ORIGINAL PAPERS

DEVELOPMENT OF TL DOSIMETERS BASED ON MTS-N (LiF:Mg,Ti) DETECTORS FOR *IN VIVO* DOSIMETRY IN A CO-60 BEAM[+)]

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INTRODUCTION

Of the three principal techniques used for *in vivo* dosimetry: ionisation chambers, semiconductor diodes and thermoluminescent dosimeters (TLDs), the last method, despite its lack of real-time readout, offers some advantages, such as elimination of supply cables or lack of temperature, dose-rate and energy dependence. Considerable experience in the production of TLDs has been gained over a period of the last 25 years or so, at the Institute of Nuclear Physics (INP) in Kraków (Niewiadomski, 1991; Niewiadomski, 1996) and in the clinical application, mainly of LiF:Cu,Ag in powder form, at the Centre of Oncology in Kraków, COK (Szymczyk et al., 1971). Sintered LiF:Mg,Ti pellets of diameter 4.5 mm and thickness 0.8 mm are being commercially produced at the INP under the trade name MTS-N (N stands for natural $^{6}Li/^{7}Li$ abundance), being for all practical applications equivalent to Harshaw-produced TLD-100 extruded ribbon chips or pellets. Most of the experience gained with TL dosimetry at the INP has been in the areas of individual and environmental monitoring. It was therefore interesting to study more closely the clinical applications of MTS-N detectors, especially for *in vitro* dosimetry in brachytherapy (Waligórski et al., 1997) and *in vivo* dosimetry of photon beams used in teleradiotherapy. The aim of this work was to develop a MTS-N-based system for *in vivo* dosimetry in a Co-60 beam, which required a more systematic study of the accuracy, repeatability, linearity of the detectors and of the angular response of the *in vivo* dosimeter , i.e. of the detector-holder assembly.

MATERIALS AND METHODS

For all measurements, one batch of 100 INP-produced MTS-N detectors was used. The standard INP pre-irradiation annealing routine of 1 hour at 400 °C, fast quench on a thick Al block, and 2 hours at 100 °C was applied. The temperature stability of an oil-heated TLD oven (Vinten, Cambridge, England) was improved and annealing of the detectors was carried out at 400 ± 0.6 °C and 100 ± 1.3 °C. The exposed detectors were read out by a computerised TL Reader-Analyser type RA'94 (Microlab, Institute of Nuclear Physics, Kraków, Poland), using a linear ramp at 3 °C/sec. The digitally-recorded glow curves were integrated over the range 150 - 270 °C. A standard IBM/PC spreadsheet (MicroSoft Excel v. 7.0) and plotting programme (Microcal Origin, v. 4.1) were used for data analysis and display.

Irradiations were carried out using a Co-60 radiotherapy unit (ALCYON II, CGR MeV, France). The detectors were irradiated in water and plexiglass phantoms at the centre of a $10 \times 10\ cm^2$ field, at a standard source-to-surface distance, SSD = 80 cm. Detectors were exposed to a calibration dose of 1 Gy in water, as measured by a 0.6 cm^3 NE 2571 thimble chamber connected to a 2590 Ionex Dosemaster (Nuclear Enterprises, Edinburgh, Scotland).

After irradiation, immediately before readout, detectors were placed at the temperature of 100 °C for 10 min. The batch of MTS-N detectors underwent six calibration exposure-readout-annealing cycles.

An *Individual Response Factor* (IRF) was ascribed to each detector on the basis of its readout after every calibration exposure, defined as follows:

$$\mathbf{IRF_{i,k} = INT_{i,k} / AVINT_k} \qquad (1)$$

where:

i is the detector number (for the batch of 100 detectors, i = 1,...100)

k is the calibration series number (since there were 6 exposure-readout cycles, k = 1,...6),

$\mathbf{IRF_{i,k}}$ is the individual response factor for the detector number i, in the calibration series number k,

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$INT_{i,k}$ is the value of integrated TL curve for the i-th detector, in the k-th calibration series, and

$$\mathbf{AVINT_k} = \frac{1}{n}\sum_{i=1}^{n} INT_{i,k}$$ is the average value of integrated TL curve for all 100 detectors in the k-th calibration series.

The *accuracy* of the detectors was determined in each cycle by evaluating the distribution of their IRFs after each exposure.

The *repeatability* or *stability* of the detectors was determined by evaluating the distribution of the IRF of each detector over a given number of cycles.

Values of accuracy and repeatability are represented by their coefficients of variation, i.e. 100%*(SD/mean), for 100 detectors, or over 6 consecutive cycles for each detector, respectively.

Linearity of the detectors was tested over the dose range 50 cGy-250 cGy in one readout cycle.

After completing the calibration exposures, *TL dosimeters*, consisting of TL detectors placed in plexiglass holders assuring electronic equilibrium, were used for determining the absorbed dose. From the readout of the value of the integrated TL curve for the i -th detector (INT_i) the *absorbed dose in water* was evaluated as follows:

$$\mathbf{D_i = INT_i * 1/K * 1/AVIRF_i} \qquad (2)$$

where:

D_i is the absorbed dose in water for the i -th detector [cGy],

K is a factor numerically equivalent to the calibration dose [TL pulses/cGy],

$AVIRF_i$ is the value of the IRF for the i -th detector averaged over *n*=6 calibration series,

$$\mathbf{AVIRF_i} = \frac{1}{n}\sum_{k=1}^{n} IRF_{i,k}$$

The *angular response* of the TL dosimeters was studied by placing the dosimeters on the surface of a water phantom and irradiating them at different beam angles, over the range from 0° (normal incidence) to 60°. The results were then represented relative to normal beam incidence.

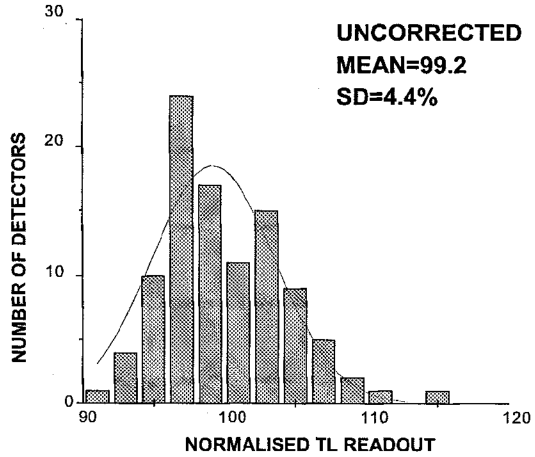
RESULTS

1. *Accuracy and Repeatability of Detectors*

Assessment of the accuracy and repeatability (or stability) of MTS-N detectors was based on a series of six consecutive exposure-readout-annealing cycles performed on a batch of 100 detectors exposed to the same calibration dose of 1 Gy.

Application of Individual Response Factors (IRF, equation 1) improved the accuracy of the batch of 100 detectors exposed to 1 Gy of Co-60, from about 4.5% („raw" value of SD, representing the manufacturer's selection accuracy, series 1, Fig.1.a) to 2.1% (series 2, IRF-corrected, Fig. 1.b).

a)



b)

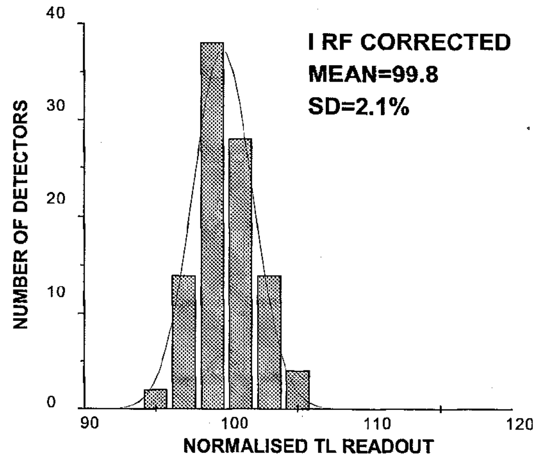
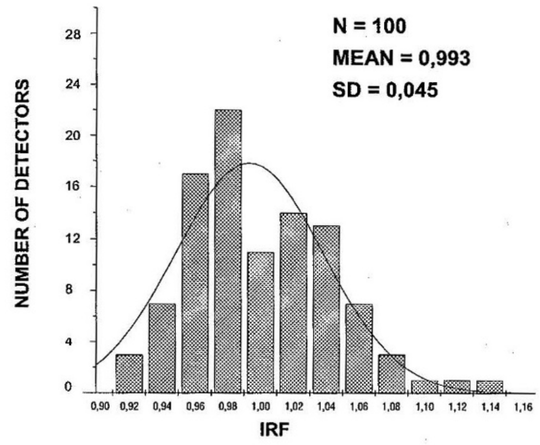


Fig. 1. Distribution of readout values for a batch of 100 detectors exposed to a dose of 1 Gy. of Co-60 gamma-rays: histogram and Gaussian; a) uncorrected, series No. 1; b) after correction of the readout of each detector by an Individual Response Factor.

To study the effect of averaging the value of IRF over the number of calibration series, an average value of IRF was calculated over six calibration series for each detector ($AVIRF_i$, cf. equation 2). Analysis of the distribution of IRF around the value 1.0 for 100 detectors in series No. 6 gave a Gaussian with an average value MEAN = 0.993 and standard deviation SD = 0.045, as shown in Fig. 2.a. The same analysis carried out for the distribution of AVIRF averaged over series 1- 6 for each of the 100 detectors yielded a Gaussian of MEAN = 0.992, and SD = 0.029 (Fig.2.b).

a)



b)

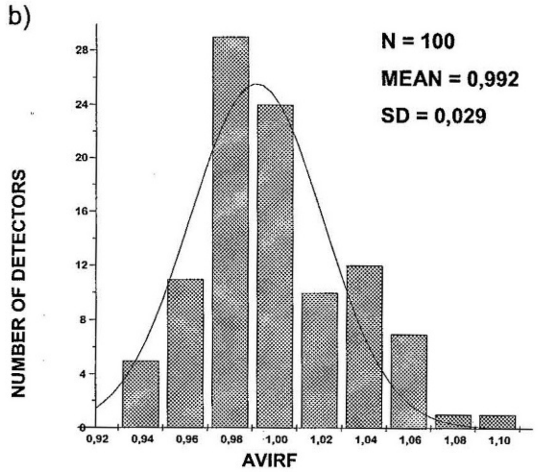


Fig. 2. Results for a batch of 100 detectors exposed to 1 Gy of ^{60}Co gamma rays: histogram and Gaussian; a) distribution of values of Individual Response Factors (IRF) in calibration series No. 6; b) distribution of values of Average Individual Response Factors (AVIRF) calculated over six calibration series, No.1 through No.6.

In order to represent the repeatability (or stability) of individual detectors, the IRF of each detector in consecutive series (1-6) was calculated according to equation (1). Next, the average relative value of IRF for a given (i-th) detector over the six series, $AVIRF_i$, was calculated as well as the standard deviation, SD, around this value. It was then possible to rank individual detectors in ascending order of repeatability (stability), as measured by this value of SD. Figure 3 shows a pie-chart distribution of the number (percentage) of detectors in given ranges of repeatability. Fig. 4 illustrates the repeatability (as represented by the relative change of IRF values over six series) for the most stable detector (No. 60), for one of average stability (No. 77) and for the least stable detector (No. 1).

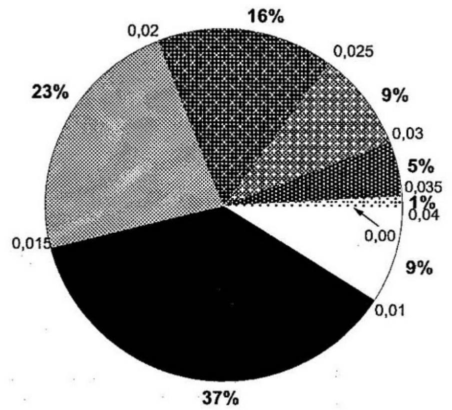


Fig. 3. Pie-diagram of detector repeatability for 100 detectors exposed to 1 Gy of ^{60}Co gamma rays, i.e. for each detector (i=1...100) a distribution of values of SD from the value of its Average Individual Response Factors (AVIRF), calculated over calibration series No.1 through No.6.

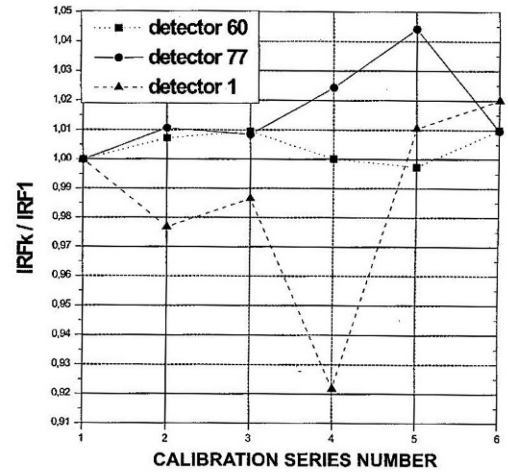


Fig. 4. Relative individual detector stability, i.e. changes of IRF in consecutive calibration series (k = 1-6) for the most stable detector (No. 60), an average one (No. 77) and the least stable detector (No. 1), relative to their IRF values in series No. 1.

2. *Linearity of detector response*

Detectors were irradiated in a single anneal-readout cycle over the dose range 50-250 cGy (50, 100, 150 and 250 cGy), 10 detectors being exposed together at each dose value (20 detectors at 100 cGy). Values of the measured dose calculated from equation (2) are shown in Fig. 5 versus the planned values. Linear regression applied in Fig. 5 to average measured dose values gave the following expression:

$$y[\mathrm{cGy}] = -5.34\ \mathrm{cGy} + 1.04\ x\ [\mathrm{cGy}] \quad (3)$$

In Table 1 the values of measured dose are compared with those calculated from equation (3) at each point. The ratios of average values of measured and planned dose are also given. Errors are given as 1 SD.

Number of detectors	Planned dose [cGy]	Measured dose [cGy]	Calculated dose [cGy]	dose (measured/ calculated)
10	50	48.2 ± 0.9	46.67 ± 2.03	1.033 ± 0.018
20	100	97.9 ± 1.4	98.68 ± 2.03	0.993 ± 0.018
10	150	150.0 ± 1.6	150.69 ± 2.03	0.995 ± 0.018
10	250	255.5 ± 2.9	254.72 ± 2.03	1.008 ± 0.018

Table 1. Values of measured dose (averaged from the TL readout of the listed number of detectors for a given planned dose), values of dose calculated from linear regression (equ. 3, see text) and ratio of measured-to calculated dose, over the dose range 50 - 250 cGy. Errors are 1 SD values.

3. *Angular response of the dosimeter*

The *in vivo* dosimeter consists of three MTS-N detectors placed centrally side-by-side in a flat cylindrical plexiglass container 5 mm thick and of diameter 20 mm. A single layer of thin Mylar foil separates the detectors from the surface of the phantom or the patient's skin. To assess the angular dependence of the dosimeter assembly, the dosimeter was placed on a water phantom and irradiated at beam angles of 0°, 15°, 30°, 45°, and 60° (0° represents normal beam incidence). At any one angle, the average value of absorbed dose from two dosimeters (six readouts of TL detectors) was calculated using equation (2) and represented relative to dose at normal (0°) beam incidence. Results are shown in Fig. 6. The relative variation in response over the range of angles of 0° through 60° remains constant within about 2% .

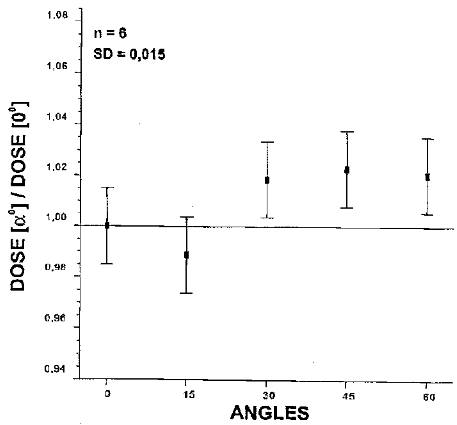


Fig. 6. Relative angular response of the *in vivo* TL dosimeters described in text, exposed to 1 Gy at beam incidence angles 0°, 15°, 30°, 45°, and 60° (0° represents normal beam incidence).; *n* is the number of detectors assigned to a given angle, average values of detector readout have been normalised to 1.0 at 0° beam incidence angle.

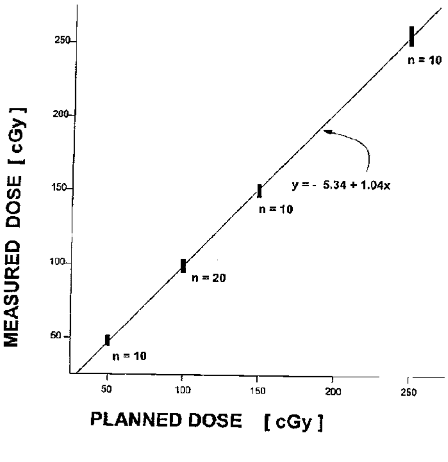


Fig. 5. Planned dose (absorbed dose in water) *versus* dose measured by the *in vivo* TL dosimeters described in text, in the range 50 cGy - 250 cGy. A straight line has been fitted by linear regression (eq. 3, see text) through the average of measured values at given doses; *n* is the number of TL detectors assigned to each given measurement point.

DISCUSSION AND CONCLUSIONS

Application of individual response factors (IRF) to solid TL detectors clearly improves the accuracy with which the detectors are able to measure absorbed dose *in vivo*, from the manufacturer's value of about 4.5% (Fig. 1.a) to 2.1% (Fig. 1.b). Another important consideration is detector repeatability, i.e. the degree to which the IRF of a given detector may vary over several irradiation-anneal cycles. It appears that averaging the value of IRF over several cycles could further improve the accuracy of dose assessment, from 4.5% (Fig. 2.a) to 2.9% (Fig. 2.b).

However, a more effective manner of improving the accuracy of dose assessment is to select a group of detectors which are stable within a set SD value, e.g. 2%. It may be seen from Fig. 3 that in our batch of 100 detectors, 69 fulfil this criterion. This kind of analysis can assure that the accuracy and reproducibility (or stability) of the selected detectors are sufficient to carry out dose measurements with a precision no worse than 2-3%.

Another important consideration is the range of linearity of the detectors. We were able to represent the dose response over the range 0.5-2.5 Gy with a linear fit sufficiently accurately for most clinical applications. As shown in Table 2, the differences between the measured values of dose and those calculated from eq. (3), also shown in relative form in Table 3, do not exceed 3% at the 0.5 Gy level and are below 1% at higher doses.

It is more practical to use a dosimeter containing three detectors, rather than one detector, except, perhaps, in special circumstances. Our plexiglass holders fulfil buildup conditions for Co-60 photons and our dosimeters show a sufficiently flat angular dependence over a range of 60^0 (Fig. 6). This is of importance for head and neck exposures, where the detector is placed at angles other than normal with respect to beam incidence.

Our *in vivo* dosimeter will now be evaluated under clinical conditions. Further work is under way to develop a system suitable for accelerator-produced photon beams.

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