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CALCULATION OF DETECTOR SPECIFIC OUTPUT CORRECTION FACTORS BASED ON IAEA-AAPM TRS-483 FOR 6 MV FLAT / UNFLAT PHOTON BEAMS AND COMPARISON WITH LITERATURE

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ABSTRACT

Purpose: In the TRS-483 recommendations, output correction factors to be applied in small fields are specified as a single value for each detector and field size, regardless of the Linac model, collimation, or filtering difference. Since Scp (Total Scatter Correction Factor) values are used in the treatment MU value calculations, if the Correction Factors due to the detector differences are entered into the TPS incompletely, the dose calculations are incorrect in the treatment plans where small fields such as SRS, SRT, SBRT are used frequently.

Methodology: In our study, correction factors for 7 different detectors and 9 different field sizes (0.5, 1.0, 2.0, 2.5, 3, 3.5, 4.0, 4.5, 5.0 cm2) were calculated using the experimental method and their compatibility with the correction factors specific to the detectors specified in the literature was investigated. 6 WFF (With Flattening Filters) and 6 FFF (Flattening Filter Free) photon output factors for IBA CC01, IBA CC04, SI Extradin A16, PTW Pinpoint 3D, PTW microDiamond, Sun Nuclear Edge and PTW SRS Diode were calculated with Daisy Chain method.

Findings and Conclusion: It was observed that ion chambers are under-response in field sizes smaller than 3 cm^2 and need a larger correction factor, whereas solid-state detectors are over-response, and need a relatively smaller correction factor than ion chambers. The percent variation between the correction factors for the two energies 6 WFF / 6 FFF was found to be a maximum of 2% for both ion chambers and solid detectors at field sizes of 2 cm^2 and larger.

Keywords: *Output Correction Factors, Small Field Dosimetry, TRS-483.*

INTRODUCTION

Radiation therapy is a treatment method that damages the DNA to prevent the growth and spread of tumour cells within the body by using ionization radiation. Its goal is to safely deliver the prescribed dose to tumour cells while limiting the radiation to healthy tissues in the surrounding area. The TCP (Tumour Control Probability) and NTCP (Normal Tissue Complication Probability) models are indications of the success rates of the planned treatment. The outcome of these models is displayed in Dose-Volume Histograms (DVH) to evaluate target and OARs (Organs at Risk) doses [1]. Scp factors obtained by output measurements of the Linear Accelerator for different field sizes are used to calculate treatment MU values. Dose calculations will be erroneous in treatment plans when small fields such as SRS, SBRT, and SRT are commonly employed if output correction factors are incorrectly or incompletely supplied owing to detector differences. Since imperfect dosimetry measurements can have major impacts, such as significantly higher doses or potentially fatal clinical results for patients, it is crucial to comprehend the physics and limitations involved. Guidelines for the application of accurate dosimetry measurements under various conditions are available, including IAEA TRS-398 [2], AAPM TG-51 [3], IAEA TRS-483 [4], and AAPM TG-155 [5].

Due to the technological breakthrough of treatment techniques like IMRT (Intensity-Modulated Radiation Therapy), VMAT (Volumetric Modulated Arc Therapy), SRT (Stereotactic Radiotherapy), SRS (Stereotactic Radiosurgery), and SBRT (Stereotactic Body Radiation Therapy), higher dose fractionations in daily clinical use have become standard procedures. While the desired dose is delivered to the target volume using these sophisticated techniques, a rapid dose gradient is created by using nonconventional fields (3 x 3 cm2), referred to as "small fields," resulting in maximum critical organ protection. On the other hand, when dealing with small fields, accurate dosimetry measurements can be significantly more challenging compared to conventional fields (large fields). These physical dosimetry issues related to small fields are clearly emphasized in the IAEA TRS-483 Code of Practice (CoP), alongside the implementations and protocols that should be taken into consideration when dealing with relative and reference dosimetry for external beam photon radiation therapy [6].

Various detectors with different physical properties suitable for small field dosimetry have been produced by manufacturers [7]. When the area of interest is a small field, suitable detectors should have minimal energy dependence to produce high SNR (Signal Noise Ratio) with noise reduction. The wall material and electrode material of the detectors should be suitable for small field measurements without causing the Volume Averaging Effect.

The assumption that the ratio of absorbed dose to measured detector readings for larger field sizes where LCPE (Lateral Charge Particle Equilibrium) is achieved, is almost equal, is also not valid for small field dosimetry measurements. Due to the volume averaging and perturbation effects of the detector employed at measurements, the measured doses must be multiplied by a coefficient depending on the characteristics of the detector used to calculate the absorbed dose. In the literature, this coefficient is known as the output correction factor, and it is provided in detailed lists in the IAEA TRS-483 Code of Practice (CoP) for different operating energies (6 and 10 MV photon beams), detectors, and devices (Linac, CyberKnife, Tomotherapy, Gamma Knife) [4, 8, 9]. An extensive number of investigations have been conducted and are continuing by many researchers to avoid erroneous dose measurements caused by the small field dosimetry challenges. Numerous detectors are commercially accessible for small field dosimetry. These comprise micro ionisation chambers, diodes, synthetic diamonds, radiochromic film, plastic scintillators, metal-oxide-semiconductor field-effect transistors, and gel dosimeters [10]. Selecting a detector for small fields relies on literature with numerous validations, particularly through Monte Carlo (MC) simulations to verify radiological equivalence to water. Bassinet et al. [11] evaluated the output factors of small photon beams using different detectors and determined their corresponding correction factors. The analysis indicated that EBT2 exhibits correction factors that are approximately equal to one. Consequently, the Gafchromic film was employed as a passive detector in the present research to derive the detector-based correction factor.

Our objective in this investigation is to compute the output correction factors for 2 photon energies of 6 MV with flattening filter, denoted by 6WFF, and 6 MV without flattening filter, denoted by 6FFF, photon beams using 7 different detector types (IBA CC01, IBA CC04, SI Extradin A16, PTW Pinpoint 3D, PTW microDiamond, SN Edge, and PTW Diode SRS) and 9 different square field sizes (0.5, 0.8, 1, 2, 2.5, 3, 3.5, 4, 5 cm2) using the experimental method, and benchmark our findings with the published parameters for each detector in the TRS-483 Code of Practice.

All measurements were done with Elekta Versa HD Linac (Elekta, Stockholm, Sweden) system equipped with an Agility 160-Leaf MLC (Multi-Leaf Collimator). The Daisy Chain technique described in the literature [4, 10] was used to determine the relative output factors and to minimize the volume averaging effect caused by the detector in small fields. In the daisy chain method, a small field dosimeter is cross calibrated with an ion chamber at an intermediate field, which was selected as 4 x 4 cm2, as it is commonly chosen. This technique also uses three different field sizes: the reference field (10 x 10 cm2), the 4 x 4 cm2 intermediate field, and the clinical small field size.

This study has three aims. 1) To calculate the detector-specific output correction factors for the detectors and field sizes we used in our study and to compare the results with the literature, 2) To calculate output correction factors for detectors and field sizes not included in the TRS-483 CoP and contribute to the literature, 3) To investigate the detectors' dose-response for two different energies (6WFF and 6FFF).

MATERIAL AND METHODS

All measurements were taken using an Elekta Versa HD Agility 160 MLC linear accelerator device. Seven different active detectors with sensitive volumes appropriate for the above mentioned small fields from various manufacturers were used to measure the output reading of 6WFF and 6FFF beams. Ion chambers with a sensitive volume of less than 0.01 cm3 are called micro ion chambers, while those with a sensitive volume of between 0.01 cm3 and 0.03 cm3 are called mini ion chambers. We used two mini ion chambers (IBA CC04 and PTW 31022 Pinpoint 3D), two micro ion chambers (Standard Imaging Extradin A16 and IBA CC01), one synthetic diamond detector (PTW 60019 micro Diamond), one n-type shielded diode detector (Sun Nuclear Edge), and one p-type unshielded diode detector (PTW 60018 SRS diode). Figure 1 summarizes the properties of the detectors. Gafchromic film was used as a reference detector with a close to unity correction factor. We evaluated the correction factors obtained for both energies (6 WFF / 6 FFF) for each detector utilising the one-tailed Student's t-test, with $p < 0.05$ considered statistically significant. The IBA Smart Scan 3D water phantom (IBA Dosimetry, Schwarzenbruck, Germany) was used for the measurements taken with active detectors and the RW3 Solid Water Phantom (PTW Freiburg, Germany) was used to measure Gafchromic films on an Elekta Versa HD. The stem effect that will occur in water phantom measurements affects the output factor and profile measurements. In order to minimize the stem effect, measurements were taken by positioning IBA CC13, IBA CC04, IBA CC01, SI Extradin A16, PTW 31022 Pinpoint 3D, and Sun Nuclear Edge diode detectors perpendicular to the beam direction, whereas PTW 60019 micro Diamond and PTW 60018 diode SRS detectors were parallel to the beam direction. Figure 2 illustrates the orientation of the detector within a water phantom in relation to the beam direction.

Manufacturer	IBA	IBA	IBA	Standard Imaging	PTW Pinpoint 3D	PTW microDiamond	PTW SRS Diode	Sun Nuclear
Model	[illegible]	[illegible]	[illegible]	[illegible]	[illegible]	[illegible]	[illegible]	[illegible]
Sensitive Volume	[illegible]	[illegible]	[illegible]	[illegible]	[illegible]	[illegible]	[illegible]	[illegible]
Central Electrode	[illegible]	[illegible]	[illegible]	[illegible]	[illegible]	[illegible]	[illegible]	[illegible]
Reference Depth	[illegible]	0.8 mm	1.2 mm	0.72 mm	2.4 mm from chamber tip	[illegible] from detector tip	0.74 mm from detector tip	[illegible]
Orientation	[illegible]	[illegible]	[illegible]	[illegible]	[illegible]	[illegible]	[illegible]	[illegible]
Nominal Voltage	[illegible]	[illegible]	[illegible]	[illegible]	[illegible]	[illegible]	[illegible]	[illegible]

Figure 1: Characteristics of the detectors used for measurements.

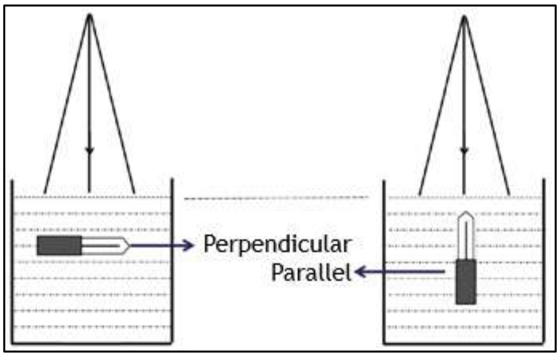

Figure 2: Detector orientation inside water phantom.

Measurements of 7 active detectors within the IBA Smart Scan water phantom

The below specified steps were carried out separately. Measurements were taken by positioning each detector in the water phantom at the EPOM (Effective Point of Measurement) as specified in the manufacturer's product data. The EPOM for the ion chambers was calculated as 0.6 radii. Each field size of 0.5, 0.8, 1, 2, 2.5, 3, 3.5, 4, and 5 cm2 at SSD was measured three times. The measurements were taken and recorded using an IBA Scanditronix Wellhöfer Dose1 electrometer at the point where the maximum signal was obtained. In our calculations, since polarization correction was applied to the ion chambers in our calculations, the ion chamber measurements were taken separately at +/- 300 Volts. The bias voltage for solid-state detectors is set to 0 volts, as described in the manufacturer's catalogues.

1) CAX (Central Axis) deviation was done with IBA myQA software for the field size of 10 x 10 cm2 at SSD: 100 cm and depth:10 cm to validate that being in the center of the irradiation field, and it was verified whether there was any deviation from the central axis and necessary shifts were made if any.

2) The inline and crossline profiles were evaluated with IBA myQA software for the field sizes of 0.8 x 0.8 cm2 at SSD: 100 cm and a depth of 10 cm following CAX deviation, and it was checked whether there was any deviation from the central axis and if required changes were made.

3) Finally, for a field size of 0.8 x 0.8 cm2, the dose was measured with the IBA myQA software at the center, + 0.2 mm, and - 0.2 mm in the inline

direction, and the point at which the signal was maximum was determined.

The measurement of the EBT 3 Gafchromic with the RW3 solid phantom

Films to be calibrated and irradiated were selected from the same lot number, 01142102, for both energies. The subsequent procedure was implemented for the calibration and readout of the irradiated films as recommended by AAPM [12].

1) Calibration films were cut into 14 equal pieces using a gladiator. Due to the directional dependencies of the EBT3 films, the films were cut into equal strips so that the long edge of the film was parallel to the long edge of the Epson Expression 10000 XL scanner, and the upper right corners were marked by specifying the dose rates to be irradiated. The calibration films were individually irradiated in 50 MU increments from 50 MU to 600 MU at SSD: 100 cm and depth dmax, placed in the center of the solid phantom, respectively. 500 MU were irradiated twice with 6XWFF. Two non-irradiated calibration films were left.

2) After the calibration films, the dose films were cut, and each area size was written in the upper right corner, taking into account the direction of the films, and then irradiated. The area sizes of 0.5, 0.8, 1, 2, 2.5, 3, 3.5, 4, 5, and 10 cm2 were irradiated 3 times with 500 MU to reduce uncertainties that may occur during the scanning process. Individually, the films were irradiated at SSD: 90 cm and depth: 10 cm by placing them in the center of the solid phantom. Since Gafchromic films are affected by UV light and are directional dependent, a very strict protocol has been followed in the process. Films were not touched with bare hands and gloves were worn. Film cutting and irradiation were done in a dark environment.

The scanning procedure of EBT 3 Gafchromic Films

Gafchromic films consist of monomer compounds that turn into polymers under radiation [11, 12]. It is advised in the literature to wait at least 24 hours for the polymerization to be finished and stabilized before reading the optical density measurements. For this purpose, after the completion of irradiations, 36 hours pass before the scanning of the 6 WFF calibration and dose films. Scans were performed in a dark environment with the Epson Expression 10.000 XL scanner. The scanner software for Windows 11 was installed on the computer, and the automatic photo correction settings were turned off by selecting Professional Mode. The driver is set to 48-bit colour, positive film, and 150 dpi resolution. Since the reading was made in the transmission mode, the films were fixed with a 3.8 mm thick 30 x 40 cm Plexiglas sheet. The scanner was warmed up for 30 minutes before the scans. By making five previews before scanning, the scanner's lamp was warmed up. To minimize the uncertainties that may occur during scanning, 3 films of each field were scanned 5 times, and the films were saved as tiffs. (Figure 3).

Figure 3: Scanning process of EBT3 Film with Epson Expression 10000XL.

ImageJ analysis software for the conversion of optical density values to absorbed dose at EBT3 Gafchromic film

Calibration films irradiated with 6WFF and saved in tiff format were split into colour channels using ImageJ software and read over a single channel (red channel). The grayscale values for each dose value were read by putting ROIs of 150 pixels in width and height in the middle of the calibration films and applying backup corrections. Based on the article by Howard et al. [13], the "Rod bard" function was used to construct the dose-response curve (calibration curve) (Figure 4). ImageJ software was used to calculate the constants a, b, c, and d. In the calculation, we used dose values as x and grayscale values read by drawing ROI as y.

$$y = d + \frac{(a-d)}{1+\left(\frac{x}{c}\right)^{b}}$$

a = 47706.03, b = 0.87108, c = 203.209, d = 4841.28

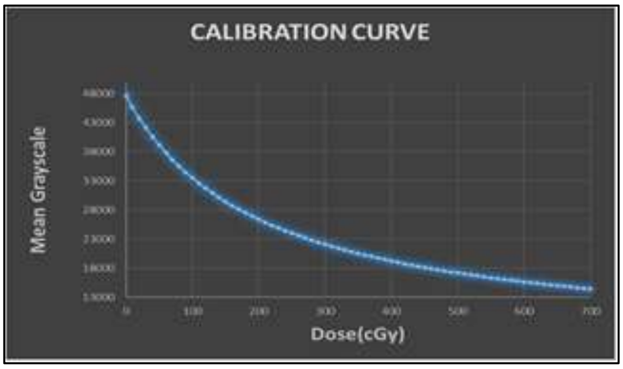

Figure 4: Calibration Curve for 6WFF Photon beams

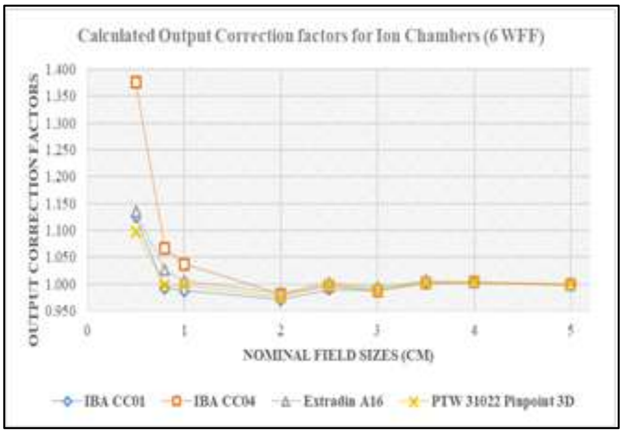

Figure 5a: Ion chambers output correction factors at 10 cm depth for 6 WFF.

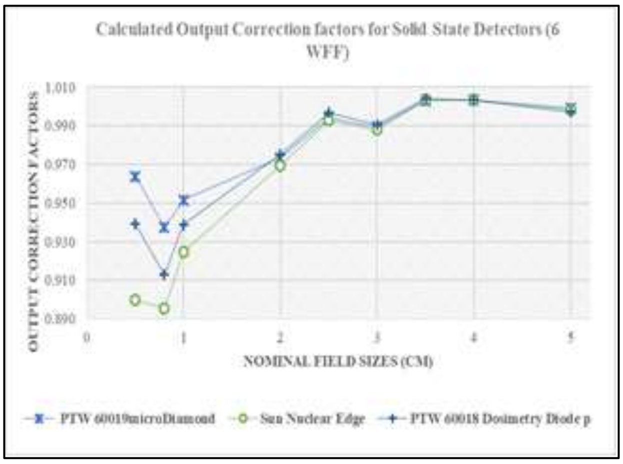

Figure 5b: Solid-state detectors output correction factors at 10 cm depth for 6WFF.

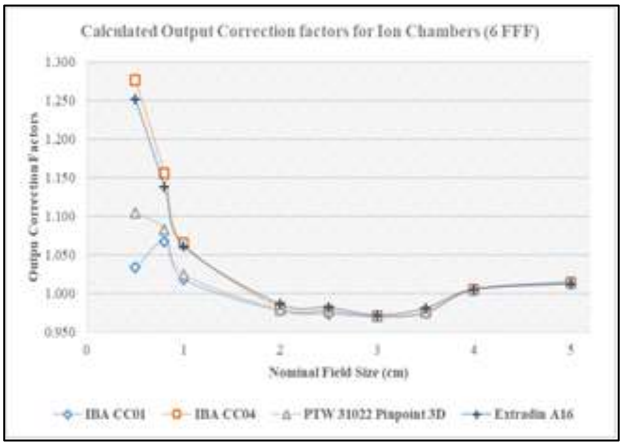

Figure 5c: Ion chambers output correction factors at 10 cm depth for 6FFF.

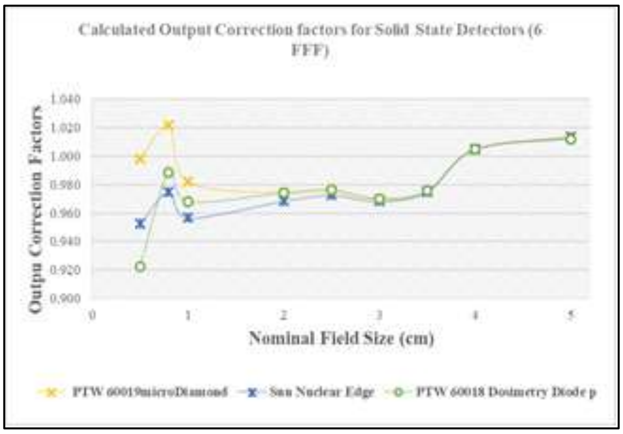

Figure 5d: Solid-state detectors output correction factors at 10 cm depth for 6 FFF.

RESULTS

In the TRS-483 CoP recommendations [4], it is stated that the output correction factors to be applied in small fields should be between 0.95 and 1.05. In our study, the output correction factors calculated using the Daisy Chain method for both energies as a consequence of the measurements for the field sizes of 1, 2, 2.5, 3, 3.5, 4, and 5 cm2 were found in the range specified in TRS-483 [4]. The correction factors for the ion chamber measurements for the two smallest field sizes, 0.5 and 0.8 cm2, were found to be greater than 1.05 as expected. This was due to issues such as the LCPE, the partial blocking of the source, perturbation effects, and the volume averaging effect, which occur when field sizes are smaller than 3 cm2. In the study of Oh et al, it was observed that solid-state detectors should be used instead of ion chambers with field sizes smaller than 3 cm2 in small field measurements [14]. For relative dosimetry, it is also recommended to use detectors whose output correction factors are almost unity, like films, scintillations, or detectors, having a low atomic number central electrode. Our results are consistent with the literature [14]. The calculated output correction factors based on the measurements for 2 different photon energies and for each detector that we studied are shown in Table 1 and Table 2.

In addition, Tyler et al. observed that the detector response in small field measurements is under-response for detectors such as ion chambers since the measurement environment is air and the density of air is lower than the density of water [15]. To put it another way, they discovered that bigger correction factors are required. Again, in their study, they found that solid-state detectors such as diamond and silicon diode detectors are over-responsive because the silicon material is denser than water. In our work, we discovered that ion chambers (IBA CC01, IBA CC04, Standard Imaging Extradin A16, PTW 31022 Pinpoint 3D) (PTW 60019 microDiamond, Sun Nuclear Edge, PTW 60018 SRS Diode) were over-response for the smallest field size (0.5 cm2). This behaviour of detectors is shown in Figure 5.a, Figure 5.b, Figure 5.c, and Figure 5.d. The results obtained with PTW 60019 microDiamond detector at a field size of 0.8 cm2 for 6FFF energy and the IBA CC01 detector at a field size of 0.8 and 1 cm2 for 6WFF energy did not match with the findings of Tyler et al [15].

The IBA CC01 was overresponsive at 6WFF, among other ion chambers. The SI Extradin A16 detector was under response, but the required correction factor was greater than with other air-filled ion chambers. Both results showed that the volume average effect may have occurred due to the electrode material of these detectors being made of steel, thus altering the output factor response of the scattered electrons. Masanga et al. found similar results for the smallest field sizes, confirming our observation [16].

According to Lechner et al.'s study, diamond and unshielded detectors require a smaller correction factor than shielded detectors [17]. Cranmer-Sargison et al. discovered that shielded detectors have a higher dose-response because of the contribution of low-energy scattered photons from the shielding, resulting in overdose-response and requiring a larger correction factor than unshielded detectors [18, 19]. The results obtained in our study are compatible with these findings for both energies (6WFF and 6FFF). In our smallest four field sizes of 0.5, 0.8, 1, and 2 cm2, we discovered that PTW 60019 microDiamond and PTW 60018 SRS diode (unshieldedtyl) detectors require a smaller correction factor than the Sun Nuclear Edge detector for both energies. The correction factor of 0.953, determined for the Sun Nuclear Edge detector at a field size of 0.5 cm2 for 6FFF energies, is the only exception.

Table 1. Detector-specific output correction factors for 6 MV flattening-filter photon beams

Detector specific output correction factors for each field size (6 MV WFF)

Field Sizes (cm2)	IBA CC01	IBA CC04	A16	PTW 31022 Pinpoint 3D	PTW 60019 micro Diamond	Sun Nuclear Edge	PTW 60018 SRS Diode
0.5	1.126	1.375	1.135	1.098	0.964	0.900	0.939
0.8	0.992	1.067	1.026	0.997	0.937	0.896	0.913
1	0.987	1.037	1.006	0.999	0.951	0.925	0.939
2	0.971	0.981	0.982	0.976	0.973	0.969	0.975
2.5	0.990	0.996	1.002	0.999	0.994	0.993	0.997
3	0.987	0.988	0.993	0.988	0.989	0.988	0.990
3.5	1.002	1.001	1.007	1.003	1.003	1.003	1.004
4	1.003	1.003	1.003	1.003	1.003	1.003	1.003
5	1.000	0.998	0.999	1.000	0.999	0.998	0.997

In our study, the largest output correction factor for our two smallest field sizes, 0.5 and 0.8 cm2, was calculated for the IBA CC04 ion chamber due to having the largest sensitive volume (0.04 cc) compared to other detectors. For field sizes of 0.5 and 0.8 cm2, it is 1.375 and 1.067 at 6WFF energies, and 1.277 and 1.156 at 6FFF energies, respectively. The values we found for both energies (6WFF and 6FFF) for each detector were evaluated by applying the one-tail student test, and $p < 0.05$ was calculated for all detectors, and no statistically significant difference was observed (Table 3). On the other hand, in the measurements obtained in Tyler et al.'s study with the Elekta linac, it's stated that the effect on output correction factors, using filters (WFF) and not using filters (FFF) in all measured field sizes, is between +/- 1% for ion chambers and +/- 1.2% for solid-state detectors [15].

The percent deviation between the 6WFF and 6FFF output correction factors we observed is a maximum of 2% for both ion chambers and solid-state detectors at field sizes of 2 cm2 and larger. This value is observed at 3% for all detectors except the Extradin A16 detector at 1 cm2 field size. For our two smallest areas, 0.5 and 0.8 cm2, the percentage deviation was a maximum of 8%. According to Tylor et al. and our findings, there may be an increase in uncertainty in the correction factors when just one value is used for both energies, as specified in TRS-483. Table 4 and Table 5 are the percent variation comparisons between our results and TRS-483. In the study by Bassinet et al., the output correction factor for the Sun Nuclear Edge detector was found to be 0.945 in the Novalis model Linac device collimated with a 4 mm circular cone at 6 FFF energy [20]. As a result of our study, the field size correction factor of 0.5 cm2 for the Elekta Versa HD device collimated with Agility 160 MLC and the Sun Nuclear Edge detector is 0.953. This result shows that the Linac device model and collimation cause an error of 1% in small fields. The study by Charles et al. confirms the findings and shows that the error due to the size of the MLC or jaw-shaped area in small fields (1 cm2 area size) is reflected in the output data as 3% [21]. Results of our experiment and literature reviews show that modeling of TPS data using the correction factors suggested in TRS-483 [4] is not sufficient, and the correction factors calculated as a result of the measurements are also affected by the device and collimation. When our findings and the study of Bozidar Casar et al. are compared, the deviation from the percentage value for 6 WFF energies varies between 2% and 5.3% in field sizes less than 2 cm2 for all detectors we used except IBA CC04 [8, 22]. At 6 FFF energy, the deviation from the maximum percentage value was 8.7% in the Sun Nuclear Extradin 16 detector. This result, on the other hand, confirms that the correction factors differ in clinical conditions even if the device and model used are the same, and therefore the collimation systems should be the same. It also confirms our observation that each clinic should calculate the detector-specific correction factors for its device instead of using the correction factors specified in the literature studies.

Table 2. Detector-specific output correction factors for 6 MV flattening-filter –free photon beams

Detector specific output correction factors for each field size (6 MV FFF)							
Field Sizes (cm2)	IBA CC01	IBA CC04	A16	PTW 31022 Pinpoint 3D	PTW 60019 micro Diamond	Sun Nuclear Edge	PTW 60018 SRS Diode
0.5	1.033	1.277	1.252	1.105	0.998	0.953	0.922
0.8	1.067	1.156	1.138	1.082	1.022	0.975	0.989
1	1.018	1.065	1.061	1.025	0.982	0.957	0.968
2	0.977	0.981	0.985	0.979	0.974	0.968	0.974
2.5	0.973	0.977	0.981	0.977	0.975	0.972	0.976
3	0.970	0.970	0.971	0.970	0.970	0.968	0.970
3.5	0.974	0.975	0.980	0.976	0.975	0.975	0.976
4	1.005	1.005	1.005	1.005	1.005	1.005	1.005
5	1.000	0.998	0.999	1.000	0.999	0.998	0.997

In the study of Azangawe et al, the output correction factor for the PTW microDiamond 60019 detector was calculated as 0.961 in the Elekta Precise for a 0.6 cm2 field size and 6WFF energy [23]. The PTW microDiamond 60019 detector correction factor found as a result of our study is 0.964 for a 0.5 cm2 field size in the Elekta Versa HD. Two different models of Linac systems belonging to Elekta are compatible with each other with a value of 0.3. The values found by Azangawe et al. and in our studies are 1,000 and 0.989, respectively, for 3.0 cm2 field sizes, and the percentage variation is 1.1%.

Table 3. One-tail t-test between 6WFF and 6FFF results

P values obtained as a result of 6 WFF and 6 FFF photon beam comparions for each detector	
IBA CC01	0.432
IBA CC04	0.435
Extradin A16	0.100
PTW 31022 Pinpoint 3D	0.252
PTW 60019 micro Diamond	0.181
Sun Nuclear Edge	0.166
PTW 60018 SRS Diode	0.361

Table 4. Percent Variation for 6 MV WFF

Percentage variation of our 6 WFF results from TRS-483 values for each detector							
Field Sizes (cm2)	IBA CC01	IBA CC04	A16	PTW 31022 Pinpoint 3D	PTW 60019 micro Diamond	Sun Nuclear Edge	PTW 60018 SRS Diode
0.5	-	-	-	-	-0.002	-	0.014
0.8	0.034	-	0.016	-	0.041	0.058	0.062
1	0.030	0.004	0.021	-	0.033	0.043	0.045
2	0.038	0.021	0.021	-	0.024	0.025	0.031
2.5	0.018	0.004	-0.001	-	0.005	0.005	0.012
3	0.021	0.012	0.007	-	0.011	0.011	0.02
3.5	-	-	-	-	-	-	-
4	0.004	-0.003	-0.003		-0.003	-0.003	0.007
5	-	-	-	-	-	-	

In the study of Azangawe et al, the output correction factor for the PTW microDiamond 60019 detector was calculated as 0.961 in the Elekta Precise for a 0.6 cm2 field size and 6WFF energy [23]. The PTW microDiamond 60019 detector correction factor found as a result of our study is 0.964 for a 0.5 cm2 field size in the Elekta Versa HD. Two different models of Linac systems belonging to Elekta are compatible with each other with a value of 0.3. The values found by Azangawe et al. and in our studies are 1,000 and 0.989, respectively, for 3.0 cm2 field sizes, and the percentage variation is 1.1%. In the study of Weber et al , the output correction factor for the PTW 60018 SRS Diode detector was found to be 0.928 at 6WFF energy for a field size of 0.5 cm2. The correction factor found as a result of our study is 0.939. The result we found with a variance of 1% is consistent with the literature [24].

Table 5. Percent Variation for 6 MV FFF

Percentage variation of our 6 FFF results from TRS-483 values for each detector							
Field Sizes (cm2)	IBA CC01	IBA CC04	A16	PTW 31022 Pinpoint 3D	PTW 60019 micro Diamond	Sun Nuclear Edge	PTW 60018 SRS Diode
0.5	-	-	-	-	-0.037	-	0.032
0.8	-0.040	-	-0.092	-	-0.047	-0.026	-0.017
1	0.000	0.023	-0.033	-	0.002	0.010	0.015
2	0.031	0.020	0.017	-	0.022	0.025	0.031
2.5	0.034	0.023	0.019	-	0.024	0.025	0.032
3	0.038	0.030	0.029	-	0.031	0.031	0.041
3.5	-	-	-	-	-	-	-
4	0.001	-0.006	-0.006		-0.006	-0.006	0.004
5	-	-	-	-	-	-	

DISCUSSION

TRS-483 [4] contains no information about each detector or field size. As can be seen in Tables 1 and 2, the correction factors for the PTW 31022 Pinpoint 3D detector are not specified in TRS-483 [4]. Again, the output correction factors of IBA CC01, IBA CC04, Extradin A16, and Sun Nuclear Edge detectors for the smallest field size of 0.5 cm2 and correction factors for all detectors for field sizes of 3.5 to 5 cm2 are not specified. We contribute to the literature with the results we found. Correction factors specified in TRS-483 [4] are specified as a single value independent of the Linac device model, collimation system (MLC or jaw), or filter. Again, correction factors for the 0.5, 3.5, and 5.0 cm2 field sizes and the PTW 31022 Pinpoint 3D ion chamber are not specified in the TRS-483 [4]. The study by Dufreneix et al. [25] shows that correcting the output factors by taking the correction factors suggested in TRS-483 [4] as reference data and uploading the beam modeling data to TPS in this way will have serious consequences in clinical practice [25].

The differences that were observed in some of our results that are different from the literature could be the use of experimental methods in our study and some possible errors due to the detection of effective point or alignment of detectors may reflect errors in our results. A large multicentre study done by Dufreneix et al. [25] confirms this observation. Still, the EBT3 Gafchromic film dosimeter requires high sensitivity and accuracy. Calibration and dose film measurements and the film readings we performed with ImageJ software may have been reflected in the output correction factors as uncertainty. Film readings and ROI selections with ImageJ are subject to error as they depend on the user performing the readings. For this purpose, we have observed that using commercial software developed for this purpose instead of using an open-source resource such as ImageJ for relative dosimeter measurement will reduce the error in output correction factors.

According to the findings of Bozidar Casar et al.'s study, a 0.1 percent volume correction should be applied to Elekta Versa HD readings taken with EBT3 film. Because volume correction was not included in our study, the output correction factors we discovered may have a margin of error of 0.1 percent [8, 21].

CONCLUSION

When comparing the correction factors we obtained as a result of our study with the correction factors suggested in TRS-483 [4], it was observed that the percent value deviation for small fields (0.5 and 0.8 cm2) is above 3.5%. As a result of our study; instead of calculating the output factors by using the correction factors suggested in TRS-483, each clinic's calculation of the detector-specific correction factor for their device and calibration conditions (SSD: 90 and depth: 10 cm) for the detector available in their clinic will reduce the error encountered in small field dosimetry, and it has been observed that it will be the most correct approach.

Scp readings are used to calculate treatment MU values. Dose calculations will be erroneous in treatment plans when small fields such as SRS, SBRT, and SRT are commonly used if output correction factors are incorrectly or incompletely supplied owing to detector differences. Although dosimetry systems appropriate for small field output measurements are used, it is beyond the scope of this study to indicate which detector is best suited for small field output measurements. More research is needed to support the output correction factor calculation of these detectors and our findings.

Conflict of Interest

There are no conflicts of interest and no acknowledgements.

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