Short Communication

Different Dosimeters/Detectors Used in Small-Field Dosimetry: Pros and Cons

Abstract

With the advent of complex and precise radiation therapy techniques, the use of relatively small fields is needed. Using such field sizes can cause uncertainty in dosimetry; therefore, special attention is required both in dose calculations and measurements. There are several challenges in small-field dosimetry such as the steep gradient of the radiation field, volume averaging effect, lack of charged particle equilibrium, partial occlusion of radiation source, beam alignment, and unable to use a reference dosimeter. Due to these challenges, special dosimeters are needed for small-field dosimetry, and this review article discusses this topic.

Keywords: *Detector, dosimeter, radiotherapy, small field dosimetry*

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Introduction

With the appearance of new techniques such as intensity-modulated radiation therapy (IMRT), volumetric-modulated radiotherapy (VMAT), stereotactic body radiotherapy (SBRT), and stereotactic radiosurgery (SRS), applying relatively small fields that are either dynamic or static is needed. For this purpose, there have been many developments in treatment machines. Small fields are usually defined between 4 cm × 4 cm and 0.3 cm × 0.3 cm.[1,2] Using such field sizes can cause uncertainty in dosimetry; therefore, special attention is required in both dose calculations and measurements. It is notable that dosimetry protocols, such as the International Atomic Energy Agency (IAEA) TRS-398,[3] have provided guidelines for a reference field size (typically 10 cm × 10 cm). However, the majority of reference condition parameters, such as perturbation correction, stopping power ratio, gradient, and fluence corrections, are not applicable to small fields. To overcome nonreference fields used by dedicated machines, the IAEA[4] has provided a framework to manage the issues related to small-field dosimetry.

There are several challenges in small-field dosimetry, including the steep gradient of the radiation field, volume averaging effect, lack of charged particle equilibrium, partial occlusion of radiation source, beam alignment, and unable to use a reference dosimeter, which will be mentioned in the next section. Due to these challenges, special dosimeters are required for small-field dosimetry which is the main subject of this review article.

Challenges in Small-Field Dosimetry

Steep gradient of the radiation field

Modern treatment techniques used in radiotherapy (such as IMRT, VMAT, and SRS) deliver the conformal dose distribution and high-dose radiation to a tumor. The high conformity of the prescribed dose with the planning target volume (PTV) can effectively kill cancerous cells while preserving the surrounding healthy tissue.[5-7]

In clinic, the dose distribution obtains using a treatment planning system (TPS). A TPS for calculation of accurate dose distribution needs to accurate input data, such as percentage depth dose (PDD) curves, profiles, and output factors. In the beam profiles, the distance between the 80% and 20% dose of the central axis defined as the penumbra region. In the penumbra region, measured dose is crucial due to the high-dose gradient. Therefore, it is necessary to use high spatial resolution detectors to obtain the accurate beam profile in the high gradient dose regions such as small fields.[7]

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Volume averaging effect

Volume averaging occurs when the dosimeter dimension is large in compared with the radiation field size. In high gradient dose regions such as small fields, the dose value changes significantly over the dosimeter's active volume. The detector reading is averaged throughout the active volume; however, only a portion of this volume is exposed to radiation.[8] Therefore, the measured beam profiles are artificially flattened.[9] Due to this effect, the dosimeter measures a lower dose than the correct value near the field center, and also this effect overestimates the dose beyond the field edge.[10]

Another important factor in volume averaging is the spreading of the penumbra, which is very important in measuring beam profiles [Figure 1].[11] Since an accurate beam profile is one of the required parameters for TPSs, these inaccuracies in measurements become a concern in commissioning and quality assurance.[11-13] Therefore, using small size detectors with a high resolution is desirable to avoid volume averaging in small photon fields.

Lack of charged particle equilibrium

If the number of charged particles leaving a volume is same with the number entering, charged-particle equilibrium (CPE) happens. In this condition, the absorbed dose is equal to the collision kerma. If the lateral range of electrons is larger than the field size, lateral electronic disequilibrium (LED) can occur.[14] In this condition, the delivered dose to the active volume of the detector is not equal to the dose created by the same electrons from the opposite edge in the lateral direction. Consequently, the anticipation of the deposited dose to the tumor is unreliable.[15]

The lateral electronic equilibrium effect is most notable when there is tissue heterogeneity, such as between lung and bone.[16] Because of the high electron range at lung tissue than that in water, the LED effect in the lung tissue leads to an increase in the size of the penumbra region, increasing the underdosage of the PTV at the edge of the radiation field.[17]

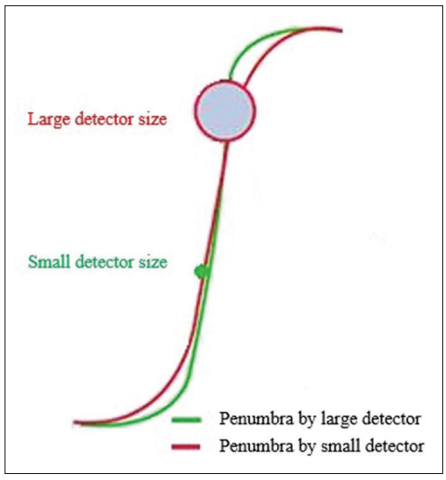

Figure 1: Volume averaging effect of dosimeters/detectors used in small-field dosimetry

In small fields, the lateral range of electrons usually is larger than the field size. Therefore, the lack of lateral CPE is important, especially in the presence of heterogeneity because the coverage of the PTV with the optimized isodose is required. Heterogeneity of the brain is not often investigated in SRS; however, in SBRT, the dose perturbations in and beyond air cavities, lung tissue, and bone must be considered[18] because neglecting the tissue heterogeneity in dose calculation may lead to errors in dose calculation and can reduce tumor control probability.[19]

Partial occlusion of radiation source

Partial occlusion of the radiation source happens because of the collimating output beam of the linear accelerator at a size approximately the same or smaller than the source size, as viewed from the detector. In this condition, only a portion of the source is seen by the dosimeter. Resultantly, the output detected will be smaller compared with that in field sizes where the detector sees the whole source.[20] When partial occlusion of the radiation source occurs, conventional methods to define the field size, such as full width at half maximum (FWHM), are inappropriate because the field size specified by FWHM is larger than the actual field size [Figure 2].[20]

Beam alignment

The correct alignment of the dosimeter is essential for small-field dosimetry, because there is no flat area (the region that includes doses over 80% of the central beam axis) in the center of small fields, in contrast to large fields. Focal spot shift and displacement in the collimator

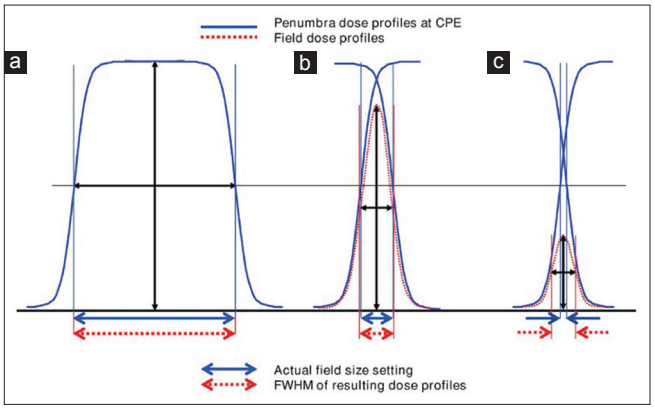

Figure 2: For sufficiently large field sizes, the full width at half maximum of dose profiles is used correctly to determine field sizes because the field borders will be at approximately 50% of the dose level (a). When the field size is of the same order as the charged particle lateral diffusion distance, the penumbra region from opposing field edges overlap, leading to a small error in determining the field size from the full width at half maximum (b), but breaks down entirely for very small fields as the obtained curve has a lower maximum and hence its half value will be pushed outward from the correct position, leading to an overestimated field size (c). Reproduced with permission from Das *et al*., 2008

rotation axis or gantry rotation axis are factors that can cause errors related to misalignments in SRS.[21] Misalignment can lead to errors in dose measurement. Paskalev *et al*. showed that a 0.2-mm error in correct alignment could lead to a 5% shift in a measured dose. To prevent these errors, it is necessary for the dosimeter to be aligned to the center of the field, so it is performed by measurement of beam profiles at several depths.[22] As a result, having a high spatial resolution detector seems to be necessary in small-field dosimetry.

Unable to use a reference dosimeter

In relative dosimetry, such as measurement of PDD and beam profiles, a reference dosimeter is required to correct the variation of the linear accelerator (linac) output. This dosimeter is usually located in the corner of the radiation field. Since there is insufficient space to insert the reference detector in small fields, the perfect solution would be correcting the fluctuations of the linac output. This problem can be solved using a monitor chamber accelerator as a reference dosimeter if this signal is available. Another way is to measure the dose without a reference dosimeter if the linac output is stable. To assess the stability of the linac output, the PDD and profile beam is measured several times. Another solution is to use a dosimeter that is located beyond the radiation field. In this condition, the noise in signals to the reference dosimeter increases because the dose rate outside of the radiation field is very low. Consequently, this effect will lead to an increase in the noise of measurements. As a result, an ionization chamber with a large active volume is required to be used as a reference detector because of their high response and low noise. Wurfel suggested using ionization chambers with active volumes larger than 2.4 cm^3 and emphasized that these detectors be located as close as possible to the beam border beyond the radiation field.[22]

Various Dosimeters Used in Small-Field Dosimetry

According to the abovementioned challenges, selecting a detector with good performance in small fields is difficult. The necessary properties of a desired detector are high spatial resolution, high signal (low noise), low energy dependence, low directional dependence, water equivalence, high stability, and easy to use clinically. Certainly, there is no standard dosimeter for small fields because no detector has all the aforementioned properties. Commonly used dosimeters in small fields are ionization chambers,[23,24] films,[25] thermoluminescent dosimeters (TLDs),[26] polymer gels,[27] metal oxide semiconductor field effect transistors (MOSFETs),[28] diamond detectors,[29] silicon diodes,[30] alanine dosimeters,[31] and Monte Carlo (MC) simulations,[32,33] among others. The advantages and disadvantages of these detectors will be discussed next.

Radiographic and radiochromic film

Film dosimeters are good detectors to measure the dose distribution in two dimensions. These detectors are divided into two categories: radiographic and Gafchromic film. Radiographic films, such as extended dose range (EDR2), have high spatial resolution that is appropriate to spatially measure the penumbra regions on beam profile curves in small fields. Nevertheless, the main problem of radiographic films is their nonconstant response with spectral variation and reproducibility.[34] Furthermore, processing conditions and the densitometer used to read out the dose influence the radiographic film response.[35,36] Perucha *et al*. reported that it is difficult to control the processing phase such that it can limit the use of radiographic films in small fields.[37]

With the development of radiochromic films, several problems related to radiographic films have been solved. The radiochromic films are self-developed and need no chemical processing to obtain an image of the radiation dose distribution.[38] These films are insensitive to ambient light and do not need a darkroom for their processing.[39] In megavoltage beam range, radiochromic films are almost tissue equivalent and reveal little energy dependence. Nevertheless, in kilovoltage beam range, these films represent varying degrees of energy dependence which it also depends on their composition.[38,40] Furthermore, this type of film can be immersed in water. Although some studies have shown that radiochromic films are a suitable detector for small-field dosimetry,[41-44] their nonlinear response in the high dose per fraction and dose rate used in SRS[45] is one of their drawbacks. Furthermore, read out the process of radiochromic films is a disadvantage; as according to the manufacturer's notes, it is necessary to wait up to approximately 48 h after exposure of the film to ensure full-color development.[39]

Diode detector

Diodes are another type of detector used in small-field ionizing radiation dosimetry. The physics and operation of these detectors have been described elsewhere.[46] The energy required to create an electron-hole pair in the silicon diodes is 3.6 eV, a value that is much smaller than the energy required to generate an ion pair in air; so the sensitivity of diodes are higher than ionization chambers. The diodes can be produced at a small size due to their high sensitivity per volume. The diodes have been widely used in small-field dosimetry due to their real-time readout, high spatial resolution, and small size.[47]

Although some studies have recommended using diodes to measure dose distribution in narrow fields, diodes have some disadvantages such as dependence on dose rate, energy, and direction.[48,49] Since the angular distribution of electrons and scattered photons alters with depth and distance from the central axis, the directional dependence is vital in measuring beam profiles and PDDs.[50] Another

disadvantage of these detectors is their energy dependency; some studies have shown an overestimation of low- and medium-energy photons. As a result, shielded diodes were designed to reduce the effect of low-energy photons.[51]

An underestimation can occur when photon scattering is poor because of the high absorptivity of the shield material.[52] In contrast, in cases without lateral electron equilibrium, silicon diodes will provide an overestimation because of the higher density of silicon compared with water.[50] Furthermore, this effect is more notable in shielded diodes due to the high-density shielding material.[53] Therefore, in small fields where the lateral electron equilibrium is degraded and there are few low-energy scatter photons, the use of unshielded diodes is recommended.

Diamond detectors

Diamond detectors are solid state and their sensitive volume is composed of natural diamond.[46] They are water equivalent due to the similarity of carbon's atomic number to tissue. Some studies in small fields have illustrated the suitability of these detectors in measuring dosimetric parameters due to their small size, high-dose response, and directional independence.[29,54-56]

However, the diamond detectors have some disadvantages. Sauer and Wilbert showed that diamond detectors have significant energy dependence. They attributed this effect to the contact material and construction of the detector.[57] Moreover, these detectors demonstrate a significant dose rate dependence and a correction factor should be considered to correct this problem.[58,59]

Metal oxide–silicon semiconductor field-effect transistor

MOSFET detectors are widely used for dosimetry in small fields because of their small active area and direct reading ability compared with some dosimeters, such as TLDs, that require prepreparation and postprocessing.[28,60-62] In the megavoltage range of energy, MOSFETs are energy independent. Furthermore, these dosimeters are dose rate independent.[63]

The major disadvantage of MOSFETs is their angular dependency that can lead to uncertainties in dosimetry.[64,65] A solution for this problem is obtaining quantitative correction factors in the commissioning stage. MOSFET represents a temperature dependency and need to the correction factor if applied at a temperature different from the temperature that are calibrated. This dependency disappears if used of the dual-MOSFET-dual-bias detector.[66]

Thermoluminescent dosimeter

TLDs are small crystals that according to thermoluminescence phenemon can measure ionizing radiation. When the crystal is heated the measured intensity of light emitted from crystal related to absorbed dose.[67] Special types of TLDs can also be applied in the dosimetry of small fields. Due to the advantages, such as high spatial resolution and dose response, TLDs provide a promising opportunity to measure the absorbed dose in a small field.[68] However, there are several drawbacks including cost, time requirement, energy dependence, long waiting periods before reading, and water nonequivalence.[68,69] Furthermore, it has been shown that TLD dosimeters are not appropriate in fields smaller than 10 mm in diameter.[70] The dependence of the TLD response on dosimeter size and beam quality has been previously studied.[71-76]

A special type of TLD is micro-TLD. They can be applied to determine the dose in a region based on their size, i.e., 1 mm × 1 mm × 1 mm. The size of these TLDs is a limitation on their accuracy in locations where the dose can vary rapidly between regions separated by only small distances.[41] Another type of TLD is TLD-100, which has a linear dose response at doses lower than 1 Gy. In addition, they are beam energy dependent; with regard to ^{60}Co, energy correction factors are 1.011 and 1.023 for 6 MV and 25 MV X-rays, respectively.[26]

Recently, optically stimulated luminescence dosimeters have become an acceptable system for dosimetry. In this system, an optical signal proportional to absorbed dose is generated when the irradiated crystal (Al2O3 doped with carbon) is exposed to light.[77] These dosimeters have dosimetric characteristics (such as linearity, dose rate, and beam energy dependence) similar to TLDs. OSL in comparison with TLD has important advantages such as high sensitivity (over the wide range of dose value and dose rate values applied in radiotherapy) and quick readout times.[78]

Gel dosimeter

Gel dosimeters are attractive detectors for the determination of dosimetric parameters in radiotherapy because of their soft-tissue equivalence and radiation direction independence. These dosimeters are considered as both a phantom and detector,[79,80] and they do not disturb the radiation field. Furthermore, they can measure three-dimensional dose distributions.[81] These detectors are divided into three categories: Fricke, polymer, and radiochromic gels.[6] Fricke gels are highly reproducible and easy to prepare than other gel dosimeters, but diffusion in the gel is a disadvantage.[82,83] Polymer gels are high-sensitivity dosimeters without any diffusion issues; however, fabrication of these detectors is difficult due to their sensitivity to the presence of oxygen.[84] Although these dosimeters cannot be introduced as a standard dosimeter due to issues concerning repeatability and their requirement for advanced data processing techniques, some reports suggest that they are suitable for measuring the relative output factor, beam profile, and dose distributions in small fields because of their high spatial resolution and

lack of issues concerning positioning.[10,29,85-87] Radiochromic gel dosimeters are insensitive to oxygen, have desirable diffusion rates, and can readout by optical methods.[6,88,89] These dosimeters are new compared to other gel dosimetry systems and need to perform further research.

Alanine

One of the techniques used for dosimetry in radiotherapy is using an alanine readout with electron paramagnetic resonance (EPR) or electron spin resonance (ESR). These dosimeters have water equivalence, energy independence, nondestructive reading, low fading, and small detector size. In addition, they have a linear dose response.[90,91] In this technique, free radicals generated from the interaction between radiation and media are detected by an amino acid, alanine; the delivery dose is then measured using EPR spectroscopy.[92] The EPR signals have to be calibrated through the ion chamber for absolute dosimetry.

Some researchers have used alanine/K-Band minidosimeters (miniALAs) to measure the dosimetric parameters in small fields and have concluded that miniALAs are suitable in determining the accurate dose.[31,93,94] Recently, alanine has been used to verify advanced radiotherapy techniques such as IMRT and radiosurgery.[95,96]

Plastic scintillation detectors

Plastic scintillation detectors (PSDs) have attractive properties including water equivalency, high spatial resolution, energy and dose rate independence, and linear dose response. Production of Cherenkov light is the main disadvantage in dosimetry with PSD-based systems. This light is generated when an optical fiber is placed in the radiation field. To solve this drawback, the light should be removed from the main signal.[97] Recently, PSDs have been used in modern radiation therapy modalities such as IMRT and SRS.[98] Morin *et al*. showed that PSDs are suitable detectors that can be introduced as reference detectors for beam characterization and quality assurance consideration in radiosurgery.[99]

Ionization chambers

Ionization chambers are used in radiation therapy dosimetry because of their excellent dose response, dose rate independence, low directional dependence, and the wide research base behind them.[100] However, problems in measurement occur when the size of these detectors is bigger than the size of the irradiated field.[101] Therefore, their application in small-field photon dosimetry is limited. The limiting factors in using the ionization chambers are the detector size and lack of lateral electronic equilibrium effects.[102] The ion chambers have an underestimating response at very small fields and this underestimating response is enhanced with the increment of the active volume chamber. Since to measure the beam profiles, especially in the penumbra region requires a high-resolution detector, it seems that ionization chambers are not well suited for small-field measurements.[22]

The pinpoint is a type of ion chamber with a tiny active volume (<0.1 cm^3) that is specifically designed for measuring relative beam profiles in small photon fields. For the measurement of absolute doses, this chamber must be calibrated against a Farmer chamber. It is noteworthy that these detectors do not have stem and polarity effects because of the very small sensitive volume.[101,103,104] The pinpoint underestimates output factors in very small fields because of volume averaging. Pantelis *et al*.[105] observed up to a 10% difference in measuring the output factor for a 5-mm beam.

Monte Carlo Simulation in Small-Field Dosimetry

MC simulation is considered as a strong and trustworthy tool when experimental measurements are not feasible spatially in small fields because beam characterization in these fields by each of the detectors is unreliable, due to volume averaging effects and lack of lateral electronic equilibrium. Using MC simulations in small fields, the dosimetric parameters (e.g., output factor, PDD curves, beam profiles) can be characterized, as well as calculating the dosimeter correction factors in predicting treatment planning requirements. The generally high level of accuracy, flexibility and the fact that the approximations employed in MC methods are far fewer than those implemented in TPS, all make MC methods attractive for use in medical physics. Furthermore, MC calculations are able to calculate doses to the media directly, thus circumventing the need for such complex corrections. This is particularly relevant for small-field dosimetry. In addition, when the complexities of small fields and proximity to inhomogeneous media are both present, as is the case for SBRT and individual beamlets in IMRT, MC methods become increasingly useful. Nevertheless, the main drawback of MC methods is the uncertainty in clinical practice due to the requirement of extensive computing time. In the other word, the greatest limitation of MC calculations is, in general, inefficiency; as this is particularly critical in a clinical environment, where it is not feasible for treatment planning to require hours or days, and commercial TPSs dubbed as MC algorithms consequently employ significant approximations to this end. While MC calculations are in general excellent in predicting measurements, one needs to be careful to understand the code and its implementation. This is particularly important as users of a commercial planning system are often not able to commission a beam model themselves but rely on the manufacturer to perform this task based on data provided by the user. This means that the user must ensure the model is actually applicable to all relevant clinical scenarios. Other issues with the use of MC are the conversion of computed tomography numbers

to materials and the fact that many systems default to a relatively large dose calculation grid which is not appropriate for small-field dose calculations.

Conclusion

The major conclusion extracted from the present study is that there is currently no dosimeter that has been all properties required for dosimetry in the small fields. Therefore, it seems to be logical to use several detectors instead of a single detector to obtain the required data for acceptance, commissioning, data entry into TPSs, and periodic quality assurance, because each detector has limitations related to themselves, for example, volume averaging in ionization chambers, energy dependency in diodes, and angular dependency in MOSFETs. In the clinic, depending on the characteristics required, a suitable dosimeter can be selected. Furthermore, in terms of sensitivity: diode and MOSFET, in terms of resolution: film, in terms of online readout: ionization chambers, MOSFET and diode and as well as from the point of view water equivalency: gel dosimeter can be considered as a good option in the small-field dosimetry.

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Conflicts of interest

There are no conflicts of interest.

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