EVALUATION OF CALIBRATION OF IN VIVO DOSIMETERS FOR ASSESSING DOSE IN RADIOTHERAPY

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Abstract

The primary purpose of in vivo dosimetry in external beam radiotherapy is to assess clinically significant differences between the planned and delivered doses, to record the dose received by individual patients, and to evaluate the "useful amount" of the given dose expressed in Gy. The in vivo dosimeters employed are p-n type semiconductors. Before these dosimeters are used clinically to estimate the dose delivered to a patient, they must undergo calibration and evaluation in terms of dosimetry. By comparing the radiation doses produced by the linear accelerator and those detected on the patient's surface with the theoretical results calculated by the treatment planning system, we obtain a clear picture of the effective radiation for external beam radiotherapy treatments. A more ambitious goal of in vivo dosimetry is to control the target dose to verify the exact amount of radiation delivered.

Key words: *in vivo dosimeters, calibration, semiconductors.*

Përmbledhje

Qëllimi kryesor i dozimetrisë in vivo në radioterapinë me rreze të jashtme është të vlerësojë ndryshimet klinikisht të rëndësishme midis dozave të planifikuara dhe të dhëna, të regjistrojë dozën e marrë nga pacientët individualë dhe të vlerësojë "sasinë e dobishme" të dozës së dhënë të shprehur në Gy. Dozimetrat in vivo të përdorur janë gjysmëpërçues të tipit p-n. Përpara se këta dozimetra të përdoren klinikisht për të vlerësuar dozën e dhënë pacientit, ata duhet t'i nënshtrohen kalibrimit dhe vlerësimit në aspektin e dozimetrisë. Duke krahasuar dozat e rrezatimit të prodhuara nga përshpejtuesi linear dhe ato të zbuluara në sipërfaqen e pacientit me rezultatet teorike të llogaritura nga sistemi i planifikimit të trajtimit, marrim një pamje të qartë të rrezatimit efektiv për trajtimet e radioterapisë me rreze të jashtme.

Një qëllim më ambicioz i dozimetrisë in vivo është të kontrollojë dozën e synuar për të verifikuar sasinë e saktë të rrezatimit të shpërndarë.

Fjalë kyçe: *dozimetra in vivo, kalibrim, gjysmëpërçues.*

Introduction

Input and output dose control provides an indirect measure of the target dose. If discrepancies are observed between the calculated and measured input and output dose values, it may indicate an incorrect target dose due to errors in monitor unit settings, radiation parameters, or unexpected machinery problems. Therefore, a more accurate determination of the target dose is essential.

Despite the recognized importance of quality control in radiotherapy, practical implementation remains incomplete due to the complexity and multiple steps involved in the radiotherapy process. While certain physical steps, such as basic dosimetry and mechanical condition checks of devices, are well-covered in quality assurance, other steps remain less clear. For instance, there is ongoing uncertainty in quality control for treatment plans, the organization, and the provision of monitoring units.

Controlling the absorbed dose in practice is critical, typically achieved by placing dosimeters on the patient's skin or within natural cavities, a method known as in vivo dosimetry. This methodology, the subject of this paper, is employed to identify errors during patient treatment, detect issues in essential procedures, evaluate the quality of specific treatment techniques, and estimate doses in cases where calculations are incorrect.(AAPM Report No 32, 1991).

This paper aims to provide information on the recommended methodology for high-energy photons, based on patient treatment conditions. It highlights the typical characteristics of detectors used in in vivo dosimetry for high-energy photons, such as semiconductors and diodes. Semiconductors are particularly noted for their relative constancy of response over time, similar to ionization chambers, despite a possible decrease in sensitivity with accumulated dose. They can be calibrated individually, although this calibration requires periodic checking more frequently than ionization chambers. Factors such as dose level and temperature further complicate their use. Nonetheless, when utilized by professionals, semiconductors have proven extremely useful in in vivo dosimetry, significantly contributing to quality assurance (IAEA, TSR No. 469, 2009).

The accuracy of calculating the dose on the skin is debatable due to the varying charges deposited by photon or electron bunches with different energies (Nordstrom, 2012). Consequently, the detector signal on the patient's surface must be converted to dose (charge-dose relation). For this purpose, two points are defined: the entrance area of the radiation beam and the exit area of the photon beam, corresponding to the input dose and the output dose, respectively. In practice, there is considerable measurement loss due to backscatter at the exit dose, necessitating correction.

In radiotherapy treatment with external radiation, skin burning is often observed. This phenomenon, warranting further study, is influenced not only by the percentage of charge deposited in the skin by high-energy particles but also by the Cherenkov effect.

The Cherenkov effect in breast radiation treatment occurs due to the interaction of high-energy radiation with the tissue, leading to the emission of visible light. Here's a detailed explanation of the phenomenon:

The Cherenkov effect, is the emission of light that occurs when a charged particle, such as an electron, travels through a dielectric medium (e.g., water or tissue) at a speed greater than the phase velocity of light in that medium. This effect is analogous to the sonic boom created by an object traveling faster than the speed of sound.

In breast radiation treatment, high-energy photons (X-rays) or electrons are directed at the breast tissue to destroy cancer cells. When these high-energy particles interact with the tissue, they can displace electrons from atoms within the tissue. These displaced electrons gain kinetic energy and travel through the tissue at high speeds (Nordstrom, 2012). If the velocity of these electrons exceeds the speed of light in the tissue (which is slower than the speed of light in a vacuum), the Cherenkov effect can occur. The Cherenkov effect can be used to visualize the radiation beam in real time, providing a way to verify that the radiation is being delivered to the correct area. This can be particularly useful for ensuring accurate treatment delivery and improving the precision of radiation therapy. Researchers are exploring the use of Cherenkov light as a non-invasive dosimetry method to measure the dose of radiation received by the tissue.

By capturing and analyzing the Cherenkov light, it may be possible to gain insights into the distribution and intensity of the radiation dose. The Cherenkov effect in breast radiation treatment is a result of high-energy

radiation causing electrons to travel through tissue at speeds greater than the speed of light in that medium, leading to the emission of visible light. This phenomenon has potential applications in visualizing and verifying radiation delivery, as well as in developing advanced dosimetry techniques to enhance the accuracy and effectiveness of radiation therapy. It is possible to check the Cherenkov effect with in vivo detectors. In fact, the detection and analysis of Cherenkov radiation can be used to improve the accuracy and effectiveness of radiation therapy. Here in vivo detectors can be utilized to observe and measure the Cherenkov effect. In vivo detectors, used for Cherenkov radiation detection, need to be carefully calibrated to ensure accurate measurements (Castro et al., 2008).

The sensitivity of these detectors should be high enough to detect the faint Cherenkov light while filtering out noise and other light sources. In vivo detectors can indeed be used to check the Cherenkov effect during radiation therapy. The use of optical fiber probes, photodiodes, PMTs, and specialized imaging systems allows for the real-time detection and analysis of Cherenkov radiation. This capability enhances the precision and accuracy of radiation therapy by providing detailed information on dose distribution and treatment delivery, ultimately improving patient outcomes.

Methodology

To achieve accurate in vivo dosimetry, it is recommended to perform separate calibrations for each radiation bundle used with the diode. Practically, it is beneficial to connect an electrometer and multiple diodes to each treatment unit, thereby limiting the number of bundles requiring diode calibration.

When using a diode for both input and output dose measurements, it is necessary to establish a calibration factor for each (IAEA, TSR No.469, 2009)

The input dose calibration factor is defined as the multiplier applied to the diode's input R_{sc} signal under reference conditions to produce the input dose. The input dose calibration factor is given by:

$$F_{entry} = \frac{D_{entry}}{R_{sc,entry}}$$

Initially, measurements were made using an ionization chamber with a field size (FS) of 10 cm x 10 cm to verify if the machine delivers the calibrated dose. The room temperature was T = 23 °C, and the room pressure was P =

1005 hPa. The source-surface distance (SSD) was set to 100 cm, and the load used was 100 monitor units (MU).

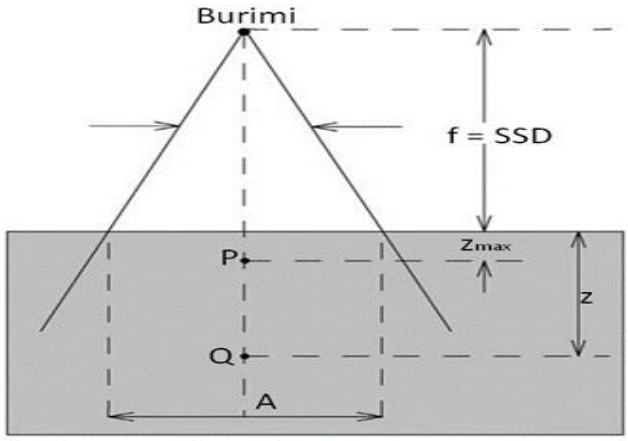

Figure 1. Geometry of measurement and evaluation of PDDs

Given that the reference depth for a 1 Gy dose is d_{ref} = 10 cm for a water phantom and 8.5 cm for PMMA (polymethyl methacrylate) material, the maximum dose occurs at a depth of 1.6 cm for a field size of 10x10 cm² for the 6 MV energy used It is important to note that the electrometer displays the charge in nanocoulombs (nC). Our task is to convert this charge to dose using the following formula:

$$D_{W,Q} = N_{D,W} * k_Q * M_Q$$

For a specific photon beam with a given Source-to-Surface Distance (SSD), the dose at a point P (at depth z_{max} in the phantom) is influenced by the field size A. The Relative Dose Factor (RDF) is calculated as the ratio of the dose at point P in the phantom, denoted as D_p (z_{max}, A, f, hν), for a field size A, to the dose at point P in the phantom for a field size of 10 x 10 cm², denoted as D_p (z_{max}, 10, f, hν) (AAPM Report 87, 2005):

$$RDF(A, h\nu) = S_{cp}(A, h\nu) = \frac{D_p(z_{max,}A, f, h\nu)}{D_p(z_{max,}10, f, h\nu)}$$

The geometry for measuring RDF (A, hν) is shown as in the figure below; (a) the geometry for measuring $D_p(z_{max,}A, f, h\nu)$ and (b) the geometry for measuring $D_p(z_{max,}A, f, h\nu)$. From the basic definitions of CF and SF we can write that RDF is the ratio:

$$RDF(10, h\nu) = \frac{D_p(z_{max,}A, f, h\nu)}{D_p(z_{max,}10, f, h\nu)} = \frac{D'_p(A, h\nu)PSF(A, h\nu)}{D'_p(10, h\nu)PSF(A, h\nu)}$$
$$= CF(A, h\nu)\ SF\ (A, h\nu)$$

or, according to Khan's notion:

$S_{c,p}(A, h\nu) = S_c\ (A, h\nu)S_p\ (A, h\nu)$

This indicates that the Relative Dose Factor (RDF) comprises two components: scattering from the collimator and scattering from the phantom.

Typical values for $RDF(A, h\nu)$, $CF(A, h\nu)$, and $SF\ (A, h\nu)$ versus field size A are illustrated in the corresponding figure. All three functions are normalized to 1 for a field size of A = 10 x 10 cm². They are greater than 1 for area sizes A > 10 x 10 cm² and less than 1 for area sizes A < 10 x 10 cm².

When additional accessories are used, such as Multi Leaf Collimators (MLCs), to shape the radiation beam on the patient's surface, resulting in a non-regular field B, so $RDF(B, h\nu)$,is approximated by the following equation:

$$RDF(B, h\nu) = CF\ (A, h\nu)SF(B, h\nu)$$

where A represents the field formed by the machine's collimator and B is the non-regular field applied to the patient's surface. It's important to note that the formation of the beam depends on the construction of collimating blocks or MLCs.

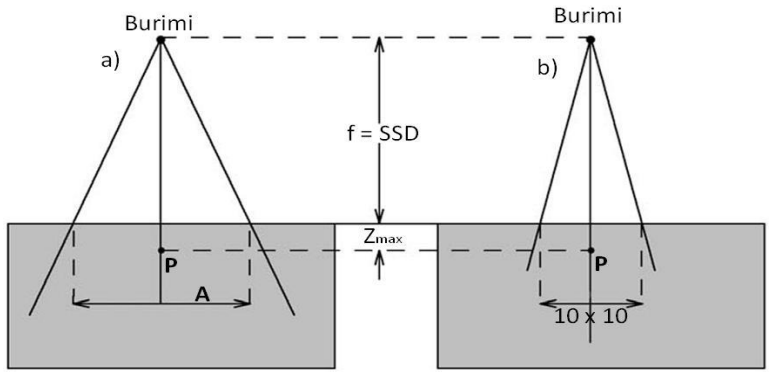

Figure 2. Geometry of measurement and evaluation of CF factor

Results

During irradiation, electron-hole pairs are produced in a range that increases with the dose level (dose per unit time). If the dose level is high, as with the pulsed radiation produced by a linear accelerator, a phenomenon called "collision" occurs. In this context, ions are produced with such high momentum that recombination cannot keep pace, leading to more charges avoiding recombination compared to lower dose levels. Consequently, the sensitivity of the diode decreases as the pulse dose for radiation decreases. This phenomenon is more evident in n-type diodes than in p-type diodes. Newer p-type detectors, with higher doping levels, exhibit limited dependence on dose level (Van Dam et al., 2006, p.12)

Given the difficulty in achieving the exact same dose level every time, diodes are calibrated to the dose level that will be used for the patient.

- Diodes used for entrance dose measurements can be calibrated at the isocenter or for a reference collimator aperture. The same calibration factor can then be applied to other collimator apertures.
- Diodes used to measure the output dose can be calibrated in the output position, for a patient of medium thickness, with the input surface beam at the isocenter and again for a reference collimator aperture.
- Diodes used for low-dose total body irradiation, where the dose level is achieved by increasing the source-surface distance (SSD), must be calibrated under these specific conditions. Using the calibration factor determined for common conditions can result in an error of up to 10%, which is highly unacceptable.

Below we are giving the values obtained during the calibration of the detectors we have available for energy 6MeV and 18MeV

Table1: Results of diode input dose measurements for energy 6 MeV:

Channels	Serial number	Charge (nC)			Calibration factor (Gy/C)
1	00083	-10.13	-10.18	-10.17	$1.448 * 10^7$
3	00082	-10.12	-10.12	-10.11	$1.453 * 10^7$
4	00078	-10.05	-10.08	-10.11	$1.458* 10^7$
2	Ionizing chamber	12.44	12.43	12.44	$5.332 * 10^7$

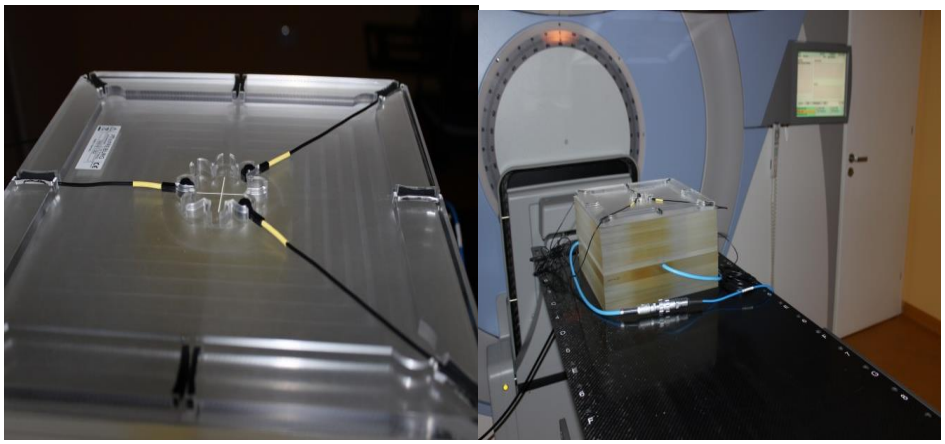

Figure 3: In vivo dosimeters diode detectors calibration set up

Figure 4 shows the plot of Percent Depth Dose (PDD) for 6 MV energy. At the maximum depth (d_{max}), which is 1.6 cm for the water phantom and 1.36 cm for PMMA, the dose reaches its peak. Given that at a depth of 10 cm the dose is 1 Gy, which is 68% of the maximum dose, the dose at d_{max} is 1.47 Gy.

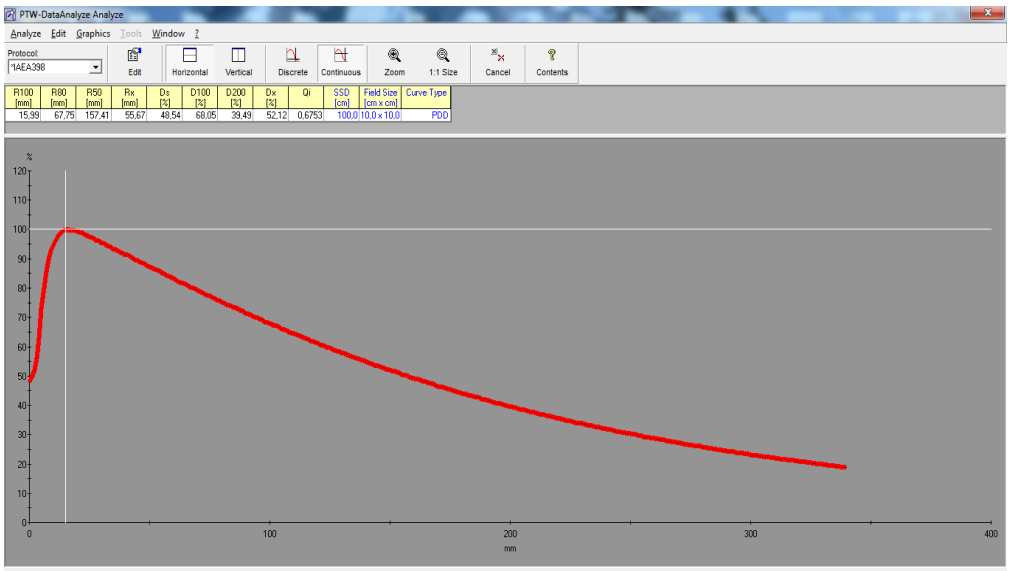

Figure 4: Depth dose percentage for 6 MeV energy for 10 x10 cm^2 field

For 18 MeV energy, the depth to reach the maximum dose (d_{max}) is 3.0 cm for the water phantom and 2.55 cm for PMMA. At a depth of 10 cm, the dose is 0.9967 Gy, which is 78% of the maximum dose. Therefore, the dose at d_{max} is 1.279 Gy. The obtained results are presented in Table 2.

Table 2: Results of diode input dose measurements for energy 18 MeV:

Channels	Serial number	Charge (nC)			Calibration factor (Gy/C)
1	00037	-15.12	-15.16	-15.16	$8.448 * 10^7$
3	00035	-15.37	-15.36	-15.36	$8.327 * 10^7$
4	00036	-9.98	-10.10	-10.25	$1.269 * 10^7$
2	Ionizing chamber	14.55	14.55	14.54	$5.332 * 10^7$

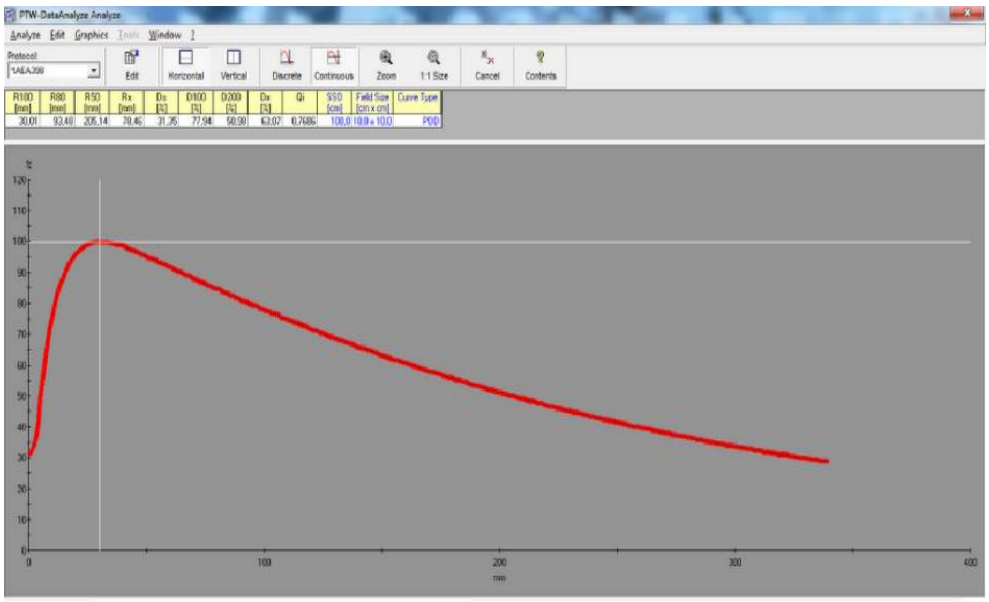

Figure 5: Depth dose percentage for 18 MeV energy for 10 x10 cm^2 field

To determine penetration, the diode is positioned on the surface of the phantom, on the inlet side of the radiation beam, and its response is compared

to a calibrated ionization chamber placed at the depth d_{max} (as shown in figure 1).

If the diode is intended for a specific treatment type, it's recommended to establish reference conditions (e.g., collimator opening, SSD) for that treatment. However, if the diode's application field is broader, such as measuring entrance dose for various patients, it's evident that reference conditions encompass a wide range of geometric parameters. In such cases, controlling the variability of this calibration factor for basic treatment techniques is advisable. For specialized techniques like total body irradiation, it's recommended to perform a separate calibration under conditions similar to those encountered during such applications (AAPM, Report 87, 2005).

Similarly, the F_{exit} gauge factor for an output dose from a diode can be determined by placing it on the output surface of the phantom and comparing its response to that of an ionization chamber positioned at d_{max} within the phantom (as depicted in figure 2).

Continuing with the measurements for the dose on the output surface, the output dose calibration factor is given by:

where the symbols, except for "output", have the same meanings as for the input dose calibration factor.

$$F_{exit} = \frac{D_{exit}}{R_{sc,exit}}$$

where the symbols, except for "output", have the same meaning as for the input dose calibration factor.

Table 3: Results of diode output dose measurements for energy 18 MeV:

Channels	Serial number	Charge (nC)			Calibration factor (Gy/C)
1	00037	-7.545	-7.605	-7.585	$9.736 * 10^7$
3	00035	-7.715	-7.725	-7.685	$9.535 * 10^7$
4	00036	-5.955	-6.000	-6.015	$1.228 * 10^7$
2	Ionizing chamber	14.49	14.48	14.51	$5.332 * 10^7$

Table 4: Results of diode output dose measurements for energy 6 MeV:

Channels	Serial number	Charge (nC)			Calibration factor (Gy/C)
1	00083	-4.115	-4.160	-4.180	$1.452 * 10^7$
3	00082	-4.215	-4.225	-4.235	$1.422 * 10^7$
4	00078	-4.110	-4.135	-4.145	$1.455 * 10^7$
2	Ionizing chamber	12.44	12.45	12.43	$5.332 * 10^7$

When considering the reference conditions for the output dose calibration factor, patient thickness becomes an additional parameter. The phantom thickness should match the thickness encountered in clinical conditions.

Another crucial parameter for diodes used to measure both doses is the ratio of the calibration factors of both doses, denoted as F:

$$F = \frac{F_{exit}}{F_{entry}}$$

Due to differences in positioning between the ionization chamber and diode for determining input and output doses, and because of the lower dose level on the output side of the radiation beam, F is expected to be greater than unity. However, for certain types of diodes surrounded by hemispherical capsules and mounted on thin plates, observed F values have been up to 1.12. These discrepancies are mainly attributed to differences in diode positioning with the ionization chamber or effects of dose level.

It's noted that such diodes must be specifically configured for measuring output doses. Additional measurements have revealed that the drop in response observed upon changing orientation, without altering distance, is the primary cause of high F values. This drop is due to imperfections in diode construction. For some diodes, using the input dose calibration factor to measure output doses is not allowed.

As the accumulated dose over time decreases the response of a diode, the calibration factor increases with dose accumulation in clinical applications. However, studies have shown that the relative increments of input and output calibration factors are equal, resulting in the F-ratio remaining relatively stable over time.

Any parameter affecting diode response may necessitate correction factors when clinical conditions differ from reference conditions. These correction factors, known as stacking factors, are multiplied by calibration factors to obtain the correct dose. If the detector response under clinical conditions is lower or higher than under reference conditions, these factors are greater or less than unity, respectively.

Conclusions

• Basic measurements for determining absorbed dose in water are typically performed at a depth of 10 cm below the water surface. This depth is chosen because electronic contamination from secondary sources is minimized, and the establishment of electronic equilibrium occurs beyond this depth (known as d_{max}). Additionally, 10 cm depth is considered the reference clinical depth.

• Surface dose deposition is primarily dependent on field size. Larger field sizes result in higher surface dose deposition by photon beams. It's advisable to limit field sizes to 20 x 20 cm^2 to avoid creating hot areas. Working with multiple fields helps achieve a more homogeneous dose distribution.

• For photon fields with standard dimensions of 10 x 10 cm^2, the depth of maximum dose (100% dose) is 16 mm for 6 MeV energy and 30 mm for 18 MeV energy.

• Absorbed dose measurement and evaluation are typically performed at the reference depth of 10 cm, with the effective measurement point of the ionization chamber positioned on the central axis of the beam. Correcting values obtained from Percent Depth Dose (PDD) curves, it's determined that at a depth of 10 cm, we have 68% of the dose for 6 MeV energy and 78% for 18 MeV energy.

• Diode dosimeters are essential for accurately measuring dose on the patient's skin surface. Precise dose measurements enable the design of treatment plans with optimal doses for the target organ.

Discussion

Regarding the publication of this article, it is important to emphasize that this is the first installment in a series dedicated to in vivo dosimetry. Future articles will delve deeper into several critical aspects, including the measurement of entrance doses at the patient's surface, quantification of the load charge or dose deposited in the skin (which contributes to skin burns during external beam therapy), and evaluation of the output dose.

Additionally, these articles will address the crucial issue of determining the effective dose deposited within the tumoral structure.

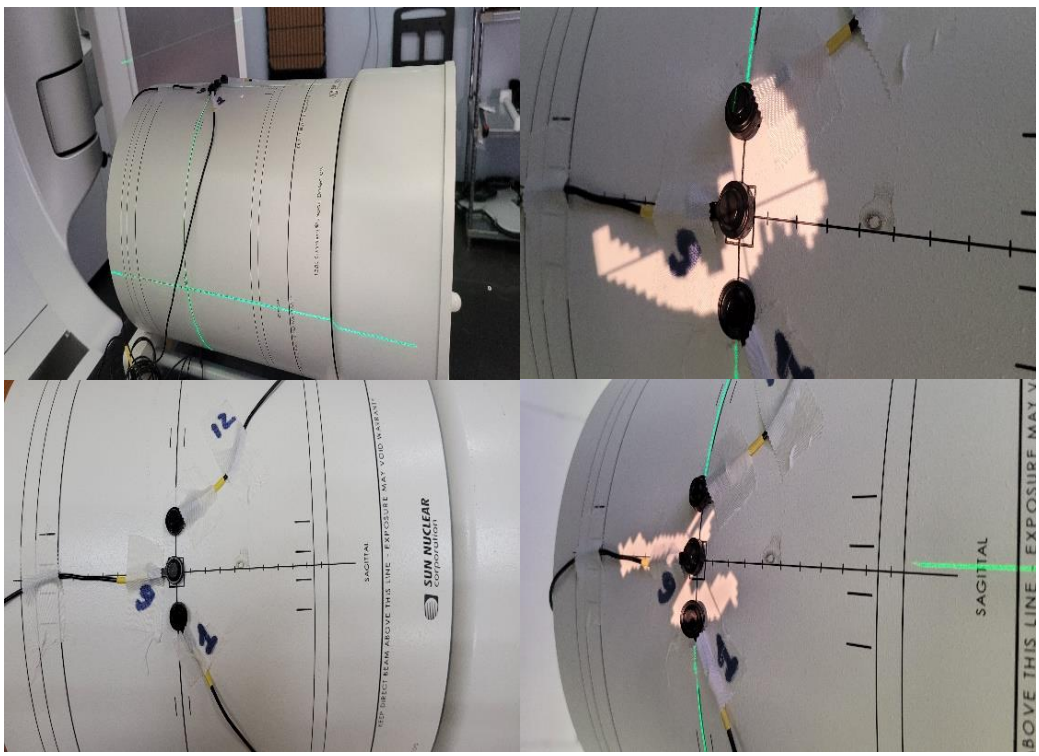

Figure 5: Set-up of in vivo detectors for field dose verification

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