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In-Vivo Dosimetry Using Thermo-Luminescent Dosimeters in a Cobalt-60 Beam In a Developing Country

[1]Samuel Y. Opoku, [2]Kofi Adesi Kyei, [3]Joel Yarney, [4]Pearl Lawson, [5]Samuel N. A. Tagoe, [6]Eric K. T. Addison

[1, 2, 5] Department of Radiography, School of Allied Health Sciences, University of Ghana
[3, 4]National Center for Radiotherapy and Nuclear Medicine, Korle-Bu, Accra, Ghana
[6]Oncology Unit, Komfo Anokye Hospital, Kumasi, Ghana

Abstract: **Radiation therapy is currently employed in managing approximately 50% of cancer patients worldwide with the aim of controlling the tumour without incurring unacceptable harm to surrounding normal tissue. It is recommended that the dose delivered to the patient be monitored with a dosimeter as part of quality assurance (QA) procedure. This however is not the case in a number of institutions, including the Oncology Unit of Ghana. Hence there is the need to perform in vivo dosimetry to check accuracy of actual absorbed dose delivered to the patient.**

The study was aimed at performing in-vivo entrance dose dosimetry using thermo-luminescent dosimeters (TLDs) to verify the accuracy of the radiation delivered to patients as compared to prescribed doses.

An experimental study design, using a convenience sampling method, was used to conduct a study with 14 patients undergoing radiotherapy for cancers of the head and neck, spinal, pelvic and breast regions at the study site. A water phantom calibration was done using the recommended IAEA method. Calibration coefficients were determined with TLDs, using a Perspex phantom to derive correction factors. Patients' doses were measured with TLDs during treatment.

The findings of the study showed that radiation dose administered to patients increased with field size but decreased with depth as established in literature. The highest frequency of percentage errors occurred between the interval of 4-4.99% and the lowest within the 5-5.99% interval. Also, the mean absolute percentage error of the measured doses from the prescribed doses was 3.01% with a standard deviation of 1.59%.

Radiation doses delivered to patients were acceptable within the recommended tolerance level of ±5%. This demonstrates the importance of in-vivo dosimetry in verifying the absorbed dose received by patients during radiotherapy.

Keywords: **Absorbed dose, Entrance dose, Radiotherapy, Perspex phantom.**

1. INTRODUCTION

Radical radiotherapy uses the biological effects of radiation to kill tumour cells with an objective of completely sterilizing tumour cells without exposing surrounding normal tissues to unacceptable injury [1, 2]. Radiation dose given to a patient is determined by the intent of the treatment (curative or palliative), irradiated tissue volume and amount of radiation received by surrounding normal tissues with consideration to the patient's performance status [3].

Even with the advanced technologies, normal tissues are often affected in an attempt to cure patients by giving a tumouricidal dose. Altering the treatment dose or modality changes the therapeutic ratio of tumour control probability (TCP) to normal tissue complication probability (NTCP) which indicates the outcome of the alteration. This calls for stringent measures to reduce the side effects on normal tissues [1].

According to the International Atomic Energy Agency (IAEA) [4], major impacts on the outcome of patient treatment could result from errors incurred during the treatment procedure. High radiation doses delivered during treatment may cause deposition of low doses to normal tissues at the edge of the radiation beam. This may result in cell mutation with a probability of developing into second cancers (i.e. while curing a primary cancer) [3]. About 0.15% mild to moderate error-induced injuries can be incurred for one treatment in radiotherapy; however some errors may escape notice particularly if they end up under-dosing the patient, resulting in serious side effects (5, 6).

As a form of quality assurance (QA) port films are usually used to monitor target volume coverage and to check normal tissue sparing. There is however a necessity to check the actual absorbed dose being delivered to the patient [7].

The radiation dose administered to a patient undergoing radiation therapy is checked by in-vivo dosimetry; it thus serves as a back-up to the clinical quality assurance (QA) programme. In comparison to other QA procedures done before treatment, it checks patient dose delivery during treatment [8]. It is employed in comparison of prescribed and delivered doses to help find patient or treatment procedure errors, to assess the accuracy of dose calculation (such as skin dose) or for validity checks of particular treatment techniques, especially when done early in treatment [7].

It is recommended that a dosimeter be used to check the first treatment dose of all patients undergoing radiotherapy. Usually, TLDs, silicon diodes, and currently metal-oxide-semiconductor field-effect transistors (MOSFET) are used for external beam in-vivo dosimetry. Dose at a point inside the patient is calculated from measurements taken with a dosimeter placed on the patient's skin [9, 10]. The American Association of Physicists in Medicine [8] affirms the same.

It has been observed that there are no recommended dosimeters (diodes or TLDs) to perform in-vivo dosimetry at the study site and that the unit relies on ionization chambers for routine machine and beam calibration quality assurance procedures. Though there is a low risk of mechanical/electrical errors for Co-60, which has a low energy (1.25MeV) as compared to other teletherapy units such as the linear accelerator (6MV), it is still necessary to check patient dose delivery [11]. Uncertainty in dose delivery to patients should generally fall within ±5% of the prescribed dose as recommended by the International Commission on Radiation Units (ICRU) (12, 13).

This study was therefore carried out to perform in-vivo dosimetry procedure to help develop a protocol for in vivo dosimetry in Ghana.

2. METHODOLOGY AND MATERIALS

An experimental study design was employed to perform in-vivo dosimetry measurements with TLDs, thus allowing control of the factors involved in the experiment. A convenience sampling method was used to obtain the sample for this study because of easy accessibility to the subjects for the study and advantage of cost and logistics [14]. It also allowed for convenient selection of the subjects on the basis of their availability, though it was non-probabilistic.

A sample of 14 patients undergoing radiotherapy for pelvic, spinal, breast and abdominal regions were used for the study. The data collected was organized and presented in graphical and tabular forms using Microsoft Excel 2010 spread sheet. Ethical clearance was sought from the Ethics and Protocol Review Committee of a higher education. Informed consent was obtained from patients and they were assured of the anonymity and confidentiality of the data obtained from them.

The following materials and equipment were used:

The GWGP 80 Cobalt 60 Teletherapy Unit; A total of 40 Lithium fluoride TLDs (10 chips and 30 rods) with polymethylmethacrylate, of 0.5cm thickness, as build-up material; A Perspex phantom consisting of a pile of twenty 30 x 30 x 1 cm^3 Perspex slabs; A solid water phantom, of dimensions 20 x 20 x 10cm^3, with a bore at 5cm depth for the ion chamber; A 0.600 cm^3 plexi Farmer Type ionization chamber(Model PTW 30001, Freiburg, Germany); UNIDOS electrometer (PTW, Freiburg, Germany); An Anaerobic Präzision barometer (Freiburg); Harshaw model 4500 manual TLD reader for thermo-luminescent readouts; Digital thermometer and Masking tape.

3. PROCEDURE

The TLDs were annealed and irradiated to the same dose. The plastic and solid water phantoms were allowed to acclimatize to the treatment room's conditions and quality assurance was performed to check for the machine's parameters before beginning any measurements. A calibration of the photon beam output was performed with a solid water phantom (as recommended by TG-51 for determination of absorbed dose to water for photon and electron beams)

using a reference standard system consisting of a cylindrical ionization chamber (Farmer Type) connected to an electrometer to record the readings/charges (outputs of the teletherapy machine) from exposure of the chamber. The reference setup was such that the beam central axis of a standard field size of 10 x 10cm^2 coincided with that inscribed on the phantom and at SSD of 80cm. The chamber was placed at a depth of 5cm inside the solid water phantom and pre-irradiated for 300s to remove traces of stray charges, and afterwards for 600s to check for stability of the electrometer readings.

Three consecutive electrometer readings each were taken for polarities of +400V, -400V and +200V to account for recombination by setting an electrometer time of 60 seconds for each voltage. Absolute measurements were also done with the ionization chamber alone inside the solid water phantom, and afterwards with a TLD placed at the central axis of the field and the ionization chamber in the same position by setting an exposure time of 60 seconds after setting the electrometer to start reading. Readout of the TLD gave absorbed dose to water value of TLD_{cal} at reference geometry. The ambient temperature and pressure were recorded during all measurements involving the ionization chamber (but not for TLDs which are not affected by these parameters), with the assumption that the phantom was in thermal equilibrium with the room conditions. The average measurement values were corrected for influence quantities $K_{T.P}$, K_{pol} and K_{sat}. With percentage depth dose (PDD) at 5 cm, corrected measurements were converted to D_{max} dose and hence I_{cal} absorbed dose to water with the ion chamber calibration factor.

Readings were then taken to obtain calibration coefficients by taping the TLDs at the entrance surface of a 30 x 30 x 25 cm^3 Perspex phantom (a pile of 25 30 x 30 x 1 cm^3 Perspex slabs) at the field centre and exposing them to gamma radiation from the Co-60 unit at various field sizes (4x4, 8x8, 10x10, 12x12, 16x16, 20x20, 25x25 cm^2) at a standard reference depth of 5cm and depths (3, 7, 11, 13, 15, 18, 20 cm) at a standard field size of 10 x 10 cm^2, to represent those normally used for treatment. The TLDs were labelled with the numbers indicated on them from the laboratory, for easy identification after exposure and readout.

A total of 14 treatment fields were obtained from 14 conveniently selected patients undergoing treatment at the time of the research. The selected patients were being treated for pelvic (prostate, cervix), breast, spinal, and head and neck cancers. The TLDs were placed on the patient at the beam central of each treatment field to measure entrance dose while patients were being treated. For head and neck, and breast cancer patients it was ensured that fields which incorporated beam splitters were not included because the beam central axis is compromised by the half beam block.

The expected dose was calculated from the patient TLD entrance dose and absorbed dose to water readings from calibration. The goal was to compare the measured and prescribed doses to detect any deviation exceeding 5% (the absolute value of +/-5%).

The calibrated ionization chamber was placed at D_{max} for the entrance dose calibration factor (A) and D_{max} from the exit side for exit dose calibration (B), with the detector at the entrance or exit surface respectively [7].

4. RESULTS

The findings include treatment parameters, TLD and electrometer readings and calibration factors. The data collected were organized and entered into Microsoft Excel 2010. The results were presented in the form of graphs and tables. The average temperature recorded during the water phantom calibration was 27.25°C with an average air pressure of 101.3kPa. An average electrometer reading of 17.17nCmin^{-1} from exposure of the ionization chamber gave a temperature and pressure corrected mean of 17.60nCmin^{-1}.

Table 1: Patient Treatment Data

Patient ID	Site	Field Treated	Treatment No.	Date of reading
1	Breast	Medial Tangential	16	6/25/12
2	Breast	Lateral Tangential	13	6/25/12
3	Cervix	Anterior	18	6/25/12
4	Cervix	Anterior	6	6/25/12
5	Breast	Medial Tangential	16	6/25/12
6	Cervix	Anterior	19	6/25/12
7	Breast	Medial Tangential	4	6/25/12
8	Cervix	Anterior	3	6/25/12

9	Prostate	Lateral Tangential	13	6/25/12
10	Larynx	Right Lateral	10	11/07/12
11	Endometrium	Anterior	2	11/07/12
12	Cervix	Anterior	23	11/07/12
13	Prostate	Anterior	24	12/07/12
14	Cervix	Anterior	1	12/07/12

Table 2: Patient TLD Entrance Dose Measurements

Patient ID	TLD No.	Treatment Depth (cm)	Equivalent Square field (cm^2)	Patients' Entrance Dose TLD Readings ($cGymin^{-1}$)	Prescribed Dose ($cGymin^{-1}$)
1	2b	5	8.9	170.70	100
2	2c	5	7.7	181.53	90
3	2d	11	16.1	1050.19	100
4	2e	10	16.1	990.73	90
5	2f	6.5	9.4	149.74	100
6	3b	10	14.4	334.24	90
7	4b	9	14.44	483.97	133
8	4d	10	15	498.06	100
9	4e	19.22	8.23	11.87	34
10	1a	7	11.9	151.89	90
11	2a	10.5	16.2	1789.63	150
12	5a	11.5	16.1	940.22	90
13	8a	11.71	9.18	96.55	64
14	9a	9.5	15.5	789.85	125

Patient TLD numbers indicating depths and equivalent square fields together with their prescribed treatment doses and raw TLD readouts obtained are also shown

Table 3: Variation of TLD Entrance Dose Readings with Field Sizes and Values for Plotting Graph for Beam Calibration on Phantom

TLD No.	Field Size(cm^2)	Phantom Entrance Dose TLD Readings ($cGymin^{-1}$)	Normalized Readings
7b	4x4	89.6389	0.5019
8b	8x8	118.7012	0.7970
3b	10x10	148.9274	1.0000
9b	12x12	110.4850	0.9519
1c	16x16	103.0363	0.6919
2c	20x20	89.7301	0.6025
3c	25x25	168.5188	1.1316

TLD readouts for phantom calibration for various field sizes at a constant standard depth of 5cm and their normalized values at a standard field size of 10cm

Table 4: Beam calibration with TLDs on phantom at various depths

TLD No.	Depth (cm)	Phantom Entrance Dose TLD Readings ($cGymin^{-1}$)	Normalized Readings
4c	3	91.4395	0.6140
3b	5	148.9274	1.0000
5c	7	168.7660	1.1332
6c	11	117.0077	0.7857
7c	13	102.2191	0.6864
8c	15	105.1018	0.7057
10c	18	171.5989	1.1522
9c	20	171.4188	1.1510

TLD readouts for phantom calibration on various depths for a constant standard field size of 10cm x 10cm and their normalized values at a standard depth of 5cm, are indicated here.

Table 5: Correction Factors for Field Sizes and Depths for Calculated TLD Dose Values

Patient ID	Patient Entrance TLD Readings ($cGymin^{-1}$)	Field Size CF_d	Depth CF_{fs}	Time Elapsed (days)	D_f	TMR	Corrected TLD Dose ($cGymin^{-1}$)
1	170.7000	0.7883	1.0044	-20	1.0072	0.8690	99.0547
2	181.5282	0.7073	1.0044	-20	1.0072	0.8617	93.7295
3	1050.1880	0.1193	1.3193	-20	1.0072	0.6956	96.9945
4	990.7302	0.1193	1.2214	-20	1.0072	0.7284	88.7062
5	149.7407	0.8053	1.1531	-20	1.0072	0.8157	95.6613
6	334.2372	0.3767	1.2214	-20	1.0072	0.7200	93.3956
7	483.9709	0.3704	1.1935	-20	1.0072	0.7545	136.1711
8	498.0564	0.2826	1.2214	-20	1.0072	0.7240	104.9563
9	11.8700	0.7492	11.0688	-20	1.0072	0.3936	32.6738
10	151.8853	0.7111	1.1750	-4	1.0014	0.8107	86.2776
11	1789.6299	0.1058	1.2586	-4	1.0014	0.7124	142.3529
12	940.2204	0.1193	1.4099	-4	1.0014	0.6790	90.0578
13	96.5459	0.7992	1.4585	-3	1.0011	0.6873	64.8340
14	789.8465	0.2058	1.2017	-3	1.0011	0.7429	121.6180

Correction factors (C_f) derived from trend equations of the polynomial graphs plotted, decay factors (D_f), and corrected patient TLD readings are shown here.

Table 6: Deviation between Prescribed and Measured Doses

Patient ID	Prescribed Dose (cGymin^{-1})	Corrected TLD Dose (cGymin^{-1})	Errors	Percentage Errors (%)
1	100	99.0547	-0.95	-0.95
2	90	93.7295	3.73	4.14
3	100	96.9945	-3.01	-3.01
4	90	88.7062	-1.29	-1.44
5	100	95.6613	-4.34	-4.34
6	90	93.3956	3.40	3.77
7	133	136.1711	3.17	2.38
8	100	104.9563	4.96	4.96
9	34	32.6738	-1.33	-3.90
10	90	86.2776	-3.72	-4.14
11	150	142.3529	-7.65	-5.10
12	90	90.0578	0.06	0.06
13	64	64.8340	0.83	1.30
14	125	121.6180	-3.38	2.71

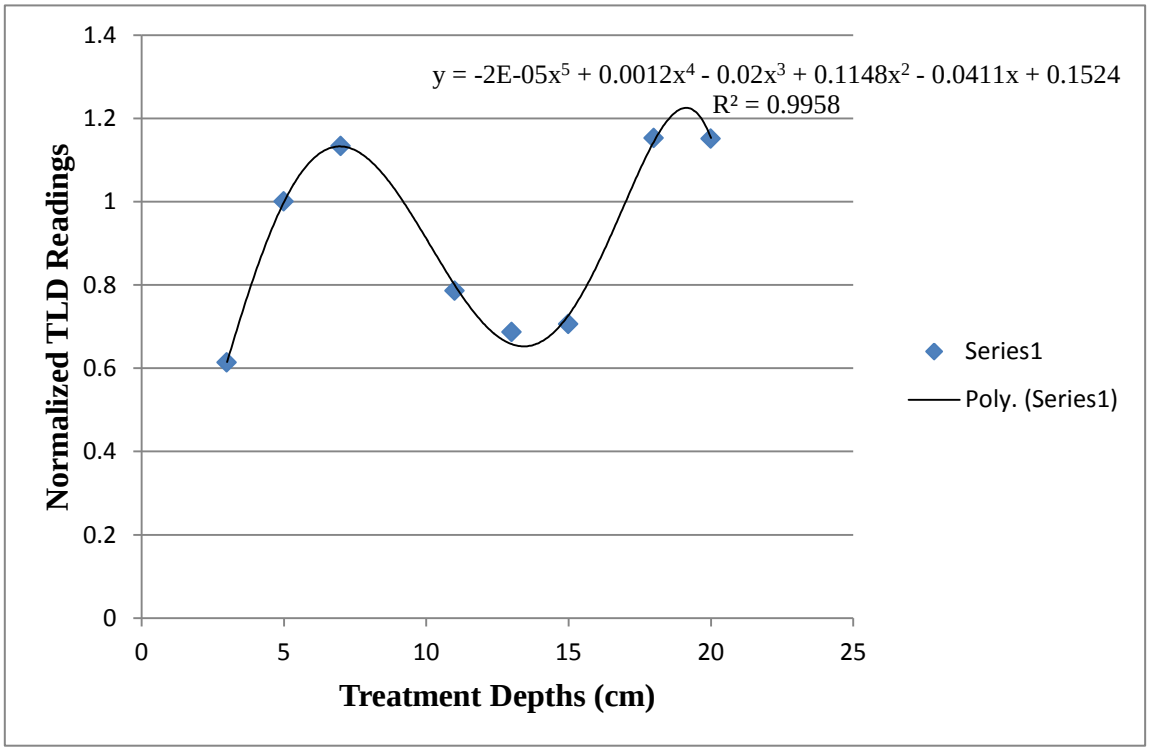

Figure 1: Normalized TLD entrance dose readings at various treatment depths

The graph of a 5th order polynomial with a strong exponential relationship between normalized TLD entrance dose readings and treatment depths for beam calibration on a phantom. Regression analysis to show how well the data was represented gave a coefficient of determination of R^2 = 0.996. The trend equation derived is shown on the graph together with the R^2 value, x representing patients' treatment depths.

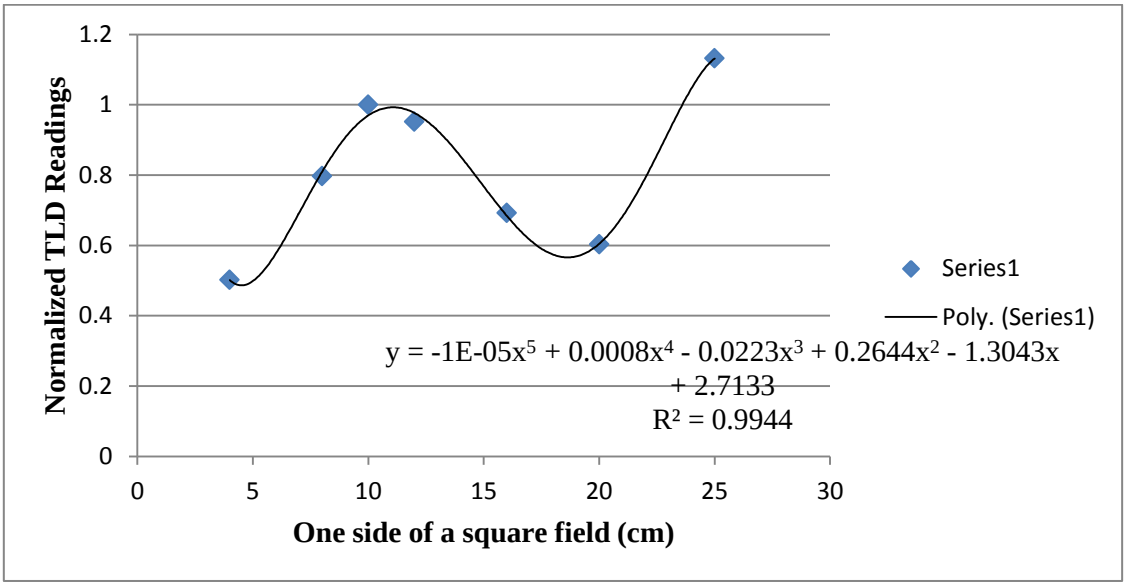

Figure 2: Normalized TLD entrance dose readings for various field sizes

The graph above shows a strong exponential relationship between the normalized TLD entrance dose readings and the equivalent square field when the Co-60 beam was calibrated with TLDs on a Perspex phantom. The regression analysis to determine how well the data was represented gave a coefficient of determination of R^2= 0.994. The trend equation of the polynomial is shown with the R^2 value, x being patients' equivalent square treatment field sizes.

The equation used for calculation of the corrected measured TLD readings is stated in equation 2.3 below:

$$\text{Corrected TLD dose} = \frac{I_{cal}}{TLD_{cal}} \times CF_{fs} \times CF_{d} \times TLD_{patient} \times 1.0125 \times D_{f} \times TMR \qquad \textbf{(2.3)}$$

5. DISCUSSION

TLDs were used in this study because they were the only dosimeters for absorbed dose measurement that could be obtained and also because they were reliable, having been in use for over 30 years though they have measurement uncertainties such as reproducibility and positioning, Co-60 dose variations reader instability, and PMT non-linearity [8, 15]. Every step in the radiotherapy procedure is a likely factor in final dose uncertainty and in-vivo dosimetry proves to be the best means of ascertaining final delivered dose accuracy. However, when errors are not detected, in vivo dosimetry indicates that treatment delivery was acceptable [8, 16].

Table 1 shows patient data taken from their folders, which were used for calculation of the measured doses. Some of the patients were in their early stages of the treatment. For example patients with identification (ID) numbers 8 and 14 at the time of study had 3 out of 25 and 1 out of 17 treatment fractions respectively. Others, such as ID 13 and 12 were almost completing, having had 24 out of 30 and 23 out of 25 fractions respectively. In-vivo dosimetry is best done in the first week of treatment for early detection and avoidance of any errors, in order to put in correct measures to ensure optimal patient treatment, as was done within the first three days in one study [8, 17].

From the data readings (Table 2) it was identified that as depth increases, the dose decreases beyond D_{max} which confirms the study carried out by Khan [18]. This is because absorbed dose increases with depth as the electrons ejected by the interaction of photons with the body travel downward, resulting in a build-up of electrons with depth but beyond D_{max} dose decreases as photon fluence, hence ejection of electrons, decreases. Also, field size increase results in absorbed dose increase, due to increased scatter radiation contribution to dose especially at greater depths beyond D_{max}. Practically, TLD readings may change slightly due to the surrounding material in their size and shape. The build-up cap takes away most of the surface dose such that the TLDs measure peak dose from D_{max}, accounting for the higher doses for some of the TLDs [19].

The percentage errors were between 0% and 5.2%, with the highest percentage error being 5.1. The highest frequency for percentage errors in measured doses occurred within the interval of 4-4.99% whereas the lowest frequency occurred in the

interval of 5-5.99%, with 0-0.99%, 1-1.99% and 2-2.99 %. The interval that had the lowest frequency (5-5.99%) had the highest measured TLD dose, a prescribed dose of 150cGy at a field size of 16.2 and a depth of 10.5cm. It can be inferred from this patient's parameters that the large field size could have contributed to the high dose, but the depth implies that dose should be low for this field.

As such the uncertainty in this measured dose could have been due to TLD reproducibility and positioning, reader instability, Co-60 dose variations and PMT non-linearity [15]. Insufficient build-up material could have also caused inaccuracy in TLD readings since the thickness was not accurately measured. Most percentage errors falling between 0-0.99%, 1-1.99% and 2-2.99 % signifies less-deviated measured individual doses according to protocol. With exception to the error of 5.1%, these values fall below the ICRU recommended action level of ±5% employed in this study . Also coefficient of determination, R^2 , at various depths showed ≈ 99.6% of variation in TLD dose reading was linked with variation in treatment depths, the remaining 0.4% probably due to other variables which have an effect on radiation dosimetry. At various field sizes, R^2, showed 99.4% of the variability in TLD reading with the remaining 0.6% probably due to other factors that affect radiation dose delivery.

These together signify that the deviation of the prescribed dose from the measured TLD readings is at an acceptable value [20]. The mean absolute percentage error in the measured doses was found to be approximately 3.01% with a standard deviation of ±1.59%. A combined uncertainty of ±3% in TLD readings, and an uncertainty in delivered dose at centres of 3% were allowed by Kron, et al [21]. Costa et al (2010) had a mean percentage deviation of measured dose from expected or prescribed dose of 99% with a standard deviation of ±2.6% and an approximated overall individual dose uncertainty of ±3% which indicated the slight discrepancy between the measured and prescribed mean doses (-1%) was caused by the dosimetric system [22]. Their work stated that individual dose measurements are allowed an overall uncertainty of less than ±3%.

Having undertaken the experimental study on in-vivo dosimetry using thermoluminiscent dosimeters on a cobalt 60 beam, the following protocol was proposed for use at the Centre.

- Anneal the TLDs for about an hour and irradiate them to the same doses. Place the TLDs in the planchet and heat at preheat and readout temperatures to erase unstable peaks and obtain dosimetric peak data.
- Allow the plastic and solid water phantoms to acclimatize to the treatment room's conditions. Perform quality assurance procedure to verify machine performance and make corrections where results fall outside acceptable range. Place water phantom on the treatment couch at SSD of 80 cm such that the beam central axis of a 10 x 10 cm^2 light field coincides with that inscribed on the phantom. Carefully insert the Farmer type ionization chamber into the slot made at one side of the phantom. Place thermometer and barometer for easy read-out and away from set-up. Connect the ion chamber to the electrometer and select the appropriate settings (chamber type, operation voltage and mode, exposure interval, and polarity).

Pre-irradiate ion chamber for 300 seconds as set on the electrometer. Find the average for three successive measurement at 60 seconds at the following voltages and polarities; +400V, -400V and +200V. Initial and final temperature and pressure readings must be recorded. Correct the average measurement value for influence quantities $K_{T.P}$, K_{pol} and K_{sat}. With percentage depth dose (PDD) at 5 cm, convert corrected measurement to depth of maximum dose (Dmax) and hence calculate $\mathbf{I}_{cal}$ absorbed dose to water with the ion chamber calibration factor.

With the ion chamber removed from its slot, carefully place a TLD at the entrance surface of the water phantom, and irradiate for 60 seconds with timer setting on the treatment console to neglect influence of transient radiation. Measure and record TLD_{cal} absorbed dose to TLD.

TLD CALIBRATION CORRECTION FACTORS CF_{FS}, CF_{D}:

Carefully arrange twenty-five (25) slabs of Perspex phantom each of dimension 30x30x1cm^3 on the treatment couch one over the other such that their center is at the beams central axis. Set the surface of the phantom at an SSD of 80 cm. At a reference depth of 5 cm, adjust the entrance surface to an SSD of 75 cm and carefully place each TLD with a polymethylmethacrylate build-up cap of 0.5 cm thickness at the center of the entrance surface of the phantom along the beam central axis. For each of the labelled TLDs, irradiate for 60 seconds each at the following square fields: 4, 8, 10, 12, 16, 20 and 15 cm^2. A plot of normalized TLD reading against varying square fields yields a polynomial function in the fifth order. For any field size, the TLD field size correction factor, CF_{fs} can be extrapolated for the function.

At a reference square field of 10 cm^2, irradiate another set of TLDs for 60 seconds each at varying depths of 3, 7, 11, 13, 15, 18 and 20 cm. Another plot of normalized TLD readings against varying treatment depth provides TLD depth correction factors, CF_d as a polynomial function in the fifth order.

PATIENT ENTRANCE DOSE TLD MEASUREMENTS, $TLD_{PATIENT:}$

Place the TLDs with build-up cap on the patient's irradiating surface along the beam central axis of each treatment field to measure patient entrance dose during treatment process.

TLD DOSE AT A DEPTH, $TLD_{D,\ D:}$

The TLD dose at any prescribed depth is calculated as a product of the measured TLD readings ($TLD_{patient}$), the calibration factor derived as the ratio of the ion chamber (I_{cal}) and TLD readings (TLD_{cal}) measured in reference geometry, and correction factors (i.e. field size (CF_{fs}) and treatment depth (CF_d)). TMR is incorporated because SAD setup is used; as well as a decay factor (D_f) to account for decay of the Co-60 and 1.0125 to account for non-isocentric teletherapy unit calibration since source to calibration point distance was 80.5cm [17].

$$\mathrm{TLD}_{D,d} = \frac{I_{cal}}{TLD_{cal}} \times CF_{fs} \times CF_d \times TLD_{patient} \times 1.0125 \times D_f \times TMR$$

The goal is to compare the measured to prescribed doses to detect any deviation. Deviation within the range +/-5% of the prescribed dose is acceptable whereas those outside this range call for investigation for necessary measures to be applied, as failure to do so would nullify the whole treatment intent.

6. CAUTIONARY MEASURES

Ensure that TLDs are protected from other irradiation and exposure to grease, dirt or humidity which may contaminate and lower their light emission properties. They should be handled and stored properly such as using in incandescent lighted rooms or placing them in envelopes or opaque casing, and performing readouts shortly after irradiation to avoid fading (loss of trapped charges by unnecessary contact with heat, light or other variables)[8]. TLDs are best used in the linear region of the dose response curve where their response is proportional to the dose they are exposed to.

Otherwise, a correction of the signal from a curve obtained from the TLD material and its reader should be done. However, in the sub-linear region approaching saturation, TLDs should not be used since they show supra-linearity [2, 8]. TLDs should be well labelled for easy identification. Treatment fields which incorporate the use of beam splitter are ignored because the beam central axis is compromised by the half beam block

7. CORRECTION FACTORS

Correction is usually not required for applied clinical dose rates, pressure and temperature (because of the high temperature used) or beam directional dependence (not even in tangential breast irradiation); except to correct for dose response non-linearity for some TLDs, which holds if only the detector calibration is done with the same beam quality as used for patient treatment [23, 24, 25]. Solid state detectors however, do not require temperature and correction.

Junell, *et al.*, [15] indicated sources of uncertainty in TLD measurements as TLD reproducibility and positioning, reader instability, Co-60 dose variations and PMT non-linearity.

8. CONCLUSION

The accuracy of the prescribed dose in comparison to that delivered to cancer patients at the study site using TLDs in a Co-60 beam was investigated. The results occurring within a range of ±5.2% showed that most percentage errors of measured individual TLD doses compared favourably with the prescribed doses. Also, the mean absolute percentage error in the measured doses was found to be approximately 3.01% with a standard deviation of approximately ±1.59% from the prescribed dose. This indicates that most of the doses delivered to patients used for this study were acceptable and within recommended tolerance range but for one of the percentage error which occurred above the standard error of ±5%. This deviation identified that other patients could be under dosed or overdosed but this was not detected since the study was conducted on a few conveniently selected subjects.

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