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ABSORBED DOSE MEASUREMENTS OF A HANDHELD 50 kV_P X-RAY SOURCE IN WATER WITH THERMOLUMINESCENCE DOSEMETERS

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Absorbed dose rate measurements of a 50 kV_p handheld X-ray probe source in a water phantom are described. The X-ray generator is capable of currents of up to 40 μA, and is designed for cranial brachytherapy and intraoperative applications with applicators. The measurements were performed in a computer-controlled water phantom in which both the source and the detectors are mounted. Two different LiF thermoluminescence dosemeter (TLD) phosphors were employed for the measurements, MTS-N (LiF:Mg,Ti) and MCP-N (LiF:Mg,Cu,P). Two small ionisation chambers (0.02 and 0.0053 cm^3) were also employed. The TLDs and chambers were positioned in watertight mounts made of water-equivalent plastic. The chambers were calibrated in terms of air-kerma rate, and conventional protocols were used to convert the measurements to absorbed dose rate. The TLDs were calibrated at National Institute of Standards and Technology (NIST) in terms of absorbed dose rate using a ^{60}Co teletherapy beam and narrow-spectrum X-ray beams. For the latter, absorbed dose was inferred from air-kerma rate using calculated air-kerma-to-dose conversion factors. The reference points of the various detectors were taken as the center of the TLD volumes and the entrance windows of the ionisation chambers. Measurements were made at distances of 3–45 mm from the detector reference point to the source center. In addition, energy dependence of response measurements of the TLDs used was made using NIST reference narrow spectrum X-ray beams. Measurement results showed reasonable agreement in absorbed dose rate determined from the energy dependence corrected TLD readings and from the ionisation chambers. Volume averaging effects of the TLDs at very close distances to the source were also evident.

INTRODUCTION

The handheld X-ray probe source for neurosurgery was introduced in the mid 1990s[1,2]. The dosimetry challenge with this device is the lack of national standards for absorbed dose for X-ray sources. Indirect means for absorbed dose calibration of this device are all that have so far been applicable. With conventional photon-emitting brachytherapy sources, a similar situation exists. However, there is a national standard for air-kerma strength for these sources[3]. The conversion between air-kerma strength and absorbed dose rate at the reference distance of 1 cm in water is accomplished with the dose rate constant, the value of which is determined by a combination of calculations and careful measurements. The 'gold standard' for such measurements is the use of small (1 mm in at least two dimensions) thermoluminescence dosemeters (TLDs)[4]. In this paper, TLD methods are applied for the first time in the dosimetry of this X-ray probe source, and compared with reference dosimetry using ionisation chambers. In a previous work[5], small plastic scintillators were also employed for the dosimetry of this source, with similar results to the present work.

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MATERIALS AND METHODS

X-ray probe source

The source used for measurements is a 50 kV_p handheld X-ray probe, manufactured by the Photoelectron Corporation. [In this paper certain commercially available products are identified by name. These identifications are for informational purposes only and do not imply that these are the best or only products available for the purpose, nor do they imply endorsement by National Institute of Standards and Technology (NIST).] The X-ray generator is capable of currents of up to 40 μA and is designed for cranial brachytherapy and intraoperative applications with applicators[6].

Automated water tank

All measurements of the X-ray probe source were performed in a specially designed automated water phantom in which the source is mounted. The tank was also equipped with waterproof detector holders made of RMI Solid Water (WT1)[7] with thin (1 mm or less) WT1 entrance windows covering the detectors. Motion of the holders was controlled such that they rotated about the source center at variable distances (1–45 mm). The probe source itself was also able to rotate under program control.

Ionisation chamber dosimetry

Reference dosimetry on the X-ray probe source was performed using two small ionisation chambers, with volumes of 0.02 and 0.0053 cm^3. Both chambers were held in the watertight mounts described above, with covering layers of 1.0 and 0.5 mm of water-equivalent plastic, respectively. The ionisation chambers were calibrated in terms of air-kerma rate using conventional protocols to convert measurements to absorbed dose rate[8–13].

TLDs and calibrations

Two TLD phosphors were employed for the measurements, MTS-N (LiF:Mg,Ti) and MCP-N (LiF:Mg,Cu,P). Both phosphors were obtained from TLD Poland, Krakow. For each phosphor, 4.5 mm diameter (0.9 mm thick and 2 mm diameter (0.5 mm thick dosemeters were used for measurements, with the smaller dosemeter used for measurements closer to the source. The chips were read out in a Harshaw 3500 TLD reader at a heating rate of 10°C s^{-1}, to 300°C for the MTS-N chips and to 240°C for the MCP-N chips. The MTS-N chips were annealed after readout at 400°C for 1 h and then after a 30 min cooling, at 100°C for 1 h. The MCP-N was simply annealed at 240°C for 10 min. For each of the determinations described below, at least six chips were irradiated individually at each irradiation depth (for the probe irradiations), energy (for the energy dependence irradiations) or dose level (for the absorbed dose calibration) to achieve sufficient measurement reproducibility. In between the above irradiations, all chips in the batch were uniformly irradiated using either $^{90}Sr/^{90}Y$ or ^{60}Co to obtain individual chip sensitivities to correct for chip-to-chip variations in response within a batch. All glow curve data was imported to Excel spreadsheets where small x-axis translations were performed to align the principal dosimetry peaks for consistent region-of-interest (ROI) analysis.

The TLDs were calibrated at NIST in terms of absorbed dose rate to water using a ^{60}Co teletherapy beam. The same TLD holders were used for these calibrations, performed at a depth of 1 cm in polystyrene with a 10×10 cm^2 field size and a 3 mGy s^{-1} absorbed dose rate.

TLD energy dependence

Experimental values for the energy dependence of the response of the various TLD phosphors employed in the study were determined by irradiations in NIST narrow spectra X-ray beams with known air-kerma rates. The same holders used for the probe irradiations were used for the energy dependence irradiations. The holders were mounted in a $30 \times 30 \times 30$ cm^3 polystyrene phantom such that the TLD and holder surfaces were flush with the phantom surface.

Absorbed dose delivered was calculated from the air-kerma calibration by using conversion coefficients between absorbed dose at a depth of 0.07 mm in water and air-kerma[14]. In addition, the energy deposition within the TLD volume was considered by an energy-dependent thickness correction. Thus the absorbed dose assigned to a TLD during an X-ray calibration was given by the following equation.

$$D(Q) = K_a\, h_K(0.07, Q) C(t, Q), \quad (1)$$

where $D(Q)$ is the assigned absorbed dose for quality Q, K_a is the delivered air-kerma, $h_K(0.07,Q)$ is the spectrum-averaged conversion coefficient between air-kerma and absorbed dose to water at 0.07 mm for quality Q, and $C(t,Q)$ is the depth-dose correction for quality Q and detector thickness t. This correction was calculated from published depth-dose data[15], which were fitted and parameterised in terms of beam half-value layer, and then integrated over the detector thickness. The various values used for these conversions are shown in Table 1. The X-ray

Table 1. Data for the conversion of air-kerma to average absorbed dose.

Beam quality (Q)	GSF catalog no.	Average energy (keV)	Conversion coefficient, h_K (0.07,Q)	Depth correction, $C(t,Q)$	
				$t = 0.9$ mm	$t = 0.5$ mm
H10	2	8.0	0.900	0.689	0.803
H15	7	11.8	0.959	0.886	0.934
H20	13	15.7	0.978	0.946	0.971
H30	26	24.2	1.080	0.992	0.997
H40	37	32.5	1.238	1.001	1.002
NS60	54	47.3	1.538	1.003	1.003
M30	21	19.7	1.001	0.946	0.971
M40	32	24.8	1.057	0.977	0.989
L40	31	23.4	1.028	0.962	0.980
L50	42	28.0	1.086	0.978	0.990

spectra used for the spectrum averaging of the published monoenergetic conversion coefficients were obtained from a catalogue of spectra[16], while the NIST and ISO X-ray beam qualities employed are described in NIST Special Publication 250–258[17].

The energy dependence of the response of the TLDs employed were expressed as the ratio of the response per unit absorbed dose for X-rays of quality Q and ^{60}Co gamma rays. For the fitting of this function, only the values for the narrow spectra shown in Table 1 were used. These values are shown in Table 2 for the two phosphor types employed. The fitted functions, as well as the data in Table 2, are shown in Figure 1, along with the ratio of the mass-energy absorption coefficients of LiF relative to water[18]. The absorbed dose interpretation of TLDs irradiated by the X-ray probe source at depth d, $D_{\text{probe}}(d)$, is determined from the response (light output), M, by the following equation.

$$D_{\text{probe}}(d) = MC_{60_{\text{Co}}}K(Q,d)\text{ECF}, \quad (2)$$

where $C_{60_{\text{Co}}}$ is the calibration coefficient for the TLD batch in units of absorbed dose per unit TLD light output, $K(Q, d)$ is the energy dependence correction for the X-ray probe spectrum at depth d, and ECF is a correction for the response of the individual TLD chip. The spectrum of the X-ray probe operated at 50 kV_p was assumed to be that of the NIST L50 X-ray beam technique, with additional water filtration for the depth of measurement. The spectra were calculated from the un(water)filtered L50 spectrum using the Institute of Physics and Engineering in Medicine (IPEM) Report 78

Table 2. Measured energy dependence of response to low energy photons in the measurement geometry.

Beam quality (Q)	Average energy (keV)	Energy dependence correction	
		MTS-N (LiF:Mg,Ti)	MCP-N (LiF:Mg,Cu,P)
H10	8.0	1.36	0.68
H15	11.8	1.36	0.75
H20	15.7	1.55	0.94
H30	24.2	1.64	1.21
H40	32.5	1.45	1.11
NS60	47.3	1.27	0.97

Table 3. X-ray probe depth and energy corrections.

Depth (mm)	Chip thickness (mm)	Probe depth and energy correction, $K(Q, d)$	
		MTS-N (LiF:Mg,Ti)	MCP-N (LiF:Mg,Cu,P)
3	0.5	0.69	1.055
5	0.5	0.67	0.98
10	0.5	0.66	0.93
15	0.5	0.66	0.92
20	0.5	0.67	0.91
25	0.5	0.67	0.91
30	0.5	0.67	0.92
30	0.9	0.69	0.92
35	0.9	0.69	0.92
40	0.9	0.69	0.92
45	0.9	0.69	0.92

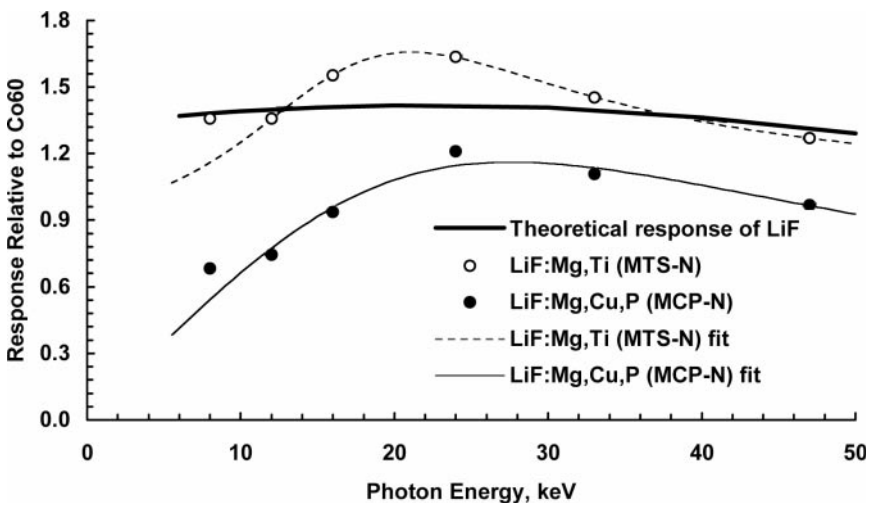

Figure 1. Measured TLD energy dependence of response relative to ^{60}Co gamma rays.

Spectrum Processor[19], and then used to obtain spectrum-averaged values for the energy dependence. The values used for these $K(Q, d)$ corrections for the various depths and phosphors employed are given in Table 3.

Probe irradiation geometry

The TLDs were held in watertight mounts made of water-equivalent plastic with a covering layer of 0.5 mm of water-equivalent plastic. Individual TLDs were mounted in solid water holders with holes milled out such that the TLD surfaces were flush with the holder surface. The holders are the same shape as the 0.0053 cm^3 ionisation chamber (PTW Model 34013) so they fit in the watertight mounting in the automated water tank. Irradiations were performed using integrated current to yield absorbed doses within the range of absorbed dose over which the TLDs were calibrated.

RESULTS

The TLD results are shown in Figure 2, compared with the reference dosimetry performed with the PTW 34013 ionisation chamber. Except at distances <10 mm where the TLD-measured dose rate was less than for the ion chamber, agreement was within the measurement uncertainties. It is likely that this exception was due to volume averaging effects within the TLD at these close distances

Table 4. Measurement of uncertainties.

Component	Uncertainty (%)	
	Type A	Type B
A. TLD calibration with ^{60}Co		
TLD response after individual response correction	4–8	
Absorbed dose rate in ^{60}Co beam at 1 cm depth		3
Combined standard uncertainty	5–8.5	
B. Energy dependence correction		
TLD response after individual response correction	1–4	
Air-kerma rate in calibration fields	0.19	0.37
Absorbed dose conversion from air-kerma		5
Depth-dose correction		2
Fitting error		2
Assumption of L50 spectrum equivalent to X-ray probe spectrum		3
Combined standard uncertainty	6–7.2	
C. Absorbed dose rate from X-ray probe		
TLD response after individual response correction	1–6	
TLD calibration with ^{60}Co (see A above)	5–8.5	
Energy dependence correction (see B above)	6–7.2	
Combined standard uncertainty	7.8–12.7	

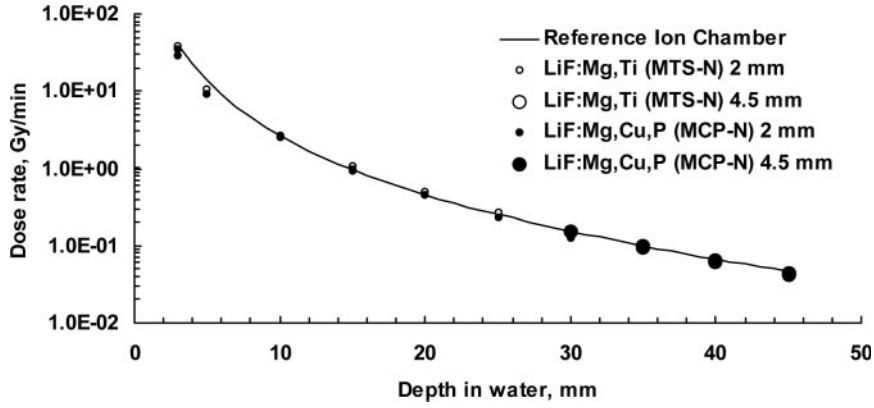

Figure 2. Measured depth-dose profile for the 50 kV X-ray probe source.

where the dose rate gradient is extremely high. These effects are not accounted for by the simplified model employed in the energy dependence determination above (Equation 1).

The TLD measurements are estimated to have a combined standard uncertainty (1 SD) between ~8 and 13%. Components in this assessment are shown in Table 4. The Type A uncertainties shown in each of the three parts of the table for TLD response after individual response correction are representative of the actual measurement results from more than six replicates. In general, the larger TLDs yielded better reproducibility, and the ~10 times more sensitive MCP-N phosphor yielded better reproducibility than the MTS-N phosphor.

CONCLUSIONS

Absorbed dose rate from a 50 kV_p handheld X-ray probe source has been measured using TLDs with energy dependence corrections determined by measurements in calibrated X-ray fields. It is anticipated that these methods, as well as the results using other suitable measurement systems such as small volume liquid ionisation chambers, sensitive radiochromic film and novel micro scintillator systems, will lead to the development of an absorbed dose standard for low-energy photon-emitting brachytherapy sources.

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