PIXEL-BASED CORRECTION METHOD FOR GAFCHROMIC®EBT FILM DOSIMETRY

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Received April 28, 2010
Accepted for Publication October 14, 2010

In this paper, a new approach using a pixel-based correction method was developed to fix the non-uniform responses of flat-bed type scanners used for radiochromic film dosimetry. In order to validate the method's performance, two cases were tested: the first consisted of simple dose distributions delivered by a single port; the second was a complicated dose distribution composed of multiple beams. In the case of the simple individual dose condition, ten different doses, from 8.3 cGy to 307.1 cGy, were measured, horizontal profiles were analyzed using the pixel-based correcton method and compared with results measured by an ionization chamber and results corrected using the existing correction method. A complicated inverse pyramid dose distribution was made by piling up four different field shapes, which were measured with GAFCHROMIC®EBT film and compared with the Monte Carlo calculation; as well as the dose distribution corrected using a conventional method. The results showed that a pixel-based correction method reduced dose difference from the reference measurement down to 1% in the flat dose distribution region or 2 mm in a steep dose gradient region compared to the reference data, which were ionization chamber measurement data for simple cases and the MC computed data for the complicated case, with an exception for very low doses of less than about 10 cGy in the simple case. Therefore, the pixel-based scanner correction method is expected to enhance the accuracy of GAFCHROMIC®EBT film dosimetry, which is a widely used tool for two-dimensional dosimetry.

KEYWORDS : EBT Film, Flat-Bed Scanner, Non-Uniform Dose Correction

1. INTRODUCTION

Film has been an important dosimeter in many fields that provide two-dimensional dose distribution. Radiographic film has been used for many years since it provides high spatial resolution with relatively low cost. However, the radiographic films have limitations in their response to megavoltage (MV) beams. Due to the AgBr component included in film, radiographic film overestimates the dose with increased depth of measurements and/or field size as the low-energy Compton scattered photons increase. Therefore, an additional correction procedure, such as using a lead filter, is required for accurate dosimetry.

As a result this shortcoming, demand is growing nowadays for radiochromic films for the confirmation of delivered dose distributions within a phantom. Radiochromic film has advantages over radiographic film, which includes lower sensitivity to the photon beam energy and extended applicability to a wide range of absorbed doses [1]. In particular, Gafchromic External Beam Therapy (GAFCHROMIC®EBT) radiochromic films (International Specialty Products, Wayne, NJ) composed of tissue-equivalent materials ($Z_{eff}^{EBT} = 6.98$, when compared to $Z_{eff}^{Water} = 7.3$) are significantly less sensitive to low-energy scattered photons. In addition, the film's self-developing property, which does not require any chemical or physical processing procedure, reduces the uncertainty associated with the processing condition. This chemical free processing provides environment-friendly dosimetry. Also, insensitivity to room light allows for GAFCHROMIC®EBT film to be cut into a desired shape and inserted completely into a phantom. The waterproof characteristic of GAFCHROMIC®EBT also allows for in-water film measurement. With these advantages, therefore, GAFCHROMIC®EBT film has been rapidly introduced in recent years for various dosimetric studies, such as radiotherapy and radiation protection.

Scanning is a requisite method for acquisition of dose information from film. Hence, accurate readout of film data without distortion is a critical aspect of EBT film dosimetry [2]. However, most commercially available scanners used for readout of EBT film employ a fluorescent lamp as a light source, which leads to light scattering on the active layer of the film [3]. This problem results in a significant distortion in dose conversion with non-uniform responses, even though the uniform dose is originally irradiated onto the film [4]. Accordingly, several studies have attempted to resolve this problem. The conventional method used to remedy this problem is based on correction factors (CF) determined from the dose profile measured by an ionization chamber [5], which is considered as a standard dosimeter. This CF curve, which is a function of scanner position, as well as delivered dose, is fitted with a second order polynomial for scanner position for several dose levels. However, this approach only accurately approximates real data within the central half area of the scanner. As a result, errors become significant in the side regions of the film. For this reason, except for in the central region, prediction of real incident doses from the decoded scan values can be very challenging.

Therefore, we investigated a new approach for the correction of the non-uniform response of a flat-bed type scanner in the whole area of the scanner. This can enhance the applicability of the EBT film to broad radiation beams. In addition, user-friendly GUI-based software for facilitation of this correction procedure has been developed, and the accuracy of the developed method was finally evaluated by comparison of the corrected results with those of ionization chamber measurements, Monte Carlo simulation, and conventional correction methods.

2. DOSE CORRECTION ALGORITHM

Our approach employs measured beam profiles, evaluates the response, and corrects over/under responses at each pixel position. The schematic diagram describing the procedure is presented in Figure 1. For evaluation of the non-uniform response pattern of the scanner, beam profile measurements at various dose levels were performed, since the response of the scanner varied with exposed doses as well as with the scan position [5]. GAFCHROMIC®EBT radiochromic films were irradiated from 0 cGy to 307.1 cGy with multi-dose steps (see Section 2.2). Each of the films was then scanned with a flat-bed type scanner, which used a fluorescent light source with a broadband emission spectrum and a linear charge coupled device (CCD) array detector [6]. Due to the aforementioned characteristics of the scanner, significant non-uniformity was observed along the horizontal direction, which resulted in overestimation of dose [7]; hence, a new correction algorithm was designed for appropriate consideration of the observed response properties of the scanner.

2.1 Calibration

In order to use EBT films for absolute or relative dosimetry, it is necessary to establish a dose response

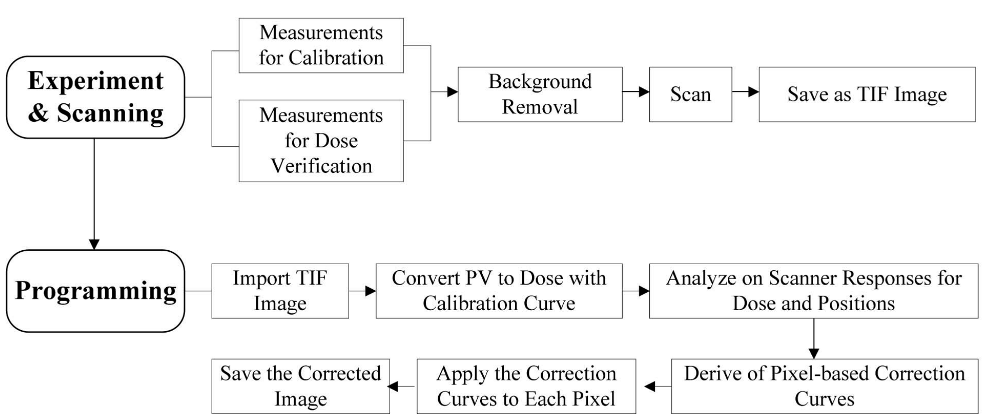


Fig. 1. Summary of Non-Uniform Response Correction Procedure. For Evaluation of Non-Uniform Response Pattern of Scanner, Calibration and Beam Profile Measurement were Performed Using EBT Film. Measured EBT Films Which were Removed Its Background were Scanned and Saved as Two-Dimensional TIF Format Array. The TIF File was Imported into the Developed GUI-Based Software. In the Software Tow-Dimensional TIF Data were Converted into Dose by Applying a Calibration Curve. The Correction Value at Each Pixel was Computed Based on the Pixel Position and Delivered Dose, and then the Computed Value was Applied to Every Pixel of the Film. Finally, Corrected Film Data were Saved in Designated Folder

relationship over the range of doses that will be encountered during exposure of the radiation. The response curve is a function of the sensitivity of the film as well as of the characteristics of the digitization system [8]. Thus, the calibration process was independently performed ahead of the film measurements and analysis.

To avoid inter-batch response variation of EBT films, which is known to be approximately 1% [9], films from the same batch-numbered packet were used for calibration. First, the film was placed inside a solid water phantom (Gammex, Middleton, WI, USA) and positioned at a depth of 5 cm, with a source surface distance (SSD) of 100 cm. With a linear accelerator (Primus, Siemens Medical Systems, Concord, CA, USA), a series of photon beams of 6 MV X-rays with the size of 5 $\times$ 5 cm^2 were irradiated from 0 cGy to 500 cGy with seventeen dose levels onto the films. At this time, the delivered doses were calculated using percent depth dose (PDD) data at a depth of 5 cm and total scatter factors of the machine (St) presented as follows: Dose = MU $\times$ PDD /100 $\times$ St, where MU is the Monitor Unit set by the machine. Daily output variation was also corrected. Following irradiation, the films were scanned by avoiding dose distortion, according to the procedure described in Section 2.3, and the calibration curve was acquired with a pixel value ranging from 50417 to 19552, corresponding to the dose range from 0 cGy to 500 cGy, as shown in Figure 2.

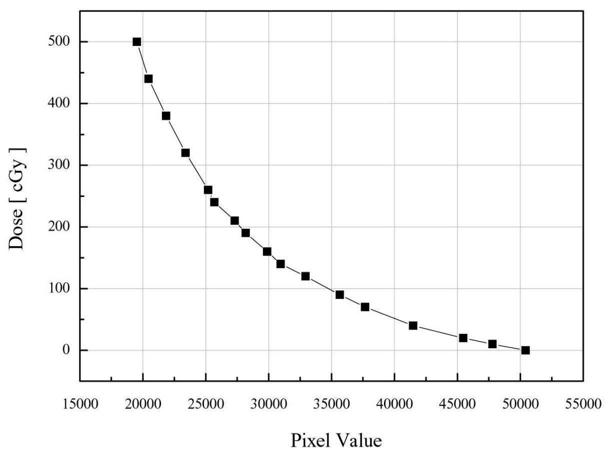


Fig. 2. Calibration Curve for the Response of GAFCHROMIC®EBT Film, Given in Terms of X-Ray Absorbed Dose in Solid Water Phantom Vs. Pixel Value

2.2 Dose Measurements Using Film

Prior to the evaluation of dose distribution, which was a composite of multiple levels of a single dose, individual dose profiles were tested for confirmation of the accuracy of the correction algorithm. Less than 3% of error is required in order to use film with an error tolerance of 3% for IMRT plan verification [9, 10-12]. Thus, we obtained dose data with ten dose-steps (8.3, 16.6, 33.2, 49.8, 74.7, 99.6, 124.5, 174.3, 224.1, 307.1 cGy). For preparation of the films, two sheets of EBT film with dimensions of 203 mm $\times$ 254 mm were individually cut into five pieces of strips along the horizontal direction (see Figure. 3 (b)). Because the five film pieces (total

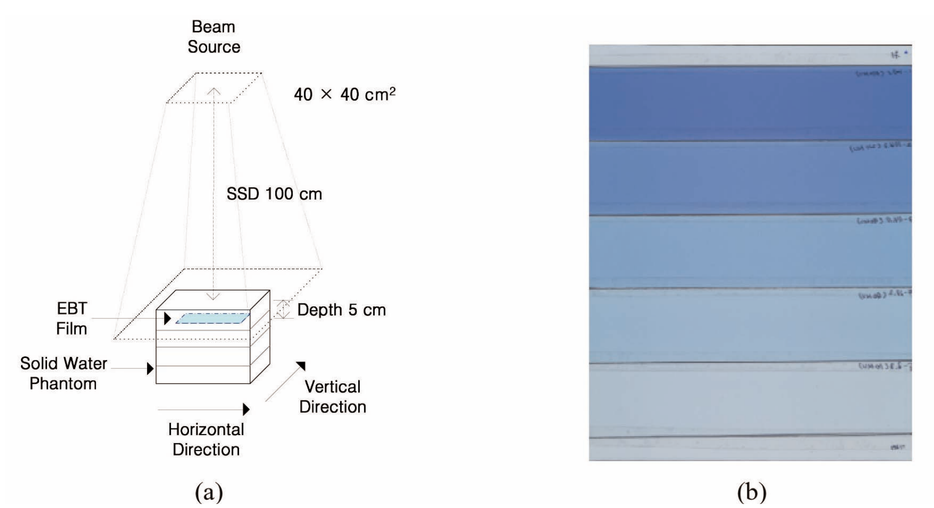


Fig. 3. (a) Experimental Scheme. Each Striped-Film was Placed at the Center of the Phantom Plate, and 40 $\times$ 40 cm^2 Beams were Irradiated. (b) EBT Film Arrangement on the Scanner Plate for Obtaining the Correction Curve and for Verification of the Corrected Doses. Two Sheets of Film with Dimension of 203 mm $\times$ 254 mm were Individually Cut into Five Stripes along the Horizontal Direction

203 mm × 225 mm) can cover the practically used area of the scanner plate, the uppermost and bottommost parts of the film were not used in measurements. Each film piece was then placed on the center of the phantom, and 6 MV X-ray beams with dimensions of 40 × 40 cm^2 were irradiated, as shown in Figure 3 (a). Other geometrical setup conditions for beam measurements were the same as those used for the calibration procedures described in Section 2.1.

2.3 Scanning Procedure and Image Acquisition

Because considerable errors can be accumulated during the scanning procedure, the scanning process is very important and challenging in radiochromic film dosimetry. For this reason, scannings were very carefully performed according to the guidelines of manufacturers and of other studies [8, 13-14].

Since the color density of the EBT film grows with time [1, 15-16], scans were performed one day after irradiation using FILMQA™ software (3cognition LLC, Great Neck, NY). In order to reduce uncertainty from the environmental conditions, the films were stored in a place in which temperature and humidity were automatically controlled at 21 ± 2 °C and 50 ± 10%, respectively. Also, the films were kept in a dark container without exposure to ambient light, as recommended by the manufacturer, International Specialty Products (ISP) [8].

The scanner (Epson Expression 1680 Pro, Epson Seiko Corporation, Nagano, Japan) was set to professional mode, with a resolution of 72 dpi (approximately 0.3528 mm per pixel). Using FILMQA software, the transmission images were saved in 48 bit red-green-blue (RGB) mode TIF format without compression. Since the maximum absorption of the EBT film is at 636 nm wavelengths in the red region of the visible spectra, and since the use of this absorption peak is known to improve image quality [17], only red channel data of 16 bits in the range of 0 to 65535 of pixel value was saved and used. Ahead of scanning, the scanner window was wiped clean with alcohol and the outer frame of the scanner plate was marked for identification of the center point of the plate to enhance film positioning reproducibility. Since the response of the scanner is stabilized after the third scan [18-19], data from the first three scans were discarded in image acquisition in order to avoid the warm-up effect of the scanner lamp. Background subtraction was conducted by scanning of an unexposed film from the same batch-number packet. For each scan, the scan position was carefully determined to avoid uncertainty resulting from position dependency of scanner-read dose. As shown in Figure 4, each film was scanned in the central region of the scanner plate, where image distortions are negligible.

2.4 New Correction Algorithm

2.4.1 Non-Uniform Response of the Scanner

The calibration curve was applied to the scanned values

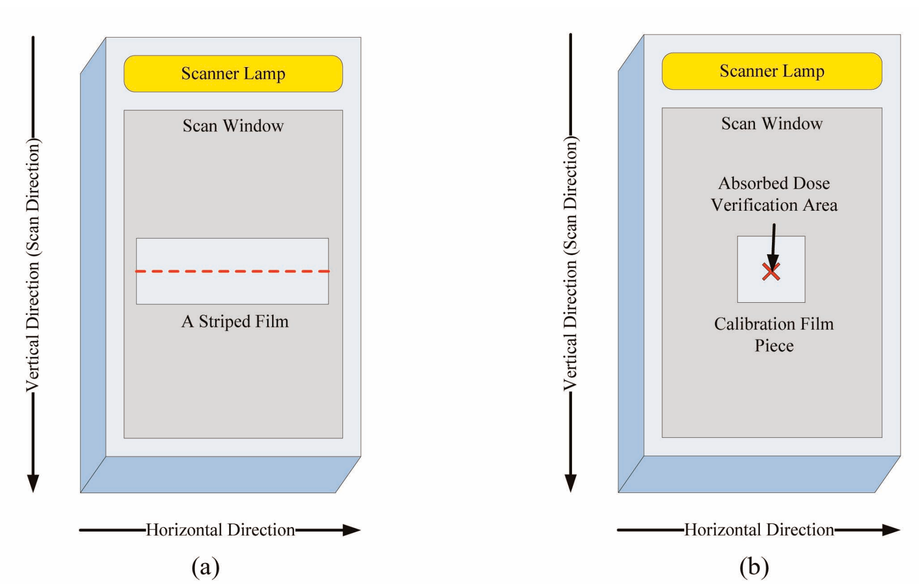


Fig. 4. Illustrations of Film Alignment on Scanner Window. Each Film was Positioned at Center of Scanner Plate, where Image Distortions are Negligible. (a) Profile of Striped Film was Verified along Film Center Indicated with Red Line. (b) In Calibration, Absorbed Dose was Verified from Central Region of Calibration Film Piece

of the striped-films, and pixel values were converted to doses [cGy]. Dose data were saved as MAT-file format with dimensions of 720 × 576 pixels to be read by in-house developed correction software (see Section 2.4.3). Normalized horizontal pixel values before conversion into doses for various dose levels are shown in Figure 5. Pixel values monotonically decrease along the horizontal direction, from the center to the edge of the scanner plate for all of the tested dose levels. The difference between maximum and minimum doses read by the scanner gradually increases with the increase of the dose delivered. This phenomenon is caused by the fact that the active layer of the radiochromic film is composed of particles dispersed in a matrix, each with a different refractive index. Due to the finite wavelength of the scanner's light source, illumination at the center of the film is greater than that at the sides of the film when the light diffuses [7].

2.4.2 Pixel-Based Correction Curve

A pixel-based correction curve, f_n, is a function of two variables that relate the real doses delivered to the medium and the scanner-read doses; this curve depends on each pixel position. Thus, there are 576 different functions for correction of a matrix composed of 576 horizontal pixels, in which pixel position and delivered dose are two independent variables. This function was derived from horizontal-axis profiles measured from the film pieces. When uncorrected doses read by the scanner were plotted against the irradiated dose, the function f_n had to be a linear function with coefficient 1 if no distortion was involved. However, as shown in Figure 6, the correction function gradually deviated from linear as the pixel was positioned farther from the center of the scanner plate and doses increased. Therefore, we assumed that f_n was a quadratic function ($f_n(x) = ax^2 + bx + c$, *where*, x = *irradiated dose*, $f_n(x)$ = *scanner-read dose at n-th pixel*), and found the three coefficients by using a least square fitting method. During this process, we used the dose irradiated to the center of the film. The flatness of the beam profile, which was less than or equal to 2%, was not considered in this process.

Figure 6 shows the pixel-based correction curves at the three sampled positions (5-th, 287-th, and 570-th pixels, positioned on the left-side, center, and right-side of the scanner, respectively) among 576 horizontal pixels. The solid black line indicates the ideal case, in which the scanner-read doses are completely equal to the irradiated doses ($y = x$). According to these correction curves, for 300 cGy, the scanner-read dose at the scanner center (287-th pixel) showed a 1.3% difference (304 cGy). On the other hand, the scanner readings at the 5-th and 570-th pixels over-evaluated the true dose (300cGy) by about 10% (331 cGy) and 20% (359 cGy). Using the known function, the original doses can be inversely estimated using $f^{-1}(y) = x$. If the dose of 307 cGy is read at the 5-th, 287-th, and 570-th pixel, the original doses at each pixel could be interpreted as approximately 263 cGy, 303 cGy, and 281 cGy, respectively. Thus, doses distorted due to the light scattering effect can be corrected by application of the pixel-based correction curve to the scanner reading along the whole horizontal-axis pixel.

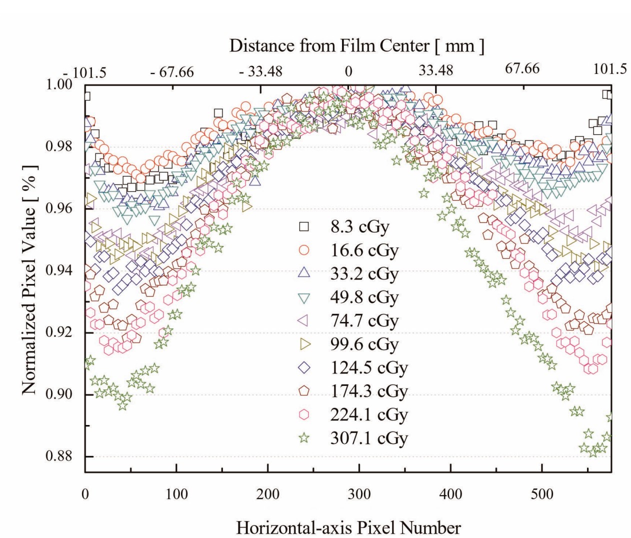


Fig. 5. Normalized Horizontal Profiles at Various Doses Read by the Scanner Ranging from 8.3 cGy to 307.1 cGy. Pixel Values Monotonically Decrease along Horizontal Direction, from Center to Edge of Scanner Plate

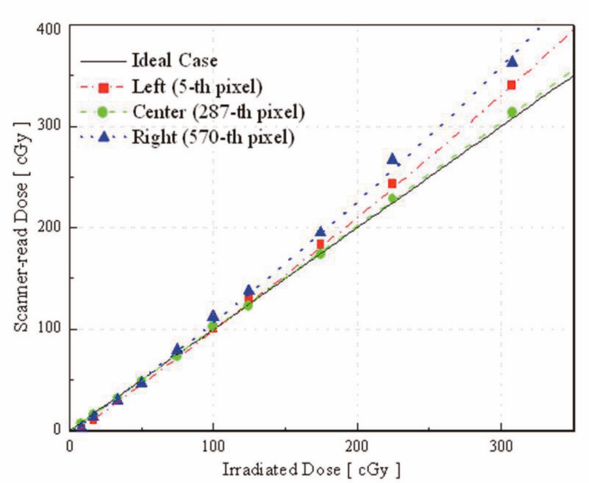


Fig. 6. Scanner Response Curve for Irradiated Doses at Three Sampled Pixels. Solid Line Indicates Ideal Case ($y = x$). Dotted Lines Have Three Coefficients with a Least Square Fitting Method According to Triangle, Square, and Circle Points Read by the Scanner

2.4.3 Development of GUI-Based Correction Software

GUI-based user-friendly software was developed with MATLAB™ to facilitate the correction procedures, as shown in Figure 7. The front panel of this software is divided into three major regions, *i.e.*, data import, calibration, and conversion. The main function of the first region is to import the raw film data, of which the background is then removed by using scan data of an unirradiated film with the same batch number, and to decode pixel values. Next, these pixel values are converted to doses in the calibration region. A variety of functions can be used in the fitting of data points from linear to n-th order polynomials. Finally, the pixel-based correction curve is applied to the data in the last region. In addition, film images of before and after the correction are shown on the panel, and corrected dose data are saved in the designated directory.

3. ACCURACY VALIDATION OF CORRECTION CURVES

3.1 Individual Dose

First, to validate the accuracy of the developed method, simple profiles were tested. For that purpose, distorted doses decoded from the scanner were corrected separately with the pixel-based curve and CF-based algorithms, according to procedures detailed by Menegotti *et al* [5]. Also, corrected doses were compared with profiles measured by an ionization chamber (sensitive volume of 0.03 cm^3, type 31015, PTW-Feiburg, Germany). In this case, the corrected profiles should be equal to the profiles measured by the ionization chamber because the ionization chamber is a standard dosimeter in the measurement of profiles with a large sized beam. Dose profiles were measured in a water phantom at the scaled depth of 5.05 cm; the interval for each data point was 0.1 mm using the continuous scan mode. The other measurement condition was the same as that of the film strip (see Section 2.2).

Table I shows the average difference between dose profiles after correction and measurement by the ionization chamber for a variety of delivered dose levels. Profiles of the horizontal axis for low (8.3 cGy), mid (124.5 cGy), and high (307.1 cGy) doses are shown in Figure 8. In the case of a very low dose, such as 8.3 cGy, the corrected profiles for both methods were not very different. Differences between results for the the ionization chamber method and the other methods were 5.59% (scanner read), 5.55% (CF based correction), and 4.99% (pixel based correction). This error is potentially due to the fact that the EBT film becomes less sensitive for lower irradiated doses [9]. Due to this property, the inherent uncertainty is high at low doses, and dose correction using any method is challenging. Furthermore, the R-squared value, which is a statistical measure of wellness of approximation to real data, is an important factor in evaluating data set. R-squared value is a descriptive

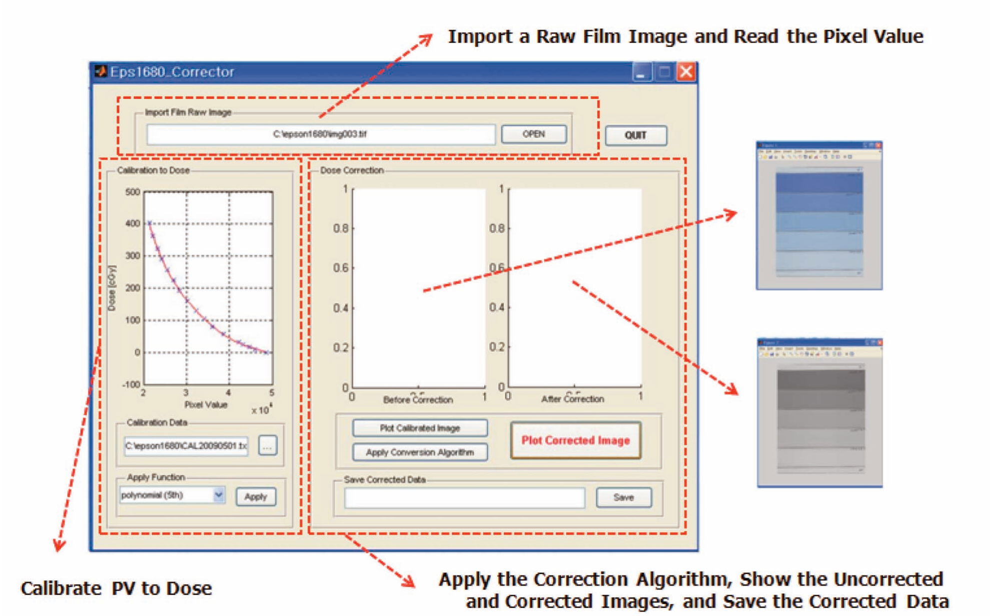


Fig. 7. Newly Developed GUI-Based Software to Facilitate Correction Procedure, Which Includes Data Import, Calibration, and Conversion Regions Based on Pixel-Based Correction Method

measure between zero and one. The closer the R-squared value is to one, the greater the ability of that model to predict a trend correctly. However, the value was very low for low dose regions (approximately 0.4 for 8.3 cGy, and 0.6 for doses under 30 cGy) while the R-squared value was ranged from 0.8 to 0.9 for higher dose regions. Therefore, the CF-based method was not accurate in the low dose regon. For this reason, the estimated regression line was less confident, and the error of the profile corrected with conventional methods grew significantly, up to 5.57%. However, profiles corrected by the conventional method agreed within 3% of error in doses above 33.2 cGy. Use of the pixel-based curve reduced the dosimetric error to less than 1% in the whole dose region, except for the aforementioned very low dose region (under about 10 cGy).

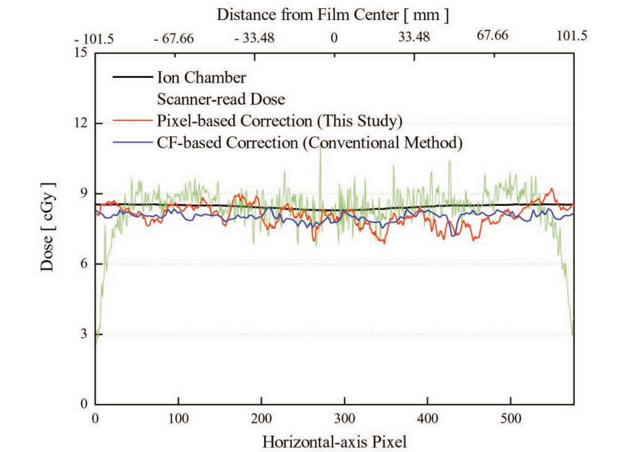
3.2 Step-Wise Dose

Steep dose gradients are frequently seen in measurements of dose distribution in radiation therapy, when multiple doses are irradiated, such as in IMRT. Until now, analysis of dose distributions with steep gradients has been challenging; however, due to the fine special resolution, film is the best dosimeter. In order to validate that the developed correction method is capable of interpreting the dose distribution correctly with steep gradients, a step-wise dose distribution was measured and corrected. For evaluation of the accuracy of the correction, the dose distribution was compared with the

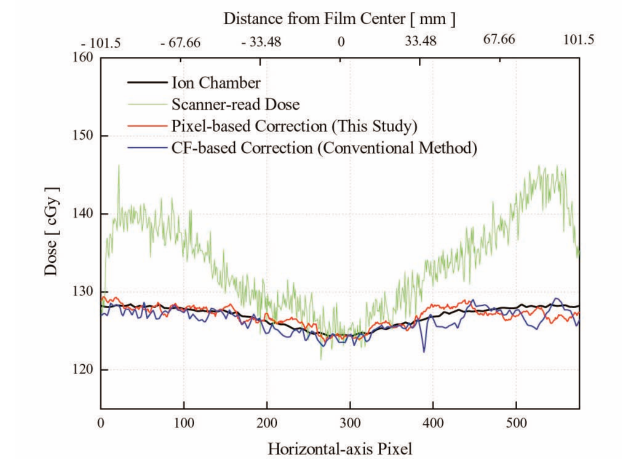
Table 1. Average Differences Compared with Profiles Measured by Ionization Chamber

Irradiated Dose *[cGy]	Scanner-read dose [%]	CF-based Correction (Conventional Method) [%]	Pixel-based Correction (This Study) [%]
8.3	5.59	5.55	4.99
16.6	1.72	5.57	0.45
33.2	3.03	1.91	0.10
49.8	3.19	0.98	0.31
74.7	4.17	2.13	0.18
99.6	5.29	2.18	0.87
124.5	4.86	0.50	0.02
174.3	6.24	1.60	0.49
224.1	7.30	2.26	0.67
307.1	8.15	1.23	0.12

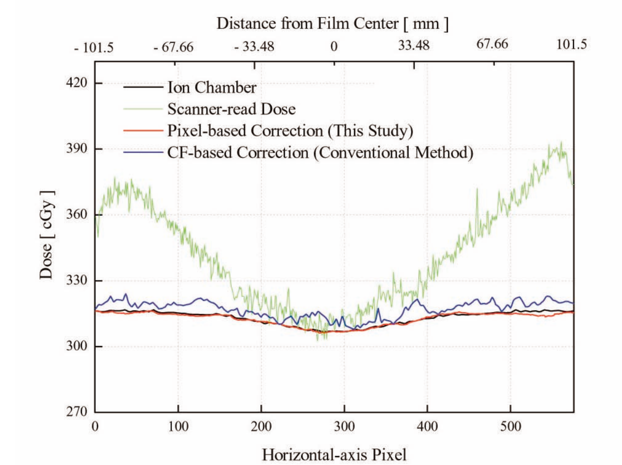
*Irradiated dose means the dose at the center of the irradiated field



(a) Low (8.3 cGy) Dose

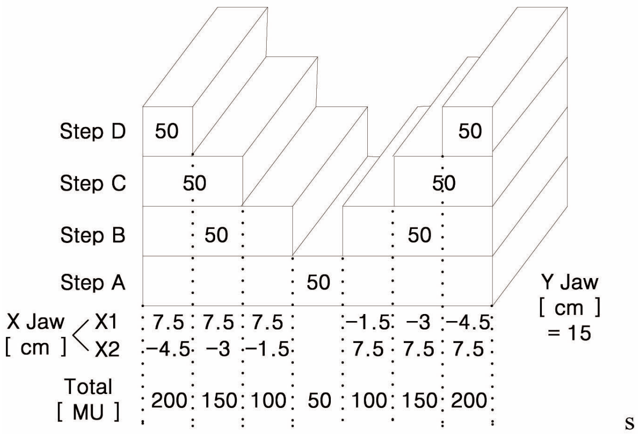


(b) Mid (124.5 cGy) Dose

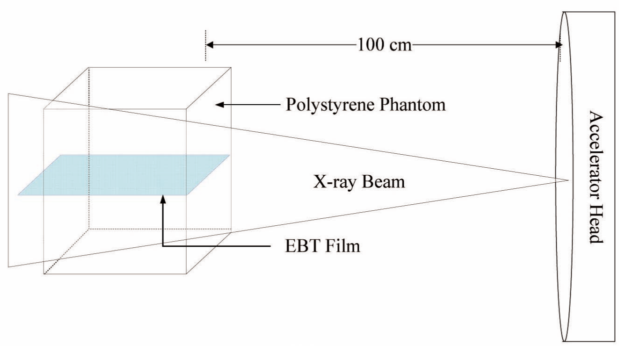


(c) High (307.1 cGy) Dose

Fig. 8. Comparisons of Horizontal Profiles for Scanner-Read Dose, Ionization Chamber, and Actual Incident Dose Corrected by Applying Pixel-Based and CF-Based Method According to Incident Dose Level ((a) Low (8.3 cGy), (b) Mid (124.5 cGy), and (c) High (307.1 cGy) Doses)



(a)



(b)

(c)

Fig. 9. (a) Irradiation Conditions for Inverse Pyramid-Shaped Beam with Four-Step of Doses by Varying Beam Sizes ($X \times Y$ cm^2), (b) Experimental Configuration for Inverse Pyramid-Shaped Beam Measurement. Film was Located Parallel to the Beam Axis. (c) Irradiated Film with Symmetric Form of Inverse Pyramid Structure

calculated data using the BEAMnrc and the PMCEPT Monte Carlo code. The Monte Carlo code has the advantage of fine spatial resolution compared with commercially available 2-dimensional dosimeters, which have 5-7 mm of detector spacing. The BEAMnrc and PMECPT Monte Calro codes were commissioned against the ion chamber measurements as well as against the Dosxyznrc data for the beam sizes of 3×3 cm^2, 9×9 cm^2, 12×12 cm^2, 20×20 cm^2, which covers the beam sizes of our study. Dose distribution corrected by the conventional method was also compared.

An inverse pyramid-shaped dose profile was obtained by irradiation of four-step doses with different field sizes. Film was located parallel to the beam axis. As shown in Figure 9 (a), 50 MU was delivered by each field to the EBT film within the solid water phantom by variation of the field sizes ($X \times Y$ cm^2), e.g., 15×15 cm^2 (Step A), 6×15 cm^2 (Step B), 4.5×15 cm^2 (Step C), 3×15 cm^2 (Step D). For each field, the X-jaw was opened (X1 : X2) at 7.5 cm : 7.5 cm (Step A), 7.5 cm : -1.5 cm (Step B), 7.5 cm : -3 cm (Step C), 7.5 cm : -1.5 cm (Step D), -1.5 cm : 7.5 cm (Step D), -3 cm : 7.5 cm (Step C), and -1.5 cm : 7.5 cm (Step B) with the Y field size fixed at 15 cm. Total accumulated doses from the largest to the smallest steps were 200 MU, 150 MU, 100 MU, and 50 MU. The profile of the inverse pyramid had regions of steep dose gradients with a maximum of 20 cGy/mm.

Average difference of the measured profiles without scanner correction and the Monte Carlo (MC) calculation was approximately 16%, as presented in Figure 10. In particular, regions with the greatest gradient (outer boundaries; from 1_{st} to 100^{th} and from 476^{th} to 576^{th} pixels)

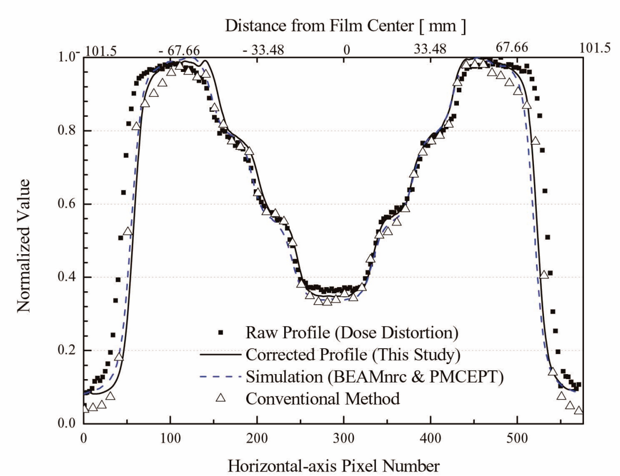


Fig. 10. Inverse Pyramid Beam Profiles for Scanner-Read and Corrected Profiles with Pixel-Based (This Study) and Conventional Correction Method. They were Also Compared with MC Calculation (BEAMnrc & PMCEPT Code)

represented a discrepancy of approximately 60%. Mean discrepancy in the in-field region decreased to approximately 2%; however, the difference in the outer boundary was reduced to approximately 6% after correction with the conventional method. However, the pixel-based curve reduced the average difference in the in–field region to approximately 1% and in the boundaries with the greatest gradient to approximately 3%. As aforementioned, the conventional CF method only approximates real data well in the central region of the scanner plate when fitted with a second order polynomial; it is shown to have a limitation in correcting of distorted doses in the side regions, with over 3% tolerance. In addition, the low R-squared value of the CF curve in a wide range of low doses makes correction of distortion in the penumbra area difficult; indeed, the error in this penumbra area (1_{th} to 40^{th} and 540^{th} to 576^{th} pixels) increased slightly from the original 39% to 40% after correction.

4. CONCLUSION

In this study, a pixel-based correction curve was developed for correction of the non-uniform response of a flat-bed type scanner used for radiochromic film dosimetry; correction curve performance was evaluated with conditions for single and composite dose delivery. In addition, user-friendly GUI-based software was developed for easy performance of the correction procedure. In the case of single beam delivery, the method developed in this study showed accuracy through reduction of dosimetric error to 1%, except for situations with very low doses of less than approximately 10 cGy. The average discrepancy between the measured profile and the MC calculation in complicated dose distributions was also reduced from approximately 16% to 1%, except for regions of steep dose gradient, such as boundaries. The conventional method using polynomial fitting was shown to have limitations in correction of dose distortion in the side region of the scanner. Also, increased error was detected in a relatively wide range of low doses due to polynomial fitting and low R-squared values with the conventional method. The newly developed method has resolved the aforementioned limitations of the conventional method, which, in turn, can extend the applicability of the EBT film dosimetry for more diverse cases that require larger radiation fields together with higher confidence levels of dosimetry. Also, the time it took for film analysis using the developed method was comparable to that found when using the conventional method, including the help of the developed software. Therefore, compared to the conventional method, the pixel-based correction curve is expected to enhance the accuracy of GAFCHROMIC®EBT film dosimetry, and can be applied to various dosimetric studies with diverse dose distributions.

ACKNOWLEDGEMENTS

This work was supported by the Nuclear R&D Program under the auspices of the Korean Government (MEST) (No. 20100018224) This work was also supported by the Proton Engineering Frontier Project (PEFP).

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