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ScienceDirect
Radiation Measurements 43 (2008) 879–882
Radiation Measurements
www.elsevier.com/locate/radmeas

Evaluation of the relative effectiveness of LiF-based TL detectors for electron radiotherapy beams over the energy range 6–20 MeV

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Abstract

We have performed systematic measurements of the efficiency, $C_E = \langle \mathrm{TLO}_\gamma \rangle / \langle \mathrm{TLO}_E \rangle$, of MTS-N (LiF:Mg,Ti) and LiB ($Li_2B_4O_7$: Mn, Si) thermoluminescence (TL) detectors exposed to electron radiotherapy beams of nominal energy 6, 9, 12, 16 and 20 MeV relative to Co-60 or 6 MV photon beams ($\langle \mathrm{TLO} \rangle$ is the average TL output of a batch of detectors exposed to a given dose of photon beam radiation or to a beam of electrons of energy E). The TL detectors were sintered pellets of 4.5 mm diameter and ca. 0.8 mm thickness. Detectors were exposed in water to 2 Gy at respective energy-dependent depths $d_{\max}$ applied in clinical dosimetry. The obtained results were re-calculated versus mean electron energy in the beam at depths $d_{\max}$. We found that, for clinical purposes, both types of TL detectors show no energy dependence: presented as average values over the investigated energy range, for MTS-N detectors $\langle C_E \rangle = 1.07 \pm 1.1\%$ and for LiB $\langle C_E \rangle = 1.03 \pm 1.7\%$. Thus, while no correction is required for dose values estimated by LiB TLDs, the under-response of MTS-N detectors exposed to a beam of electrons has to be corrected by about 7%. At mean energy of electrons in therapeutic electron beams below about 5 MeV a decline in the response of both types of TL detectors is observed.

Keywords: TL; LiF:Mg,Ti; MTS-N; Cu,P(MCP-N); LiB ($Li_2B_4O_7$: Mn, Si); Relative efficiency; Electron radiotherapy beams

1. Introduction

Thermoluminescence (TL) detectors are currently applied in interstitial, in-phantom and *in vivo* clinical dosimetry, mainly for quality control in photon beam radiotherapy. There is an interest in extending the clinical applications of TL dosimetry to electron beams. However, unlike in MV photon beams over regions of electron equilibrium, the response of a TL detector placed in a phantom exposed to a beam of electrons in the MeV energy range may depend on several factors, such as the mean electron energy at the phantom and detector surfaces, detector size, density of the detector and of the phantom medium or depth in the phantom at which the detector is irradiated. The mean energy of electrons in a radiotherapy beam, $\bar{E}_z$, decreases with beam penetration depth, z, according to the relationship:

$$\bar{E}_z = \bar{E}_0(1 - z/R_\mathrm{p}), \tag{1}$$

where $\bar{E}_0$ is the mean entrance energy of the beam and R_p is its penetration or extrapolated range (for a discussion, see Klevenhagen, 1993).

Earlier studies of the energy dependence of the relative response of LiF-based TL detectors after electron beam irradiations gave inconsistent results (Klevenhagen, 1993). Robar et al. (1996) and Ginjaume et al. (1999) reported no energy dependence over nominal beam energy range of 6–18 MeV. Robar et al. (1996) measured and calculated relative effectiveness of TLD-100 close to that of megavolt photon beams. Our interest in this subject was also stimulated by our own work on verifying external electron beam therapy planning systems using an anthropomorphic phantom in which lithium borate TLDs

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doi:10.1016/j.radmeas.2007.11.056

were used (Waligórski et al., 2006), where, to our knowledge, no such data were available.

We represent the relative effectiveness of a TL detector, C_E, as the ratio

$$C_E = \langle \mathrm{TLO_{ref}} \rangle / \langle \mathrm{TLO}_E \rangle, \quad (2)$$

where $\langle \mathrm{TLO_{ref}} \rangle$ and $\langle \mathrm{TLO}_E \rangle$ are the average values of TL readouts for a party of detectors exposed together to a dose of 2 Gy in a reference beam (Co-60 γ-rays or 6 MV X-rays) and to 2 Gy in an electron beam of nominal energy E, at its respective value of maximum dose depth, $d_{\max}$, recommended by clinical dosimetry protocols (IAEA, 1997). In our systematic study, we have so far measured C_E for LiF:Mg,Ti and $Li_2B_4O_7$: Mn, Si TL detectors, both in the form of solid pellets of diameter 4.5 mm and thickness ca. 0.8 mm. These types, and other LiF-based detectors (e.g. LiF:Mg,Cu,P), are commonly applied in clinical dosimetry.

2. Materials and methods

MTS-N (TLD Poland, IFJ PAN Kraków) and LiB (Alnor, Sweden) TL detectors were used in our study. Both were in the form of sintered pellets of 4.5 mm diameter and ca. 0.8 mm thickness.

MTS-N (natural LiF:Mg,Ti) TL detectors were annealed for 1 h at 400 °C and for 2 h at 100 °C and next cooled rapidly to room temperature. Additionally, after exposure, the detectors were annealed for 10 min at 100 °C to eliminate unstable low-temperature peaks. MTS-N detectors were read out with a HARSHAW 3500 reader (50–280 °C glow-curve integration) without nitrogen gas flow.

LiB (natural $Li_2B_4O_7$: Mn, Si) sintered detectors were read out using a DOSACUS hot-gas reader (N_2 at 300 °C, pre-heat 1.5 s, readout 10.5 s, anneal 40 s). Other than within detector readout, no further annealing was applied.

Individual response factors (IRF) were established for each detector:

$$\mathrm{IRF}_i = \langle \mathrm{TL_{ref}} \rangle / \mathrm{TL}_i, \quad (3)$$

where $\langle \mathrm{TL_{ref}} \rangle$ is the mean value of readouts obtained for the whole party of detectors and TL_i is the value read out for the ith detector in this party. IRFs were determined using reference beams (Co-60 γ-rays or 6 MV X-rays) at 2 Gy and the stability of the IRF of each detector verified over about 10 reference exposure–readout cycles. Detectors with unstable IRFs over these cycles (SD > 5%) were not used in measurements of C_E.

In measurements of C_E of MTS-N detectors, 98 most stable detectors were selected after 10 IRF evaluation cycles, of which 18 detectors were used in reference irradiations while five groups of detectors (16 per group) were irradiated by electron beams of nominal energies at depths $d_{\max}$ listed in Table 1. All irradiations were performed on the same day.

Due to the limited number of LiB detectors available (30), 20 most stable detectors were selected after seven IRF evaluation cycles and used in measurements of C_E in batches of 10 detectors per beam energy in electron beam exposures, as listed in Table 1, both batches being exposed on the same day.

Table 1
Nominal energy of electron beams and respective depths in water, $d_{\max}$, at which detectors were exposed

Nominal energy (MeV)	$d_{\max}$ (cm)
6	1.4
9	2.0
12	3.0
16	3.0
20	2.4

For both types of TL detectors, all measurements of C_E were repeated three times.

A CLINAC 2300 medical accelerator was used to deliver electron beams of nominal energies 6, 9, 12, 16 and 20 MeV. Detectors were exposed in water, in small sealed water-tight polyethylene bags from which air was evacuated to the extent possible.

3. Results

The measured values of relative efficiency C_E of MTS-N and LiB TL detectors are listed in Table 2. Percent errors at each energy were calculated as square roots of sums of standard deviations obtained in each of the three experiments squared. In this table the energy of electrons in the beam is given either as its “nominal” (accelerator setup) value or as the “extrapolated” value, i.e. value of mean energy of electrons in a radiotherapy beam, $\bar{E}_z$, at the depth $d_{\max}$ corresponding to the given nominal energy, and calculated from Eq. (1) using values of $d_{\max}$ and nominal values of beam energy listed in Table 1.

The energy dependences of the relative effectiveness C_E of MTS-N and LiB detectors at 2 Gy, versus mean electron energy $\bar{E}_z$ in the beam at $d_{\max}$, are plotted in Fig. 1. Error bars shown represent those listed in Table 1.

4. Discussion

In view of the complex dependence of the electron energy response of TL detectors on their irradiation conditions (Mobit et al., 1996), considerable care was exercised in our measurements. Multiple cycles of photon exposures of our detectors served not only to eliminate detectors with unstable IRFs, but also stabilised the detectors themselves. Each value of C_E given in this study is a result of three independent experiments. In calibration runs where X-ray beams from the CLINAC 2300 accelerator were used, the daily variation in the accelerator output was accounted for. The onset of supralinearity in MTS-N detectors at 2 Gy was also corrected for, following measurements of X-ray dose response over the range of 0.5–5 Gy.

Our results demonstrate that for clinical purposes, the energy response of both types of TL detectors may be assumed to be constant. If presented as average values over the investigated energy range, for MTS-N detectors $\langle C_E \rangle = 1.07 \pm 1.1\%$ and for LiB $\langle C_E \rangle = 1.03 \pm 1.7\%$ (1 SD). Thus, while no correction is required for dose values estimated by LiB TLDs, the under-response of MTS-N detectors exposed to a beam of electrons

Table 2
Relative efficiency at 2 Gy measured for MTS-N and LiB TL detectors exposed to electron radiotherapy beams of energy given as "nominal" (accelerator setup) or "extrapolated" (mean electron energy at d_{max}) values

Energy (MeV)		Relative efficiency C_E			
Nom.	Ext.	MTS-N		LiB	
		C_E	δC_E (%)	C_E	δC_E (%)
6	3.1	1.054	1.65	1.024	1.38
9	4.9	1.082	1.63	1.055	1.48
12	6.1	1.078	1.62	1.045	1.57
16	10.0	1.067	1.59	1.012	1.29
20	15.4	1.071	1.61	1.031	1.26

For the calculation of percent errors, see text.

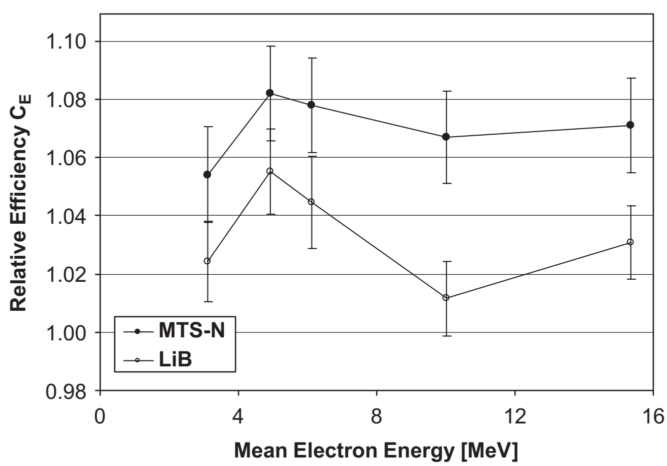

Fig. 1. Relative efficiency C_E at 2 Gy measured for MTS-N and LiB TL detectors versus mean energy $\bar{E}_z$ of electrons at d_{max} in a radiotherapy beam. Errors plotted represent those listed in Table 1.

should be upwards-corrected by about 7%. However, this potential advantage of LiB detectors over MTS-N may be offset by the distinct difference in fading characteristics (up to 5% in 30 days for LiB, below 1% per year for MTS-N) or TL light spectra (around 600 nm for LiB, against 400 nm for MTS-N) between these detectors.

Results of our measurements of C_E appear to show more variation with mean electron energy than those measured by Robar et al. (1996) and by Mobit et al. (1996), calculated by Mobit (2002), for TLD-100 rods and chips, and measured by Ginjaume et al. (1999) for ^{7}LiF: M, Cu, P chips (Chinese TLD-700H). The value of $C_E = 1$ at mean electron energies above 7 MeV measured and calculated for TLD-100 (Robar et al., 1996; Mobit et al., 1996; Mobit, 2002) differs from ours ($C_E = 1.07$) for MTS-N pellets. We note that measurements and calculations of other authors were carried out in solid (Perspex or polyethylene) phantoms, while our exposures were performed in water, and that measurements of Ginjaume et al. were referred to their 10 MeV electron beam. The generally calculated fall-off of relative effectiveness at mean electron energies below 5 MeV appears to be supported by our experimental results. We also support the calculations of those authors, whereby at the distal end of the electron beam (where electrons in the beam reach their final range), the observed output of the TL detector is due predominantly to the sizeable contribution of bremsstrahlung photons, leading to an apparent enhancement of the measured value of C_E. It will be interesting to determine whether the under-response we found for LiF:Mg,Ti will also hold for the much more sensitive LiF:Mg,Cu,P (MCP) material, in the form of sintered pellets or TLD foils (Olko et al., 2007), which we are currently studying.

5. Conclusions

We have measured the efficiency, C_E, at 2 Gy, of MTS-N (LiF:Mg,Ti) and LiB ($Li_2B_4O_7$: Mn, Si) TL detectors exposed in water to electron radiotherapy beams of nominal energy 6, 9, 12, 16 and 20 MeV relative to Co-60 or 6 MV photon beams. Our measured values of C_E averaged over the entire 6–20 MeV electron beam energy range are $\langle C_E \rangle = 1.07 \pm 0.01$ and 1.03 ± 0.02 (1 SD), respectively, for MTS-N and LiB detectors. These values can be used to correct the measured TL signal in clinical applications of these detectors for electron fields over 6–20 MeV beam energy range. Similar measurements for MCP-N detectors and 2-D foils (both LiF:Mg,Cu,P) are under way.

Acknowledgements

The authors are grateful to Dr. Simo Hyödynmaa, Tampere University Hospital (now at Kuopio University), Finland, for providing them with the LiB detectors, and to Mr. Pawel Zawadzki, now at COOK, for his contribution in performing the measurements of MTS-N detectors within his M.Sc. thesis at the University of Mining and Metallurgy, AGH, in Krakow.

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