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

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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.

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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