sciendo
Polish Journal of Medical Physics and Engineering
The Journal of Polish Society of Medical Physics
ISSN 1898-0309, doi: 10.2478/pjmpe-2024-0018
September 2024
Vol 30, Issue 3

Scientific Paper

Evaluation of the usefulness of small detectors made of Gafchromic EBT3 film for measurements in areas with high dose gradient and without charged-particle equilibrium (CPE)

Paweł WOŁOWIEC [1,ABCDE], Paweł Franciszek KUKOŁOWICZ [2,ACDEF,*], Krzysztof BULIŃSKI [1,B]

[1]*Medical Physics Department, Holycross Cancer Center, Kielce, Poland*

[2]*Medical Physics Department, Maria Sklodowska-Curie National Research Institute of Oncology, Warsaw, Poland*

[*]*Corresponding author: Prof. Paweł Franciszek Kukołowicz; pawel.kukolowicz@nio.gov.pl*

(received 12 February 2024; accepted 17 June 2024)

Abstract

Introduction: In some clinical cases a full therapeutic dose needs to be delivered in the area close to the skin surface where a high dose gradient and there no charged-particle equilibrium (CPE) exists. The accuracy of dose distribution calculations performed in this region with the treatment planning system is limited. In this work we investigated the usefulness of small pieces of Gafchromic EBT3 film for measurements of the absolute dose value in the area close to the skin surface.

Material and methods: The Gafchromic EBT3 film detectors of size 1.0 cm x 1.5 cm were prepared. The film samples were calibrated in 6 MV photon beam (Elekta Versa HD). Calibration was performed in a dose range of 0 – 250 cGy. Films were scanned using the EPSON EXPRESSION 10000 XL flatbed scanner in 48-bit RGB mode, with a resolution of 72 dpi. ImageJ software was used to calculate the dose. Triple-channel film dosimetry was applied. The uncertainty of the dose measurement method was estimated. Film measurements were compared with the dose measurements using ionization chamber. The conformity of measurements was assessed using the metrological compliance test.

Results: The relative differences between dose measurements using Gafchromic EBT3 film detectors and ionization chamber for a single square photon beam were -0.8% and 0.3% for a depth of 0.5 cm and 5.0 cm (CPE) respectively. The values of the metrological compatibility test factor ζ were 0.3 and 0.1 respectively. The maximum relative differences for dynamic beams were 0.9% and -1.0% for a depth of 0.5 cm and 5.0 cm respectively. Metrological compatibility test showed also good agreement (ζ=0.3).

Conclusions: Small film detectors made of Gafchromic EBT3 film allow for the accurate dose measurements in the high dose gradient region and without CPE. They can be used to validate the calculation of the treatment planning systems also for VMAT techniques.

Keywords: Gafchromic films; build-up region measurements; accuracy; verification of treatment planning system calculations.

1. Introduction

Verification of the dose distributions computations accuracy performed with the use of computer treatment planning systems is an essential element of acceptance tests and approval procedures before clinical application.[1] With the development of treatment techniques it is necessary to implement new verification methods. It is a challenging task in case of dynamic treatment techniques such as volumetric modulated arc therapy (VMAT), especially when the accuracy of dose calculations needs to be determined in a dose build-up regions where charged-particle equilibrium (CPE) doesn't exist and where there is a high dose gradient.[2,3] In most clinical cases the accuracy of the calculations in dose build-up regions is not of great clinical importance. However, for patients irradiated in the head and neck area, for some of them it is necessary to deliver full therapeutic dose in the area close to the skin surface, In such cases, knowledge about the accuracy of dose calculations in this area plays an important role. Useful detector for the verification of dose distribution calculation in the area close to the patient surface should be small, it should be possible to place it at any depth, also in the buildup region. Moreover, it should be possible to measure the absolute dose and it should have a high spatial resolution. A promising detector for this task is the Gafchromic EBT3 film.[4]

Authors' contribution:
A – Research concept and design, B – Collection and/or assembly of data, C – Data analysis and interpretation, D – Writing the article, E – Critical revision of the article, F – Final approval of the article.

So far, Gachromic films have been used mainly for the verification of the IMRT and VMAT treatment plans before beginning of patient's irradiation.[5] The usefulness of films for quality control of tomotherapy treatment plans has been demonstrated,[6,7] as well as for dosimetry and quality assurance in stereotactic radiosurgery (SRS) and stereotactic body radiation therapy (SBRT) performed with medical linear accelerators. Gafchromic films have also been used for measurements of patients skin dose resulting from the kilovoltage imaging systems (kV CBCT) mounted on the medical linear accelerators.[8] Gafchromic films are also used in interventional radiology.[9]

In this work, we investigated the usefulness of small detectors made of Gafchromic EBT3 film for measurements of the absolute dose value in areas without CPE and in areas of high dose gradient.

2. Materials and methods

2.1. Gafchromic EBT3 radiochromic films

The films consist of 28 μm thick active layer, coated between two layers of matte-polyester substrates with a thickness of 125 μm. The films are made of materials equivalent to soft tissue and have a response that is nearly independent of the electromagnetic radiation energy in the rage of 100 keV to 18 MeV (<5% difference in optical density change).[10] Their dynamic dose range is from 0.1 to 20 Gy, while the optimal dose range declared by the manufacturer is 0.2 to 10 Gy. Film response is practically independent of the ambient temperature and atmospheric pressure, as well as the angle of incidence of the radiation beam.

Under the influence of ionizing radiation, the active component of the film polymerizes and turn blue with an absorption maximum of approximately 633 nm. Application of a marker dye in the active layer of the film allows for the correction of film heterogeneity by means of multichannel dosimetry and reduces the film sensitivity to visible light (especially UV radiation). Gachromic EBT3 films allow for fractionated dose measurements. Films can be cut to any shape. The great advantage of Gafchromic EBT3 films is that they are very thin (thickness ~278 μm) which results in negligible attenuation of the megavoltage ionizing radiation beam, and high spatial resolution of the measurement in the direction perpendicular to the film.

The small film detectors made of 8" × 10" (20.3 cm × 25.4 cm) Gafchromic EBT3 film sheets for dose measurements in the dose buildup region were investigated. Film sheets were cut into small samples of 1.0 cm × 1.5 cm. Such dimensions were chosen in order to avoid the influence of delamination of the film at the edges during cutting on the measurement result. That size enabled placing a square region of interest (ROI) of 10 pixels × 10 pixels in the center of a sample with sufficient dimensions for the measurement, and large enough to mark the sample and its orientation.[11] To avoid the flatbed CCD scanner lamp effects (film orientation) which affects the readout result, the rectangular shape of the film samples was chosen.

2.2. Film calibration

2.2.1. Film irradiation

Film calibration was performed in a solid slab phantom (RW3 by PTW-Freiburg), made of polystyrene (C_8H_8) containing 2% (±0.4%) of titanium dioxide (TiO_2). The physical density of the RW3 plates is 1.045 g/cm^3, and their electron density is 1.012 times higher than electron density of water. Gafchromic EBT3 film samples were placed in the phantom, in the central axis of the beam, at a 5 cm depth, in a radiation field of 10 cm x 10 cm. The distance from the radiation source to the surface of the RW3 plates was 90 cm. The films were irradiated with a 6 MV photon beam generated by the Elekta Versa HD medical linear accelerator. To monitor accelerator dose output and to measure the dose delivered to film samples, a reference 0.6 ccm Farmer type ionization chamber (model 30013 by PTW-Freiburg) was additionally placed at a depth of 10 cm. The dose was read on a UNIDOS electrometer, model 10001, by PTW-Freiburg. The dose to the film samples placed at a depth of 0.5 cm and 5.0 cm was calculated from the dose measured by an ionization chamber at a depth of 10 cm. The conversion was performed on the basis of previously experimentally determined coefficients. The coefficients were determined as the quotient of the ionizing chamber signal at a given depth (0.5 cm or 5.0 cm) to the signal measured at a depth of 10 cm (for the same measurement geometry and the same environmental conditions). The influence of the presence of film samples above the ionization chamber on the measurement result with the ionization chamber was also checked and taken into account. Ionization chamber dose measurements were performed according to the IAEA TRS 398 report.[12]

Calibration was performed in the range of fractional doses used in conventional teleradiotherapy. The calibration points were selected based on the recommendations of the manufacturer, that is 6 – 8 points for a typical situation (including unexposed film) with doses in geometric progression. A total of 8 points were obtained on the calibration curve. Film samples were irradiated with doses of 0, 10, 25, 50, 100, 150, 200 and 250 cGy. The actual dose output of the accelerator was taken into account during all measurements.

To diminish uncertainty of the film readout, six film samples were irradiated simultaneously for one point of the calibration curve.[13] The films were placed in the center of the square field of 10 cm side and arranged so as to form a square with a side of 3 cm. The maximum difference of the dose values in the area where the irradiated film samples were located was 0.3%.

The darkening of the films was read at least 24 hours after exposure to irradiation. The time was chosen so that the polymerization process taking place in the active layer of the

film, induced by ionizing radiation, was completely finished. Thanks to this, the influence of errors related to the dynamics of the polymerization process can be neglected.[14]

The films handling procedure followed the recommendations of the AAPM TG-55 report.[15] All films were stored in the same place, under the same conditions, in order to limit the influence of storage conditions on the final measurement result. Films were handled in such a way as to minimize the exposure of samples to visible/ultraviolet light.

2.2.2. Film scanning procedure

The film samples were scanned in transmission mode using the EPSON EXPRESSION 10000 XL flatbed scanner, recommended by the Gafchromic EBT3 films manufacturer. The original EPSON Scan software provided with the scanner was used for scanning. All image processing options, color corrections and all available filters in the software have been turned off in order to get raw images. Scanning was performed in transmission mode, in 48-bit RGB color mode (16-bit per color), with a resolution of 72 dpi. Scanned images were saved as TIFF (tagged image file format).

The scanner was always turned on 30 minutes before the beginning of films scanning. Then, five blank scans were performed in order to stabilize the working conditions of the scanner lamp, to minimize the effect of temperature changes on the scanner performance.[16]

In order to minimize the influence of scanner's inhomogeneity on the measurement result and avoid the need of applying corrections, the films were placed in the center of the scanning area, along the direction of the lamp movement, always in the same place, using a previously prepared template.[17] The film samples were placed in a horizontal orientation. The long side of the film was set perpendicular to the direction of the lamp's movement (landscape orientation).

The analysis of faulty pixels was also performed for the area of the scanner table where the films were placed. For this purpose, five consecutive blank scans were performed. On the obtained images, in the area where the film samples were placed, a histogram of the pixel values was made. Defective pixels have been recognized as pixels whose value differs by more than two standard deviations from the mean value. The obtained average number of faulty pixels from all images was 0.1%. This proved that there were no artifacts in this area that could significantly affect the measurement result and that the scanner is operating properly.

2.2.3. Analysis of film images

ImageJ software[18] used to analyze the obtained images of scanned film samples using scripts written by the first author of this work. Each scanned image was divided into three separate red (R), green (G), and blue (B) color channels, each with 16-bit depth. In the center of each film sample, separately on each color channel, a region of interest (ROI) of 10 pixels × 10 pixels (3.5 mm × 3.5 mm) was placed, from which the readout was made. The area of interest was placed at least 3 mm from the edge of the sample to avoid edge artifacts resulting from cutting the film sheet.[19]

In order to remove noise, before readout, each individual color channel of the obtained images was separately processed before readout using a median filter with diameter of 2 pixels.[13]

The conversion of the pixel values read from the three color channels (R, G, B) into the dose delivered to the film sample was carried out with the use of experimentally determined calibration curves, using a hand-written script in the ImageJ software.

2.2.4. Determination of the calibration curves

The calibration procedure was based on the methodology described in other studies.[14] The applied multichannel procedure allowed to reduce the influence of dose-independent factors on the signal, such as non-uniform thickens of the active film layer, inhomogeneity or artifacts of the scanner.[20] Detailed description of the calibration procedure is given in **Appendix 1**. The calibration curves are presented in **Figure 1**.

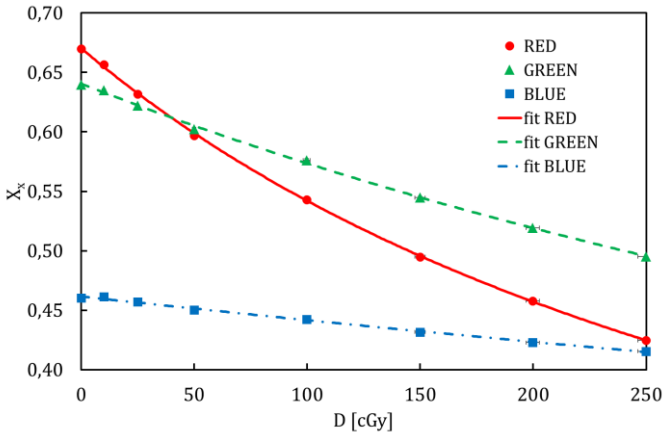

Figure 1. Calibration curves for small samples of Gafchromic EBT2 film, for three color channels, in the dose range including fractional dose.

2.3. Dose calculation procedure – triple-channel film dosimetry

Calculation of the doses delivered to the film samples was done by means of self-written script in ImageJ software.[18] The procedsure was based on methodology described by Andre Micke.[14,20] A detailed description of the dose calculation procedure is enclosed in **Appendix 2**. The script was written in order to find the minimum of the function Ω (**Equation 1**), separately for each pixel on the image. The Ω equation contains the sum of the squared noise differences (Δd_x, where x = R, G, B) defined as the quotient of the scanned film optical density $d_{x,scan}(D)$ and the film optical density calculated from the calibration curve ($d_{x,cal}(D)$). The optical density values $d_{x,scan}(D)$ were calculated based on the pixel values (PV_x) read from the images of the scanned film samples (**Equation 5**).

The optical density values $d_{x,cal}(D)$ were calculated based on the previously determined calibration curves (**Equation 6**).

$$\Omega(D) = (\Delta d_R - \Delta d_B)^2 + (\Delta d_B - \Delta d_G)^2 + (\Delta d_G - \Delta d_R)^2 \rightarrow min \quad \text{Eq. 1}$$

The dose was read from a 10 pixels x 10 pixels ROI placed in the center of the film sample image. **Figure 2** shows the solution of the Ω equation as a function of dose, for film pieces irradiated with doses of 50 cGy and 150 cGy.

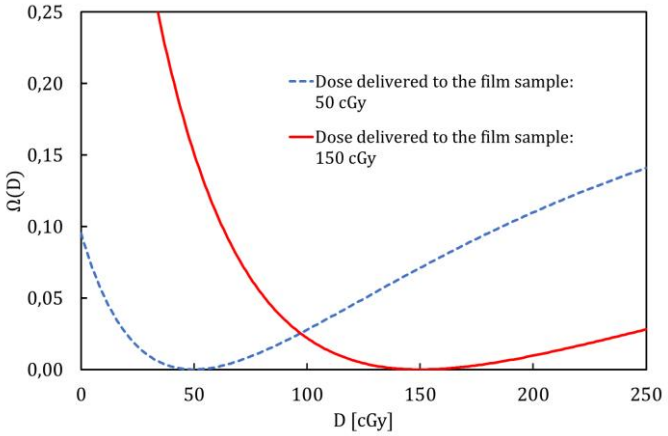

Figure 2. The results of calculations of the equation Ω as a function of dose, fort two film samples irradiated separately with the dose of 50 cGy and 150 cGy.

2.4. Standard uncertainties of measurements

The uncertainty of the dose measurement using small samples of the Gafchromic EBT3 film was determined based on the JCGM 100:2008 report entitled „Evaluation of measurement data - Guide to the expression of uncertainty in measurement".[21] Since the measured dose value is calculated with the application of numerical methods, using self-written algorithm, its standard uncertainty was assessed by the type B method. For this purpose, film samples utilized to obtain calibration curves were employed (42 samples in total). Based on the calibration films images, the dose values were calculated using the methodology described in section 2.3. The obtained values were compared with the values of the reference doses measured with the ionization chamber during calibration. The combined standard uncertainty of the dose measured using ionization chamber during calibration was calculated based on the JCGM 100:2008 report[21] and the IAEA-TECDOC-1585 report[22] and is 1.5%.

Based on the differences between the calculated doses and actual doses delivered to the film samples, the mean value of these differences and the standard deviation of a single measurement were calculated. The mean difference between the dose measured by the ionization chamber and the dose measured with small film detectors was -0.2%. Standard deviation of a single measurement was 2.9%. This value was taken as the total relative standard uncertainty of the dose measured by means of small film samples, using the method presented in this article. This value is close to standard uncertainties of the dose measurement using conventional in vivo dosimetry system.[23]

2.5. The accuracy assessment of measurements using small film detectors

In order to validate the described method of the dose measurement with the use of small Gafchromic EBT3 film detectors, measurements were made in the RW3 slab phantom, for a single square photon beam, at a depth of 5 cm – in the area of charged-particle equilibrium (CPE). Then, in the same phantom, for the same photon beam, measurements were made in the buildup area at a depth of 0.5 cm (high dose gradient) where the charged-particle equilibrium didn't exist. Finally, measurements were made for the beam shaped by the dynamic movement of the collimator leaves, also at two depths of 0.5 cm and 5.0 cm (a high dose gradient and no CPE at both depths).

2.5.1 Measurements in the RW3 slab phantom for a single square photon beam

Measurements were made using small film samples, in simple geometry, for a single square photon beam, in the RW3 slab phantom, at two depths. One depth was in the area of slow dose decrease (where there is CPE) and another one was in the buildup region (CPE doesn't exist). The films were irradiated on the Elekta Versa HD medical linear accelerator, placed between the RW3 plates at a depth of 5.0 cm (CPE) and 0.5 cm (no CPE) – separate measurements. For this purpose, a perpendicular 6 MV photon beam, with dimensions of 10 cm × 10 cm was used. The distance between the ionizing radiation source and the surface of the phantom (SSD) was 90 cm. The appropriate number of monitor units (MU) was applied to deliver a dose of 200 cGy at a depth of 5.0 cm. The accelerator dose output monitoring and the actual dose delivered to film samples measurements (D_{ic}) were handled in the same way as in section 2.2.1.

Finally, the dose measured with the use of small samples of Gafchromic EBT3 films and the dose calculated based on the ionization chamber measurement were compared, independently for the two depths. The results are shown in **Table 1**.

Table 1. The results of the comparison between the dose measurements using Gafchromic EBT3 small film detectors (D_{film}) against the doses calculated based on ionization chamber measurements (D_{ic}), at a depth of 0.5 cm and 5.0 cm, in a solid slab RW3 phantom (PTW-Freiburg), in conditions similar to the films calibration conditions.

d [cm]	D_{ic} [cGy]	$u(D_{ic})$ [cGy]	$u_r(D_{ic})$	D_{film} [cGy]	$u(D_{film})$ [cGy]	$u_r(D_{film})$	$D_{film}-D_{ic}$ [cGy]	$(D_{film}/D_{ic}-1)$	$\zeta(D_{film,p}-D_{ic})$
0.5	198.9	3.1	1.5%	197.2	5.7	2.9%	-1.7	-0.8%	0.3
5.0	201.1	3.1	1.5%	201.7	5.8	2.9%	0.5	0.3%	0.1

Table 2. The results of the comparison between the dose measurements using Gafchromic EBT3 small film samples (D_{film}) against the doses calculated based on the ionization chamber measurement (D_{ic}), at the depth of 0.5 and 5.0 cm, in the solid slab RW3 phantom (PTW-Freiburg), for the VMAT treatment plan reduced to 0° gantry angle.

d [cm]	D_{ic} [cGy]	$u(D_{ic})$ [cGy]	$u_r(D_{ic})$	D_{film} [cGy]	$u(D_{film})$ [cGy]	$u_r(D_{film})$	$D_{film}-D_{ic}$ [cGy]	$(D_{film}/D_{ic}-1)$	$\zeta(D_{film,p}-D_{ic})$
0.5	240.9	3.7	1.5%	243.1	7.0	2.9%	2.2	0.9%	0.3
5.0	239.2	3.7	1.5%	236.8	6.9	2.9%	-2.4	-1.0%	0.3

2.5.2. Measurements in the RW3 slab phantom for dynamic beams

To create more complex measurement conditions a treatment plan made in volumetric modulated arc therapy technique (VMAT) was utilized. In VMAT technique dose distribution modulation takes place, among other, through the dynamic movement of the leaves of the multi-leaf collimator, variable dose rate and variable rotation speed of the accelerator gantry. The VMAT technique was used to obtain a dose distribution with areas of high dose gradient and without CPE. The treatment plan in the VMAT technique was carried out in the Philips Pinnacle treatment planning system for the anthropomorphic phantom of the head and neck. The plan consisted of three arcs (6 MV photon beams) which were to deliver a dose of 200 cGy to the isocenter. Then the plan was transferred to the CT of the RW3 homogeneous slab phantom in the treatment planning system (as in section 2.5.1) and was recalculated. All arcs were brought to 0° gantry angle, maintaining the dynamic fields shaped by the multi-leaf collimator (**Figure 3**). The dose at the isocenter for the RW3 slab phantom was 251 cGy. Measurements were made for one treatment fraction. Irradiation and measurements were done in the same way as in the section 2.5.1. The only difference was that a Markus flat-parallel ionization chamber, model 23343, from PTW-Freiburg, was used to monitor a dose output of the accelerator and to measure the actual dose delivered to the film samples. The results are shown in **Table 2**.

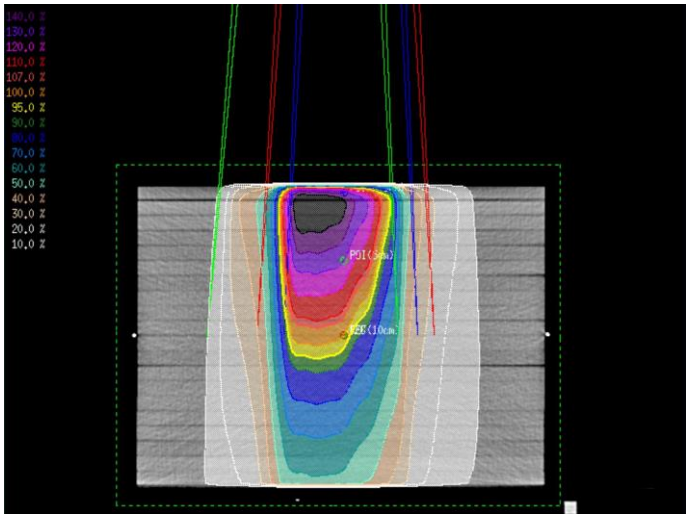

Figure 3. Dose distribution of the VMAT treatment plan consisting of three arcs (6 MV photon beams) brought to 0° gantry angle, calculated in the RW3 slab phantom (PTW-Freiburg) in the Pinnacle, Philips treatment planning system.

2.6. Conformity assessment of the results

The conformity of measurements was assessed using the metrological compliance test. For two uncorrelated variables, the quantity ζ was determined[24]:

$$\zeta(x_1 - x_2) = \frac{|x_1 - x_2|}{\sqrt{u^2(x_1) + u^2(x_2)}} \leq 2 \qquad \text{Eq. 2}$$

where x_1, x_2 are the values of the variables, and $u(x_1)$, $u(x_2)$ are their standard uncertainties.

In order to conclude that two variables are compatible (at 95% confidence level), the absolute difference in their values divided by the root of the sum of the squares of their standard uncertainties should be less than or equal to 2.

In this work, for each pair of two compared quantities, the variables x_i (i = 1, 2) will be the dose measured using Gafchromic EBT3 film small samples (D_{film}) and the dose calculated based on the ionization chamber measurement (D_{ic}). Their standard uncertainties $u(x_i)$ will be combined standard uncertainties calculated according to the methodology given in the JCGM 200:100 report (description in section 2.4).

3. Results

3.1. Results of the measurements in the RW3 slab phantom for a single square photon beam

Table 1 shows the results of dose measurements using Gafchromic EBT3 film samples made in RW3 slab phantom (D_{film}) and the dose obtained based on the ionization chamber measurement (D_{ic}) for a single square photon beam (section 2.5.1). The table presents the dose values D_i (i = film; ic), their combined standard uncertainties $u(D_i)$ (calculated according to section 2.4 and the JCGM 100:2008 report[21]), as well as the relative uncertainties $u_r(D_i)$ expressed in percentages. For the compared dose values, their absolute differences, relative differences and ζ factor of the metrological compatibility test (**Equation 2**, section 2.4) were calculated. Measurements were made at depth of 0.5 and 5.0 cm.

3.2. Results of the measurements in the RW3 slab phantom for dynamic beams

Table 2 shows the comparison results of dose measurements using Gafchromic EBT3 small film samples (D_{film}) against the dose obtained from the ionization chamber measurement (D_{ic}). Measurements were made in the RW3 slab phantom (section 2.5.2) for the treatment plan prepared in the VMAT technique (three arcs reduced to the 0° gantry angle). The table present the dose values D_i (i = film; ic), their combined standard uncertainties $u(D_i)$ (calculated according to section 2.4 and JCGM 100:2008 report[21]), as well as relative uncertainties $u_r(D_i)$ expressed in percentages. For the compared dose values, their absolute differences, relative differences and the ζ factor of the metrological compatibility test were calculated (**Equation 2**, section 2.6).

4. Discussion

4.1. Calibration curves

The analysis of the calibration curves for the three color channels shown in **Figure 1** allows to conclude that the rational function represented by the **Equation 4** describes the measurement results very well. The R^2 parameter was 1.000, 0.999, 0.996 for the red, green and blue channels respectively. **Figure 1** also show that film response expressed as the normalized pixel value (X_x) is not linear, and that the slope of the curves for all channels decreases.

4.2. Measurements in RW3 slab phantom – comparison with an ionization chamber

Measurements carried out with the Gafchromic EBT3 small film samples, made in the RW3 slab phantom (PTW-Freibrug), proved that the method presented in this study is sufficiently accurate to be applied in practice. This is shown by the comparison between the dose measurements using Gafchromic EBT3 small film detectors (D_{film}) against the doses calculated based on the ionization chamber measurement (D_{ic} – the actual dose that films were irradiated) presented in **Table 1**. The maximum difference was -0.8% and corresponds with the measurement made at 0.5 cm depth (dose buildup area, no CPE). The difference at a depth of 5.0 cm was 0.3%. The values of the metrological compatibility test factor ζ were 0.3 and 0.1 for a depth of 0.5 cm and 5.0 cm respectively. The results in **Table 1** show that small Gafchromic EBT3 film detectors allow for reliable dose measurement at various depths, also in the area without charged-particle equilibrium.

The comparison results for a more complex situation are presented in **Table 2**. There are the results of measurements and calculations for the treatment plan prepared in the VMAT technique, consisting of three arcs of 6 MV photon beams, transferred to the RW3 slab phantom and reduced to the 0° gantry angle (section 2.5.2). The total dose distribution was obtained by the continuous, dynamic movement of the collimator leaves. The dose measured using small film detectors (D_{film}) was compared to the dose measured with the ionization chamber (D_{ic}). The absolute dose value differences at both depths are at a similar level (0.5 cm: 0.9%, 5.0 cm: -1.0%) and agree well with the ionization chamber measurements (differences 2.2 cGy and -2.4 cGy) – as it was in the case of a single photon beam. However, it should be noted that neither the films nor the ionization chamber were in the area of the uniform dose. The isodose distribution in **Figure 3** shows that in the area where the detectors were placed (depths of 0.5 cm and 5.0 cm – film samples, 10 cm – ionization chamber), there was a dose gradient which is the sum of multiple dose distributions for multiple irradiation segments. Despite this, the metrological compatibility factor showed good agreement between the results ($\zeta = 0.3$).

5. Conclusions

More and more complex techniques of photon beam radiotherapy are used in cases where irradiation structures are located at very small depths. In such situations, it is very important to know the accuracy of calculation algorithms in treatment planning systems, especially in the areas of high dose gradient and without charged-particle equilibrium. For this reason, appropriate tools are necessary to determine the accuracy of radiotherapy, regardless of the treatment technique or measurement conditions. The results obtained by the authors indicate that small film detectors made of Gafchromic EBT3 film can be good detectors for this purpose. They are very thin and practically do not attenuate the photon beam, they can be cut to any shape and have no energy dependence. They enable direct dose measurement in the areas where they are placed. They allow for accurate dose measurements in the area of high dose gradient and without CPE. They can also be used to validate the calculation of the treatment planning systems, especially for complex irradiation techniques (such as VMAT) and in conditions where the use of others commercially available detectors would not be possible.

References

1. Miften M, Olch A, Mihailidis D, et al. Tolerance limits and methodologies for<scp>IMRT</scp>measurement-based verification QA Recommendations of AAPM Task Group No. 218. Medical Physics. 2018;45(4). doi:10.1002/mp.12810
2. Oinam AS, Singh L. Verification of IMRT dose calculations using AAA and PBC algorithms in dose buildup regions. J Applied Clin Med Phys. 2010;11(4):105-121. doi:10.1120/jacmp.v11i4.3351
3. Wang L, Cmelak AJ, Ding GX. A simple technique to improve calculated skin dose accuracy in a commercial treatment planning system. J Applied Clin Med Phys. 2018;19(2):191-197. doi:10.1002/acm2.12275
4. Cao Y, Yang X, Yang Z, et al. Superficial dose evaluation of four dose calculation algorithms. Radiation Physics and Chemistry. 2017;137:23-28. doi:10.1016/j.radphyschem.2016.02.032
5. Bennie N, Metcalfe P. Practical IMRT QA dosimetry using Gafchromic film: a quick start guide. Australas Phys Eng Sci Med. 2016;39(2):533-545. doi:10.1007/s13246-016-0443-0
6. Borca VC, Pasquino M, Russo G, et al. Dosimetric characterization and use of GAFCHROMIC EBT3 film for IMRT dose verification. J Applied Clin Med Phys. 2013;14(2):158-171. doi:10.1120/jacmp.v14i2.4111
7. Marrazzo L, Zani M, Pallotta S, et al. GafChromic® EBT3 films for patient specific IMRT QA using a multichannel approach. Physica Medica. 2015;31(8):1035-1042. doi:10.1016/j.ejmp.2015.08.010
8. Aland T, Moylan R, Kairn T, Trapp J. Effect of verification imaging on in vivo dosimetry results using Gafchromic EBT3 film. Physica Medica. 2016;32(11):1461-1465. doi:10.1016/j.ejmp.2016.10.020
9. Kidoń J, Polaczek-Grelik K, Gołba K, Wojakowski W, Ochała A. Calibration of Gafchromic XR-RV3 film under interventional radiology conditions. Polish Journal of Medical Physics and Engineering. 2021;27(2):165-173. doi:10.2478/pjmpe-2021-0020
10. Chełmiński K, Bulski W, Georg D, et al. Energy dependence of radiochromic dosimetry films for use in radiotherapy verification. Reports of Practical Oncology & Radiotherapy. 2010;15(2):40-46. doi:10.1016/j.rpor.2010.02.003
11. GAFChromic™ EBT3 film specifications, http://www.gafchromic.com/documents/EBT3_Specifications.pdf
12. IAEA 2006 Technical Reports Series No. 398, Absorbed Dose Determination In External Beam Radiotherapy (Vienna: International Atomic Energy Agency).
13. Wołowiec P, Kukołowicz PF. The analysis of the measurement uncertainty with application of small detectors made of Gafchromic EBT films for the range of doses typical for in vivo dosimetry in teleradiotherapy. Radiation Measurements. 2016;92:72-79. doi:10.1016/j.radmeas.2016.08.001
14. Lewis D, Micke A, Yu X, Chan MF. An efficient protocol for radiochromic film dosimetry combining calibration and measurement in a single scan. Medical Physics. 2012;39(10):6339-6350. doi:10.1118/1.4754797
15. Niroomand-Rad A, Blackwell CR, Coursey BM, et al. Radiochromic film dosimetry: Recommendations of AAPM Radiation Therapy Committee Task Group 55. Medical Physics. 1998;25(11):2093-2115. doi:10.1118/1.598407
16. Matney JE, Parker BC, Neck DW, Henkelmann G, Rosen II. Evaluation of a commercial flatbed document scanner and radiographic film scanner for radiochromic EBT film dosimetry. J Applied Clin Med Phys. 2010;11(2):198-208. https://doi.org/10.1120/jacmp.v11i2.3165
17. Devic S, Seuntjens J, Sham E, et al. Precise radiochromic film dosimetry using a flat-bed document scanner. Med Phys. 2005;32(7Part1):2245-2253. doi:10.1118/1.1929253
18. Schneider CA, Rasband WS, Eliceiri KW. NIH Image to ImageJ: 25 years of image analysis. Nat Methods. 2012;9(7):671-675. https://doi.org/10.1038/nmeth.2089
19. Dąbrowski R, Drozdyk I, Kukołowicz P. High accuracy dosimetry with small pieces of Gafchromic films. Reports of Practical Oncology & Radiotherapy. 2018;23(2):114-120. doi:10.1016/j.rpor.2018.01.001
20. Micke A, Lewis DF, Yu X. Multichannel film dosimetry with nonuniformity correction. Med Phys. 2011;38(5):2523-2534. doi:10.1118/1.3576105
21. JCGM 100:2008, Evaluation of measurement, data—Guide to the expression of uncertainty in measurement, https://www.bipm.org/documents/20126/2071204/JCGM_100_2008_E.pdf
22. IAEA-TECDOC-1585, Measurement Uncertainty, 2008, https://www-pub.iaea.org/MTCD/Publications/PDF/te_1585_web.pdf.
23. Essers M, Mijnheer B. In vivo dosimetry during external photon beam radiotherapy. International Journal of Radiation Oncology*Biology*Physics. 1999;43(2):245-259. doi:10.1016/s0360-3016(98)00341-1
24. Gromczak K, Ostrowska K, Gąska A, Sładek J. The universal validation algorithm of coordinate measuring methods. Measurement Automation Monitoring. 2016;62(4):120-124.

Appendix 1. Calibration curves determination

From region of interest (ROI) of the size 10 pixels × 10 pixels, placed in the center of the irradiated film samples images, the mean pixel value MPVx and the standard deviation SDx were read (where x corresponds to the R, G, B color channel). The read mean pixel values were normalized to the maximum possible pixel value for a 16-bit image (i.e. 2^{16} = 65536), thus obtaining normalized pixel values X_x, separately for each of the color channel.

$$X_x = \frac{MPV_x}{2^{16}} \quad \text{Eq. 3}$$

The film's response to ionizing radiation is well described by rational functions. Therefore, calibration data was fitted using the following formula:

$$X_x(D) = \frac{a_x + b_x \cdot D}{c_x + D} \quad \text{Eq. 4}$$

where $X_x(D)$ is the normalized pixel value for the color channel x (x = R, G, B), D is the dose delivered to the film sample, a_x, b_x i c_x are the calibration curve fitting parameters. Thus, three calibration curves (red, green and blue channel) were obtained.

Figure 1 shows the calibration curves for the individual red, green and blue color channels. They are represented as normalized pixel values of the color channels (X_x) as a function of the dose delivered to the film (D). The calibration curves shown were determined in a dose range of 0 to 250 cGy, covering the doses received by the patient during a single treatment session.

The points in the **Figure 1** corresponds to the normalized pixel values (X_x) calculated according to the **Equation 3**, based on the mean pixel values (MPV_x) read from the calibration films irradiated with the specified doses (D). The points are the values obtained during the calibration measurements. The standard uncertainties are not visible in the graph due to the fact that they are small and overlap with symbols. The lines show fitting curves calculated according to the **Equation 4**. The fitting accuracy described by the R^2 parameter was 1.000, 0.999, 0.996 for the red, green and blue channel, respectively.

The pixel values read from the film images can be converted to the optical density using the formula:

$$d_{x,scan} = -log\left(\frac{MPV_x}{2^{16}}\right) = -log(X_x) \quad \text{Eq. 5}$$

where $d_{x,scan}$ is the optical density of the film for the channel x (x = R, G, B). In this way, calibration data were obtained in the form of doses delivered to the film samples (D) and optical density values for individual color channel ($d_{x,scan}$), which were calculated based on the normalized pixel values (X_x) obtained from the images of the irradiated film detectors.

The optical density values for a given color channel can also be calculated based on the obtained calibration curves:

$$d_{x,cal}(D) = -\log\left(\frac{a_x + b_x \cdot D}{c_x + D}\right) \quad \text{Eq. 6}$$

Appendix 2. Dose calculation procedure – triple-channel film dosimetry

The triple-channel method of dose calculation procedure in this article, using the three image channels of the film, was based on the methodology described by Andre Micke.[14,20]

The calibration was done so that it reflects the average response of the scanner-film system to ionizing radiation. The measured (calculated) optical density of the film used for the measurement consists of the dose-dependent part of the signal represented by the calibration function and the part of the signal originating from other factors which cause disturbances, such as e.g. different thickness of the active film layer, scanner inhomogeneity or noise. This can be represented by the formula:

$$d_{scan}(D) = d_{cal}(D) \cdot \Delta d \quad \text{Eq .7}$$

where $d_{scan}(D)$ is the scanned optical density of the film, $d_{cal}(D)$ is the optical density of the film during calibration, D is the dose delivered to the film and Δd are disturbances.

Therefore, for each color channel (x = R, G, B) it can be written:

$$\Delta d_x = \frac{d_{x,scan}(D)}{d_{x,cal}(D)} \quad \text{Eq. 8}$$

Disturbances Δd_x are not dose dependent and they are independent on the color channel (i.e. the wavelength of scanner lamp light). Otherwise, Δd_x should be the same for each color channel. Therefore, finding the minimum of the equation (9) is an equivalent to finding dose value closes to the dose actually delivered to the film.

$$\Omega(D) = (\Delta d_R - \Delta d_B)^2 + (\Delta d_B - \Delta d_G)^2 + (\Delta d_G - \Delta d_R)^2 \rightarrow min \quad \text{Eq. 9}$$