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# Small Field Dosimetry: What Have We Learnt?

Indra J Das[1, a, *)] , Johnny Morales[2,b)] , and Paolo Francescon [3,c)]

[1]*Department of Radiation Oncology, New York University Langone Medical Center, 160 East, 34th St, New York, NY 10016, USA*
[2]*Department of Radiation Oncology, Chris O'Brien Lifehouse, Missenden Road, Camperdown, NSW 2050, Australia*
[3]*Department of Radiation Oncology, Ospedale Di Vicenza, Viale Rodolfi, Vicenza 36100, Italy*

[a)]indra.das@nyumc.org
[b)]johnny.morales@lh.org.au
[c)]paolo.francescon@ulssvicenza.it

**Abstract.** Small field x-ray beam dosimetry is difficult due to a number of challenges that include a lack of lateral electronic equilibrium, source occlusion, high dose gradients, and detector volume averaging. This has become more apparent with a rapid increase in the number of treatment machines that deliver small x-ray fields which are used for precision radiotherapy techniques such as stereotactic radiosurgery (SRS) and SBRT. A large body of literature is now available for dosimetry in small fields, but with a lot of contradiction. There is also a large collection of micro-detectors that are being advocated for dosimetry. This review provides an update on small field dosimetry, recommendations for measurements and updates on recent commercial detectors on the market. It is recommended that detectors that are small volume and tissue equivalent are best suited for small field dosimetry which are plastic scintillators, synthetic diamond detectors and possibly Gafchromic films.

## INTRODUCTION

Evolution in technology has changed radiation therapy to a highest degree of sophistication and complexity that needs to be properly understood. Advances in stereotactic radiosurgery (SRS), stereotactic body radiotherapy (SBRT), for cranial and extra-cranial lesions and intensity modulated radiation therapy (IMRT) and volumetric modulated radiotherapy (VMAT) use relatively small fields (<3 cm) that are static or dynamic. This has created many innovations in treatment machines like various designs of gamma knife, CyberKnife, Tomotherapy and linear accelerators that deliver relatively small fields either in specialized cones, iris or multileaf collimators (MLC) from 10 mm leaf width to now 2.5 mm in micro-MLC. Traditional radiation oncology fields that are commissioned span from 3x3 $cm^2$ to maximum (40x40 $cm^2$) that has been described by Das et al (1). Dosimetry in small fields is complicated but relatively new as described by Das et al (2). Over a decade this has become one of the important topic to understand due to many incidences that has created national news (3, 4).

Most modern radiation therapy modalities use sub-centimeter field sizes where dosimetry is uncertain. The difficulty of dosimetry lies in the electron transport created by the photon interaction with medium. Das et al [5] have provided definition of small fields that depends on dose disequilibrium, source size and more so the selection of detector. In the past SRS dosimetry had been uncertain for small fields by as much as 14% among institutions and detectors (5). Typically published data from major institutions have been used as gold standard for dedicated devices (6) however this had large errors as we did not know the physics of small fields. The dosimetry protocols like the IAEA TRS-398 (7) and AAPM TG-51 (8) provided guidelines for reference field size which is typically 10x10 $cm^2$. Most reference conditions parameters such as stopping power ratio, perturbation correction, fluence and gradient corrections are not applicable to small fields. To overcome non-reference fields by specialized machines, the International Atomic Energy Agency (IAEA) has provided a framework of an international approach to deal with the issues in small field dosimetry (9). In the same time frame, AAPM formed a task group (TG-155) to provide relative dosimetry in small field (10).

At the same time, the manufacturers have started marketing a range of micro-detectors for use without having characterized data in small X-ray fields. This paper discusses clinical aspect of small field dosimetry, its development over a decade and a review of the current status of the literature. Most importantly, it provides a rational for appropriate choice of detector, perturbations, in various machine and energy for accurate dosimetry.

## IAEA Approach

In the extension of TRS-398 (7) dosimetry protocol for the small fields, the IAEA provided a formalism and necessary correction factors to account for possible changes in detector response for the determination of reference dose in situations when the standard reference field 10x10 cm$^2$ condition cannot be realized as the case for modern technological machine that has machine specific reference field ($f_{msr}$).

$$D_{w,Q_{\mathrm{msr}}}^{f_{\mathrm{msr}}} = M_{Q_{\mathrm{msr}}}^{f_{\mathrm{msr}}} \bullet N_{D,w,Q_o} \bullet k_{Q,Qo} \bullet k_{Q_{\mathrm{msr}},Q}^{f_{\mathrm{msr}},f_{\mathrm{ref}}} \tag{1}$$

$D_{w,Q_{\mathrm{msr}}}^{f_{\mathrm{msr}}}$ is the absorbed dose at a reference depth in water in the absence of the detector at its point of measurement in a field size specified by $f_{\mathrm{msr}}$ and beam quality $Q_{\mathrm{msr}}$. The notation "msr" stands for machine specific reference. *M* is the measurement reading by the detector (corrected for variations in environmental conditions, polarity, leakage, stem correction and ion recombination corrections). The notation $f_{\mathrm{ref}}$ denotes the conventional reference field in dosimetry protocols for which the calibration coefficient of an ionization chamber in terms of absorbed dose to water is provided by a standard laboratory and Q is the beam quality of $f_{\mathrm{ref}}$. The $f_{\mathrm{msr}}$ is the machine-specific-reference field, $N_{D,\mathrm{w}}$ is the chamber specific calibration coefficient in terms of absorbed dose to water for $^{60}$Co, and $k_{Q,Qo}$ is chamber specific beam quality calibration factor. Unfortunately the last term in Eq 1 the k values are not readily available and there is some variability in the literature. In coming years, this will be finally formalized in various published task group reports.

For small fields, it is obvious that the ratio of the detector reading is not equivalent to the ratio of doses and so a correction factor, k is needed that was introduced by Alfonso et al (9) as described in Eq. 2.

$$\frac{D_{w,Q_{clin}}^{f_{clin}}}{D_{w,Q_{msr}}^{f_{msr}}} = \left[\frac{M_{Q_{clin}}^{f_{clin}}}{M_{Q_{msr}}^{f_{msr}}}\right] k_{Q_{clin},Q_{msr}}^{f_{clin},f_{msr}} \tag{2}$$

where *D* is dose, *M* is detector reading, *f* is field, *clin* is clinical, *msr* is machine specific reference, *w* is water and *Q* is beam quality and *k* is correction factor that depends of *Q, f*, detector and machine (focal spot). For the sake of simplicity, Francescon et al (11) defined $k_\Omega$ as below:

$$\mathrm{k}_\Omega \equiv \mathrm{k}_{Q_{clin},Q_{msr}}^{f_{clin},f_{msr}} = \left(\frac{D_{\mathrm{w},Q_{\mathrm{clin}}}^{f_{\mathrm{clin}}}}{D_{\mathrm{w},Q_{\mathrm{msr}}}^{f_{\mathrm{msr}}}}\right)\left(\frac{M_{Q_{\mathrm{msr}}}^{f_{\mathrm{msr}}}}{M_{Q_{\mathrm{clin}}}^{f_{\mathrm{clin}}}}\right) \tag{3}$$

This equation is same as equation 1 it is simplified to $k_\Omega$ due to many functional form with dependence on many parameters as shown below that had been subject of intense research over a decade. Hence acquiring the values for $k_\Omega$ (*machine, focal spot, detector, detector orientation, depth,* $f_{clin}$) is subject of many international task groups. However, some observations based on published data are presented this paper.

Ionization chambers had been the backbone for radiation dosimetry. However, when field size deceases to smaller the range of charged particle, electronic equilibrium cannot be established and Bragg-Cavity theory cannot be used. There had been surge of literature (12-19) searching the solution in such situation. We will not dwell on cavity theory in this paper, however we refer to various published papers on this subject. For ionization chambers, $k_\Omega$ is derived as;

$$k_{\Omega} \equiv k_{Q_{clin},Q_{msr}}^{f_{clin},f_{msr}} = \frac{\left[\left(\frac{\overline{L}}{\rho}\right)_{air}^{w} \cdot P_{fl} \cdot P_{grad} \cdot P_{stem} \cdot P_{cell} \cdot P_{wall}\right]_{f_{clin}}}{\left[\left(\frac{\overline{L}}{\rho}\right)_{air}^{w} \cdot P_{fl} \cdot P_{grad} \cdot P_{stem} \cdot P_{cell} \cdot P_{wall}\right]_{f_{msr}}} \quad (4)$$

Where $L/\rho$ is restricted stopping power, and various other parameters are perturbations due to components of the chamber. It is noted that $P_{grad} \equiv P_{\rho} \cdot P_{vol}$ that accounts for the perturbation due to density and volume (20).

In recent years there had been significant developments in designing and marketing micro-detectors whose definition has been described by Das et al (1). However, proper characterization is subject of active research. Some of the data and choice of detectors will be discussed in this paper.

## Detectors

There are now an increasing number of available detectors on the market for small field dosimetry. These include miniature ionization chambers, diodes, synthetic diamonds, radiochromic film, plastic scintillators, MOSFETS, gel dosimeters and more. Each of the detectors will have a number of characteristics which make them suitable for small field dosimetry. The choice of a particular detector should consider the following: ready availability, multiple independent validations within the literature and with a full validation using Monte Carlo techniques to check for radiological water equivalence. In either situation, detector orientation plays an important role in dose and profile measurements that had been described by Franscescon et el (11) for percent depth dose (PDD), tissue maximum ratio (TMR) and off axis ratio (OAR). There is also a list of detectors and associated $k_{\Omega}$ values tabulated by Azangwe et al (21).

For the Tomotherapy unit where $f_{ref}$ and $f_{msr}$ are relatively large, the data on $k_{\Omega}$ varies to a very little extent. The $k_{\Omega}$ for tomotherapy as shown by Sterpin et al (22, 23) is nearly close to 1.0±0.02. For other treatment units generating small fields, the data should be carefully examined since it may be determined for isocentric or fixed SSD mode or at various depths.

## CyberKnife

The correction factor, $k_{\Omega}$ for small fields in CyberKnife have been extensively studied for various types of detectors (24-34). Francescon et al (33, 35) recently published two studies for the CyberKnife system, on the corrections needed for various detectors for use in small field dosimetry. This study provided an updated set of $k_{\Omega}$ values for a series of detectors used for measurements in the CyberKnife system at an SSD of 800 mm. These results show that micro ionization chambers tended to under-respond at small field sizes, the diodes had an over-response and that the corrections tend to unity as field size increases, are consistent with those reported in previous studies. The results also indicate that the plastic scintillator detector (PSD), Exradin W1, has a $k_{\Omega}$ correction factor close to unity. However it is important to note that this is based on Monte Carlo calculations for the geometry and the study stated that there was no experimental investigation in relation to the Cerenkov light emitted in the optical fiber for this detector (36).

In a recent experimental study by Chalkley and Heyes (32) the performance of the PTW 60019 microDiamond detector was very good for relative dosimetry measurements. This was based on a comparison with measured factors determined using PTW 60017 and 60018 diodes and PTW 31014 PinPoint chambers with application of Monte Carlo derived correction factors based on the study by Francescon et al (35) Similarly, PDD measurements with the microDiamond detector were in good agreement with those measured with the ionization chambers. However it should be noted that there was broadening of the penumbra for profile when the 60019 was oriented stem parallel to the incident beam.

Monte Carlo simulation cannot provide data with different dose rate. Thus experimental data from various detectors was provided by Francescon et al (35). This data indicated that dose rate in Cyberknife does not play any role in small field dosimetry as $k_{\Omega}$ values are relatively identical for several detectors.

## Gammaknife

Benmakhlouf et al (37) published a study to determine output correction factors for a number of commercial detectors for the Leksell Gamma Knife, by Monte Carlo calculations and measurements. The detectors which yielded the smallest corrections for the smallest field sizes were the PTW micro Liquid Ionization Chamber (MLIC) and the PTW 60019 microDiamond detector. Several recent publications comparing detectors for use for Gammaknife beam dosimetry have recommended use of the IBA SFD, the PTW microDiamond and Gafchromic radiochromic film as the detectors of choice for fields down to 4 mm diameter.(38)

## Linear accelerators

Francescon et al (27, 35, 39) provided detailed information on $k_\Omega$ factor variability for multiple linear accelerator and detector configurations down to a field size of 5 mm square. Similar data has been also been reported by other groups (16, 19). Figure 1 shows data for Siemens and Elekta machine for various detectors. Note that variation in $k_\Omega$ is dependent on type of detectors as shown in Table 1a) and 1b). For small fields it varies from 0.92 to 1.14 with a total of 22%. Similar data is also provided for Varian machine along with other vedor by Liu et al(40)who indicated that a single set of data can be used for machines, type of collimation and depths within ±2% tolerable uncertainity. This aspect will be discussed later.

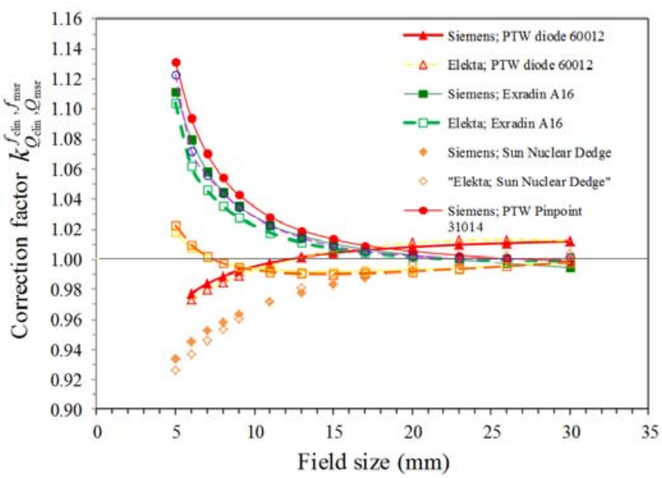


**FIGURE 1.** The factor ($k_\Omega$) for various machines and detectors versus field size. Redrawn from Francescon et al (41).

**TABLE 1 (a)** Values of $k_\Omega$ calculated as the mean over all the values obtained by changing the linac model (Siemens Primus™ 6 MV and Elekta Synergy® 6 MV), the radial FWHM and energy of the electron source as presented by Francescon *et al*.(41)

| Detectors | Field Size | | | | | |
|---|---|---|---|---|---|---|
| | 0.5x0.5 ($cm^2$) | 0.75x0.75 ($cm^2$) | 1.0x1.0 ($cm^2$) | 1.25x1.25 ($cm^2$) | 1.5x1.5 ($cm^2$) | 3.0x3.0 ($cm^2$) |
| PTW 60012-60017 | 0.968±0.003 | 0.984±0.002 | 0.995±0.001 | 1.001±0.001 | 1.006±0.001 | 1.013±0.001 |
| Sun Nuclear Dedge | 0.932±0.003 | 0.951±0.002 | 0.967±0.001 | 0.978±0.003 | 0.986±0.002 | 1.001±0.002 |
| IBA SFD | 0.972±0.003 | 0.996±0.002 | 1.007±0.001 | 1.011±0.001 | 1.015±0.001 | 1.017±0.001 |
| Exradin D1V | 0.980±0.003 | 0.994±0.002 | 0.999±0.001 | 1.000±0.001 | 1.005±0.001 | 1.012±0.001 |
| Exradin A16 | 1.112±0.018 | 1.044±0.007 | 1.020±0.001 | 1.007±0.001 | 1.002±0.001 | 0.999±0.001 |
| PTW PinPoint 31014 | 1.128±0.018 | 1.053±0.007 | 1.024±0.002 | 1.010±0.001 | 1.005±0.001 | 1.000± 0.001 |
| PTW MicroLion | 1.023±0.009 | 0.997±0.002 | 0.993±0.001 | 0.992±0.001 | 0.994±0.001 | 0.998±0.001 |
| Exradin W1 PSD | 0.998±0.003 | 0.995±0.002 | 0.996±0.002 | 0.996±0.002 | 0.994±0.002 | 0.994±0.002 |

**TABLE 1 (b)** Monte Carlo calculated, $k_{\Omega}$ for 6 MV beam of a Varian IX machine. The uncertainity is 0.15% as presented by Benmakhlouf *et el.*(42)

| Detectors | Field Size | | | | |
|---|---|---|---|---|---|
| | 0.5x0.5 ($cm^2$) | 1x1 ($cm^2$) | 2x2 ($cm^2$) | 4x4 ($cm^2$) | 10x10 ($cm^2$) |
| PTW 60016 (shielded-photon diode) | 0.910 | 0.956 | 0.996 | 0.998 | 1.000 |
| PTW 60017 (Unshielded electron-diode) | 0.945 | 0.992 | 1.016 | 1.014 | 1.000 |
| PTW PinPoint 31014 (parallel) | 1.102 | 1.001 | 1.003 | 1.004 | 1.000 |
| PTW PinPoint 31014 (perpendicular) | 1.147 | 1.010 | 1.000 | 1.001 | 1.000 |
| PTW T31018 (MicroLion) | 1.011 | 0.992 | 1.003 | 1.003 | 1.000 |
| PTW T60003 (Diamond) | 1.002 | 0.997 | 1.008 | 1.005 | 1.000 |
| IBA PFD | 0.947 | 0.951 | 0.983 | 0.991 | 1.000 |
| IBA SFD | 0.980 | 1.016 | 1.023 | 1.021 | 1.000 |
| IBA CC01 (parallel) | 1.050 | 1.003 | 1.002 | 1.000 | 1.000 |
| IBA CC01 (Perpendicular) | 1.081 | 0.996 | 1.000 | 1.000 | 1.000 |

Bassinet et al (43) has presented correction factors ($k_{\Omega}$) for a range of detectors used for a Varian linear accelerator operating with 6 MV x-rays with MLC shaped fields down to 6×6 $mm^2$. Their reference dose values were taken to be the mean dose reading from Gafchromic EBT2 film and small LiF TLD cubes. Their work demonstrated that for passive dosimeters an excellent agreement was observed (better than 2%) for all relative output factor measurements. Furthermore, in their work correction factors for active detectors were determined from the mean experimental output factors measured by passive detectors.

Papaconstadopoulus et al (44) performed a full Monte Carlo model of a Novalis linear accelerator, using the BEAMnrc Monre Carlo code and a full geometric model of various detectors including the PTW microDiamond, the PTW microLion and the W1 PSD detector. They found that the values of $k_{\Omega}$ for the microDiamond varied by up to 1.4% down to a field size of 0.5 cm which confirms that it is radiologically water equivalent with minimal correction factors even at very small field sizes. Figure 2 also shows a significant variation in response when the W1 PSD is orientated in the perpendicular direction and indicates that the W1 detector is under-responding relative to the dose in water at the same point. This effect should be investigated in any PSD measurements otherwise incorrect doses may be measured and lead to incorrect values of $k_{\Omega}$ particularly at the smaller field sizes.

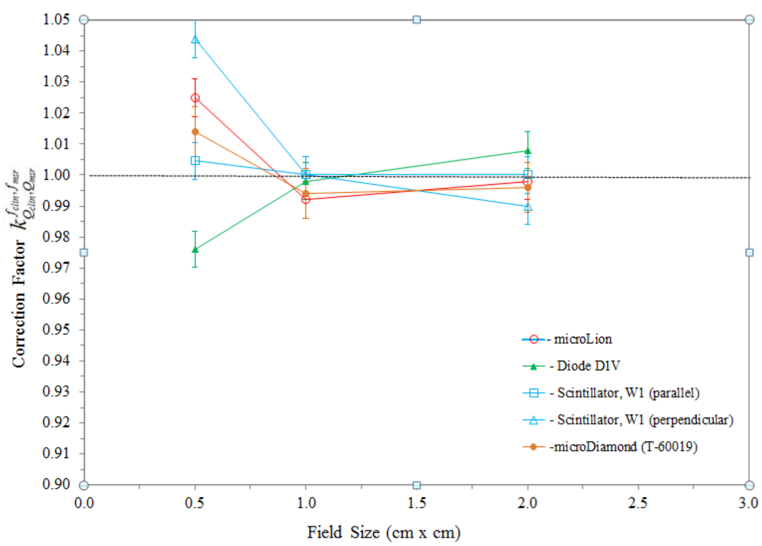

**FIGURE 2.** Correction factor, $k_{\Omega,}$ for several modern detectors on a Novalis linac operating at 6 MV and with an $f_{msr}$ is 10×10 $cm^2$.

## Discussions

From the vast amount of data on for various machines (various focal spot), detector, depth and field size, the data in Fig 3 was compiled from various references (31, 34, 42, 45, 46), shows the variation of detector response. A clear picture emerges that there are certain detectors whose k values are close to unity even for very small field sizes of 5 mm. These recommended detectors are the PTW 60019 microDiamond, the Standard Imaging W1 PSD, the PTW micro LIC and Gafchromic EBT2/EBT3 radiochromic film.

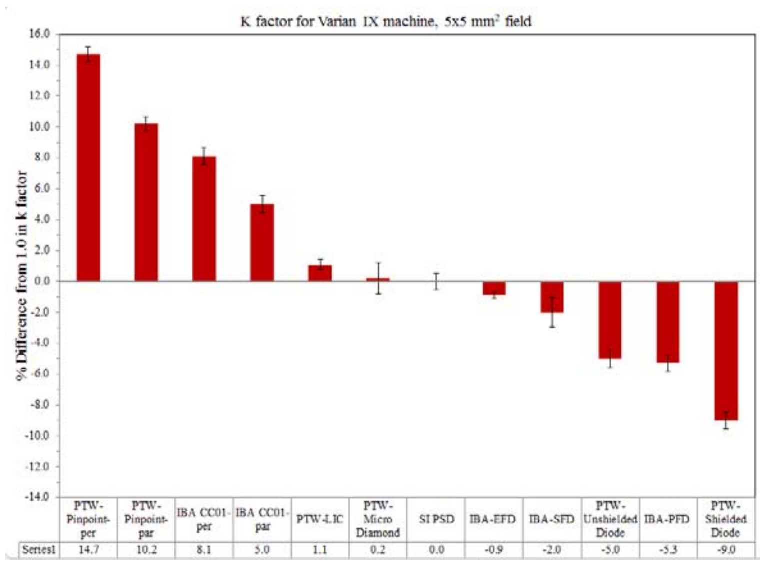


**FIGURE 3**. Difference of k values from unity is plotted for various detectors used in small fields for 5×5 $mm^2$ from a 6 MV linear accelerator. There are a limited number of detectors with k values that are ±2% compared to unity and can be recommended for use.

The focal spot or source size is an important consideration in small fields (2). There are various methods to measure source size (47-50), however it is beyond the scope of most clinical physicists. In such situation question lies as how sensitive is source size with respect to $k_{\Omega}$ of a detector. The answer is shown in Fig 4 analyzed from data from Moigner et al (34). It shows that for a given cone, source size does not play any important role (Fig 4). The same finding was also echoed by Liu et al (40) who viewed this question in terms of type of machine, collimation and depth of measurement and found that data can be interchangeable used within ±2% limit from one system to other. Czarnecki et al (51) provided conclusive data indicating that $k_{\Omega}$ is relatively independent of focal spot of various detectors very similar to data as shown in Figure 7. Sam et al (52) provided extensive Monte Carlo data for very small fields with variable focal spot. The data (PDD, Profiles) are only different for extremely small fields and can be ignored for modern machines and within the accuracy needed. Scott et al (53) provided extensive data for the effect of source occlusion in small fields and its impact of profile and output. It was noted that effect is only pronounced for larger source size and very small fields (≥5 mm).

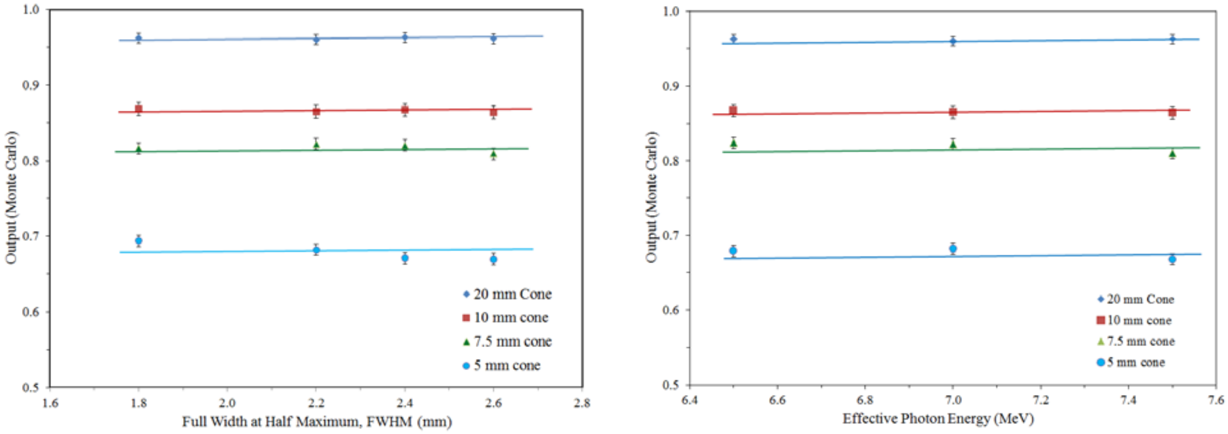


**FIGURE 4**. Variation of output for (a) focal spot (FWHM) for a 7 MeV effective photon energy and (b) effective photon energy for a FWHM of 2.2 mm for CyberKnife simulated using Monte Carlo. Data adapted from Moigner et al (34). These data are in agreement with other published data (40) for Elekta and Varian machines, indicating that machine variation for same energy can be ignored within the limit of simulation and measurement accuracy.

Even though, data for K is being published, but due to uncertainty, Kamio et al (57) provided a unique method to limit the field size and detector combination for correction less condition. The recommendations for clinical work are to use a second detector with small correction factors for all measurements. For example, this could mean using radiochromic film and one of the unshielded diodes as the secondary measurement chain. While peer reviewed published data is available as discussed in this paper, the results can be variable due to fine details in the measurements or the simulation. For example, output factor can be measured at $d_{max}$, 5 cm or 10 cm depth in SSD or SAD mode. The SAD vs SSD data can be variable and one should be careful in comparing results as differences of several percent will occur just due to geometrical setup differences.

For example data provided by Benmakhlouf et al (42) is for 100 cm SSD at 10 cm depth whereas data from Franscescon et al (41) is at SAD. Machine variation or source size is not significant as originally thoughts. Data from Moignier et al (34) shows that output is relatively constant for the various energy and focal spot.

## Summary

Despite size of focal spot or source size, that produces source occlusion with small fields which impact dosimetric parameters, its impact in modern machines having submillimeter source size (58) for selected detector is minimum. Microdetectors can be classified in two groups regular and suitable. It is found that suitable class of microdetectors is PTW microLion, PTW microdiamond, the Exradin W1 PSD, Gafchromic EBT2/3 films and the D1V diode. In general, we can conclude that detectors that are small volume and water equivalent are best suited for small field dosimetry. The microLion has been discontinued by manufacturer and is not available commercially. For the Gafchromic EBT2/3 films, proper precautions and experience is needed for achieve comparable data (59-61). User should also evaluate the criterion for correction-less dosimetry if there is any doubt as discussed by Kamio and Bouchard (57).

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