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Optically stimulated Al_2O_3:C luminescence dosimeters for teletherapy: $H_p(10)$ performance evaluation

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HIGHLIGHTS

- We evaluate the performance of Al_2O_3:C OSL dosimeters using Co-60 irradiations.
- The reproducibility (< 2%), ~0.5% signal reduction and dose linearity up to 50 mSv were reported.
- We determine the $H_p(10)$ measurements accuracy (< 7%) using the trumpet curve analysis.

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ABSTRACT

The performance of optically stimulated luminescence dosimeters (OSLDs, Al_2O_3:C) was evaluated in terms of the operational quantity of $H_P(10)$ in Co-60 external beam teletherapy unit. The reproducibility, signal depletion, and dose linearity of each dosimeter was investigated. For ten repeated readouts, each dosimeter exposed to 50 mSv was found to be reproducible below 1.9 ± 3% from the mean value, indicating good reader stability. Meanwhile, an average signal reduction of 0.5% per readout was found. The dose response revealed a good linearity within the dose range of 5–50 mSv having nearly perfect regression line with R^2 equals 0.9992. The accuracy of the measured doses were evaluated in terms of operational quantity $H_P(10)$, wherein the trumpet curve method was used respecting the 1990 International Commission on Radiological Protection (ICRP) standard. The accuracy of the overall measurements from all dosimeters was discerned to be within the trumpet curve and devoid of outlier. It is established that the achieved OSL Al_2O_3:C dosimeters are greatly reliable for equivalent dose assessment.

1. Introduction

For the treatment of tumors, external beam radiotherapy (EBRT) also called teletherapy uses an externally generated high energy electron, heavy-ion, X-ray or γ-ray beam. Among the distinguished γ-emitters (such as ^{137}Cs, ^{60}Co and ^{152}Eu) used in EBRT, ^{60}Co is the most widely exploited radioactive source due to its high specific activity, relatively long half-life, suitable energy range of the emitted photons and modes of production. It is worth mentioning that the implementation of high energy γ-ray in radiotherapy offers several benefits including low radiation side scatters and lesser skin dose. These notable features of high energy γ-ray are attributed to the formation of electronic equilibrium at greater depth (Richardson et al., 1954). Whilst to ensure the human safety, it is essential to keep account of the dose levels absorbed by the patient because of the high radiation dose utilized in radiotherapy. Earlier studies on the dose estimation outside the primary beam of radiation therapy sources revealed that the scattered radiation is strongly dependent on the field size and patient to beam axis distance (Van der Giessen and Hurkmans, 1993). Thus, precise monitoring of dose became a mandatory safety feature.

In external radiotherapy, the typical doses of interest can be as much as 20 Sv or greater (dose fractionated for each cycle), while the

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amount of radiation dose received at any point by the patient's body outside the direct photon beam could originate from the scattered radiation (Van der Giessen and Hurkmans, 1993). Thus, in personal dosimetry (EBRT unit) the radiation exposure to the patient is not the only major concern but the equivalent absorbed dose by the personnel is also a primary fret. In both on and off modes of the teletherapy machine, some radiations always escape from the unit. Generally, the amount of escaped radiation during off mode remains below about 0.01 mSv/h at a distance of 100 cm away from the source. According to ICRP regulations, the average leakage from the source head should be less 0.02 mSv/h at 100 cm of source to surface distance (SSD) (Podgorsak, 2005). Regarding radiation protection, personal dosimetry is definitely reliable for the measurement of the operational quantity $H_P(10)$ which can be translated in term of equivalent absorbed dose to the skin. Over the years, different types of detectors have been tested towards personal dosimetry for high energy photons (1 mSv up to 100 mSv) dose delivery (Arib et al., 2014; Guimarães and Okuno, 2003; Ikmal et al., 2016; Kadir et al., 2013; Saraví et al., 2007; Stadtmann et al., 2014), and for quality assurance in diagnostic X-rays (Green et al., 1999). Moreover, the variability of reported dose equivalents in various studies enforced us for further evaluation of the dose distribution in teletherapy procedure. Such scrutiny would provide some supplementary data for consistent comparison and assessment of the operational quantity $H_P(10)$.

As per ICRP guidelines for personal dosimeter testing, the use of a square slab polymethylmethacrylate (PMMA) phantom having 30 × 30 cm^2 entrance surface and made of tissue-equivalent material is necessary. Furthermore, the dosimeters response relating to radiation energy being the vital characteristic must be tested to determine their performance efficiency (IAEA, 1999). The dose dependence of certain luminescence signal within the dose region of interest is the basis for radiation dosimetry implementation, where the dosimeter's dose response must be preferably linear and reproducible (Yukihara and McKeever, 2011). According to IAEA and ICRP guidelines, personnel who perform the teletherapy should not receive a dose exceeding 50 mSv to the whole body per year (IAEA, 1996; ICRP 1992). Individual doses below 50 mSv from external radiation sources can also be measured using sensitive dosimeters to ascertain the natural background contribution (Rivera-Montalvo, 2016).

This allowed us to select the exposed doses in the range of 5–50 mSv for assessing the deep dose ($H_p(10)$) performance of the InLight OSLD implemented in teletherapy unit. Presently, the commercially available OSLDs (Al_2O_3:C from Landauer Inc.) have radically aid the dosimetry services. Compared to the well-known thermoluminescence dosimeters (TLDs), the Al_2O_3:C OSLDs possess some notable attributes such as multiple re-analysis, short readout time, high sensitivity and low fading (McKeever and Moscovitch, 2003). Besides, the performance of such dosimeters have been tested in different modalities with special emphasis towards clinical therapy applications (Butson et al., 2017; Conheady et al., 2015; Hu et al., 2009; Viamonte et al., 2008) due to the presence of inherent operational risk in these procedures.

In this view, we took an attempt to test the performance of InLight OSLDs and the accuracy of measured $H_P(10)$ at delivered doses below 50 mSv in ^{60}Co teletherapy unit. Radiation exposures were subjected to a water phantom to determine the reproducibility and signal depletion of the individual dosimeters after repeated readouts, and dose linearity response. Results were analyzed, discussed, and compared.

2. Experimental methods

2.1. Materials

All irradiations were performed using Eldorado-8 unit located at the Secondary Standard Dosimetry Laboratory (SSDL) Malaysian Nuclear Agency. The Eldorado-8 unit is a teletherapy source that uses ^{60}Co radionuclide sealed inside a cylindrical stainless steel capsule where any leakage of radioactive material is prevented. The Cobalt-60 source has activity of 37 GBq and decays by emitting two γ-ray photons with energies of 1.17 MeV and 1.33 MeV, resulting to mean energy of 1250 keV (Ikmal et al., 2016). Experiments were conducted using the InLight OSLD (Landauer Inc., Glenwood, IL, USA) which is comprised of a sensitive disk shaped element of diameter 5 mm and thickness of 0.2 mm made of carbon-doped aluminium-oxide (Al_2O_3:C). In the present work, InLight microStar (Landauer Inc.) was used for absorbed dose measurements. The details of the readout process using the microStar reader and specifications of InLight dosimeters are documented elsewhere (Dunn et al., 2013; Jursinic, 2007, 2010). A water phantom of volume (30 cm × 30 cm × 15 cm) was used for various irradiations. All OSLDs were annealed before irradiations using a portable optical annealer (Pocket Annealer produced by Landauer Inc.). The Pocket Annealer could reset only one InLight dosimeter at a time and cleared doses up to 0.1 Gy.

The dose delivered to the OSLDs was calculated using the following expression that relates the γ-radiation exposure (X) to the air kerma (K_{air}) and amount of the electric charge (Q) liberated through the ionization of air of mass m (Lamperti and O'Brien, 2001):

$$X = \frac{dQ}{dm} = K_{air}\frac{1}{(W/e)}(1-g) \tag{1}$$

where dQ is the sum of the electrical charges on all ions produced in air when all the electrons liberated by γ-ray photons in a volume element of air of mass dm are completely stopped, g is the fraction of the initial kinetic energy of secondary electrons dissipated in air (which is 0.32% for ^{60}Co gamma-ray source), and W/e is the mean energy per unit charge expended in air (which is 33.97 J/C) (Lamperti and O'Brien, 2001). The value of K_{air} was estimated via (Ikmal et al., 2016; Kadir et al., 2013):

$$K_{air} = Q_t N_k K_{TP} \tag{2}$$

where Q_t is the average charge per unit time, N_k is the calibration factor of the ionization chamber and K_{TP} is the correction factor.

2.2. Reproducibility, reader stability and signal depletion characterization

The reproducibility and OSL signal depletion of individual dosimeters were tested after a single exposure. Three OSLDs were irradiated with 50 mSv from Co-60 at 2 m SSD on the water phantom over a field size of (10 cm × 10 cm). Each OSLD was repeatedly read 30 times after 24 h of post-irradiation. The reproducibility for each OSLD was determined in terms of coefficient of variation of the first ten readings, while the deviation of each reading from the mean value was used to characterize the reader stability. Each reading process involved insertion, reading, and removing the dosimeter from the reader. The signal depletion was determined from the 30 successive readouts using the model proposed by Jursinic which is provided below through Eqs. 3 and 4 (Jursinic, 2007).

$$S(n) = S(n=0)^{*}f^{n} \tag{3}$$

where $S(n)$ defines the signal on nth reading, $S(n = 0)$ is the putative signal before any reading depletion, and f is the faction by which the signal was reduced per reading.

For $f = 1 - a$ and $f^n = (1 - a)^n$, simplifying Eq. (3) yields:

$$S(n) = S(n=0)^{*}(1-an) = S(n=0) - S(n=0)^{*}an \tag{4}$$

where $-S(n = 0)\times a$, and $S(n = 0)$ are the slope and the intercept, respectively and a is a parameter given by $a = 1- f$.

2.3. Dose linearity

Fifteen OSLDs were prepared for this test and divided into five groups of three. Irradiations were performed on the water phantom at fixed conditions of 10 cm × 10 cm field size and 200 cm SSD. These five groups of OSLDs were irradiated with 5, 10, 15, 20 and 50 mSv