

## 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

### 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.<sup>[1,2]</sup> 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,<sup>[3]</sup> 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<sup>[4]</sup> 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.<sup>[5-7]</sup>

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.<sup>[7]</sup>

This is an open access journal, and articles are distributed under the terms of the Creative Commons Attribution-NonCommercial-ShareAlike 4.0 License, which allows others to remix, tweak, and build upon the work non-commercially, as long as appropriate credit is given and the new creations are licensed under the identical terms.

For reprints contact: reprints@medknow.com

**How to cite this article:** Parwaie W, Refahi S, Ardekani MA, Farhood B. Different dosimeters/detectors used in small-field dosimetry: Pros and cons. *J Med Sign Sens* 2018;8:195-203.

**Wrya Parwaie<sup>1</sup>,  
Soheila Refahi<sup>2</sup>,  
Mahdiah Afkhami  
Ardekani<sup>3</sup>,  
Bagher Farhood<sup>4</sup>**

<sup>1</sup>Department of Medical Physics and Biomedical Engineering, School of Medicine, Tehran University of Medical Sciences, Tehran, <sup>2</sup>Department of Medical Physics, Faculty of Medicine, Ardabil University of Medical Sciences, Ardabil, <sup>3</sup>Department of Radiology, Faculty of Para-Medicine, Hormozgan University of Medical Sciences, Bandar Abbas, <sup>4</sup>Department of Radiology and Medical Physics, Faculty of Paramedical Sciences, Kashan University of Medical Sciences, Kashan, Iran

W. Parwaie

ORCID ID

<https://orcid.org/0000-0003-3839-7050>

B. Farhood

ORCID ID

<https://orcid.org/0000-0003-2290-7220>

#### Address for correspondence:

Dr. Bagher Farhood,  
Department of Medical Physics  
and Radiology, Faculty  
of Paramedical Sciences,  
Kashan University of Medical  
Sciences, Kashan, Iran.  
E-mail: bffarhood@gmail.com

Website: www.jmss.mui.ac.ir

DOI: 10.4103/jmss.JMSS\_3\_18

## 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.<sup>[8]</sup> Therefore, the measured beam profiles are artificially flattened.<sup>[9]</sup> 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.<sup>[10]</sup>

Another important factor in volume averaging is the spreading of the penumbra, which is very important in measuring beam profiles [Figure 1].<sup>[11]</sup> 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.<sup>[11-13]</sup> 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.<sup>[14]</sup> 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.<sup>[15]</sup>

The lateral electronic equilibrium effect is most notable when there is tissue heterogeneity, such as between lung and bone.<sup>[16]</sup> 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,

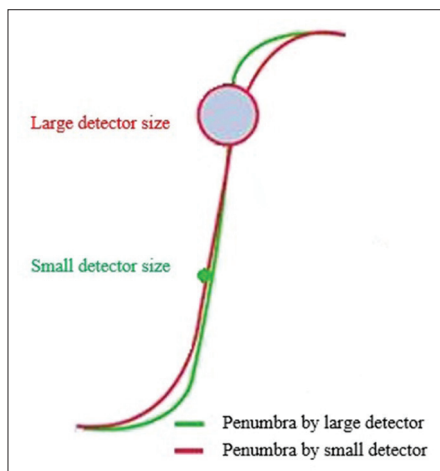


Figure 1: Volume averaging effect of dosimeters/detectors used in small-field dosimetry

increasing the underdosage of the PTV at the edge of the radiation field.<sup>[17]</sup>

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<sup>[18]</sup> because neglecting the tissue heterogeneity in dose calculation may lead to errors in dose calculation and can reduce tumor control probability.<sup>[19]</sup>

## 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.<sup>[20]</sup> 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].<sup>[20]</sup>

## 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

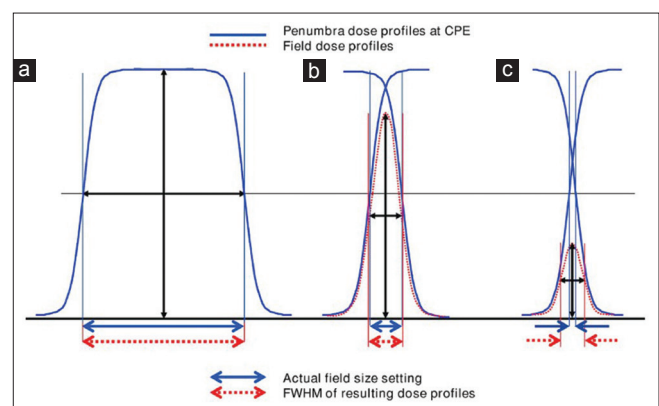


Figure 2: For sufficiently large field sizes, the full width at half maximum of dose profiles is used correctly to determine field sizes because the field borders will be at approximately 50% of the dose level (a). When the field size is of the same order as the charged particle lateral diffusion distance, 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.<sup>[21]</sup> 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.<sup>[22]</sup> 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<sup>3</sup> and emphasized that these detectors be located as close as possible to the beam border beyond the radiation field.<sup>[22]</sup>

### 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,<sup>[23,24]</sup> films,<sup>[25]</sup> thermoluminescent dosimeters (TLDs),<sup>[26]</sup> polymer gels,<sup>[27]</sup> metal oxide semiconductor field effect transistors (MOSFETs),<sup>[28]</sup> diamond detectors,<sup>[29]</sup> silicon diodes,<sup>[30]</sup> alanine dosimeters,<sup>[31]</sup> and Monte Carlo (MC) simulations,<sup>[32,33]</sup> 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.<sup>[34]</sup> Furthermore, processing conditions and the densitometer used to read out the dose influence the radiographic film response.<sup>[35,36]</sup> 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.<sup>[37]</sup>

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.<sup>[38]</sup> These films are insensitive to ambient light and do not need a darkroom for their processing.<sup>[39]</sup> 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.<sup>[38,40]</sup> 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,<sup>[41-44]</sup> their nonlinear response in the high dose per fraction and dose rate used in SRS<sup>[45]</sup> 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.<sup>[39]</sup>

### 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.<sup>[46]</sup> 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.<sup>[47]</sup>

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.<sup>[48,49]</sup> 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.<sup>[50]</sup> 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.<sup>[51]</sup>

An underestimation can occur when photon scattering is poor because of the high absorptivity of the shield material.<sup>[52]</sup> In contrast, in cases without lateral electron equilibrium, silicon diodes will provide an overestimation because of the higher density of silicon compared with water.<sup>[50]</sup> Furthermore, this effect is more notable in shielded diodes due to the high-density shielding material.<sup>[53]</sup> 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.<sup>[46]</sup> 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.<sup>[29,54-56]</sup>

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.<sup>[57]</sup> Moreover, these detectors demonstrate a significant dose rate dependence and a correction factor should be considered to correct this problem.<sup>[58,59]</sup>

### 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 preparation and postprocessing.<sup>[28,60-62]</sup> In the megavoltage range of energy, MOSFETs are energy independent. Furthermore, these dosimeters are dose rate independent.<sup>[63]</sup>

The major disadvantage of MOSFETs is their angular dependency that can lead to uncertainties in dosimetry.<sup>[64,65]</sup> 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.<sup>[66]</sup>

### Thermoluminescent dosimeter

TLDs are small crystals that according to thermoluminescence phenomenon can measure ionizing radiation. When the crystal is heated the measured intensity

of light emitted from crystal related to absorbed dose.<sup>[67]</sup> 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.<sup>[68]</sup> However, there are several drawbacks including cost, time requirement, energy dependence, long waiting periods before reading, and water nonequivalence.<sup>[68,69]</sup> Furthermore, it has been shown that TLD dosimeters are not appropriate in fields smaller than 10 mm in diameter.<sup>[70]</sup> The dependence of the TLD response on dosimeter size and beam quality has been previously studied.<sup>[71-76]</sup>

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.<sup>[41]</sup> 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 <sup>60</sup>Co, energy correction factors are 1.011 and 1.023 for 6 MV and 25 MV X-rays, respectively.<sup>[26]</sup>

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 (Al<sub>2</sub>O<sub>3</sub> doped with carbon) is exposed to light.<sup>[77]</sup> 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.<sup>[78]</sup>

### 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,<sup>[79,80]</sup> and they do not disturb the radiation field. Furthermore, they can measure three-dimensional dose distributions.<sup>[81]</sup> These detectors are divided into three categories: Fricke, polymer, and radiochromic gels.<sup>[6]</sup> Fricke gels are highly reproducible and easy to prepare than other gel dosimeters, but diffusion in the gel is a disadvantage.<sup>[82,83]</sup> 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.<sup>[84]</sup> 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.<sup>[10,29,85-87]</sup> Radiochromic gel dosimeters are insensitive to oxygen, have desirable diffusion rates, and can readout by optical methods.<sup>[6,88,89]</sup> 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.<sup>[90,91]</sup> 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.<sup>[92]</sup> 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.<sup>[31,93,94]</sup> Recently, alanine has been used to verify advanced radiotherapy techniques such as IMRT and radiosurgery.<sup>[95,96]</sup>

### 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.<sup>[97]</sup> Recently, PSDs have been used in modern radiation therapy modalities such as IMRT and SRS.<sup>[98]</sup> 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.<sup>[99]</sup>

### 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.<sup>[100]</sup> However, problems in measurement occur when the size of these detectors is bigger than the size of the irradiated field.<sup>[101]</sup> 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.<sup>[102]</sup> 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.<sup>[22]</sup>

The pinpoint is a type of ion chamber with a tiny active volume ( $<0.1 \text{ cm}^3$ ) 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.<sup>[101,103,104]</sup> The pinpoint underestimates output factors in very small fields because of volume averaging. Pantelis *et al.*<sup>[105]</sup> 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.

## Financial support and sponsorship

None.

## Conflicts of interest

There are no conflicts of interest.

## References

- Alagar AG, Mani GK, Karunakaran K. Percentage depth dose calculation accuracy of model based algorithms in high energy photon small fields through heterogeneous media and comparison with plastic scintillator dosimetry. *J Appl Clin Med Phys* 2016;17:132-42.
- Das IJ, Francescon P, Ahnesjö A, Aspradakis MM, Cheng CW, Ding GX, *et al.* Small fields and non-equilibrium condition photon beam dosimetry: AAPM Task Group Report 155. *Med Phys* 2014 (in review).
- Andreo P, Burns DT, Hohlfield K, Huq MS, Kanai T, Laitano F, *et al.* Absorbed Dose Determination in External Beam Radiotherapy: An International Code of Practice for Dosimetry Based on Standards of Absorbed Dose to Water. IAEA TRS 398; 2000.
- Alfonso R, Andreo P, Capote R, Huq MS, Kilby W, Kjäll P, *et al.* A new formalism for reference dosimetry of small and nonstandard fields. *Med Phys* 2008;35:5179-86.
- Farhood B, Khezerloo D, Zadeh TM, Nedaie HA, Hamrahi D, Khezerloo N, *et al.* Evaluation of the effect of temperature variation on response of PRESAGE® dosimeter. *J Cancer Res Ther* 2017;13:118-21.
- Khezerloo D, Nedaie HA, Takavar A, Zirak A, Farhood B, Movahedinejad H, *et al.* PRESAGE® as a solid 3-D radiation dosimeter: A review article. *Radiat Phys Chem* 2017;141:88-97.
- Arnfield MR, Otto K, Aroumougama VR, Alkins RD. The use of film dosimetry of the penumbra region to improve the accuracy of intensity modulated radiotherapy. *Med Phys* 2005;32:12-8.
- García-Vicente F, Delgado JM, Peraza C. Experimental determination of the convolution kernel for the study of the spatial response of a detector. *Med Phys* 1998;25:202-7.
- Duggan DM, Coffey CW 2<sup>nd</sup>. Small photon field dosimetry for stereotactic radiosurgery. *Med Dosim* 1998;23:153-9.
- Calcina CS, de Oliveira LN, de Almeida CE, de Almeida A. Dosimetric parameters for small field sizes using fricke xylenol gel, thermoluminescent and film dosimeters, and an ionization chamber. *Phys Med Biol* 2007;52:1431-9.
- Laub WU, Wong T. The volume effect of detectors in the dosimetry of small fields used in IMRT. *Med Phys* 2003;30:341-7.
- Bedford JL, Childs PJ, Nordmark Hansen V, Mosleh-Shirazi MA, Verhaegen F, Warrington AP, *et al.* Commissioning and quality assurance of the pinnacle(3) radiotherapy treatment planning system for external beam photons. *Br J Radiol* 2003;76:163-76.
- Al-Najjar WH, Guru Prasad S, Parthasaradhi K, Bloomer WD, Nanda RK. Dosimetric aspects of small circular fields of 10 MV photon beam. *Med Dosim* 1998;23:39-42.
- Andreo P, Benmakhlouf H. Improved reference and relative dosimetry of small radiation therapy photon beams. Stockholm: Strålsäkerhetsmyndigheten; 2014.
- Mesbahi A. The effect of electronic disequilibrium on the received dose by lung in small fields with photon beams: Measurements and Monte Carlo study. *Iran J Radiat Res* 2008;6:71-7.
- Fu W, Dai J, Hu Y. The influence of lateral electronic disequilibrium on the radiation treatment planning for lung cancer irradiation. *Biomed Mater Eng* 2004;14:123-6.
- Tsiakalos MF, Theodorou K, Kappas C, Zefkili S, Rosenwold JC. Analysis of the penumbra enlargement in lung versus the quality index of photon beams: A methodology to check the dose calculation algorithm. *Med Phys* 2004;31:943-9.
- Stathakis S, Esquivel C, Quino LV, Myers P, Calvo O, Mavroidis P, *et al.* Accuracy of the small field dosimetry using the Acuros XB dose calculation algorithm within and beyond heterogeneous media for 6 MV photon beams. *Int J Med Phys Clin Eng Radiat Oncol* 2012;1:78-87.
- Vergote K, De Deene Y, Claus F, De Gerssem W, Van Duyse B, Paelinck L, *et al.* Application of monomer/polymer gel dosimetry to study the effects of tissue inhomogeneities on intensity-modulated radiation therapy (IMRT) dose distributions. *Radiother Oncol* 2003;67:119-28.
- Das IJ, Ding GX, Ahnesjö A. Small fields: Nonequilibrium radiation dosimetry. *Med Phys* 2008;35:206-15.
- Khan FM, Gibbons JP. *Gibbons, Khan's the Physics of Radiation Therapy*. Philadelphia, PA: Lippincott Williams & Wilkins; 2014.
- Würfel JU. Dose measurements in small fields. *Med Phys Int J* 2013;1:81-90.
- Stasi M, Baiotto B, Barboni G, Scielzo G. The behavior of several microionization chambers in small intensity modulated radiotherapy fields. *Med Phys* 2004;31:2792-5.
- Prezado Y, Martinez-Rovira I, Thengumpallil S, Deman P. Dosimetry protocol for the preclinical trials in white-beam minibeam radiation therapy. *Med Phys* 2011;38:5012-20.
- Wilcox EE, Daskalov GM. Evaluation of GAFCHROMIC EBT film for cyberknife dosimetry. *Med Phys* 2007;34:1967-74.
- Peña-Jiménez S, Lárraga-Gutiérrez JM, García-Garduño OA, Gamboa-deBuen I. Characterization of TLD-100 micro-cubes for use in small field dosimetry. *AIP Conf Proc* 2014;1626:168-70.
- Parwaie W, Yarahmadi M, Nedaie HA, Zahmatkesh MH, Barati AH, Afkhami M. Evaluation of MRI-based MAGIC polymer gel dosimeter in small photon fields. *Int J Radiat Res* 2016;14:57-63.
- Kurjewicz L, Berndt A. Measurement of Gamma Knife® helmet



- factors using MOSFETs. *Med Phys* 2007;34:1007-12.
29. Pappas E, Maris TG, Zacharopoulou F, Papadakis A, Manolopoulos S, Green S, *et al.* Small SRS photon field profile dosimetry performed using a PinPoint air ion chamber, a diamond detector, a novel silicon-diode array (DOSI), and polymer gel dosimetry. Analysis and intercomparison. *Med Phys* 2008;35:4640-8.
  30. Reggiori G, Mancosu P, Suchowerska N, Lobefalo F, Stravato A, Tomatis S, *et al.* Characterization of a new unshielded diode for small field dosimetry under flattening filter free beams. *Phys Med* 2016;32:408-13.
  31. Ramírez JV, Chen F, Nicolucci P, Veliz DA, Baffa O. Tissue Interfaces Dosimetry in Small Field Radiotherapy with Alanine/EPR Minidosimeters and Monte Carlo-PENELOPE Simulation, ISSD; 2014.
  32. Yarahmadi M, Allahverdi M, Nedaie HA, Asnaashari K, Vaezzadeh SA, Sauer OA, *et al.* Improvement of the penumbra for small radiosurgical fields using flattening filter free low megavoltage beams. *Z Med Phys* 2013;23:291-9.
  33. Benmakhlouf H, Johansson J, Paddick I, Andreo P. Monte carlo calculated and experimentally determined output correction factors for small field detectors in leksell gamma knife perfexion beams. *Phys Med Biol* 2015;60:3959-73.
  34. Zhu XR, Jursinic PA, Grimm DF, Lopez F, Rownd JJ, Gillin MT, *et al.* Evaluation of kodak EDR2 film for dose verification of intensity modulated radiation therapy delivered by a static multileaf collimator. *Med Phys* 2002;29:1687-92.
  35. Haus AG, Rothenberg LN. Advances in film processing systems technology and quality control in medical imaging. *Med Phys* 2001;28:1813.
  36. Pai S, Das IJ, Dempsey JF, Lam KL, Losasso TJ, Olch AJ, *et al.* TG-69: Radiographic film for megavoltage beam dosimetry. *Med Phys* 2007;34:2228-58.
  37. Perucha M, Sánchez-Doblado F, Leal A, Rincón M, Arráns R, Núñez L, *et al.* Investigation of radiosurgical beam profiles using monte carlo method. *Med Dosim* 2003;28:1-6.
  38. Devic S. Radiochromic film dosimetry: Past, present, and future. *Phys Med* 2011;27:122-34.
  39. Niroomand-Rad A, Blackwell CR, Coursey BM, Gall KP, Galvin JM, McLaughlin WL, *et al.* Radiochromic film dosimetry: Recommendations of AAPM Radiation Therapy Committee Task Group 55. American Association of Physicists in Medicine. *Med Phys* 1998;25:2093-115.
  40. Bekerat H, Devic S, DeBlois F, Singh K, Sarfehnia A, Seuntjens J, *et al.* Improving the energy response of external beam therapy (EBT) GafChromic™ dosimetry films at low energies ( $\leq 100$  keV). *Med Phys* 2014;41:022101.
  41. Wong CJ, Ackerly T, He C, Patterson W, Powell CE, Qiao G, *et al.* Small-field size dose-profile measurements using gel dosimeters, gafchromic films and micro-thermoluminescent dosimeters. *Radiat Meas* 2009;44:249-56.
  42. García-Garduño OA, Lárraga-Gutiérrez JM, Rodríguez-Villafuerte M, Martínez-Dávalos A, Celis MA. Small photon beam measurements using radiochromic film and monte carlo simulations in a water phantom. *Radiother Oncol* 2010;96:250-3.
  43. Palmer AL, Dimitriadis A, Nisbet A, Clark CH. Evaluation of gafchromic EBT-XD film, with comparison to EBT3 film, and application in high dose radiotherapy verification. *Phys Med Biol* 2015;60:8741-52.
  44. Yarahmadi M, Nedaie HA, Allahverdi M, Asnaashari KH, Sauer OA. Small photon field dosimetry using EBT2 Gafchromic film and Monte Carlo simulation. *Int J Radiat Res* 2013;11:215-24.
  45. Ramani R, Lightstone AW, Mason DL, O'Brien PF. The use of radiochromic film in treatment verification of dynamic stereotactic radiosurgery. *Med Phys* 1994;21:389-92.
  46. Rosenfeld AB. Electronic dosimetry in radiation therapy. *Radiat Meas* 2006;41:S134-53.
  47. Griessbach I, Lapp M, Bohsung J, Gademann G, Harder D. Dosimetric characteristics of a new unshielded silicon diode and its application in clinical photon and electron beams. *Med Phys* 2005;32:3750-4.
  48. Jursinic PA. Angular dependence of dose sensitivity of surface diodes. *Med Phys* 2009;36:2165-71.
  49. Eklund K, Ahnesjö A. Modeling silicon diode dose response factors for small photon fields. *Phys Med Biol* 2010;55:7411-23.
  50. Westermark M, Arndt J, Nilsson B, Brahme A. Comparative dosimetry in narrow high-energy photon beams. *Phys Med Biol* 2000;45:685-702.
  51. Mancosu P, Reggiori G, Stravato A, Gaudino A, Lobefalo F, Palumbo V, *et al.* Evaluation of a synthetic single-crystal diamond detector for relative dosimetry on the leksell gamma knife perfexion radiosurgery system. *Med Phys* 2015;42:5035-41.
  52. McKerracher C, Thwaites DI. Assessment of new small-field detectors against standard-field detectors for practical stereotactic beam data acquisition. *Phys Med Biol* 1999;44:2143-60.
  53. Lee HR, Pankuch M, Chu JC, Spokas JJ. Evaluation and characterization of parallel plate microchamber's functionalities in small beam dosimetry. *Med Phys* 2002;29:2489-96.
  54. Heydarian M, Hoban PW, Beddoe AH. A comparison of dosimetry techniques in stereotactic radiosurgery. *Phys Med Biol* 1996;41:93-110.
  55. Das IJ, Downes MB, Kassae A, Tochner Z. Choice of radiation detector in dosimetry of stereotactic radiosurgery-radiotherapy. *J Radiosurg* 2000;3:177-86.
  56. Das IJ, Morales J, Francescon P. Small field dosimetry: What have we learnt? In Medical Physics: Fourteenth Mexican Symposium on Medical Physics. AIP Conf Proc 2016;1747:060001.
  57. Sauer OA, Wilbert J. Measurement of output factors for small photon beams. *Med Phys* 2007;34:1983-8.
  58. Laub WU, Kaulich TW, Nüsslin F. A diamond detector in the dosimetry of high-energy electron and photon beams. *Phys Med Biol* 1999;44:2183-92.
  59. Rustgi SN, Frye DM. Dosimetric characterization of radiosurgical beams with a diamond detector. *Med Phys* 1995;22:2117-21.
  60. Chuang CF, Verhey LJ, Xia P. Investigation of the use of MOSFET for clinical IMRT dosimetric verification. *Med Phys* 2002;29:1109-15.
  61. Francescon P, Cora S, Cavedon C, Scalchi P, Reccanello S, Colombo F, *et al.* Use of a new type of radiochromic film, a new parallel-plate micro-chamber, MOSFETs, and TLD 800 microcubes in the dosimetry of small beams. *Med Phys* 1998;25:503-11.
  62. Francescon P, Cora S, Cavedon C, Scalchi P. Application of a monte carlo-based method for total scatter factors of small beams to new solid-state micro-detectors. *J Appl Clin Med Phys* 2009;10:2939.
  63. Reft CS. The energy dependence and dose response of a commercial optically stimulated luminescent detector for kilovoltage photon, megavoltage photon, and electron, proton, and carbon beams. *Med Phys* 2009;36:1690-9.
  64. Gopiraj A, Billimagga RS, Ramasubramanian V. Performance characteristics and commissioning of MOSFET as an *in-vivo* dosimeter for high energy photon external beam radiation therapy. *Rep Pract Oncol Radiother* 2008;13:114-25.
  65. Kumar AS, Sharma SD, Ravindran BP. Characteristics of mobile

- MOSFET dosimetry system for megavoltage photon beams. *J Med Phys* 2014;39:142-9.
66. Soubra M, Cygler J, Mackay G. Evaluation of a dual bias dual metal oxide-silicon semiconductor field effect transistor detector as radiation dosimeter. *Med Phys* 1994;21:567-72.
  67. Podgorsak EB. *Radiation Oncology Physics*. Vienna: International Atomic Energy Agency; 2005.
  68. Zhang B, Zhu J, Li Y, Chen S, Chen L, Liu X, *et al.* Feasibility of lateral dose profile measurements in a small field using TLDs. *Phys Med Biol* 2015;60:N47-57.
  69. Pham C. Characterization of OSLDs for use in small field photon beam dosimetry. Master Thesis the University of Texas. 2013.
  70. Massillon-Jl G, Cueva-Prócel D, Díaz-Aguirre P, Rodríguez-Ponce M, Herrera-Martínez F. Dosimetry for small fields in stereotactic radiosurgery using gafchromic MD-V2-55 film, TLD-100 and alanine dosimeters. *PLoS One* 2013;8:e63418.
  71. Mobit P, Agyingi E, Sandison G. Comparison of the energy-response factor of LiF and Al<sub>2</sub>O<sub>3</sub> in radiotherapy beams. *Radiat Prot Dosimetry* 2006;119:497-9.
  72. Mobit PN, Mayles P, Nahum AE. The quality dependence of LiF TLD in megavoltage photon beams: Monte carlo simulation and experiments. *Phys Med Biol* 1996;41:387-98.
  73. El-Faramawy NA, Göksu HY, Panzer W. Thermoluminescence dosimetric properties of a new thin beta detector (LiF: Mg, Cu, P; GR-200F) in comparison with highly sensitive Al<sub>2</sub>O<sub>3</sub>:C beta dosimeters. *J Radiol Prot* 2004;24:273-82.
  74. Naqvi SA, D'Souza WD, Earl MA, Ye SJ, Shih R, Li XA, *et al.* Using a photon phase-space source for convolution/superposition dose calculations in radiation therapy. *Phys Med Biol* 2005;50:4111-24.
  75. Chen SW, Wang XT, Chen LX, Tang Q, Liu XW. Monte carlo evaluations of the absorbed dose and quality dependence of Al<sub>2</sub>O<sub>3</sub> in radiotherapy photon beams. *Med Phys* 2009;36:4421-4.
  76. Wang X, Zhu J, Chen S, Tang Q, Liu X. Monte-Carlo simulations of Al<sub>2</sub>O<sub>3</sub> dosimetry in uniform MV photon beams: Influence of field and detector size. *Radiat Meas* 2012;47:501-3.
  77. Pradhan AS, Lee JI, Kim JL. Recent developments of optically stimulated luminescence materials and techniques for radiation dosimetry and clinical applications. *J Med Phys* 2008;33:85-99.
  78. Bhatt BC. Thermoluminescence, optically stimulated luminescence and radiophotoluminescence dosimetry: An overall perspective. *Radiat Prot Environ* 2011;34:6-16.
  79. Venning AJ, Nitschke KN, Keall PJ, Baldock C. Radiological properties of normoxic polymer gel dosimeters. *Med Phys* 2005;32:1047-53.
  80. Fong PM, Keil DC, Does MD, Gore JC. Polymer gels for magnetic resonance imaging of radiation dose distributions at normal room atmosphere. *Phys Med Biol* 2001;46:3105-13.
  81. Farhood B, Abtahi SM, Geraily G, Ghorbani M, Mahdavi SR, Zahmatkesh MH. Dosimetric characteristics of PASSAG as a new polymer gel dosimeter with negligible toxicity. *Radiat Phys Chem* 2018;147:91-100.
  82. Pedersen TV, Olsen DR, Skretting A. Measurement of the ferric diffusion coefficient in agarose and gelatine gels by utilization of the evolution of a radiation induced edge as reflected in relaxation rate images. *Phys Med Biol* 1997;42:1575-85.
  83. Baldock C, Harris PJ, Piercy AR, Healy B. Experimental determination of the diffusion coefficient in two-dimensions in ferrous sulphate gels using the finite element method. *Australas Phys Eng Sci Med* 2001;24:19-30.
  84. Guo PY, Adamovics JA, Oldham M. Characterization of a new radiochromic three-dimensional dosimeter. *Med Phys* 2006;33:1338-45.
  85. Young Nam K, Byung Ock C, Hong Seok J, Chul Seung K, Seok Hyun S, Hun-Joo S, *et al.* Development of BANG-3 (R) polymer gel dosimetry system in small radiosurgical fields. *J Korean Phys Soc* 2011;59:3422.
  86. Kawamura H, Shinoda K, Fuse H, Terunuma T, Miyamoto K, Sakae T, *et al.* Comparison between polymer gel dosimetry and calculated dose with small field in stereotactic irradiation. *J Phys Conf Ser* 2013;444:012031.
  87. Shih TY, Wu J, Shih CT, Lee YT, Wu SH, Yao CH, *et al.* Small-field measurements of 3D polymer gel dosimeters through optical computed tomography. *PLoS One* 2016;11:e0151300.
  88. Khezerloo D, Nedaie HA, Farhood B, Zirak A, Takavar A, Banaee N, *et al.* Optical computed tomography in PRESAGE® three-dimensional dosimetry: Challenges and prospective. *J Cancer Res Ther* 2017;13:419-24.
  89. Khezerloo D, Nedaie HA, Takavar A, Zirak A, Farhood B, Banaee N, *et al.* Dosimetric properties of new formulation of PRESAGE® with tin organometal catalyst: Development of sensitivity and stability to megavoltage energy. *J Cancer Res Ther* 2018;14:308-13.
  90. Ramirez JV, Marques T, Chen F, Nicolucci P, Baffa O. Percentage Depth Dose Curves Comparison between L-Alanine Minidosimeters, Radiographic Film and PENELOPE Monte Carlo Simulation for a 60Co Beam, *Proceed. IRPA*, 12; 2012.
  91. Anton M, Kapsch RP, Krauss A, von Voigts-Rhettz P, Zink K, McEwen M, *et al.* Difference in the relative response of the alanine dosimeter to megavoltage x-ray and electron beams. *Phys Med Biol* 2013;58:3259-82.
  92. Helt-Hansen J, Rosendal F, Kofoed IM, Andersen CE. Medical reference dosimetry using EPR measurements of alanine: Development of an improved method for clinical dose levels. *Acta Oncol* 2009;48:216-22.
  93. Ramirez JV, Chen F, Nicolucci P, Baffa O. Dosimetry of small radiation field in inhomogeneous medium using alanine/EPR minidosimeters and PENELOPE Monte Carlo simulation. *Radiat Meas* 2011;46:941-4.
  94. Abrego FC, Calcina CS, de Almeida A, de Almeida CE, Baffa O. Relative output factor and beam profile measurements of small radiation fields with an L-alanine/K-band EPR minidosimeter. *Med Phys* 2007;34:1573-82.
  95. Budgell G, Berresford J, Trainer M, Bradshaw E, Sharpe P, Williams P, *et al.* A national dosimetric audit of IMRT. *Radiation Oncol* 2011;99:246-52.
  96. Garcia T, Lacornerie T, Popoff R, Lourenço V, Bordy JM. Dose verification and calibration of the Cyberknife® by EPR/alanine dosimetry. *Radiat Meas* 2011;46:952-7.
  97. Archambault L, Beddar AS, Gingras L, Roy R, Beaulieu L. Measurement accuracy and cerenkov removal for high performance, high spatial resolution scintillation dosimetry. *Med Phys* 2006;33:128-35.
  98. Klein DM, Tailor RC, Archambault L, Wang L, Therriault-Proulx F, Beddar AS, *et al.* Measuring output factors of small fields formed by collimator jaws and multileaf collimator using plastic scintillation detectors. *Med Phys* 2010;37:5541-9.
  99. Morin J, Beliveau-Nadeau D, Chung E, Seuntjens J, Therriault D, Archambault L, *et al.* A comparative study of small field total scatter factors and dose profiles using plastic scintillation detectors and other stereotactic dosimeters: The case of the CyberKnife. *Med Phys* 2013;40:011719.
  100. Low DA, Moran JM, Dempsey JF, Dong L, Oldham M. Dosimetry tools and techniques for IMRT. *Med Phys* 2011;38:1313-38.



101. Low DA, Parikh P, Dempsey JF, Wahab S, Huq S. Ionization chamber volume averaging effects in dynamic intensity modulated radiation therapy beams. *Med Phys* 2003;30:1706-11.
102. Pappas E, Maris TG, Papadakis A, Zacharopoulou F, Damilakis J, Papanikolaou N, *et al.* Experimental determination of the effect of detector size on profile measurements in narrow photon beams. *Med Phys* 2006;33:3700-10.
103. Shimono T, Koshida K, Nambu H, Matsubara K, Takahashi H, Okuda H, *et al.* Polarity effect in commercial ionization chambers used in photon beams with small fields. *Radiol Phys Technol* 2009;2:97-103.
104. Agostinelli S, Garelli S, Piergentili M, Foppiano F. Response to high-energy photons of PTW31014 PinPoint ion chamber with a central aluminum electrode. *Med Phys* 2008;35:3293-301.
105. Pantelis E, Antypas C, Petrokokkinos L, Karaiskos P, Papagiannis P, Kozicki M, *et al.* Dosimetric characterization of CyberKnife radiosurgical photon beams using polymer gels. *Med Phys* 2008;35:2312-20.

---

## BIOGRAPHIES



**Wrya Parwaie** was born in Sanandaj, Iran, in 1981. He had educated in Applied Physics at Kurdistan University in 2005. He received the Master degree in Medical physics at Tehran University of Medical Sciences in 2013. He is now Ph.D Candidate of Medical Physics at Tehran University of Medical Sciences. His research interests include Radiotherapy, Dosimetry and Monte Carlo Simulation.

**Email:** werya.parwaie@gmail.com



**Soheila Refahi** received the M.S degree in Medical Physics at Tabriz University of Medical Sciences. She has finished his PhD in Medical Physics at Tehran University of Medical Sciences in 2014. She is currently an Associate Professor at Ardabil University of Medical Sciences in Medical Physics Department. Involved research topics include Radiobiology and Medical images.

**Email:** soheila52@yahoo.com



**Mahdieh Afkhami Ardekani** was born in Bandar Abbas, Iran. She had educated in Medical Physics at Tehran University of Medical Sciences. She is currently a faculty member at Hormozgan University of Medical Sciences in Radiology Department. Involved research topics include Radiotherapy, dosimetry and Medical images

**Email:** m.afkhami87@gmail.com



**Bagher Farhood** was born in West Azerbaijan, Iran, in 1989. He received the B.Sc. degree from Babol University of Medical Sciences, Iran on Radiotherapy in 2012. He also obtained M.Sc. and Ph.D. degrees from the Mashhad University of Medical Sciences, Iran and Tehran University of Medical Sciences, Iran, in 2015 and 2018, respectively. He is currently an Assistant Professor at Kashan University of Medical Sciences, Iran in Departments of Medical Physics and Radiology. His research interests include Radiotherapy Physics, Clinical Radiotherapy, Ionizing Radiation Dosimetry, Radiobiology, and Monte Carlo Simulation

**Email:** bffarhood@gmail.com