Available online at www.sciencedirect.com

ScienceDirect

journal homepage: http://www.elsevier.com/locate/rpor

Original research article

The semiconductor diode detector response as a function of field size and beam angle of high-energy photons

Tomasz Koper, Anna Kowalik, Sebastian Adamczyk*

Dep. of Medical Physics, Greater Poland Cancer Centre, Garbary 15, Poznan, Poland

ARTICLE INFO

Article history:
Received 18 August 2016
Received in revised form
19 September 2016
Accepted 20 December 2016
Available online 21 April 2017

Keywords:
In vivo dosimetry
Calibration factor
Semiconductor detectors

ABSTRACT

Aim: The measurements of semiconductor diode detector response as a function of field size and beam angle of high-energy photons.
Background: In vivo dosimetry plays an important role in the therapeutic process of the patient. Because of the different orientation of the beam relative to the patient and different sizes of irradiation fields, it is extremely important to take into account the response of the detector depending on the angle and the size of the beam.
Materials and methods: In this study we used a 30 cm × 30 cm × 25 cm PMMA slab phantom. On the surface of the phantom, various semiconductor detectors were placed sequentially in two configurations, angle and tilt.
Results: For the measurements of the calibration factor based on the different value of the angle, the correction coefficient value was close to 1.00 for smaller values of the angle for all the detectors used in the energy range of 6–12 MV. For the measurements, the calibration factor based on the size of the field of irradiation to the value of the correction coefficient is 1.00 for the field of 8 cm × 8 cm and 10 cm × 10 cm. With the increase field size, the correction factor shows a linear relationship in the direction of value less than 1.00.
Conclusion: Flat Detectors – used for both photon beams generated by the accelerating potential of 6 MV and 20 MV show a greater angular dependence than the cylindrical detectors. Also, the repeatability of measurements made using the flat detector is less as evidenced by larger standard deviations for the results.

1. Background

Radiotherapy is one of the main treatments for cancer patients. The basis for the effectiveness of the radiotherapy is to ensure consistency between the dose that the patient received during the irradiation and the one that was planned. It is essential to assure that the tumour target actually receives the prescribed dose. If the dose is insufficient, the treatment may not be effective. Likewise, if the dose is too high, it may affect the surrounding healthy tissues. For this reason, the role of in vivo dosimetry is crucial.

* *Corresponding author.*
E-mail address: annakowalik@poczta.onet.pl (A. Kowalik).
http://dx.doi.org/10.1016/j.rpor.2016.12.004
1507-1367/

Radiation therapy is designed to accurately deposit a specified dose to the irradiated volume. In accordance to the recommendations of the International Commission on Radiation Units and Measurements (ICRU) formulated in Report No. 24, accuracy should be within ±5%. Compliance with these guidelines is not easy to achieve even in optimal conditions due to numerous factors that can cause partial errors.[1] In this sense, in vivo dose measurements play a key role in radiotherapy because this allows us to verify that the dose actually delivered to the patient during radiotherapy complies with the planned dose.

The purpose of in vivo dosimetry is to verify that the treatment is carried out as prescribed. Together with other treatment verification tools used in radiotherapy, in vivo dosimetry constitutes a part of the quality management system in a radiotherapy department. It is a suitable method both to monitor delivery and to detect various errors early in the course of treatment. It may help to limit escalation of an error to subsequent treatment session for a particular patient and to avoid systematic errors affecting many patients. Even if no errors are detected, the in vivo measurement provides a treatment record confirming that the dose was delivered correctly within the expected tolerance.

The function of in vivo dosimetry is to measure the actual dose that a patient receives during a therapeutic session, and this is achieved using detectors. However, given that the treatment plan is composed of several radiation beams oriented at various angles relative to the patient, and the irradiation fields are of varying sizes, the detector response may change as a function of beam angle and field size. There are only a few studies on this issue—one performed by the European Society for Radiotherapy and Oncology (ESTRO)[6] and one by the American Association of Physicist in Medicine (AAPM).[13]

The AAPM report reported that in vivo dosimetry can be used to identify major deviations in treatment delivery in radiotherapy. It is known that the diode response varies significantly with the treatment beam setup. It has been found that it is necessary to play additional correction factors to take into account their response as a function of source-to-surface distance (SSD), field size, wedle field, and diode orientation.[9–11]

In order to fill this knowledge gap in the literature, we conducted the study presented here. The aim of this study was to examine the influence of field size and the incidence angle of the high-energy photon beam generated by a linear accelerator on the semiconductor detector response. It is necessary to provide quality assurance of semiconductor detectors and to make sure that a right correction factor is used.

2. Aim

The measurements of semiconductor diode detector response as a function of field size and beam angle of high-energy photons.

3. Material and methods

All measurements were performed on Clinac 2300 C-D\S (Varian Medical Systems Palo Alto, CA; USA) equipped with two high-energy photon beams – 6 MV and 20 MV.

3.1. *Semiconductor detectors*

Two different types of detectors were used for the experiment: cylindrical (Sun Nuclear Corporation, Melbourne, FL, USA) and flat detectors (PTW, Freiburg, Germany). Both detector types have the same purpose: to measure the dose of irradiation that the patient receives during the treatment session, thus allowing for assessment of conformity between the measured and planned dose. These detectors are designed to be both highly sensitive and stable through the use of integral build-up shields. Apart from promoting equilibrium and eliminating electron contamination, the shields are engineered to provide optimum transverse angular response at the corresponding energy range. The detectors are placed directly on the patient's body surface for the measurement. Flat detectors and cylindrical detectors may be used for various irradiation techniques including total body irradiation (TBI). The active surface of the cylindrical detectors (Sun Nuclear Corporation, Melbourne, FL, USA) amounts to 1.5 mm^2. To perform precise measurements, it is necessary to select the appropriate detector model for the energy range. Thus, for photons with energies ranging from 1 to 4 MeV, the blue detector is used. For 6–12 MeV and for 15–25 MeV, the yellow and red detectors, respectively, are utilized.[2]

Flat detectors (PTW, Freiburg, Germany) have a semicircular shape, approximately 12 mm in diameter, with a flat bottom that helps them to conform well to the body of the patient. There are three different types of flat detectors for photon beams with a maximum energy of 1.25 MeV (60 Co) to 25 MeV and one type for the measurement of electrons. The active area of the flat detectors is 1.0 mm^2.[3,4]

4. PMMA phantom

A solid phantom—the PMMA slab phantom (PTW, Freiburg, Germany)—was used. The radiation is absorbed and dispersed in the phantom in a similar manner as occurs in human tissue because the density is similar ($\rho \approx 1.18$ g/cm^3). The total dimensions of the entire phantom (all slabs) are 30 cm × 30 cm × 30 cm. Each slab is made very precisely, keeping the tolerance of ±0.1 mm thickness.[3] The phantom is designed to allow for the measurement of photon radiation energy generated by the accelerating potential in the range of 70 kV to 50 MV and electron beam energy levels ranging from 1 to 50 MeV. To ensure dispersion, the secondary plates are placed well below the radiation detector.

5. PTW farmer cylindrical ionization chamber

The cylindrical ionization chamber ("Farmer" type) is a waterproof chamber used most often in clinical dosimetry to measure water or a solid material with a photon radiation generated by the accelerating potential of 30 kV to 50 MV and an electron beam with a nominal energy of 10–45 MeV. The cavity length is 24 mm, an inner diameter of 6.25 mm, and an outer wall thickness (graphite) of 0.37 mm. The active volume of the chamber is 0.6 cm^3. Numerous additional chambers have a

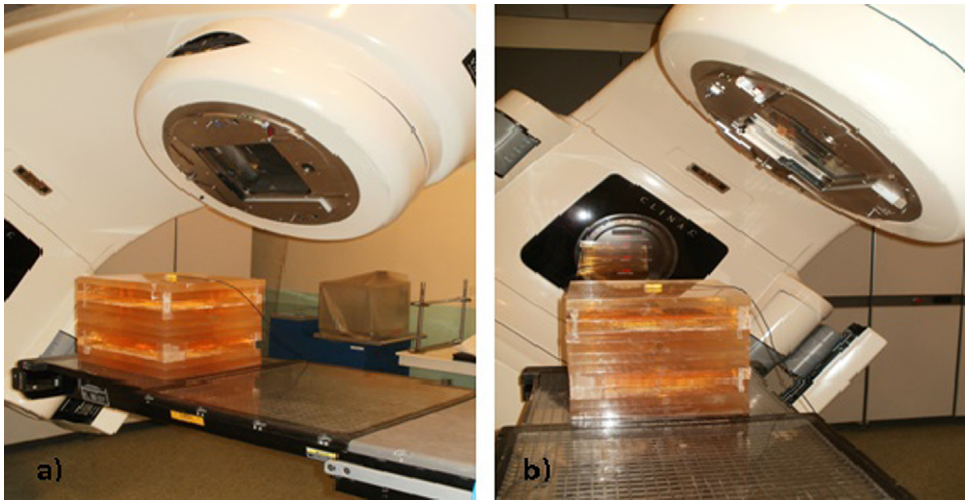

Fig. 1 – Semiconductor detectors on the surface of the PMMA phantom in two measurement configurations: (a) "angle", (b) "tilt".

special build up overlay made of plastic. These chambers are designed to ensure electron equilibrium. The Farmer type ionization chamber can be used to measure both in absolute and relative terms.[5,9,12]

5.1. *Semiconductor detector response as a function of incidence angle of the radiation beam*

To obtain the detector calibration factor for 6 MV and 20 MV photon beam quality, the following setup was used:

1. PMMA slab phantom with dimensions of 30 cm × 30 cm × 25 cm;
2. Sun Nuclear cylindrical solid-state detectors dedicated for the energy range of 6–12 MV and 15–25 MV;
3. PTW flat solid-state detectors dedicated for the energy range of 6–12 MV and 15–25 MV;
4. Dosimeter connected with semiconductor detectors (PTW Vivodos).

Measurements were performed with the gantry and collimator angle set to 0° for a radiation field size of 10 cm × 10 cm and the SSD (Source to Surface Distance – the distance between the radiation source and the surface of the PMMA phantom) of 100 cm. In each repetition the exposure time was 100 MU. The authors defined 23 different incidence beam angles for which measurements were performed.

Measurement of the input dose is not only important to assure a match between the prescribed and received doses, but it also can help to verify patient alignment accuracy, beam stability, and the beam parameters.[5] In addition, simultaneous measurements of the input and output doses are used to determine the dose at the centre of the irradiated area, and provide an opportunity to evaluate the precision of the radiotherapy process.[5]

Each detector was placed on the surface of the phantom in the central axis of the beam (CAX) in two different configurations, one called the "angle" where the long axis of the detector (and cable) is parallel to the gantry rotation axis (Fig. 3a) and the other denominated "tilt" where the long axis of the detector (and cable) is perpendicular to the gantry rotation axis (Fig. 3b). Both configurations are shown in Fig. 1. A total of four sets of measurements were performed (for both detector models in two different configurations).

5.2. *Solid-state detector calibration factor as a function of the radiation field size*

To measure the calibration factor according to the size of the radiation field applied, we used the following materials:

1. PMMA slab phantom measuring 30 cm × 30 cm × 25 cm;
2. Cylindrical solid-state detectors for the energy ranges of 6–12 MV and 15–25 MV (Sun Nuclear);
3. Flat solid-state detectors for the energy ranges of 6–12 MV and 15–25 MV (PTW);
4. Unidos and Vivodos dosimeters (PTW);
5. Cylindrical ionization chamber – "Farmer" type (PTW);
6. Thermometer and barometer (Testo AG; Lenzkirch, Germany).

The ionization chamber was placed in the CAS in the PMMA phantom at a depth of 5 cm for 6 MV photons and at 10 cm for 20 MV photons. On the upper surface of the phantom we determined the CAX where detectors were placed in sequence (after the dose was measured with an ionization chamber). The measurement setup is shown in Fig. 2.

Gantry and collimator angles were set to 0° and the SSD was 100 cm. The thermometer and barometer were used to measure and correct for the temperature and pressure in the therapeutic bunker. Exposure time was 100 MU.

There were 11 field sizes for which the measurements were performed, as follows: 5 cm × 5 cm, 8 cm × 8 cm, 10 cm × 10 cm, 12 cm × 12 cm, 15 cm × 15 cm, 18 cm × 18 cm,

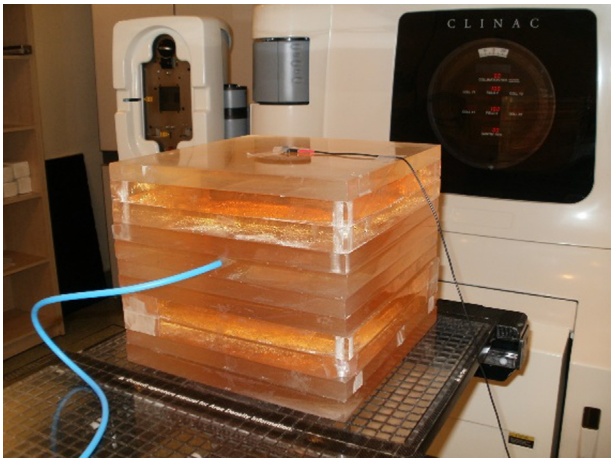

Fig. 2 – The setup used for the measurements of solid-state detector calibration factor as a function of radiation field size.

20 cm × 20 cm, 25 cm × 25 cm, 30 cm × 30 cm, 35 cm × 35 cm and 40 cm × 40 cm.

For each of the defined fields, three consecutive dose measurements were performed using a "Farmer" ionization chamber. In the next step, semiconductor detectors were placed on the surface of the phantom (in the CAX) and three measurements were taken for each of the defined field sizes.

For each field size, we calculated the ratio of the mean dose measured by the semiconductor detector to the mean dose measured by the ionization chamber. The results were then normalized to the reference field (10 cm × 10 cm). This step provides the coefficients needed to correct the calibration factor determined under standard conditions (0° gantry and collimator angle with a field size of 10 cm × 10 cm).

6. Results

6.1. *Semiconductor detector response as a function of the incidence angle of the radiation beam*

Fig. 3 shows a plot graph depicting the correction factors used for the two different types of semiconductor detectors and the two orientations (tilt and angle) as a function of gantry angle. The measurements shows that the correction factor is close to unity for smaller gantry angles, ranging from −30° to +30° for all detectors used. For the flat detectors, a notable deviation from 1.00 towards higher values (for angles of incidence of radiation <−30° and >+30°) can be observed, reflecting a greater angular dependence of these detectors versus the cylindrical detectors. The correction factors for cylindrical detectors in angle configuration indicate the closest match with reference value of 1.00 in the full range of gantry angles.

For the tilt configuration, we observed a sharp drop in dose for both detector types and thus a large change in the correction factor at a gantry angle of 90°. The difference between the value measured at 90° and the reference value (1.00) results from the cable position with respect to the beam axis. It should be noted that before reaching the active volume of the solid-state detector, the photon radiation must overcome the signal cable.

All measurements described above for the photon beam were repeated for 20 MV energies. Fig. 4 presents the correction factors determined for both the cylindrical and flat detectors. As is readily evident in Fig. 4, the correction factors for the cylindrical detector for 20 MV photons in the angle orientation are close to the reference value. However, in the tilt orientation, the correction factor is lower than unity for gantry angles ranging from −90° to 0°. Moreover, the behaviour is opposite in the range from 0° to 80°, where the correction factor is greater than the reference value.

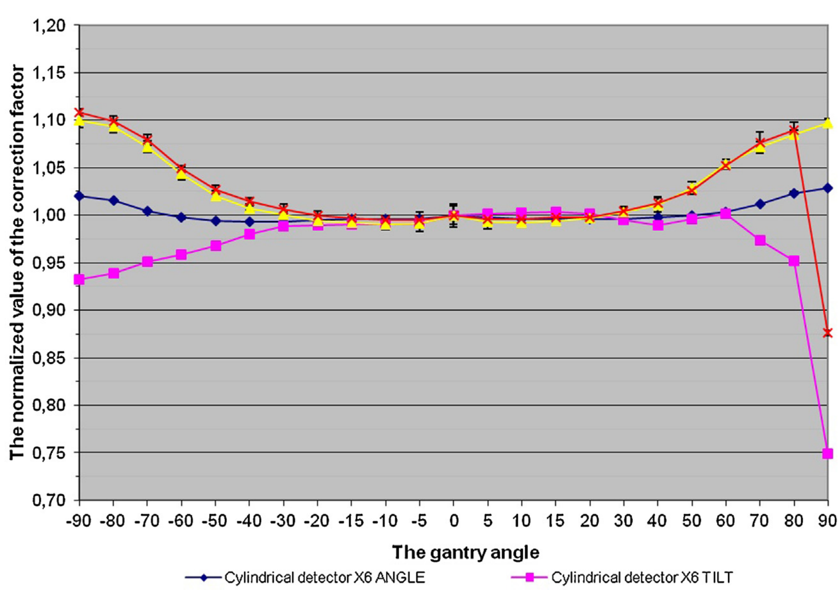

Fig. 3 – Normalized correction factors for various gantry angles for both types of semiconductor detectors and both configurations (tilt and angle) for a 6 MV photon beam.

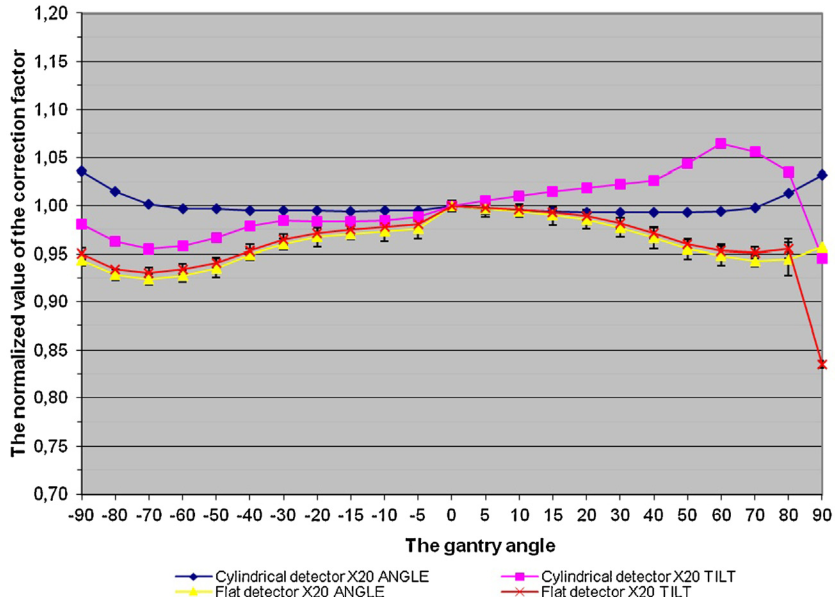

Fig. 4 – Normalized correction factors for various gantry angles for both types of semiconductor detectors and both configurations (tilt and angle) for 20 MV photon beam.

For flat detectors, the correction factor was lower than the reference value, which means a noticeable decrease in measured dose, in both the tilt and angle orientation for gantry angles between −90° and 0° as well as from 0° to 80°. As mentioned above, for the 20 MV photon beam, we also observed a significant decrease in the correction factor for a beam incidence angle of 90° for both types of detectors in the tilt orientation.

The standard deviation (SD) of the correction factor for cylindrical detector measurements ranged from 0.000–0.003 (6 MV photon beam) and 0.000–0.002 (20 MV beam). The corresponding SD of correction factors for flat detectors ranged from 0.001–0.011 (6 MV) and 0.000–0.017 (20 MV).

6.2. *Dependence of the calibration factor on field size*

Fig. 5 shows a comparison of the correction factors for the flat and cylindrical semiconductor detectors for various field sizes. All correction factors were normalized to the 10 cm × 10 cm field size. As Fig. 7 shows, the correction factor was closest to 1.00 for the field size of 8 cm × 8 cm. As field size increases, the correction factors show a linear decrease, reaching 0.950 for the flat detectors (20 MV) and 0.979 for cylindrical detectors (20 MV) for the 40 cm × 40 cm field size. For the smallest field size (5 cm × 5 cm), the correction factor ranged from 1.020 to 1.040 for both detector types.

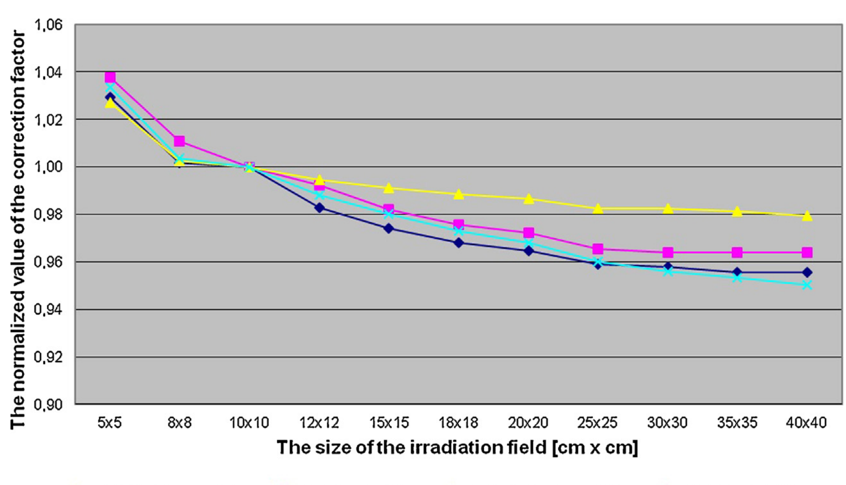

Fig. 5 – Normalized value of correction factors at various radiation field sizes and for both semiconductor detector types at 6 MV and 20 MV.

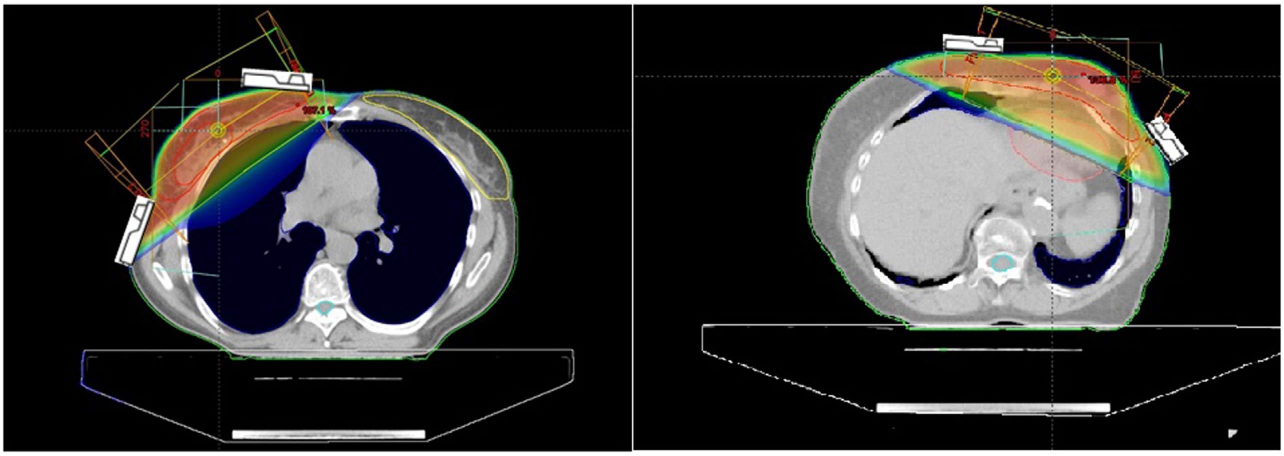

Fig. 6 – Two example scans of in-vivo dosimetry for breast cancer treatment.

7. Discussion

In vivo dosimetry plays an essential role in assuring the effectiveness and safety of radiotherapy treatment, and for this reason, multiple dose measurements are performed during the treatment. In this study, we used a PMMA slab phantom and two different types of semiconductor detectors to investigate how the response of the semiconductors varied according to the field size and beam angle. The main findings of this study were that flat detectors show a greater angular dependence than cylindrical detectors. Moreover, measurements performed with the flat detectors were less repeatable, as evidenced by a larger standard deviation for the results. These results seem to indicate that cylindrical detectors may deliver more accurate results.

During radiotherapy, particularly in modern techniques such as intensity-modulated radiotherapy (IMRT), a large number of different beam orientations and field sizes are employed to deliver high doses to the target while minimizing the radiation dose to healthy tissues. In order to measure the actual dose being delivered under these conditions, it is essential to take into account the variations in detector response as a function of the gantry angle and the field size. For certain cancers such as lung or breast cancer, the therapeutic fields are split into two smaller fields due to technical limitations, and this splitting also affects the detector response. Measurement of doses in small fields is difficult due to the detector size, which may extend outside the field or be located on the edge of the irradiation field, which results in significant differences in the measured dose.

The geometry of the irradiation field also has a very large impact on the in vivo dose measurement. Typically, in the case of patients with breast cancer, the treatment plan consists of two opposing fields. The geometry of the irradiation field for these patients often causes difficulties associated with large slanting beam inputs, which prevents the precise placement of the detector in the beam axis. Semiconductor detectors have an angular relationship with large bundle squint inputs, as shown in Fig. 6. When the angle is large, then the measured dose should be modified according to a correction factor. The difficulties in dose measurement of high incident beam angles are related to the design of semiconductor detectors and build-up.

In our study, we found that, for 6 MV photons, both flat and cylindrical detectors in tilt orientation, at a 90° gantry angle,

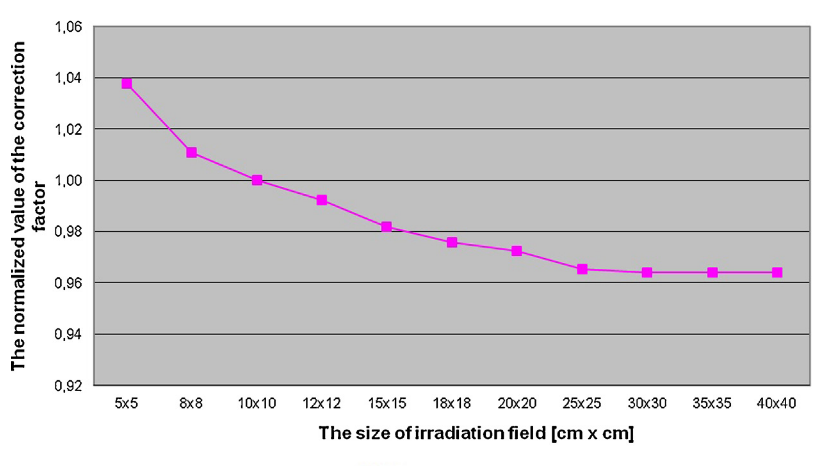

Fig. 7 – Correction factors for the 11 different irradiation field dimensions assessed for an X6 flat solid state detector.

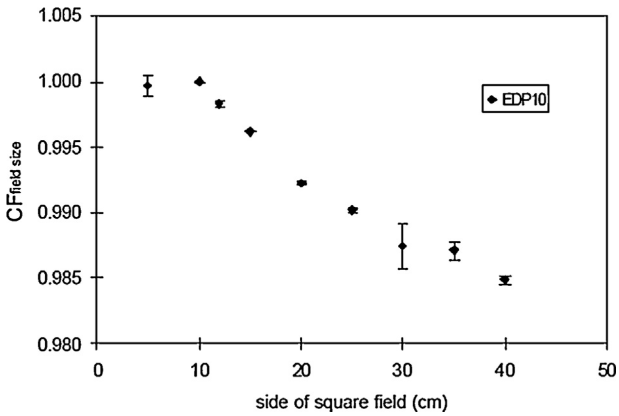

Fig. 8 – Correction factors reported by Huyskens et al.[6] for various irradiation field dimensions for the 4X6 flat solid state detector.

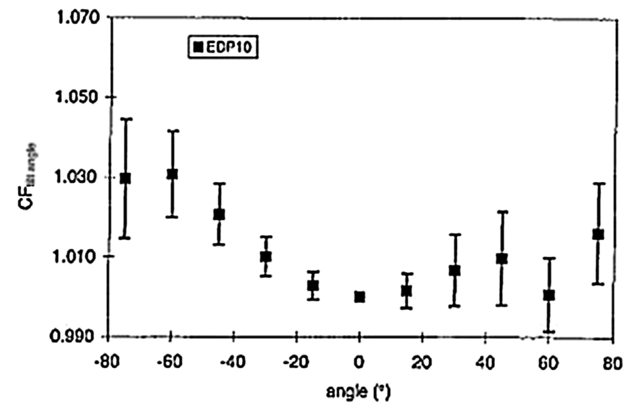

Fig. 10 – The values of the correction factors for different values of the angles of the head in the case of a flat solid state detector configuration X6 tilt.[6]

the correction factor was significantly reduced (0.75 and 0.88 for cylindrical and flat detectors, respectively). The behaviour of both types of detectors was similar when the photon energy generated by the accelerating potential of 20 MV was applied (0.94 and 0.84, respectively). This decrease is due to the position of the detector signal cable relative to the beam axis, which is an important cause of inaccurate measurements. Our study showed clearly that it is crucial to distinguish the beam angle correction factors, not only between different types of detectors but also between different detector orientations (tilt or angle).

We also found that the impact of field size on the correction factor was large. There was a significant correlation between the correction factor and the size of the radiation field. For fields smaller than 10 cm × 10 cm, the correction factor ranged from 1.00 to 1.04, thus indicating a possible overestimation of the semiconductor-measured dose by up to 4%. For larger fields, the difference was relatively smaller, ranging from 0.95 to 1.00; this result indicates a maximum possible underestimation of the measured dose of 5%.

We compared our results with those reported by Huyskens et al. in an ESTRO guidelines publication.[6] The figures below show the results of measured correction factor as a function of the radiation field size in the case of a flat detector for 6 MV photons. Fig. 7 shows the results of our measurements while Fig. 8 shows the results reported by Huyskens and colleagues. The detector response as a function of the irradiation field size in both cases is comparable for the whole range of measured values. Both studies reveal that the correction factor is highly dependent on the size of the irradiation field.

Our results were also similar to those reported by Huyskens and colleagues with regards to the relationship between the correction factor and the angle of incidence of the radiation beam.[6] Fig. 9 summarizes the results of measurements

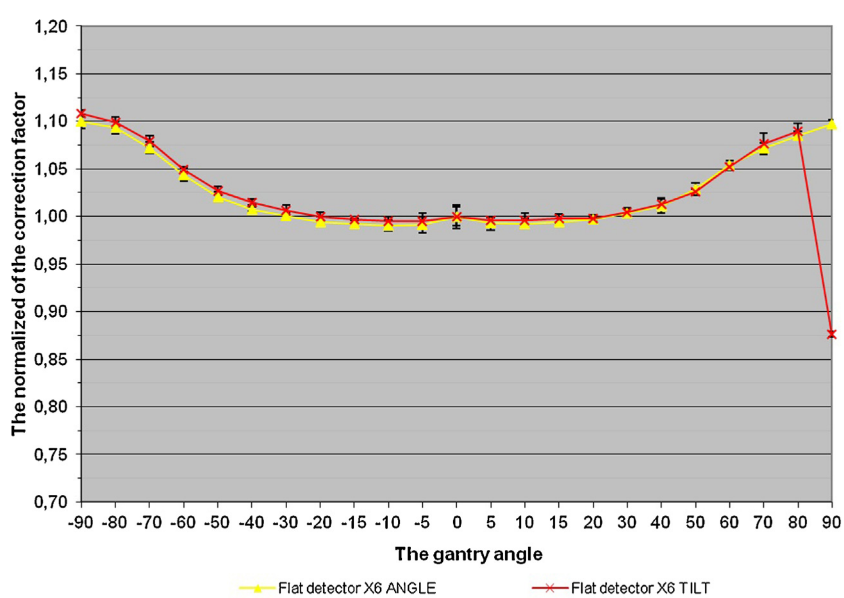

Fig. 9 – Normalized correction factors for different gantry angles for flat semiconductor detector and both configurations (tilt and angle) for 6 MV photon beam.

obtained for the flat detector used for 6 MV photon beam in both orientations (tilt and angle). Fig. 10 shows the corresponding results published by Huyskens et al.[6]

In both configurations, the detector correction coefficient for smaller values of the gantry angle (−20° to +20°) is close to the reference value (1.00). For larger gantry angle values, the correction factor increased significantly.

In 2005, the AAPM published guidelines (Report #87) on the use of diode in vivo dosimetry for patients receiving EBRT.[7] Part of that report described the angular dependence of semiconductor detectors, with results that were similar to those obtained in our study. The authors used detectors dedicated for the energy range of 16–25 MV, which—as in our study—were placed in two configurations (angle and tilt). For angles ranging from −15° to +15°, the value of the correction factors for both detector configurations were close to the reference value (1.00). For other angles (from −30° to −60° and +30° to +60°), there is an increase of the correction factor to about 1.02 for tilt orientation and to about 1.05 for angle orientation.

8. Conclusions

The main findings of this study were that flat detectors show a greater angular dependence than cylindrical detectors. Flat Detectors – used for both photon beams generated by the accelerating potential of 6 MV and 20 MV show a greater angular dependence than the cylindrical detectors. Also, the repeatability of measurements made using the flat detector is lower as evidenced by larger standard deviations for the results. Moreover, measurements performed with the flat detectors were less repeatable as evidenced by a larger standard deviation for the results. These results seem to indicate that cylindrical detectors may deliver more accurate results.

Based on our results, we believe that several conclusions can be drawn regarding the impact of field size and incidence angle on semiconductor detector response. First, detector response is highly affected by the signal cable connecting the detector and the dosimeter when the radiation has to pass through it. Second, flat detectors, due to their construction, are more prone to changing response as a function of beam angle, regardless of the photon beam energy applied. Third, dose measurements obtained for flat detectors exhibit higher standard deviations with respect to the measurements obtained for a cylindrical detector, which may indicate that the reproducibility of measurements is smaller. Finally, the correction factor has a linear relationship with respect to changes in the size of the irradiation field. We conclude from the results of this work that high-energy build-up diodes (Sun Nuclear Corporation) can be used for in vivo dosimetry in the entire megavoltage energy range used in radiotherapy. We suggest from our results that cylindrical detectors may be more accurate and do not need to use a lot of correction factors to calculate the in vivo dose than flat detectors. However, a calibration factor of the diode system needs to be checked periodically because of the well-known damage with the X-ray energy.[8]

Conflict of interest

None declared.

Financial disclosure

None declared.

REFERENCES

1. IAEA International Atomic Energy Agency. *Absorbed dose determination in external beam radiotherapy. An international code of practice for dosimetry based on standards of absorbed dose to water. Technical Reports series No. 398*. Vienna: IAEA; 2000. p. 1.
2. Sun Nuclear. *ISORAD detector user's guide. Photon detector for in vivo dosimetry*; 2008.
3. PTW Freiburg: www.ptw.com, daty wejścia: 19, 21.03.2011r.
4. PTW Freiburg. *User manual in-vivo semi-conductor probe*; 2002.
5. PTW Freiburg. *User manual VIVODOS. Firmware version 2.30*; 2002.
6. Huyskens DP, Bogaerts R, Verstraete J, et al. *Practical guidelines for the implementation of in vivo dosimetry with diodes in external radiotherapy with photon beams (entrance dose)*. ESTRO; 2001, 74–75, 80–81.
7. American Association of Physicists in Medicine. *Diode in vivo dosimetry for patients receiving external beam radiation therapy*. USA: Medical Physics Publishing; 2005. p. 42–3.
8. Colussi VC, Beddar AS, Kinsella TJ, Sibata CH. In vivo dosimetry using a single diode for megavoltage photon beam radiotherapy: implementation and response characterization. *J Appl Clin Med Phys* 2001;**2**(fall (4)).
9. Fontenla DP, Yaparpalvi R, Chui CS, Briot E. The use of diode dosimetry in quality improvement of patient care in radiation therapy. *Med Dosim* 1996;**21**:235–41.
10. Alecu R, Loomis T, Alecu J, Ochran T. Guidelines on the implementation of diode in vivo dosimetry programs for photon and electron external beam therapy. *Med Dosim* 1999;**24**:5–12.
11. Zhu XR. Entrance dose measurements for in vivo diode dosimetry: comparison of correction factors for two types of commercial silicon diode detectors. *J Appl Clin Med Phys* 2000;**1**:100–7.
12. Rutonjski L, Petrović B, Bauca M, Teodorović M, Čudić O, Basarić B. In vivo dosimetry as a part of quality assurance program for breast cancer irradiation. *Phys Med* 2014;**30**(Suppl. 1):e75.
13. Al Kattar Z, El Balaa H, Zahran S. Uncertainties of in vivo dosimetry using semiconductors. *J Life Sci* 2015;**9**:120–6.