Physica Medica 64 (2019) 40–44

Contents lists available at ScienceDirect

Physica Medica

journal homepage: www.elsevier.com/locate/ejmp

Original paper

Dose measurements nearby low energy electronic brachytherapy sources using radiochromic film

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ARTICLE INFO

Keywords:
Electronic brachytherapy
Radiochromic film dosimetry

ABSTRACT

Purpose: We investigate the effect of the GafChromic™ film EBT3 model absorbed dose energy response when used for dose measurements around low-energy photon sources. Monte Carlo based correction procedure in synergy with appropriate calibration curves was shown to provide more accurate absorbed dose (either relative or absolute). An assessment was made of possible dose errors that might be encountered if such energy dependent response is ignored.

Methods: We measured PDDs in water from a Xoft 50 kVp source using EBT3 film, and compared to PDD measurements acquired with a PTW-TN34013 parallel-plate ionization chamber. For the x-ray source, we simulated spectra using the EGSnrc (BEAMnrc) Monte Carlo code, and calculated Half Value Layer (HVL) at different distances from the source in water. Measurement strips of EBT3 film were positioned at distances of 2–6 cm from the Xoft source in a water phantom using a custom-made holder and irradiated simultaneously.

Results: Our results show that film calibration curves obtained at beam qualities near the effective energy of the Xoft 50 kVp source in water lead to variation in absorbed dose energy dependence of the response of around 5%. However, if the calibration curve was established in an MV beam quality, the error in absorbed dose could be as large as 20%.

Conclusion: Accurate dose measurements using radiochromic films at low photon energies require that the radiochromic film dosimetry system be calibrated at appropriate corresponding low energies, as large absorbed dose errors are expected when film calibration is performed in MV beam qualities.

1. Introduction

Due to its high resolution and near tissue equivalence, radiochromic films have been used for two-dimensional dose measurements in both clinical and research irradiations for more than two decades [1,2]. Various radiochromic film based reference dosimetry systems have been suggested using relatively inexpensive flatbed document scanners [3] as read-out technology. Depending on the calibration curve function used, response of the film could be non-linear (if optical density is used) or, as of recently, a much easier quantity to use for relative dose measurements with a linear dose response function, has been the normalized pixel value, or PV_{norm} [4]. In either case, when using optical density or normalized pixel value for the determination of dose in absolute terms, the calibration curve (in the case of OD) or the calibration coefficients (in the case of PV_{norm}) have to be determined during the calibration process. For relative dose measurements, on the other hand, if one decides to use OD as the quantity in which the detector signal is expressed, a dose determination in absolute terms will be necessary first due to non-linear OD response with dose. Alternatively, since PV_{norm} (under specific conditions) has a linear response with dose, a simple division by the same function at the reference point will represent relative dose. However, the latter statement has some limitations; it is correct only if the response function does not change with beam quality, or if the beam quality does not change within the measuring plane in which the film strip is positioned. It is very well known [5,6] that the response of the EBT3 film is practically energy

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https://doi.org/10.1016/j.ejmp.2019.05.017
Received 26 February 2019; Received in revised form 24 May 2019; Accepted 25 May 2019
Available online 22 June 2019

independent down to 100 keV. This in turn means that in a very broad range of beam qualities (100 keV effective energy – MV beams) one can use the simplified method for relative dose measurements based on PV_{norm}. However, for photon beam qualities below 100 keV effective energy, response of the EBT3 film starts to depend on beam quality [6] and using only one low photon energy or MV beam calibration curve could lead to systematic errors. Similar applies when PV_{norm} is used for relative dose measurements.

Measurements of dose distributions of low energy photon beams represent a challenge when a detector with energy dependent response is used [7]. In this work we studied dose measurements using the EBT3 model GafChromic™ film in the vicinity of 50 kVp Xoft source. Strips of film were placed together with source inside a water phantom at several distances in order to measure the percent depth dose curve (PDD). Separately, we used Monte Carlo simulations to determine spectra at these particular measurement points and we calculated corresponding half value layers (HVL) for such beams. We have also generated a family of air kerma calibration curves for various low energy photon beams, and for each measurement point, we used the appropriate calibration curve to convert film readings into absolute dose. We also measured PDD for the same source using a parallel-plate ionization chamber within the INTRABEAM (Zeiss) water phantom. Both measurements were compared to TG-43 data for Xoft 50 kVp source [8].

2. Materials and Methods

Because of their high spatial resolution, near tissue equivalence and their relatively beam quality-independent dose response, radiochromic films have been used extensively for absorbed dose measurements in the regions with high dose gradients. We investigated the possibility of using the EBT3 model GafChromic™ film for dose measurements in the vicinity of Xoft Axxent Electronic Brachytherapy (eBT) System (iCad, Inc., Sunnyvale, CA). However, due to significant changes in absorbed dose response with beam qualities below 40 keV effective energy [6], photon spectra from the Xoft 50 kVp source inside the water phantom were determined. At each measurement point, Monte Carlo simulations provided the corresponding photon fluence spectrum (and hence HVL and/or effective energy), and the appropriate dose calibration curve for a given beam quality was used to correct the film measurements. In essence this work relies on the fact that the film dosimeter can be considered a “photon detector” (or “kerma detector”), which is justified by the fact that in this energy region, the electron range is of the order of only 10 s of microns and the signal in the film sensitive layer is only determined by photon interactions within the film sensitive layer.

2.1. PDD measurements

2.1.1. Ionization chamber measurements

Percent depth dose measurements were performed using a soft x-ray parallel-plate ionization chamber (PTW-TN34013) positioned in a self-shielded water phantom originally designed for the INTRABEAM System (Carl Zeiss Meditech AG, Jena, Germany). The ionization chamber position was fixed within a waterproof cover at two possible locations, one at 0° and 90° to the source (see Fig. 1a). For the ion chamber used, there is a distance between entrance foil and chamber top of 0.155 mm. Chamber holder within the INTRABEAM phantom (made of Solid Water™) has a thickness of 1.018 mm and there is an air gap of 0.5 mm. So, the total distance from the outside plate of the chamber holder to the effective point of measurement for the ion chamber was 1.673 mm, which was used to determine the effective measurement point within the INTRABEAM phantom.

Depth dose measurements were performed by mounting the Xoft source in a custom holder attached to a DoseView 1D scanning arm (Standard Imaging, Middleton, USA), which was then used to position the source at defined distances from ionization chamber. The source is positioned within plastic sheath with the cathode located at 2.6 mm from the surface of the source assembly in both tip-end and side-end directions. Therefore, the smallest distance between the source and effective measurement point of the ion chamber is 4.3 mm. The very same value of 2.6 mm was used as an offset for the source when positioning film pieces. In the case of the EBT3 film model used, the effective measurement point (middle of the sensitive layer, Ref. [3]) is 0.176 mm (thickness of Polyester scaled by density). Therefore, in the case of film measurements, the physical distance between the tip of the source assembly, and surface of the film was 2.8 mm.

Ion chamber measurements were taken at 2–6 cm distance from the source, and averaged readings from five repeats at each measurement distance were normalized to 2 cm depth. The first measurement depth was chosen as a clinically relevant one, as the distance from the bare source to the surface of commonly used applicators is 2 cm. Also, we wanted to minimize measurement positional uncertainties within steep dose gradient close to the source. Finally, ion chamber measurements have demonstrated a small difference in PDD values for Tip-End and Side-End geometries, which disappeared beyond 10 mm distance from the source. In this work, all results are reported for Tip-End measurement geometry, as indicated in Fig. 1a. At each measurement point, uncertainty of the measured signal (calculated as standard deviation from five measurements) was 1% and therefore total uncertainty of calculated PDD values is 2%.

2.2. Film measurements

There are two ways to obtain a relative dose distribution, including percent depth doses (*PDDs*) with radiochromic films. The measured depth signal is commonly normalized to the value at the reference point to obtain *PDD*. Taking ratios of signals only for a non-linear dose response function (*netOD*, for example), will result in higher relative doses. Recently, several functional forms have been reported [4,9,10] that can make radiochromic film response linear with dose, which significantly simplifies measurement of relative doses when radiochromic films are used.

In this work, we will use normalized pixel value of the green color channel from transmission images obtained using Epson Expression 11,000 XL flatbed document scanner, as suggested by Aldelaijan and Devic [4]:

$$PV_{norm} = \left(\frac{PV^{after}_{unexposed}}{PV^{after}_{exposed}} \right) - 1 \quad (1)$$

where $PV^{after}_{unexposed}$ represents the average pixel value over an ROI of unexposed (control, or normalization) film strip and $PV^{after}_{exposed}$ the average pixel value over the ROI of the exposed film, both readings extracted from the image scanned 24 h after irradiation.

While the use of linear response function (PV_{norm}) will alleviate the problem of a non-linear response when measuring relative dose distributions, the problem of the beam quality change remains if one wants to measure *PDD* as the beam quality changes with distance away from the source.

To correct for the beam quality dependence of dose response for the EBT3 film model, we generated several calibration curves at various beam qualities, ranging from 50 kVp (HVL = 0.16 mm Al) to 120 kVp (HVL = 4.19 mm Al) in beams of the XSTRAHL *ortho*-voltage unit. The 50 kVp beam quality was used with three beam filtrations: inherent, 0.2 mm Al and 0.5 mm Al, as indicated in Table 1. We also included a 6 MV calibration curve obtained from a linac beam for comparison. The list of beam qualities with their corresponding HVLs and effective energies are given in Table 1.

For low energy photon beams, strips of film were irradiated in air to predefined values of air kerma in air and using the AAPM TG-61 document [11] these values were converted into dose to water in air. The irradiation setup used to calibrate film strips in air has been described in detail by Bekerat et al. [6]. For the 6 MV beam quality, strips

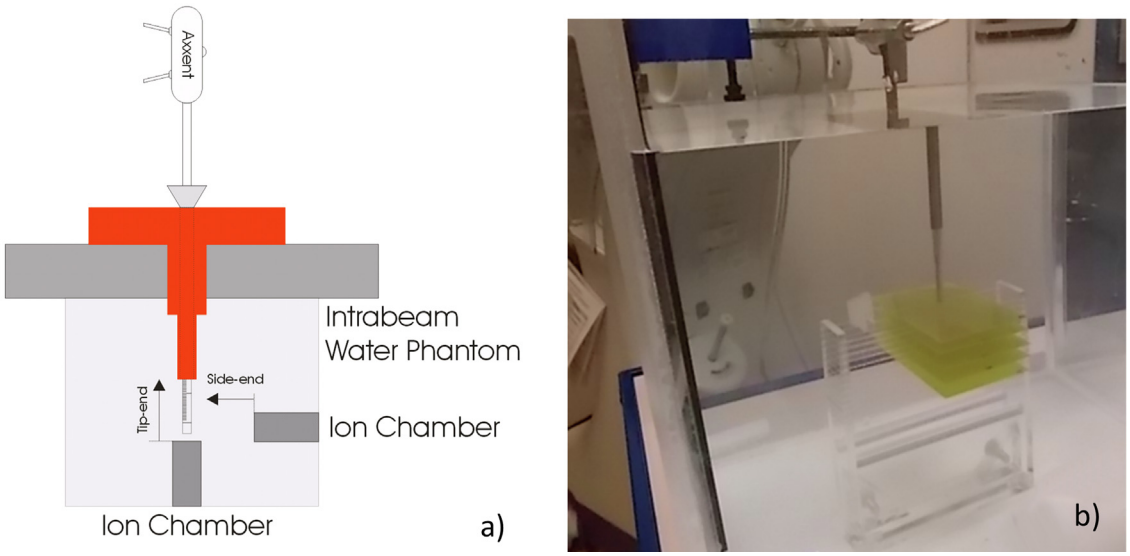

Fig. 1. Experimental setups for *PDD* measurements in water: a) using parallel-plate ionization chamber within IntraBeam Water Phantom (arrows indicate two directions of motion of the source with respect to the ion chamber); b) using EBT3 model GafChromic™ film for tip-end setup.

Table 1
Beam qualities used for film calibration.

Beam	HVL (mm Al)	E_{eff} (keV) [NIST]
50 kVp	0.16	12.7
50 kVp–0.2 mm Al	0.42	16.4
50 kVp–0.5 mm Al	0.74	19.9
70 kVp	1.33	24.8
80 kVp	2.18	29.5
120 kVp	4.19	38.7

of film (2.5″ × 4″ in size) were irradiated in Solid Water™ at depth of 10 cm (where the beam is more uniform than at z_{max}) in accordance to reference radiochromic film dosimetry protocol described in Reference 3. During the reference calibration process, the maximum dose for all beam qualities was set to 15 Gy. Irradiated film strips were only scanned 24 h after irradiation, together with an unexposed film strip, cut from the very same sheet to obtain a normalization pixel value. Film signals were scored over five Regions of Interest (ROIs) 2 mm × 2 mm in size (10 pixels × 10 pixels, from images scanned at 127 dpi = 0.2 mm/pixel) as shown in Ref. [4]. Corresponding values of dose to water were plotted as a function of PV_{norm} for every beam quality.

The same film dosimetry protocol was used for the *PDD* measurement in water. Strips of film, 1″ × 2″ in size, were positioned at different depths (2, 3, 4, 5, 6 cm) from the Xoft source in water (Fig. 1b). A special holder, built from pairs of fish-strings was made to keep film strips in place. All five strips were irradiated simultaneously, and the dose we aimed to deliver at the nearest film strip (positioned 2 cm from the tip of the source) was 10 Gy. For a given setup and the source used, this time amounted to 5.6 min. Once the five film strips were irradiated, they were removed from and control-film strip was placed in the phantom. The control film strip was left to stay in water without irradiating it for the same amount of time as measuring strips, and this film strip served as both a signal normalizing film and control film [12]. For each film strip PV_{norm} was calculated using Eq. (1) and by using the appropriate calibration curve (for a given beam quality at a given depth) the absolute dose was obtained. Film pieces were positioned with respect to the source assembly taking into account the offset of 2.8 mm, as described earlier. For comparison, *PDDs* were normalized to dose value at 2 cm distance from the source (same depth in water as in the case of the ionization chamber measurements described above). Due to beam hardening effects we were not able to take advantage of simply dividing chosen film response values (PV_{norm}) but we had to determine absolute dose at each measurement point. Response to radiation was sampled from each film piece over relatively smaller ROI than in the case of calibration film pieces (1 mm × 1 mm), which were also distributed over smaller area on the central axis, to minimize impact of radiation beam non-uniformity. Since the uncertainty of radiochromic film dosimetry protocol using PV_{norm} as a response function is 2% [4] the overall uncertainty of the PDD values obtained from film measurements is 3% of the measured relative dose.

2.3. Monte Carlo simulations

Monte Carlo simulations of photon energy spectra were performed using BEAMnrc within the EGSnrc system [13]. The source was defined as a 50 kVp photon point source in air and the corresponding energy spectrum at “zero depth” is shown in Fig. 2 and was obtained from measurements reported by Liu et al. [14]. Transport cutoffs of 1 keV kinetic energy for photons and electrons were used. In addition to the standard EGSnrc parameters, the PRESTA-II electron step algorithm,

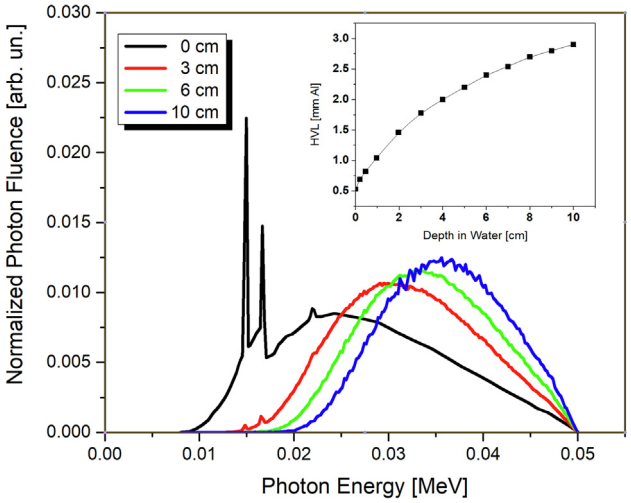

Fig. 2. Monte Carlo simulations of the Xoft 50 kVp source spectra at different distances from the source in water. The spectra are normalized to unit area. Inset represents HVL values calculated using Eq. (2) from the same spectra.

Rayleigh and bound Compton scattering including atomic relaxation were enabled.

The input spectrum, the "zero depth" spectrum in air, was transported through slabs of liquid water and phase-space files were scored at depths ranging from 2 cm to 10 cm from the source using $10 \times 10\ cm^2$ scoring regions. Energy spectra were extracted from the central region (radius of 1 cm) of the phase-space files using the BEAMDP tool within the BEAMnrc package.

The spectra were used to determine HVL (expressed in mm Al) by iteratively solving for the thickness of material required to reduce the air kerma to half its initial value. Air kerma was calculated as per Eq. (2), where K_i and K_f are the un-attenuated and attenuated air kerma, respectively and N represents the number of photons of energy E. Data for attenuation coefficients in Al and mass-energy absorption coefficients in air are obtained from Hubbell and Seltzer [15].

$$\frac{K_f}{K_i} = \frac{\sum_0^{E_{max}} E \bullet N \bullet \left(\frac{\mu_{en}}{\rho}\right)_{air}(E) \bullet e^{-\mu(E)x}}{\sum_0^{E_{max}} E \bullet N \bullet \left(\frac{\mu_{en}}{\rho}\right)_{air}(E)} \quad (2)$$

3. Results and discussion

Fig. 2 shows results of simulated spectra in water using the EGSnrc (BEAMnrc) Monte Carlo code. The spectra are shown for several depths in water from the source for illustrative purposes and clearly demonstrate a beam hardening effect. As an inset in Fig. 2 we present HVL (mm Al) values calculated using Eq. (2) from the simulated spectra, which were then converted to an effective energy (Table 2). As can be seen in the inset (and from Table 2), the HVL changes from 0.53 mm Al at the surface of the source to 2.91 mm Al at a distance of 10 cm from the 50 kVp Xoft source primary spectrum.

Values for HVL given in Table 2 were used to select the appropriate calibration curve at each measurement point within the water phantom. According to Table 2, within the measurement range (2–6 cm) HVL values vary from 1.468 mm Al to 2.403 mm Al (corresponding to 25.6–30.7 keV effective energy).

Fig. 3 shows EBT3 film model calibration curves in terms of dose to water as a function of normalized pixel value (PV_{norm}) for various beam qualities listed in Table 1. Calibration curves are actually straight lines, as the PV_{norm} response is linear with dose. We also added the calibration curve for the 6 MV photon beam for comparison. Fig. 3 also confirms previously published data on EBT film models (5, 6) indicating a reduced sensitivity for photon beam qualities of effective energies below 100 keV.

By comparing Tables 1 and 2, one may observe that in the range of film measurements (2–6 cm) the appropriate calibration lines are the 70 kVp and 80 kVp in Fig. 3. It is of note that in this region, the same signal could produce a relative dose measurement error of 5% if the beam hardening effect is ignored. Currently, Monte Carlo simulations are not readily available to all clinicians, leading to a widespread lack of appreciation for the amount of beam hardening in water phantoms around 50 kVp sources. As the HVL of the source is measured in air, Fig. 3 also suggests that if the calibration line of such beam quality is used for measurements in water an error of more than 25% could be encountered. Also, should one choose to simplify the problem by using a calibration line obtained for more readily available MV beam, one could stumble upon an error of 20%. In the case of relative dose measurements for our experimental geometry (distances from the source ranging from 2 to 6 cm), simple division of the response function that linearizes dose response (PV_{norm}) will result in an error of less than 5%, depending on which point was used for normalization.

Table 2
Calculated HVLs for the photon spectra simulated at different distances from the Xoft source in water and their corresponding effective energies.

Distance	HVL (mm Al)	E_{eff} (keV)
0 cm	0.531	17.8
2 cm	1.468	25.6
3 cm	1.783	27.5
4 cm	2.029	28.7
5 cm	2.231	29.7
6 cm	2.403	30.7
10 cm	2.914	33.1

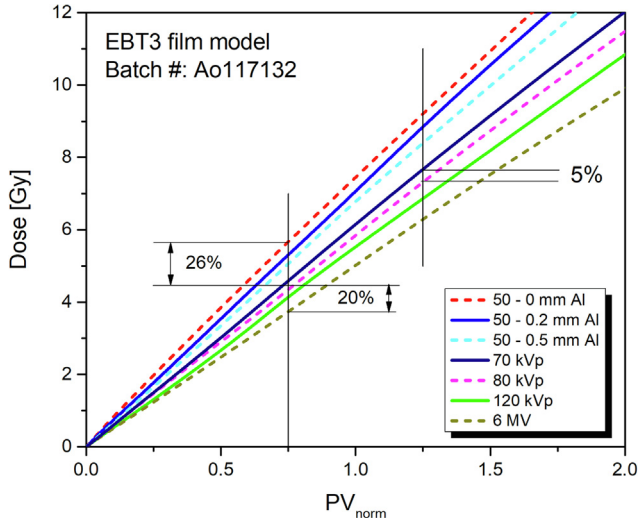

Fig. 3. Calibration curves for EBT3 model GafChromic™ film for different beam qualities.

Using the effective energies for given film measurement points within the water phantom and using calibration lines given in Fig. 3 for each measurement point measured film response (PV_{norm}) was converted to absolute dose using the curve plotted in Fig. 4. For a measured PV_{norm} value, at a given depth (3 cm in the example given in Fig. 4), corresponding dose values were taken for different beam qualities from the calibration lines given in Fig. 3. By using a spline interpolation, we obtained a continuous curve, which was subsequently used to obtain an absolute dose value for a given effective energy at 3 cm depth of our 50 kV_p beam in water. The very same procedure was repeated for other film measurement points, and relative doses were calculated by

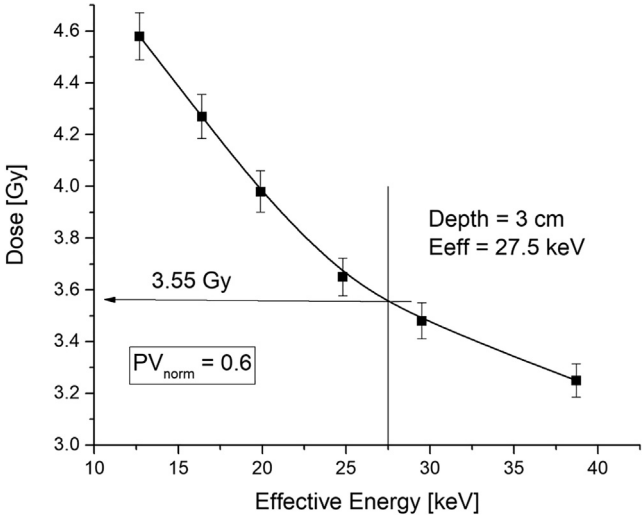

Fig. 4. Improving doe measurement accuracy by using appropriate beam quality calibration curve. For a given beam quality (27.5 keV effective energy) at a given measurement depth (3 cm) measured response (PV_{norm} = 0.6) was used to reconstruct Dose vs Effective energy curve, which was used to determine dose for a measured signal.

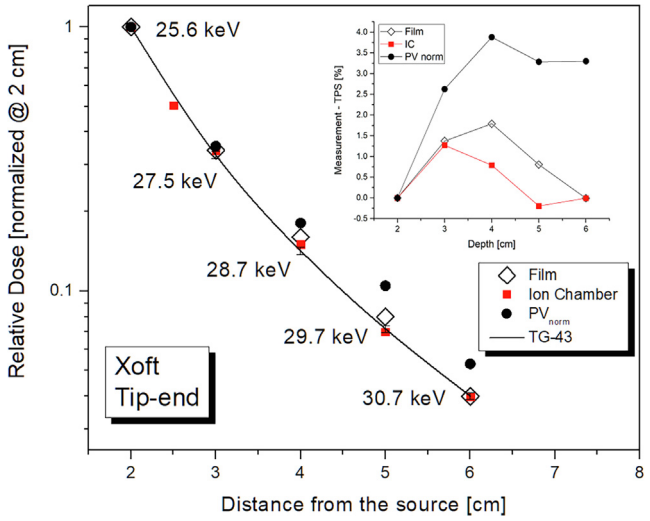

Fig. 5. Comparison of *PDD* measurement results using film (diamond) and ion chamber (square) with TG-43 data provided by manufacturer of the source. Circle represents data obtained by simply dividing PV_{norm} a function that is linear with film dose response. Inset represents percentage difference between measurements and TG-43 based *PDD* curve.

normalizing such obtained doses to the point 2 cm from the source. Dose points in Fig. 4 (for a given measured PV_{norm}) have measurement uncertainty of 2% and the corresponding error bars correspond to one sigma uncertainty.

Fig. 5 shows results of film (diamond) and ionization chamber (square) *PDD* measurements compared to TG-43 data (solid line) provided by the manufacturer [8]. All measurements were normalized at 2 cm depth. At each depth, we also indicate the corresponding effective photon energies. Circles represents data obtained by simply dividing the data of PV_{norm} at each measurement point by PV_{norm} at 2 cm distance from the source (chosen to be reference point), a method that ignores the variation in beam quality within the measuring region. One sigma uncertainty for film measurements amounts to 3%, while for the ion chamber measurements and simple PV_{norm} division amounts to 2% and are all smaller than points. The inset in Fig. 5 represents percentage difference between measurements and TG-43 based *PDD* curve. While the relative dose difference for ionization chamber and beam quality-corrected film measurements are within 2% from the TG-43 data, a simple signal division (PV_{norm}) results in an error of the order of 4% as expected based on data presented in Fig. 3. Figure 5 also reveals that not only the beam quality change would hamper the use of simple PV_{norm} division for relative dosimetry, but the uncertainty vs. relative dose error too, as the error amounts to 4% (inset in Fig. 5) while the uncertainty is only 2%, which means that the error would be significantly underestimated.

4. Conclusions

Accurate dose measurement using radiochromic films in low photon energies require the radiochromic film dosimetry system to be calibrated at beam qualities corresponding to measurement points as significant beam hardening effect occurs as a function of depth in water for the 50 kVp source. Use of film calibration in air for the 50 kVp source when measuring absorbed dose in phantom may result in an error of up to 25%. If, on the other hand, one would use calibration performed in MV beam, the resulting error in measured dose could reach 20%. Although the use of a linear dose response function could make relative dose measurements with radiochromic film easier, taking only ratio of readings (PV_{norm}), which ignores the beam hardening effect, will results in a systematic error of 5%.

Acknowledgement

This work was supported in part by the Natural Sciences and Engineering Research Council of Canada contract No. 386009.

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