Volume 5 No.1, January 2015 ISSN 2224-3577
International Journal of Science and Technology

http://www.ejournalofsciences.org

Accuracy of Using In-Vivo Dose Verification with Diodes for Different Sites

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ABSTRACT

In radiotherapy, treatment success depends on accurate dose delivery to the target, and should be adequately taken into consideration. One of the effective methods for ensuring accurate dose delivery and as a quality check tool is *in-vivo* dosimetry using diodes during radiotherapy. *In-vivo* dosimetry measurements were performed with diode detectors calibrated against a reference ionization chamber for 6 MV and 15 MV for photon beams and 6 MeV, 10 MeV and 15 MeV for electron beams. A total of 101 cancer patients were considered for this study, with 420 field measurements made with low and high energy diode detectors for prostate, cervix, breast and head and neck cancer cases. All patients had their treatments planned with an Ocentra Masterplan Treatment Planning System v. 4.1 and 4.3, and treated with either 6MV and 15MV or a mixture of both energies using an Elekta Synergy platform Linear accelerator. The diodes measured the entrance doses in each field in the D_{max} depth and were compared with the calculated dose from the TPS at D_{max} depth for every single beam. All treatment cases including the wedged fields were considered appropriately and assigned to a 10 % action level. Approximately 95.59 % of the 136 measurements for the head and neck treatments were within the SGMC ±10 % action level. In the case of the cervical cancer treatments, 97.87 % of 47 measurements made were within the ±10 % action level, whereas the breast and prostate treatments produced 89.06 % (64 measurements) and 96.53 % (173 measurements) respectively within the set action level. For the Head and Neck treatment, the average discrepancy, δ_{avg} for 136 measurements (N) was 2.16 %, while the corresponding standard deviation (**σ)** was 4.61%.The results for prostate treatments (N=173) indicated an average discrepancy, δ_{avg} of 4.57 % and a corresponding **σ** of 3.65%. For cervix (N=47), an average discrepancy, δ_{avg} of 4.69 % was obtained with a corresponding value **σ** of 2.58 %, while 5.02 % and 5.19 % was obtained for δ_{avg} and σ respectively for the breast cancer patients. A greater percentage of the observed discrepancies were well within the set tolerance level. It is however recommended that the positioning of diode on patient's skin, and the angular sensitivity of the diodes should be reconsidered. It is also recommended that, a more accurate calculation of expected diode values be done, especially for fields which pass through the table. Further studies are being carried out to find the existence of a cluster of larger deviations at a particular gantry angular range. By these, several efforts would then be made to achieve action levels of ±5 %.

Keywords: *Diodes, in-vivo dosimetry, entrance dose, radiotherapy, dose verification*

1. INTRODUCTION

The success of radiotherapy depends on accurate dose delivery to the target and minimizing doses to the surrounding healthy tissues. One of the effective methods for accurate dose delivery and quality check is *in-vivo* dosimetry using diodes during radiotherapy treatment. *In-vivo* dosimetry provides dosimetric information regarding actual treatment delivery, and is understandably considered an indispensable quality assurance procedure and a safety measure in the treatment process [1-3]. *In-vivo* dosimetry, recommended by various national and international organizations including the International Commission of Radiological Protection (ICRP) [4], can be carried out at several levels. Two different goals can be identified: the measurement of doses to organs at risk that are difficult to calculate (such as the dose to eyes and gonads) and the verification of the delivered dose in order to improve treatment accuracy and to minimize the risk of dose misadministration [5]. These measurements are compared to the planned doses specified by the oncologist and calculated by the Treatment Planning System (TPS) for the target and critical organs (e.g. rectum or spinal cord). In this way, set-up calculation, motion or transcription errors that may have gone unnoticed during pre-treatment checks, can be recovered before dose delivery. In the absence of errors, routine *in-vivo* dose measurements indicate that the treatment was delivered correctly.

Diodes are basically small detectors attached to a long wire, and are used to measure the dose being received in 'real time', whilst a patient is undergoing radiotherapy treatment. They are normally attached to the patient's body with adhesive tape at specific points, where the treatment beam enters the body. Many professionals acknowledge their importance because they have the potential to detect any errors that may have slipped through the quality safety net [3].

While errors in the delivery of radiation therapy are rare and usually result in little or no injury to the patient, the real danger is if an error in administration goes undetected. This may result in healthy tissues being exposed to unnecessary levels of radiation or the tumour site not receiving full effect of therapy. According to previous studies, a severe misadministration may

result in radiation necrosis to vital organs or structures and can be fatal.

In recent publications, several radiotherapy reports have been given on erroneous exposure of patients to radiations. In a report by Cohen et. al [9], a cobalt unit for radiotherapy at a hospital in Ohio was initially calibrated correctly. It is further stated that the between 1974 -1976, the physicist had failed to perform regular measurements (calibrations and QA), but relied on estimations of the decay of the source to predict the dose rate and calculate the treatment time, rather than calculating the decay. The errors in predicting the dose rate resulted in the dose-rate being under-estimated by 10% to 45%, which translates to the patients receiving corresponding overdoses of 10% to 55%. Eventually, it was revealed that 426 patients received significant overdoses as a result.

In another incidence also reported by the IAEA [12], a report is made on an erroneous use of a treatment planning system. In their publication, Ash et. al [10] report that a hospital in the United Kingdom relied on manual calculations for the correct dose to be delivered to the tumour. In that report, the distance correction factor was erroneously applied twice for all patients treated isocentrically, or at non-standard SSD. This error caused patients to receive doses lower than prescribed doses. The underdose varied between 5 and 35%, and in the end, it was revealed that, of 1045 patients whose calculations were affected by the incorrect procedures, 492 developed local recurrences that could be attributed to the error.

In a report by IAEA [11, 12], A cobalt source was exchanged for a new one in 1996 in a hospital in Costa Rica. At the subsequent calibration, the medical physicist incorrectly interpreted 0.3 minutes as being 30 seconds (instead of the correct interpretation of 18 seconds). As a consequence, the absorbed dose rate of the new source was underestimated, resulting in treatment times being overestimated by 66%. Severe reactions by some of the treated patients led to investigation, which revealed overexposures involving 115 patients. Two years after the event, at least 17 of the overexposed patients died.

In another report by IAEA [13], on July 6, 2006, the French Nuclear Safety Authority was informed that 23 patients treated by external beam therapy for prostate cancer between May 2004 and May 2005 were overexposed to levels between 7% and 34% more than the prescribed radiation dose. At least 16 of the patients developed acute complications (rectal inflammation/burns), and at least 1 patient died as a result of the overexposure. An investigation revealed that the TPS simulation was performed with static wedges, but the accelerator was set with dynamic wedges. Another highly publicised case of radiation misadministration resulting in fatality occurred in Glasgow, Scotland in 2006 [6]. In that incident, a young patient, Lisa Norris, received a 58% higher dose than prescribed to her craniospinal area. An autopsy revealed that her tumour was still present despite radiation therapy.

With the numerous reported cases of radiation accidents worldwide, it is increasingly becoming incumbent that *in-vivo* dosimetry is necessarily practised in radiotherapy departments to improve accuracy in the dose delivery, and reduce radiation misadministration.

With the *in-vivo* entrance dose measurements, one can expect to avoid most of these accidents, by detecting equipment related errors; Changes in the dose delivered per monitor unit; Incorrectly aligned wedge filter or other accessories; Beam parameters out of tolerance (e.g. flatness, energy); Human errors in data generation, data transfer and treatment set-up; Incorrect setting of monitor units; missing or incorrectly positioned wedge filter (but not a reversed wedge); Wrong choice of energy; positioning discrepancies between treatment planning and delivery (e.g. SSD, beam geometry, the use of a wrong table height when applying wedged lateral fields); treatment data for the wrong patient selected; and miscalculation of the entrance dose by the TPS [5].

The International Commission on Radiation Units and Measurements (ICRU) has recommended that radiation dose must be delivered to within 5% of the prescribed dose [7, 8]. Moreover, in a recent publication by the IAEA (2013), an appropriate goal is to be able to use a tolerance level of 5% for simple treatments, with a level of 7% for situations such as breast treatments and other treatments where measurement complications exist. However, it is recommended that, although in the initial stages of the introduction of *in-vivo* dosimetry the tolerance levels may need to be higher, every effort should be made to achieve tolerance levels of about 5% by a process of progressive elimination of identified causes of dose differences [5].

This paper seeks to compare the entrance doses derived from the signal of the diode detectors placed on the skin with the theoretical values as calculated by the TPS under set tolerance values.

2. MATERIALS AND METHODS

Treatment Unit and Detectors

Two photon diode detectors (P10 and P20) manufactured by IBA which were calibrated against an ionization chamber were used in this study. Both diodes are made with a hemispherically shaped build-up cup of polyacetal, with 10 mm build-up thickness (for P10 diode) and 20 mm build-up thickness (for P20 diode). This build-up thickness has the advantage of minimizes the need for correction factors for field size and SSD and greatly simplifies the interpretation of the readings. The hemispherical shape of the integral build-up minimizes the directional dependence as well as the perturbation [14]. Both diodes have smaller temperature dependency of 0.1–0.3 %/degree. The diodes had been tested for consistency using 0^0,90^0 and 180^0

gantry angles for the 6 MV and 15 MV photon energies, and for 100 MU as shown in appendix B.

At our centre, quality assurance (QA) are performed on daily, weekly, monthly and yearly basis to check machine fluctuations from baseline. The checks involve dosimetry, mechanical and safety checks. These measurements ensure that the system is working as intended.

The entrance dose, $D_{entrance}$ is defined as the dose at depth of maximum dose for the corresponding energy. The diode reading that is expected for each treatment field is given by the treatment planning system at Dmax. The TPS uses the collapse cone algorithm in calculation for doses and equivalent path length for inhomogeneity corrections. Dmax for 6MV photon is at 1.6cm and that for 15Mv is at 2.5cm.

The diodes were placed in the field based on the radiation type as well as its energy (low or high) being used in the treatment of the patient at the time.

- For 6 MV photon energy, the P10 diode was used.
- For 15 MV photon energy, the P20 diode was used.

The detectors were connected to the Apollo 5 electrometer, and all measurements were performed in photon radiation beams generated by Elekta Synergy Platform accelerator.

In-Vivo Measurement

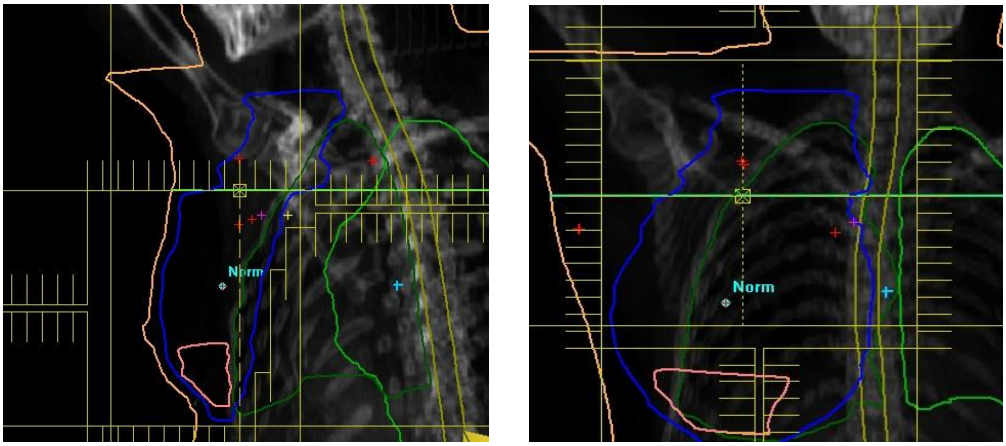
To perform the *in-vivo* dose measurements, immediately after the patient was set up and before starting treatment for all radiation fields, the diode was carefully taped to the patient. This was done in two ways, depending if it's a symmetric or asymmetric field.

For Symmetric Fields:

After satisfactory patient set up, the diodes were placed on the crosswire at the central beam with adhesive tapes over them for support.

For Asymmetric Fields:

After satisfactory patient setup, the diodes were placed 2 cm from the field edge along one of the cross wires. On the other hand, if these were closer to the edge than 2 cm, the diodes were placed centrally into the field.



a) Beam eye-view of asymmetric field **b) Beam eye-view of symmetric field**

Figure 1: Beam eye-view of symmetric and asymmetric fields

In positioning the diode, the standard position taken for the diode was the central axis of the beam. This point is easily identified, and the dose at this point is easily calculated. It was important that the diode is firmly attached to the patient, and was done with adhesive tape. In the treatment cases where it was difficult to attach the diode at the isocentre due to certain conditions, it was appropriate that we choose a part of the field outside the isocentre where it was favourable. However, care was taken that the diode position was not closer than 2 cm to the beam edge. In the case of a head and neck patient being treated in an immobilization mask, the diode was placed on the surface of the mask shell. However, care was taken that the dose was calculated for the correct position by considering the inverse square correction factor when the field goes through the couch, and the diode was placed on the surface of the couch.

For breast cases, the uncertainties resulting from the angular dependence of the beam was analysed. In these cases, a measurement point was found which could be uniquely defined and at which the expected dose could be calculated. If for some reason the diode could not be placed on the beam axis and a wedge was used, the diode was moved away from the beam axis

in the unwedged direction so that the influence of the wedge could be more easily calculated.

The dose measured by the diode detector is the maximum of the Percentage Depth Dose (PDD) or Tissue Maximum Ratio (TMR), and is referred to the corresponding maximum depth, $d_{\max}$ in water. For a 6 MV beam, the depth corresponding to the dose at $d_{\max}$ is equal to 1.5 cm depth, and for 15 MV beam, the depth corresponds to 2.5 cm depth.

The diode values (per field for each treatment case) on the electrometer for the respective energies being used was recorded in an oncology patient record system (MosaiQ) together with their calculated deviations in percentage (%).

In the situations when the beam passed through the couch, the measurement readings were multiplied by appropriate correction factor (inverse square factor) and compared to the values calculated by the planning system. Furthermore, the discrepancy δ between the *in-vivo* measured entrance dose ($Dose_{diode}$) and expected dose ($Dose_{\exp ected}$), as computed by treatment planning system was calculated for each patient and for each field, according to the relation:

$$\delta(\%) = \frac{Dose_{diode} - Dose_{\exp ected}}{Dose_{\exp ected}} \times 100$$

δ was calculated for each patient in each measurement session to estimate the difference between the actually measured, and the TPS calculated dose at the central beam. Finally, we calculated the mean dose discrepancy (δ_{mean}) for each patient's treatment case, and its standard deviation (σ).

In this study, action level is defined as an acceptable level for the percentage difference between the expected TPS dose and the measured diode dose. All treatment cases including the wedged fields were considered appropriately and assigned to a ±10% action level taking into account the accuracy and reproducibility of measurements.

If the discrepancy exceeded the action level, a thorough investigation of treatment parameters was performed together with a scrupulous review of the treatment plan. In this investigative process, all the parameters of treatment such as treatment planning steps, data transfer, patient set-up, choice of right diode corresponding to energy of radiation field, diode set-up, and SSD correction, were strongly verified.
The entrance dose measurements were performed during treatment sessions for 101 patients with prostate, cervix, breast, and head and neck tumours. The dose calculations were performed using a CT-based three dimensional treatment planning system (TPS).

In-Vivo Phantom Measurements

The accuracy of diode measurements was evaluated by using solid water phantoms. This was done under two conditions. In the first measurement, the diode was placed on the couch for fields coming from the posterior direction and thus exiting through the couch. The couch attenuation correction factor was applied and compared to that of diode reading. The second measurement was done with the diode placed on the phantom and various beams (e. asymmetrical and wedged fields from the TPS) exposed and compared with diode readings.
Also in some cases where the measurements did not agree within the action level set for patient *in-vivo*, an anthropomorphic phantom was used for the investigation. This was to check whether the system had a problem. In this case, CT images were taken for the specific localization of the phantom and sent to the treatment planning room (TPR), and a similar treatment plan was generated using the Ocentra Masterplan TPS. The treatment for that plan was then delivered to the phantom, and the "*in-vivo"* measurement was performed. The measured dose values were then compared with the calculated doses.

3. RESULTS

In-vivo measurements

A detailed result of the *in-vivo* measurements on patients with the diodes are presented in Table 1 for the different patient irradiation sites.
The values of mean (δ) and standard deviation (σ) of the distribution of discrepancies between the measured and expected entrance doses are presented in Table 1 together with the percentage of measurements (%N) for which the discrepancy was within the ±10% tolerance level.

Table 1: Discrepancies between the measured and TPS entrance Dose

SITE	Cases	N	δ_{avg}%	σ%	%N ($\lvert\Delta\rvert \le 10$%)
Head & Neck	32	136	2.16	4.61	95.59%
Prostate	41	173	4.57	3.65	96.53%
Cervix	12	47	4.69	2.58	97.87%
Breast	16	64	5.02	5.19	89.06%
Overall	101	420	3.87	4.30	95.24%

N represents the number of measurements, σ is the standard deviation of the distribution, δ_{avg} is the average discrepancy in percentage and $\%N\ (\lvert\Delta\rvert \le 10\%)$ is the percentage of measurements within the ± 10% tolerance level. The measurements are sorted according to the site: cervix, head and neck, breast, and prostate measurements.

A total of 101 treatment cases were considered for this study, and 420 field measurements were made with diodes for prostate, cervix, breast and head and neck cancer patients. All the patients had their treatments planned with the Ocentra Masterplan TPS.

The external beam irradiation technique for the tumour regions like cervix, head and neck had a number of combination of treatment fields such as anterior posterior (AP), posterior anterior (PA), right lateral (RL) and left lateral (LL) open beam combined with two opposing lateral wedge fields. A collapsed-cone convolution algorithm was used to calculate dosage, with a dose grid size spacing of 0.3 × 0.3 mm. For the breast cases dynamic/motorized wedges were used for missing tissue compensation. All the patients included in this study were positioned in supine position. For each patient, computed tomography (CT) scans were acquired using Somatom Emotion CT scanner. The Axis Distance (SAD) technique was used in all cases, and the dose prescription was made at the isocentre. Patient treatment were delivered using the Elekta Synergy platform Linear accelerator equipped with MLC to execute 3DCRT.

Figure 2, 3, 4 and 5 show the first standard deviation for each patient. In Figure 2, the data for 12 patients with cervix cancer are provided. In Figure 3, data for 16 breast cancer patients are shown, while in Figure 4 and Figure 5, data for 41 prostate cancer patients and 32 head and neck cancer patients are displayed respectively.

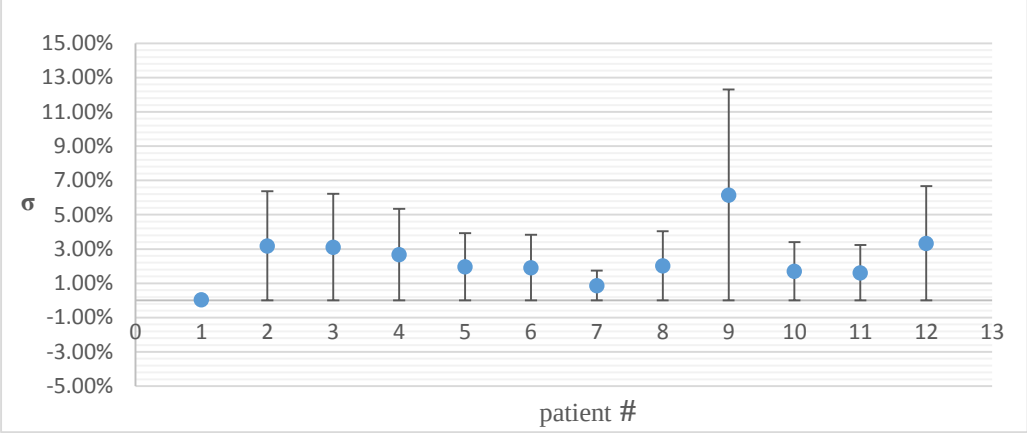


Figure 2.Standard deviation in dose for cervix cancer treatments using Elekta Synergy Platform Linac

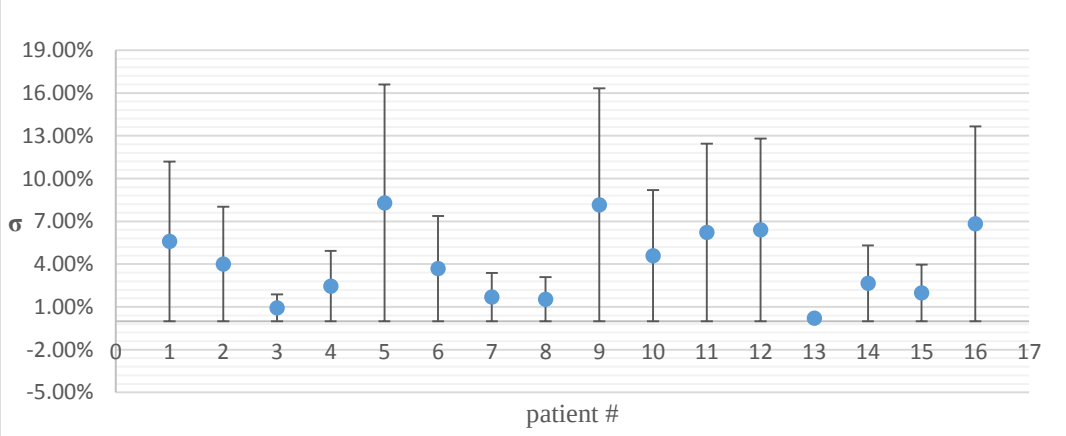


Figure 3: Standard deviation in dose for breast cancer treatments using Elekta Synergy Platform Linac

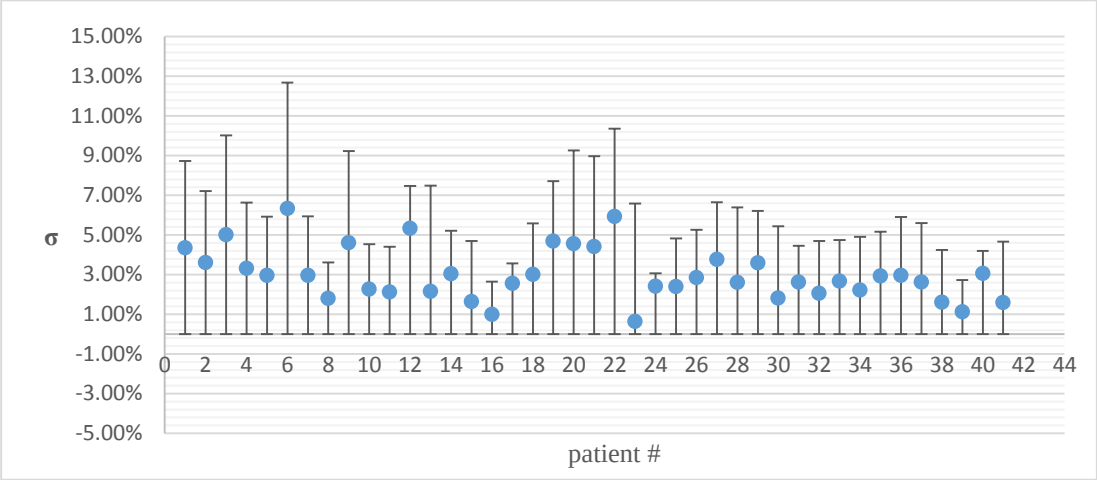


Figure 4: Standard deviation in dose for prostate cancer treatments using Elekta Synergy Platform Linac

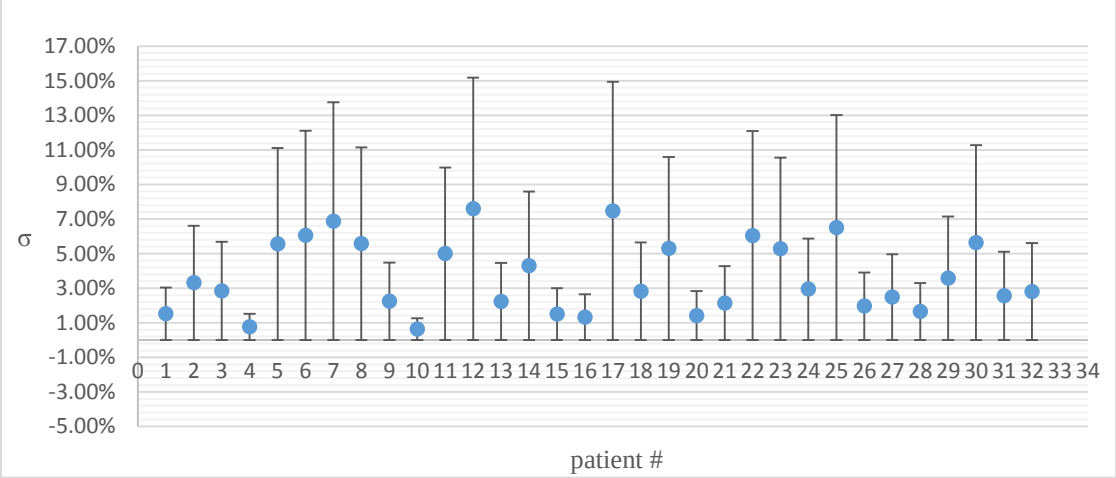


Figure 5: Standard deviation in dose for Head and Neck cancer treatments using Elekta Synergy Platform Linac.

Tables 2, 3, 4 and 5 of Appendix A show the average discrepancy for each patient, averaging the values on all the fields used in treating the patient. It has to be pointed out that, on these tables (Tables 2 to 5), only average values are presented.

4. DISCUSSION

In-vivo dosimetry results for patients with head and neck, cervix and prostate cancer have shown better results compared to patients with breast. Out of the 64 field measurements made for the breast cancer patients, seven (7) fields had their discrepancies outside the ±10% action level set. For the prostate cases (173 field measurements) and head and neck cases (136 field measurements), six (6) fields discrepancies outside the ±10% action level was recorded for each case. The cervix treatments however recording only one (1) field discrepancy outside the set action level, indicating a higher degree of accuracy.

The study revealed that, in the case of the prostate measurements, the higher discrepancy in the doses for the particular fields (exceeding the action level) may have been due the reason that the isocentre very close to the jaws and multi-leaf collimator of the LINAC machine. As a result, scatter from the jaws and the multi-leaf collimator could possibly contribute to the high dose to the diode, hence a probable higher discrepancy of the dose. In the case the breast treatment, the higher dose discrepancy observed in three of the fields may possibly have been due to the effect of wedge on the diode readings. During the breast cancer

treatment, precise positioning of the diode turned out to be a problem, especially in combined wedged and oblique fields. This had been due to the inability to place the diodes accurately at the central beam line.

Also, in a scrupulous investigation made, it was clearly revealed that some of the dose discrepancies observed may possibly have been the signal variation of diodes with tilt angles. During the in-vivo measurements, large angles of incidence were sometimes observed, especially during breast cancer measurements. In Table 3 of Appendix A, the case of patients #16 provides an example. In this case, some of the frontal fields had angle of incidence close to 60^0. Owing to difficulties in finding the actual angle of incidence, no correction factor was used, consequently influencing the accuracy of the in-vivo dosimetry, and the likelihood to lead to an observed high discrepancy.

It was also observed that, during some of the treatment sessions, the diodes were slightly displaced as a result of loosening of the adhesive tape used. The diodes therefore had recorded doses outside the isocentre, leading to some of the observed discrepancies.

As indicated in Table 1, approximately 95.59 % of the 136 measurements for the head and neck treatments were within the SGMC ±10 % action level. In the case of the cervical cancer treatments, 97.87 % of 47 measurements made were within the ±10 % action level, whereas the breast and prostate treatments produced 89.06 % (64 measurements) and 96.53 % (173 measurements) respectively within the set action level.

For the Head and Neck treatment, the average discrepancy, δ_{avg} for 136 measurements (N) was 2.16%, while the corresponding standard deviation (σ) was 4.61%. These discrepancies are lower than those observed in the other treatment cases. The results for prostate treatments (N=173) indicate an average discrepancy, δ_{avg} of 4.57% and a corresponding σ of 3.65%. For cervix (N=47), an average discrepancy δ_{avg} of 4.69% was obtained with a corresponding value σ of 2.58%, while 5.02% and 5.19% was obtained for δ_{avg} and σ respectively for the breast cases.

The analysis of 420 measurements showed 95.24% accuracy in dose delivery, and delivered an overall mean of δ = 3.87%, and the standard deviation of σ = 4.30%. This indicates that the combined uncertainty of the treatment delivery and in-vivo dosimetry at the Sweden-Ghana Medical Centre is 4.30%.

5. CONCLUSION

In summary, *in-vivo* dosimetry is an effective method to detect errors in radiotherapy, to assess clinically relevant differences between the prescribed and delivered doses, to reduce potential harm to patients, and to fulfil requirements set forth by national and international regulations. In this study, much greater percentage of the observed discrepancies are well within the set tolerance level. A greater percentage of the observed discrepancies were well within the set tolerance level. It is however recommended that the positioning of diode on patient's skin, and the angular sensitivity of the diodes should be reconsidered. It is also recommended that, a more accurate calculation of expected diode values be done, especially for fields which pass through the table. Further studies are being carried out to find the existence of a cluster of larger deviations at a particular gantry angular range. By these, several efforts would then be made to achieve action levels of ±5 %.

Funding

This work was supported by the Radiological and Medical Sciences Research Institute, and the Sweden-Ghana Medical Centre.

APPENDIX A: AVERAGE DISCREPANCIES IN DOSE AND CORRESPONDING STANDARD DEVIATIONS

Table 2: Average discrepancies in dose and corresponding standard deviations for cervix cancer treatments

Patient #	Average discrepancy δ_{avg}	Standard Deviation σ
1	6.64%	0.04%
2	4.39%	3.18%
3	6.22%	3.11%
4	5.61%	2.67%
5	5.95%	1.96%
6	4.42%	1.91%

7	6.97%	0.875
8	3.73%	2.02%
9	4.35%	6.15%
10	2.40%	1.70%
11	3.71%	1.62%
12	2.75%	3.33%

Table 3: Average discrepancies in dose and corresponding standard deviations for breast cancer treatments

Patient #	Average discrepancy δ_{avg}	Standard Deviation σ
1	6.20%	5.60%
2	1.03%	4.01%
3	6.30%	0.94%
4	7.76%	2.46%
5	0.63%	8.30%
6	3.66%	3.68%
7	3.41%	1.69%
8	1.37%	1.55%
9	0.68%	8.16%
10	5.02%	4.60%
11	6.60%	6.22%
12	7.37%	6.40%
13	7.15%	0.21%
14	3.86%	2.66%
15	5.47%	1.98%
16	9.67%	6.82%

Table 4: Average discrepancies in dose and corresponding standard deviations for prostate cancer treatments

Patient #	Average discrepancy δ_{avg}	Standard Deviation σ
1	6.02%	4.37%
2	5.10%	-
3	1.67%	5.01%
4	4.72%	3.32%
5	5.82%	2.96%
6	2.99%	6.34%
7	6.57%	2.97%
8	3.43%	1.81%
9	9.98%	4.61%

10	6.69%	2.27%
11	5.73%	2.14%
12	6.13%	5.33%
13	5.98%	2.16%
14	5.09%	3.06%
15	7.39%	1.65%
16	6.39%	1.00%
17	5.00%	2.56%
18	7.39%	3.01%
19	5.46%	4.70%
20	5.92%	4.56%
21	5.68%	4.42%
22	2.86%	5.94%
23	3.38%	0.64%
24	6.15%	2.43%
25	7.62%	2.40%
26	0.28%	2.86%
27	6.70%	3.78%
28	7.80%	2.61%
29	0.10%	3.61%
30	6.08%	1.83%
31	2.23%	2.63%
32	5.70%	2.07%
33	2.23%	2.68%
34	1.93%	2.23%
35	2.03%	2.94%
36	9.80%	2.97%
37	4.32%	2.63%
38	2.46%	1.61%
39	1.77%	1.12%
40	2.47%	3.07%
41	2.24%	1.59%

Table 5: Average discrepancies in dose and corresponding standard deviations for Head and Neck cancer treatments

Patient #	Average discrepancy δ_{avg}	Standard Deviation σ
1	-0.88%	1.52%
2	-1.60%	3.31%
3	4.15%	2.84%
4	2.03%	0.76%
5	3.21%	5.55%
6	1.69%	6.05%

7	-2.05%	6.87%
8	0.94%	5.57%
9	4.30%	2.24%
10	1.94%	0.63%
11	-1.83%	4.99%
12	-1.60%	7.59%
13	6.85%	2.23%
14	2.73%	4.30%
15	4.09%	1.50%
16	4.33%	1.32%
17	-1.28%	7.46%
18	2.27%	2.82%
19	1.43%	5.29%
20	3.86%	1.41%
21	-4.59%	2.13%
22	3.42%	6.04%
23	0.08%	5.27%
24	4.23%	2.94%
25	5.30%	6.51%
26	3.97%	1.96%
27	1.43%	2.48%
28	6.70%	1.65%
29	4.98%	3.58%
30	3.49%	5.64%
31	4.08%	2.55%
32	-1.77%	2.80%

APPENDIX B: DIODE CONSISTENCY MEASUREMENTS

Table 6: Diode Consistency test for 6 MV photon.

DOSE MEASUREMENTS	GANTRY ANGLE		
	0^0	90^0	180^0
1	1.00	1.00	1.03
2	1.00	1.00	1.03
3	1.00	1.00	1.03
4	1.00	1.00	1.03
5	1.00	1.00	1.03
6	1.00	1.00	1.03

Table 7: Diode Consistency test for 15 MV photon

DOSE MEASUREMENTS	GANTRY ANGLE		
	0^0	90^0	180^0
1	0.99	1.00	1.03
2	0.99	1.00	1.03
3	0.99	1.00	1.03

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