0.3 mm thick of Al_2O_3:C OSL material encased in a light-tight 10 × 10 × 2 mm^3 plastic cassette (Fig. 1). The RPC has considerable experience working with these particular dosimeters, having tens of thousands of individual dosimeters currently in circulation as part of its external beam audit program.

After they were irradiated, the nanoDots were read using the microStar reader system (Landauer, Inc., Glenwood, IL).

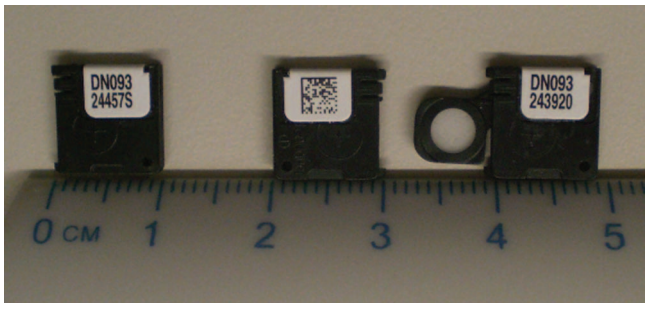

FIG. 1. Three nanoDot OSLDs. Each cassette includes a white disk of active OSL material (right).

Each dosimeter was inserted into the reader; then, the tray holding the active OSL material was pushed out of the cassette and illuminated by a light-emitting diode array. A photomultiplier tube collected the photons emitted by the dosimeter due to photostimulation. The reader also includes two optical filters having a combined peak sensitivity of 420 nm, which is near the dominant emission band of Al_2O_3:C OSL material, positioned in front of the PMT to filter light emitted by the LED array. The entire reading procedure took approximately 7 s for a single nanoDot.

Along with simplicity in readout, OSLDs have several other desirable features. As mentioned above, the near-planar geometry of the nanoDot packaging (active dosimeter thickness of approximately 0.3 mm) reduces the volume-averaging effect in the area of steep spatial dose gradients inherent to brachytherapy sources. OSLDs have also been found to be dose-rate independent.[8–10] The RPC has found the dosimeters to be unaffected by typical variations in temperature and humidity encountered during shipping.[11] Lastly, OSLDs can be bleached and reused, extending their useful life. The RPC's standard operating procedure is to reuse nanoDots up to a cumulative dose of 10 Gy, and that procedure was followed in this work as well.

2.B. Phantom

We designed and manufactured the phantom prototype with the goals and requirements of a remote audit program in mind. Because the phantom was intended to be mailed to institutions, the physical size had to be small in order to reduce shipping expenses yet large enough for the source-to-detector distance to be clinically relevant. However, this meant that the phantom would likely be too small to provide full scatter conditions, which Perez-Calatayud estimated to be a water sphere with a 40 cm radius[12] for ^{192}Ir. Lack of full scatter was therefore addressed by introducing a correction factor into the dose calculation. Such a correction is only possible if the setup and irradiation geometry is fixed and unchanging, as it was using the phantom described here.

Another consideration in the design of the audit tool was the phantom medium. We chose to make the phantom out of a solid material rather than create one designed to be filled with water. This choice reduced the overall complexity of the phantom and the opportunity for error on the part of a physicist performing the audit with only written instructions to follow. Furthermore, OSLD nanoDots are not completely waterproof so the use of a solid phantom reduced the chances that they might be damaged. Lastly, manufacturing the phantom from a solid block of plastic material increased the durability of the tool which was important because it was designed to be shipped repeatedly.

The phantom included a single channel for source placement; a single source channel was sufficient to audit the source strength yet also minimized treatment planning complexity and the time required to perform the audit. The channel was designed to be long enough to include several dwell positions so that a constant isodose line across the OSLDs could be generated. This was important because a varying

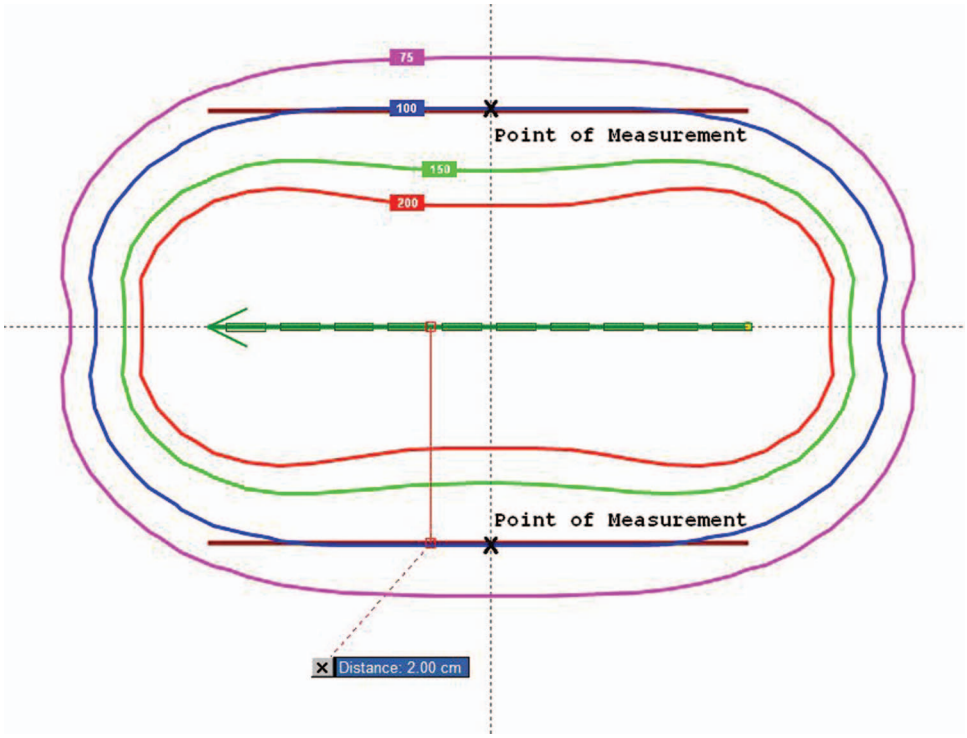

FIG. 2. The coronal plane isodose distribution from a treatment plan created to perform characterization irradiations. One OSL dosimeter was positioned at each point of measurement. The arrow down the center represents ten dwell positions with 5 mm center-to-center spacing.

isodose profile in the region of the dosimeters might have introduced additional uncertainty.

The phantom was also designed to have slots for two dosimeters, arranged symmetrically on either side of the central channel. This arrangement was necessary to reduce measurement uncertainty due to minor variations in the lateral positioning of the source within the catheter. To quantify this uncertainty, we performed TG-43 calculations under the assumptions that the diameter of the channel and the widths of the dosimeter slots varied from specification by up to $\pm 5\%$. A catheter with negligible wall thickness was also assumed. Both of these assumptions were conservative, because the phantom was manufactured using computer-controlled milling equipment with a stated tolerance of ± 0.025 mm and any catheter chosen for measurement purposes will of course have a finite thickness. Thus, the TG-43 calculation results represent a worst-case scenario in the dose measurement error owing to the lateral movement of the source within the channel.

2.C. System characterization and dose measurement

To characterize the dose measurement abilities of the phantom and OSLDs, a number of irradiations were performed using a Nucletron microSelectron v2 ^{192}Ir HDR source (Nucletron, Veenendaal, Netherlands) and a Varian VariSource VS2000 ^{192}Ir HDR source (Varian Medical Systems, Inc., Palo Alto, CA). First, a simple treatment plan was created which featured ten dwell positions in a single channel, similar to the proposed configuration to be used during actual remote audits. The dwell positions were spaced 5 mm apart and dwell times were chosen to create a flat isodose line 2 cm away from the channel and parallel to it. This distance corresponded to the position of the detectors in the phantom prototype. By scaling the dwell times uniformly, any dose could be delivered to the point of measurement using this same treatment plan. The plan with isodose lines is illustrated in Fig. 2.

First, we quantified the dose measurement robustness of the system given errors in distal/proximal source positioning of up to 1 cm. The basic treatment plan described above was delivered with the catheter fully inserted into the phantom. Then, to simulate potential setup error, the same plan was delivered with the catheter withdrawn from the end of the channel by distances of 2.5 mm, 5.0 mm, 7.5 mm, and 10.0 mm. We also performed TG-43 calculations using the same source positioning to confirm the measured results.

Next, the nonlinearity in the response of the OSLDs at various doses was measured, because Al_2O_3:C OSLD sensitivity is known to be dependent on dose.[8,13] The Nucletron HDR source was used to irradiate 78 dosimeters to doses ranging from 50 cGy to 400 cGy. For each dosimeter, the nominal dose delivered was divided by the total photon counts obtained during the readout of that particular OSLD. This quantity was known as the dose response for an individual dosimeter and had units of dose per count. Then, the dose response of each nanoDot was plotted against the nominal dose delivered to it and then a linear regression was applied to determine the relationship between nanoDot response and dose when irradiated in the phantom. Finally, the dose linearity correction factor k_L was calculated by normalizing this linear fit to 1.000 at a nominal dose of 100 cGy, which was the dose that

institutions will be asked to deliver as part of a future remote audit program.

An additional correction was required to convert OSLD readings to dose to water, the medium specified by the American Association of Physicists in Medicine's Task Group 43 report.[14] This correction addressed the energy-dependent response of Al_2O_3:C, the lack of full scatter conditions owing to the small size of the phantom, the difference in dosimetric properties between the phantom medium and water, and the angular dependence of the nanoDot OSL dosimeters.[15] Because the proposed audit tool featured a fixed geometry and was only intended for use with ^{192}Ir HDR brachytherapy sources, a single correction factor to account for all these various effects was determined. This correction factor was designated the block correction factor, or k_B.

To quantify the block correction factor, we made 20 OSLD measurements with the Nucletron HDR source and 10 OSLD measurements with the Varian HDR source. Here and throughout this work, a measurement was defined as the average readings of a pair of OSL dosimeters irradiated in the phantom at the same time. The following formalism was used to calculate the dose from an OSLD reading:

$$\text{Dose} = \text{reading} \times \text{ECF} \times \text{sensitivity} \times k_F \times k_L \times k_B, \tag{1}$$

where *reading* was the average raw OSLD reading, *ECF* was the element correction factor specific to the calibration of each individual nanoDot, *sensitivity* was the system sensitivity (dose per photon counts) established under reference ^{60}Co irradiation conditions, k_F was a correction for signal fading with time after irradiation, and k_L and k_B were as described above. Once we measured the linearity correction factor k_L as described in the previous section and recorded the delivered dose, Eq. (1) was solved to determine k_B:

$$k_B = \frac{\text{Dose}}{\text{reading} \times \text{ECF} \times \text{sensitivity} \times k_F \times k_L}. \tag{2}$$

Prior to making measurements of k_B, a source strength calibration traceable to the National Institute of Standards and Technology (NIST) was performed on each source. An HDRC-1 well-type ionization chamber (Precision Radiation Measurement, Inc.) was connected to a MAX-4000 electrometer (Standard Imaging, Middleton, WI). We then inserted a 6 French endobronchial catheter all the way into the well chamber. The HDR source stepped through seven dwell positions near the middle of the chamber in 5 mm increments. At each position, the ionization current was allowed to stabilize for approximately 5 s and then recorded. This was repeated three times and the current at each position averaged. We recorded the maximum average current and then calculated the air-kerma strength of the source following the formalism of DeWerd and Thomadsen:[16]

$$S_K = I \times P_{\text{elec}} \times C_{T,P} \times N_{RK} \times A_{\text{ion}} \times P_{\text{ion}}, \tag{3}$$

where S_K is the air-kerma strength, I is the measured ionization current, P_{elec} is the electrometer scale correction factor, $C_{T,P}$ is the temperature and pressure correction, N_{RK} is the ADCL-provided calibration coefficient for the well chamber, A_{ion} is the ADCL-provided correction for collection efficiency at the time of calibration, and P_{ion} is the correction for collection efficiency at the time of measurement.

We then used BrachyVision version 8.9.15 (Varian Medical Systems, Inc., Palo Alto, CA) to perform a TG-43 calculation of the dose to water at the point of measurement in the phantom using the calibrated air-kerma strength, clinical TG-43 parameters,[14] and the treatment plan described above. The block correction factor, which encompassed a variety of individual corrections unique to nanoDot OSLDs and the phantom irradiation geometry as described above, was then calculated by dividing the treatment planning system-reported TG-43 dose to water by a fully corrected OSLD *reading* [Eq. (2)]. In effect, k_B is the ratio of the quantity of interest (TG-43 dose to water) to the quantity that was measured (dose to a polystyrene phantom).

2.D. Trial remote audits

As a proof of concept and to test the accuracy of the proposed phantom, four institutions with Nucletron HDR sources and four institutions with Varian HDR sources agreed to perform trial remote audits. Each institution was provided with the phantom preloaded with two OSL dosimeters ready for irradiation. Each institution was also sent a form on which to record various machine, source, and plan characteristics as well as a page of written instructions. The instructions directed the participants to deliver 100 cGy to a line 2 cm away from the phantom channel using ten dwell positions spaced 5 mm apart. Participants were further encouraged to place the phantom in any orientation and location they found convenient during irradiation. Lastly, physicists participating in the trial remote audits were asked to report the amount of time they spent performing the audit, including the time needed to create an appropriate treatment plan, set up the phantom, irradiate the dosimeters, and gather all necessary information for reporting to the RPC.

After each institution returned the phantom, we read the OSLDs, applied the measured correction factors, and averaged the two readings together to determine the RPC-measured dose. This dose was then compared to each institution's treatment planning system-reported dose at the point of measurement (i.e., the center of the nanoDot cassette).

3. RESULTS

3.A. Phantom design

The finalized phantom prototype is shown in Fig. 3. The phantom is a solid $8 \times 8 \times 10$ cm^3 block of polystyrene (density 1.04 g/cm^3). A single 2 mm-diameter channel, sized to admit a 6 French or smaller catheter, extends 8.5 cm from the center of one of the 8×8 cm^2 faces lengthwise into the phantom. The phantom also has two slots for nanoDot dosimeters, located one on each side of the channel, 2 cm away from it laterally and centered along the 5 cm active source length. The slots are oriented such that the "face" of each nanoDot is

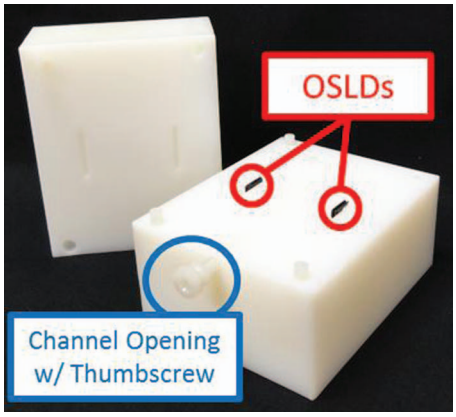

FIG. 3. The phantom prototype, separated into its two sections to show loaded nanoDots. The thumbscrew at front secures a catheter in the central channel.

parallel to the channel. The phantom can be separated lengthwise into two pieces for ease of loading and unloading the dosimeters.

The TG-43 calculations performed showed that assuming channel and dosimeter slot variations of up to ±5% and a negligible catheter wall thickness, doses to individual dosimeters varied by as much as 5% due to the lateral uncertainty in the source positioning. However, when two dosimeters were arranged symmetrically on either side of the channel, the average of their doses varied by only about 0.2% from an ideally positioned source. This dose variation due to lateral source positioning uncertainty, as calculated with TG-43, is illustrated in Fig. 4. These results informed the decision to include two dosimeters in the phantom design.

The results of the measurements and TG-43 calculations used to quantify the effect of setup error in the distal/proximal direction are shown in Fig. 5. An error in source positioning in the distal/proximal direction of 10 mm resulted in a dose measurement that was approximately 1% different from that achieved with ideal source positioning.

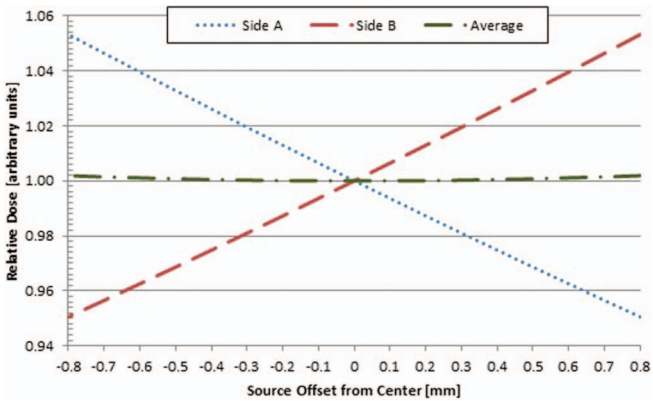

FIG. 4. The relative dose (from TG-43 calculations, Nucletron source) for two nanoDot dosimeters positioned in the two phantom slots given a lateral offset in the source positioning between them. Side A and Side B are arbitrary designations. With the source shifted ±0.8 mm from the center of the phantom, the average relative dose was 1.002. For the Varian source, the corresponding relative dose was also 1.002.

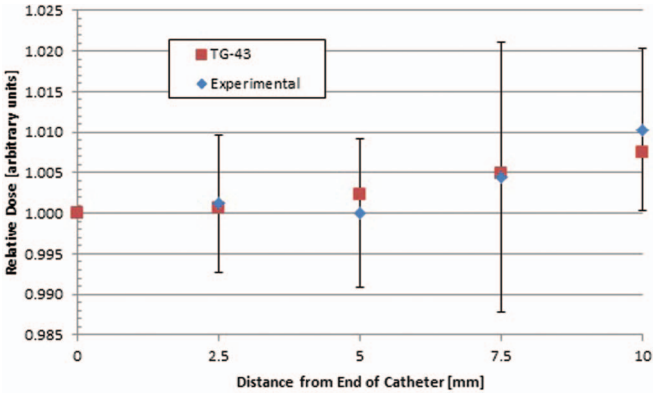

FIG. 5. Experimental and TG-43-calculated relative dose measured after incrementally retracting the catheter from the end of the channel. The relative doses at each position are normalized to the dose measured with the catheter inserted to the end of the phantom channel. Error bars represent 1 SD.

3.B. Correction factors

The linearity correction factor k_L is shown in Fig. 6. The linear best fit for k_L is given in Eq. (4):

$$k_L = \left(-9.433 \times 10^{-5}\right) \times \text{dose} + 1.009, \tag{4}$$

where *dose* is the nominal dose in cGy. The uncertainty in the linear fit in the region of 90 cGy to 110 cGy is $\sigma \approx 0.15\%$ ($k = 1$). Because 100 cGy is the target dose for the proposed remote audit program, this quantity was used as the estimated uncertainty in k_L encountered during a remote audit (see Table III).

The results of measurements of the block correction factor k_B are shown in Table I for both the Nucletron and Varian HDR sources. We measured k_B to be 1.026 [95% confidence interval (CI): 1.023–1.028] for the Nucletron source and 1.000 (95% CI: 0.995–1.005) for the Varian source. The 2.6% difference in the correction factor between the two source models was significant ($p < 0.001$) and possible explanations for this difference are discussed below.

3.C. Trial remote audits

The results of the eight trial audits are shown in Table II. The average RPC-measured dose to institution-reported dose ratio for the eight audits was 1.000 [standard deviation (SD) = 0.011]. The average ratio for the Nucletron sites was 1.004 (SD = 0.012) and for the Varian sites was 0.996 (SD = 0.010). The ratio of the RPC-measured OSLD dose to institutional treatment planning system dose was within 2% of unity for each of the eight sites that participated in the audit.

No information was collected from the participants regarding phantom orientation or location during irradiation. The instructions provided with the phantom suggested that the participants should place the phantom in whatever position they found convenient. Any differences in setup which may have occurred between the eight audits did not result in an outlying measurement.

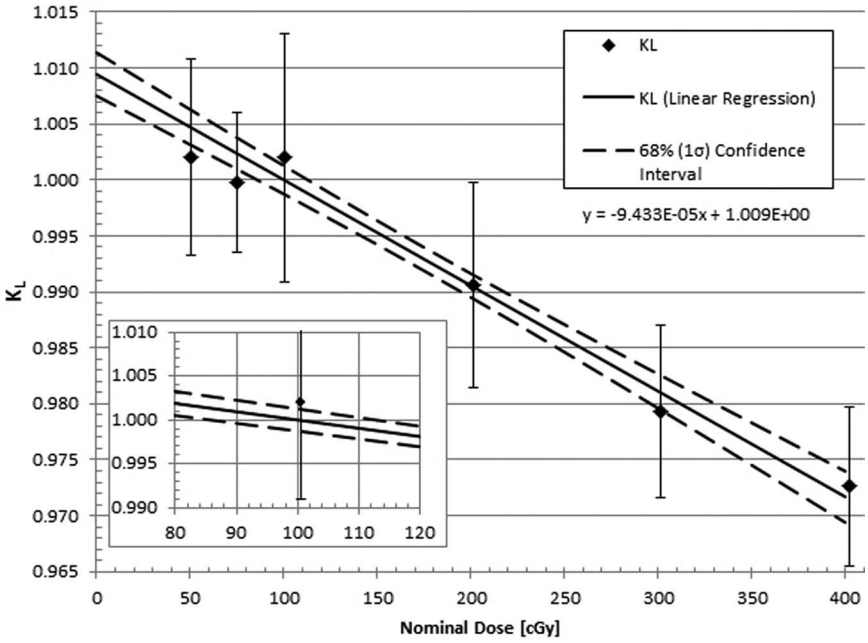

FIG. 6. The linearity correction factor. Dosimeters were irradiated to doses ranging from 50 cGy to 400 cGy and a simple linear regression applied. The inset shows the region from 80 cGy to 120 cGy. Error bars represent 1 SD.

The mean audit time of the trial remote audits as reported by the participating physicists was 69 min (range: 15–120 min; median: 60 min).

4. DISCUSSION

The linearity correction factor determined in the present study differed slightly from that used by the RPC for external beam dosimetry using the same batch of dosimeters. At 105 cGy, the difference in the two correction factors was 0.06% and at 110 cGy it was 0.08%. The difference grows as the nominal dose moves away from 100 cGy, since both factors are normalized at that point. The difference is of little consequence to the HDR audit program proposed here, which will be based on delivered doses of 100 cGy. However, the difference does indicate that the dose linearity behavior of nanoDot OSLDs may not be independent of energy or irradiation geometry. Indeed, Reft[17] found that OSLD reading per dose was not constant with photon energy. Furthermore, previous RPC dosimeter batch commissioning results indicate that the linearity correction factor k_L is OSLD batch-dependent. Hence, the current RPC practice is to determine linearity behavior on a batch-by-batch basis. For these reasons, new batches of nanoDot OSLDs introduced must be commissioned specifically for use with the HDR audit tool proposed here.

Jursinic[8] found that the sensitivity of Al_2O_3:C OSLDs under ^{192}Ir irradiation was 6% higher than that of the same dosimeters under irradiation by a 6 MV x-ray beam. Energy dependence was a major component of the block correction factors found in this work, which were both smaller in magnitude than 6%. This discrepancy is due to the block correction factor described in the present study accounting for several other effects in addition to energy dependence. The extra effects that were corrected for were lack of full scatter conditions, angular dependence of the nanoDot dosimeters, and the use of polystyrene instead of water as the phantom medium. All of these factors would be expected to reduce the measurement sensitivity of the system, somewhat counterbalancing the known energy dependence of Al_2O_3:C OSLDs.

TABLE I. Results of 20 and 10 measurements used to determine the block correction factors (k_B) for Nucletron and Varian sources, respectively, and the standard deviation of each factor.

	$k_B^{\text{Nucletron}}$	k_B^{Varian}
n	20	10
Average	1.026	1.000
SD	0.006	0.007
95% CI	1.023–1.028	0.995–1.005

TABLE II. Results of eight trial remote audits using the phantom and nanoDot OSLDs.

Source model	Trial	RPC measured dose (cGy)	Institution reported dose (cGy)	RPC dose/ institution dose	Average RPC/Inst	SD
Nucletron	1	99.9	101.0	0.989	1.004	0.012
	2	100.5	100.6	0.999		
	3	102.1	100.7	1.014		
	4	101.3	100.1	1.012		
Varian	1	100.4	99.9	1.005	0.996	0.010
	2	100.0	99.9	1.001		
	3	98.3	100.0	0.983		
	4	99.4	99.8	0.996		
Overall					1.000	0.011

We speculate that the 2.6% difference in the block correction factors of the Nucletron and Varian sources is due to differences in the sources' physical geometry. The density thicknesses of both the iridium core and the source encapsulation for the Nucletron source are greater than those of the Varian source. Previous studies have identified differences in the water-kerma[18] and emitted spectra[19] of the two sources; thus, it was not unexpected that the sources' block correction factors would differ somewhat. However, it should be stressed that the block correction factors measured in this work are only valid for two specific HDR sources — the Nucletron microSelectron v2 and the Varian VS2000. New block correction factors must be measured in order to use the tool with any other source models or substantially revised future versions of the models studied.

It is also expected that the block correction factor for each source model will be OSLD batch-dependent. Preliminary measurements using a new batch of OSLD nanoDots currently being commissioned at the RPC and the methods described in this work revealed values for k_B that were 0.9% and 0.4% higher than the values reported here for the Nucletron and Varian source, respectively. This finding is consistent with the RPC's existing practice of measuring k_L anew for each batch of nanoDots. The specific reason for the observed batch-to-batch variations is not well understood at this time.

The results obtained from eight remote audits performed in the present study are similar to those of previous RPC well chamber measurements. From 1994 to 2011, the RPC performed 193 HDR source strength audits using a well chamber during onsite dosimetry review visits to participating institutions. The average RPC-to-institution source strength ratio for those measurements was 1.009 (SD = 0.014) which was within 1% of the dose ratio for the OSLD measurement system described here, which was 1.000 (SD = 0.011). These data support the use of this new HDR remote audit tool as equivalent to performing measurements during onsite review visits.

The uncertainty in dose measurements using the phantom and OSLDs was calculated following the formalism of Eq. (1). The RPC's internal evaluation of its current OSLD program[20] provided the percent uncertainty in the first four terms in Eq. (1): reading × ECF, sensitivity, and k_F. These terms are specific to the RPC's use of nanoDot OSLDs but do not depend on the specific modality of irradiation. For measurements made in the controlled setting of its formal audit programs, the RPC has determined that $\sigma_{\text{reading}\times\text{ECF}} = 0.57\%$. This is the uncertainty in the readout of a single experimental dosimeter irradiated to an unknown dose multiplied by that dosimeter's unique correction factor. Furthermore, the corresponding uncertainty in *sensitivity* was found to be $\sigma_{\text{sensitivity}} = 0.8\%$ and the uncertainty in the fading correction factor was $\sigma_{k_F} = 0.3\%$. All uncertainties are quoted at the $k = 1$ level.

The percent uncertainty in the final two terms in Eq. (1) was determined in the present study. To determine the one standard deviation percent uncertainty in k_L, we used the 68% confidence interval in the linear regression fit of k_L in the region of 90 cGy to 110 cGy (Fig. 6). This dose range was selected because institutions participating in remote audits using the phantom will be asked to deliver 100 cGy to the dosimeters. The confidence interval in this region of the k_L linear fit was 0.15%; therefore, $\sigma_{k_L} = 0.15\%$. The standard deviations of the 20 and 10 individual measurements of k_B for the Nucletron and Varian HDR sources, respectively, were used as the uncertainty in the stated block correction factors. These were $\sigma_{k_B^{\text{Nucletron}}} = 0.6\%$ and $\sigma_{k_B^{\text{Varian}}} = 0.7\%$. Because these standard deviations in k_B arose directly from measurements, they were expected to include the relatively small uncertainty introduced by deviations in lateral source positioning described above. Thus no separate term for this particular positioning uncertainty was included in our analysis. Similarly, no term was included for differences in table or wall scatter due to institution-specific phantom setup because of the consistency of our remote audit results.

TABLE III. Uncertainty budget for dose measurements, based on the formalism in Eq. (1).

Quantity	Uncertainty (%)		Source
	Nucletron	Varian	
Reading × ECF	0.57	0.57	RPC audit programs (Ref. 20)
Sensitivity	0.8	0.8	RPC audit programs (Ref. 20)
k_F	0.3	0.3	RPC audit programs (Ref. 20)
k_L	0.15	0.15	Fit of k_L in region of 90 cGy to 110 cGy
k_B	0.6	0.7	k_B measurements (Sec. 3.B)
Total (1σ)	1.2	1.3	
Total (2σ)	2.4	2.5	

A complete uncertainty budget is given in Table III. The uncertainties in all terms in Eq. (1) were added in quadrature to determine the percent uncertainty in *Dose*, or the dose measured with the phantom and OSLDs. The overall uncertainty in dose measurements was 2.4% for Nucletron sources and 2.5% for Varian sources ($k = 2$).

Note that the analysis in Table III includes only Type A uncertainty in k_B. However, our retrospective analysis of 26 total measurements made with three different Nucletron sources and one Varian source revealed a standard deviation of 1.2%. Furthermore, the standard deviation among eight trial audit measurements was 1.1% (see Sec. 3.C). The excellent agreement between these observational results and the calculated uncertainty detailed in Table III informed our decision to forego an estimation of additional Type B uncertainties in k_B.

5. CONCLUSION

The OSLD-based HDR remote audit tool we created offers dose measurement uncertainties of 2.4% and 2.5% ($k = 2$) for Nucletron and Varian HDR sources, respectively. This level of uncertainty is sufficient to allow the RPC to establish a ±5% acceptance criterion with a confidence of >90% for remote audits in which the tool is used. Furthermore, the tool is lightweight, compact, and durable and thus may be reliably and inexpensively mailed to participating institutions. The tool is expected to become the basis for a

future official RPC audit program for ^{192}Ir HDR brachytherapy sources.

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