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A Method to Enhance the Spatial Resolution of the Two-dimensional Detector Arrays for the Precise Dose Assessment of Intensity Modulated Radiation Therapy

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Two-dimensional (2D) detector arrays are widely used in the patient-specific quality assurance of intensity modulated radiation therapy (IMRT). One of the disadvantages of the 2D detector arrays is the low spatial resolution. We proposed a new method to enhance the spatial resolution of the 2D detector arrays. We showed that this method also reduced the volume effect of a detector element. To demonstrate the method we evaluated one field of an IMRT verification plan with the 2D ion chamber array. An open field was measured and its penumbra width was evaluated to show the reduction of the volume effect of the detector. Using the proposed method, the distance between measurement points was reduced from 7.62 mm to 6.00 mm and the penumbral width was also reduced.

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I. INTRODUCTION

Patient-specific quality assurance (QA) of intensity modulated radiation therapy (IMRT) is a verification process of the individual IMRT plan before actual dose delivery [1]. Normally a two-dimensional (2D) dose distribution is measured and compared to the corresponding dose distribution calculated by the treatment planning system. The measurement is generally performed using films and ion chambers. The relative dose distribution is measured with the film and the absolute point dose is measured with the ion chamber. The overall process of the measurement using the film is a complex and time-consuming task. The film need to be developed and scanned before one gets the result. 2D detector arrays are also used to measure the 2D dose distributions [2–5]. One of the advantages of the 2D detector arrays is that the measurement process is relatively simple. One can get the result immediately after dose irradiation to the 2D detector arrays. For this reason, the 2D detector arrays are most efficient for the routine patient-specific QAs. However, one of the disadvantages of the 2D detector arrays is that they have lower spatial resolution than films. In the high dose-gradient region, the 2D detector array could not cover all the necessary data points. Therefore, the 2D detector array is generally rec-

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ommended to be used as a routine patient specific QA after the IMRT technique is commissioned using dosimeters with the higher spatial resolution such as films [6]. We noted that increasing the spatial resolution of the 2D detector array could be an alternative, but such research is rare in our knowledge. There was a study to increase the spatial resolution of the 2D ion chamber array. Spezi *et al*. proposed a multiple acquisition sequence with 2D Array Seven29 (PTW, Freiburg, Germany) to increase the spatial resolution of the 2D ion chamber array from 10 mm to 5 mm [7]. However, it asked four repeated irradiations with three table movements to increase the final spatial resolution. In this study, we proposed a novel method to enhance the spatial resolution of the 2D detector array for the patient-specific QA of IMRT with a single irradiation and no table movement. Another advantage of our method is that the volume effect of a detector element can be reduced. Because the active volume of each detector element is not a point, but a constant volume, it measures the averaged dose over the volume rather than the actual dose at the center of the volume. This is called the volume effect of detectors. Because of the volume effect, there is some error in measuring a point dose with the detector and this error becomes dominant in the high-dose gradient region. We could obtain the measured dose distribution that was less affected with the volume effect using the proposed method. We expect that the proposed method that provides the higher spatial resolution of dose assessment and the lower volume effect of the individual detectors can improve the overall quality of patient-specific QA.

II. MATERIALS AND METHODS

1. Theoretical considerations

A schematic diagram of the experimental setup for the patient specific QA of IMRT with a 2D detector array is illustrated in Fig. 1A. S indicates the radiation source, and O_1 is the point where the central axis intersects the measurement plane of the detector array. SO_1 is the source-to-detector distance (SDD). SR indicates a radiation beam path emitted from the source. SR intersects the detector plane at X_1. Black squares indicate detector elements. For the verification of the IMRT dose delivery, the 2D dose distribution on the detector plane was calculated and compared to the measurement. Normally, the measurement is performed at SDD=SO_1 which is the same configuration as the calculation. In this case the spatial resolution of the measured data is same as the spatial resolution of the detector array. If the measurement is made at the extended SDD=$SO_2 > SO_1$ as illustrated in Fig. 1B, the dose distribution along X_2O_2 normalized at O_2 is same as the one along X_1O_1 normalized at O_1 except that the spatial scale is increased by the factor $\rho \equiv SO_2/SO_1$. Because the doses at O_2

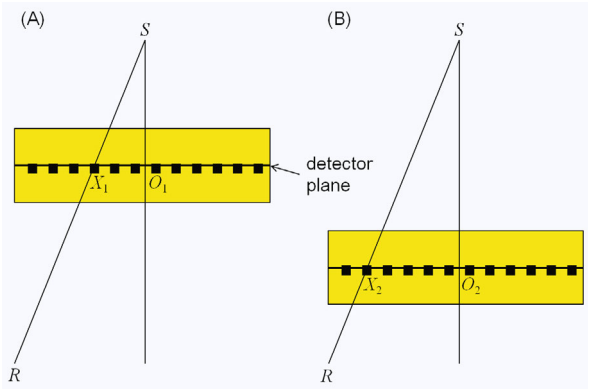

Fig. 1. (Color online) Schematic diagrams of a normal (A) and an extended SDD (B) setup for the patient specific QA of IMRT with a 2D detector array. S indicates radiation source. O_1 (O_2) is a point where the central axis intersects a measurement plane of the detector array. SR indicates a radiation beam path emitted from the source. SR intersects the detector plane at X_1 (X_2). Black squares indicate detectors.

and O_1 are related by the inverse-square-law with each other, the dose at X_1 is given by

$$D_1(X_1) = \rho^2 D_2(X_2), \tag{1}$$

where D_1 and D_2 are dose distributions along X_1O_1 and X_2O_2, respectively and $X_2 = \rho X_1$. From the measured dose distribution D_2 at the extended SDD=SO_2 the dose distribution at SDD=SO_1 can be obtained by Eq. (1). This method gives us more data points of D_1 than the normal measurement of D_1 at SDD=SO_1. As shown in Fig. 1, we have more detector elements between X_2 and O_2 than between X_1 and O_1. The distance between neighboring data points is decreased by the factor ρ.

Our method also reduces the volume effect of the individual detectors of the 2D detector array. The detailed analysis of the reduction of the volume effect is described in the followings. The dose $D_1^m(x)$ at x measured at SDD $= SO_1$ can be given by

$$D_1^m(x) = \int_{-\infty}^{\infty} D_1(u)K(x-u)du, \tag{2}$$

where $D_1(u)$ corresponds to the real dose profile at the detector position SDD $= SO_1$ and $K(x)$ is the response function of a single detector. The dose $D_2^m(X)$ at X measured at SDD $= SO_2$ is similarly given by

$$D_2^m(x) = \int_{-\infty}^{\infty} D_2(U)K(X-U)dU, \tag{3}$$

where $D_2(U)$ is the real profile at the detector position SDD $= SO_2$. Define $D_{12}^m(x)$ as

$$D_{12}^m(x) \equiv \rho^2 D_2^m(X), \tag{4}$$

where $X = \rho x$. From Eq. (3) and Eq. (1), $D_{12}^m(x)$ becomes

$$D_{12}^m(x) = \int_{-\infty}^{\infty} D_1(u)\rho K(\rho(x-u))du, \tag{5}$$

where $u = U/\rho$. The physical meaning of $D_{12}^m(x)$ becomes clear when we define new response function $K_2(x)$ as

$$K_2(x) \equiv \rho K(\rho x). \tag{6}$$

The detector with the response function $K_2(x)$ has the same properties as the detector with response function $K(x)$ except that the dimension in the x-axis is decreased by the factor ρ. The constant factor ρ in front of $K(\rho x)$ in Eq. (6) is a normalization factor. Using the definition of $K_2(x)$, Eq. (5) becomes

$$D_{12}^m(x) = \int_{-\infty}^{\infty} D_1(u) K_2(x-u) du. \tag{7}$$

The above equation implies that $D_{12}^m(x)$ is the dose at x measured at SDD = SO_1 with the detector whose response function is $K_2(x)$. Therefore we get the dose profile $D_{12}^m(x)$ at SDD = SO_1 measured with the smaller detector. $D_{12}^m(x)$ can be obtained from the measured data of $D_2^m(X)$ and Eq. (4).

2. Measurements

To demonstrate the proposed method, a patient specific QA of an IMRT plan was performed using the perpendicular field-by-field method with a 2D ion chamber array I'mRT MatriXX (IBA Dosimetry, Schwarzenbruck, Germany). The I'mRT MatriXX has 1020 vented parallel plate ion chambers arranged in a 32 by 32 grid. Each chamber has the dimension of 4.5 mm in diameter and 5 mm in height, and the sensitive volume of 0.08 cm^3. The distance between neighboring chambers is 7.62 mm. The I'mRT MatriXX has the inherent front buildup and the backscatter layer made of the water equivalent materials of about 0.3 cm and 3.5 cm, respectively. Additionally, a solid water phantom MULTICube (IBA Dosimetry, Schwarzenbruck, Germany) with the thickness of 4.7 cm was put on the MatriXX as a buildup layer. The thickness of an additional backscatter layer provided by the MULTICube was 1.4 cm as shown in Fig. 2. The IMRT plan and the corresponding QA plan were made using a treatment planning system Monaco V5.11 (Elekta, Stockhlm, Sweden) with the Monte Carlo dose calculation algorithm. The grid spacing for dose calculation was 3 mm. The QA plan was made for each field of the IMRT plan with the gantry and collimator angles set to 0. The source-to-surface distance was set to 95 cm in the QA plan so that the SDD was 100 cm. The measurement was performed using 10 MV x-ray of a linear accelerator VersaHD (Elekta, Stockhlm, Sweden) with the Agility multi-leaf collimator and HexaPOD six-dimensional (6D) couch. We only measured one field of the QA plan. The measurement was performed in two ways. First, we measured the dose distribution using the conventional method (SDD = 100 cm), which used the same configuration with the QA plan. Second, we measured the dose distribution using the proposed method. In the proposed method we extended the SDD to 127 cm and measured the dose distribution. The measured dose distribution at the SDD = 127 cm was converted using Eq. (1) (or Eq. (4)) to get the final dose distribution at the SDD = 100 cm. The value of $\rho = 1.27$ means that the distance between neighboring measurement points is decreased from 7.62 mm to 6.00 mm. The two measured dose distributions were compared to the calculated dose distribution using the Gamma index analysis with criteria (3%/3 mm). Only dose points larger than 10% of the maximum dose were included in the Gamma index analysis. Dose profiles along the inferior-superior direction from the three dose distributions were also compared.

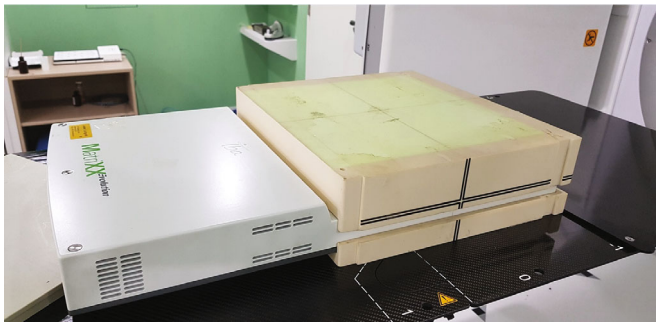

Fig. 2. (Color online) ImRT MatriXX 2D ion chamber array with solid water phantom MULTICube.

An open field with the dimension of 10 × 10 cm^2 was measured in the same measurement setup as the IMRT QA using flattened and flattening filter-free (FFF) x-rays of energy 6 MV and 10 MV to verify the reduction of the detector volume effect. For each energy, we measured the open field in two ways using the conventional (SDD = 100 cm) and the proposed method (SDD = 127 cm). As a reference dose distribution, the dose distribution of the open field was calculated for each energy using the Monaco V5.11 with the Monte Carlo dose calculation algorithm. The grid spacing for dose calculation was 3 mm. The calculations were performed in the same configuration with the conventional measurements. For each energy, dose profiles along the left-right direction at the center of the field from the two measured dose distributions were compared to the corresponding profile from the calculation. The dose profiles were centered and normalized at the center for comparison of the profiles in the penumbral region. The penumbral widths (20–80% distance) of the dose profiles were calculated and compared.

III. RESULTS AND DISCUSSION

Figure 3 shows 2D dose distributions of the IMRT field. Figure 3A indicates the calculated dose distri-

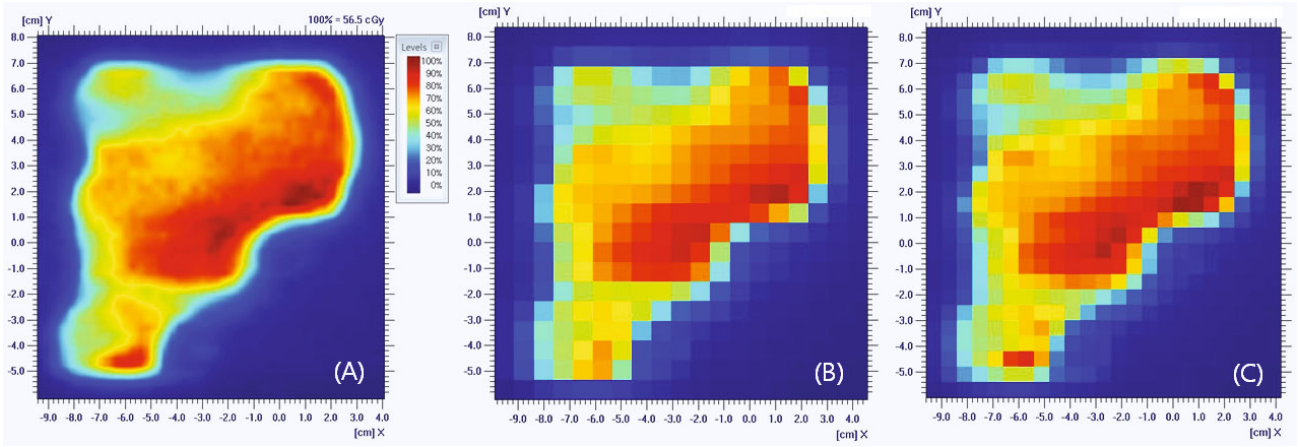

Fig. 3. (Color online) Coronal 2D dose distributions of the IMRT QA field calculated by the treatment planning system (A), and measured using the conventional (B) and the proposed (C) method.

Table 1. Penumbral width (20-80%) of the dose profiles along the left-right direction of the 10×10 cm^2 open field.

Energy	Conventional method	Proposed method	Calculation
6 MV	1.05 cm	0.87 cm	0.55 cm
10 MV	1.07 cm	0.87 cm	0.61 cm
6 MV FFF	1.17 cm	0.91 cm	0.69 cm
10 MV FFF	1.28 cm	1.18 cm	1.02 cm

bution by the treatment planning system. Figure 3B and 3C indicate the measured dose distribution using the conventional and the proposed method, respectively. The gamma index passing rates of the both measured dose distributions were 100%.

Figure 4 compares dose profiles along the y-axis (x = -5.7 cm in Fig. 3) from the IMRT QA measurements. Solid line indicates the calculated dose profile using the Monaco treatment planning system. Square dots indicate the measured dose profile using the conventional method (SDD = 100 cm), and triangle dots indicate the measured dose profile using the proposed method (SDD = 127 cm). Both measured profiles show similar trend. The number of triangle dots is 25, more than that of square dots 20. We got more measured data, which means the higher spatial resolution, using the proposed method. Needless to say, we can measure the dose distribution more precisely with more data points. As shown in Fig. 4, the proposed method could detect the first peak of the calculated profile (see the arrow) while the conventional method could not.

Figure 5 compares dose profiles of the open field for 6 MV (A), 10 MV (B), 6 MV FFF (C) and 10 MV FFF (D) x-rays. Dotted and dashed lines indicate measured dose profiles using the conventional (SDD = 100 cm) and the proposed method (SDD = 127 cm), respectively. Solid lines indicate calculated dose profiles.

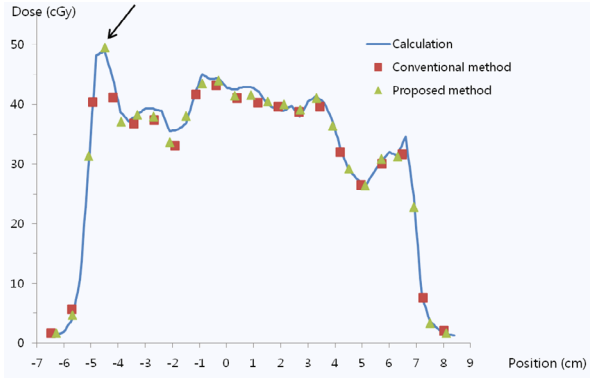

Fig. 4. (Color online) Dose profiles along the inferior-superior direction of the IMRT QA measurements. Solid line indicates the calculated dose profile using the Monaco treatment planning system. Square and triangle dots indicate dose profiles measured using the conventional (SDD = 100 cm) and the proposed (SDD = 127 cm) method, respectively. The arrow indicates the first peak of the dose profile.

The averages of the left and right penumbral width of the dose profiles are tabulated in Table 1. For all energies the punumbral widths of the measured profiles were greater than the widths of the calculated profiles. This is due to the volume effect of the ion chambers. The penumbral widths of the dose profiles measured using the proposed method were smaller than the widths using the conventional method. This shows that the volume effect observed with the proposed method is smaller than that observed in the conventional method.

Using the proposed method we could measure the dose distributions with the higher spatial resolution and the lower detector volume effect. The method needs only one measurement to get the result with the higher resolution whereas Spezi's multiple acquisition method needs multiple measurements. Our method can measure the dose distribution more precisely because the volume effect of

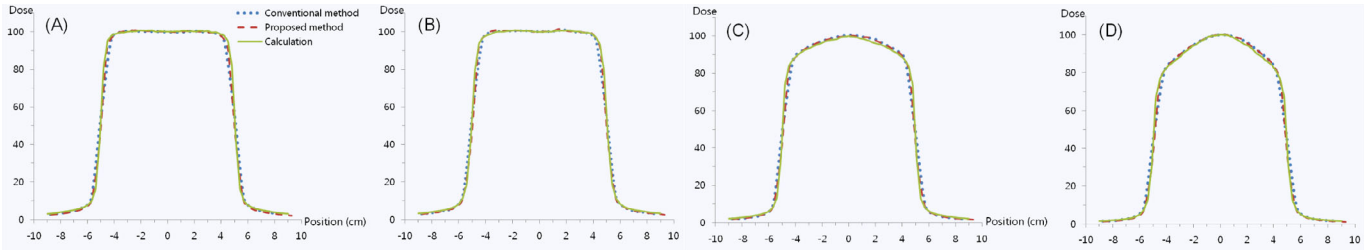

Fig. 5. (Color online) Dose profiles along the left-right direction of the open field 10 × 10 cm^2 for 6 MV (A), 10 MV (B), 6 MV FFF (C) and 10 MV FFF (D) x-rays. Dotted and dashed lines indicate dose profiles measured using the conventional method (SDD = 100 cm) and the proposed method (SDD = 127 cm), respectively. Solid lines indicate calculated dose profiles.

the detector is reduced. Theoretically, using the proposed method we can increase the spatial resolution of the 2D detector array as much as we need by increasing the SDD. Actually, however, there is a limitation to increase the SDD because of the limited table movement. For example to decrease the distance between neighboring measurement points from 7.62 mm to 5 mm, we need to measure dose distributions at the SDD = 152 cm. However, the HexaPOD 6D couch can't be lowered to the point where the SDD = 152 cm. This is why we selected the SDD = 127 cm, where the distance between neighboring measurement points becomes 6 mm. We can mitigate this limitation by setting the gantry angle to 270° and couch angle to 90° and putting the 2D detector array vertically on the couch so that the gantry is faced to the detecting surface of the 2D detector. In this setup, using the HexaPOD 6D couch we could increase the SDD upto 381 cm, which gives the distance between neighboring measurement points of 2 mm. One should be careful in setting this configuration to make the detecting surface of the 2D detector perpendicular to the central axis.

The AAPM (American Association of Physicists in Medicine) TG-218 report [8] recommended two IMRT QA delivery methods. One is the perpendicular field-by-field (PFF) method another is the true composite (TC) method. The proposed method is readily applicable to the PFF method where the gantry is faced to the detecting surface of the 2D detector. There are some reasons why the proposed method is difficult to apply to the TC method. Because the sensitivities of the detectors in the 2D detector arrays have dependence on the incident beam angles, the 2D detector arrays are typically used in the PFF method, not in the TC method [5,6,8]. There were some studies to correct this angular dependence [9–11]. Because the sensitivities of every detectors in the 2D detector array should be corrected for every gantry angles, it required extensive measurements and calculations to estabilsh the accurate correction method [11]. The TG-218 report recommened not to use the 2D detector arrays if the angular dependence is not accurately accounted for in the vendor software. Even though the angular dependence is corrected, to set the 2D detector array at the extended SDD, the detector should be moved to the different position whenever the gantry angle changes. This is time-consuming and not adquate for the routine clinical IMRT QAs. Moreover, there's a limit in the couch movement in the lateral direction. For example, the HexaPOD 6D couch can move upto about 25 cm from the center. This limits the distance of the SDD to about 125 cm when gantry angle is around 90° and 270°, which means that the smallest distance we can make between neighboring measurement points is 6.1 mm in case of the MatriXX detector. This limits the use of the proposed method. Another limitation of the proposed method is that the available field size we can cover using the 2D detector array is decreased as the SDD is increased. For example, the maximum field size that the MatriXX can cover is decreased from 24 × 24 cm^2 to 19 × 19 cm^2 when we increase the SDD from 100 cm to 127 cm. In general high spatial resolution is needed in the small field size. When an IMRT field size is larger than the maximum field size of the 2D detector array can measure, we could measure multiple times to cover the whole field and merge dose distributions to get the whole dose distribution.

IV. CONCLUSION

We proposed a novel and efficient method to enhance the spatial resolution of the 2D detector array and to measure the high dose gradient region more precisely for the accurate IMRT dose assessment in the patient-specific QA. The proposed method can provide a useful patient- specific QA procedure to improve the quality of IMRT dose assessment without additional exposure or hardware cost.

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