Deconvolution of Detector Size Effect Using Monte Carlo Simulation

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The detector size effect due to the spatial response of detectors is a critical source of inaccuracy in clinical dosimetry that has been the subject of numerous studies. Conventionally, the detector response kernel contains all the information about the influence that the detector size has on the measured beam profile. Various analytical models for this kernel have been proposed and studied in theoretical and experimental works. Herein, a method to simply determine the detector response kernel using the Monte Carlo simulation and convolution theory has been proposed. Based on this numerical method, the detector response kernel for a Farmer type ion chamber embedded in a water phantom has been obtained. The obtained kernel shows characteristics of both the pre-existing parabolic model proposed by Sibata et al. and the Gaussian model used by Garcia-Vicente et al. From this kernel and deconvolution technique, the detector size effect can be removed from measurements for 6MV, 10×10 cm^2 and 0.5×10 cm^2 photon beams. The deconvolved beam profiles are in good agreements with the measurements performed by the film and pin-point ion chamber, with the exception of in the tail region.

Key Words: Detector size effect, Deconvolution, Monte Carlo simulation

INTRODUCTION

In radiotherapy, the accuracy of the absorbed dose in areas where the gradient of the radiation field is large depends strongly on the spatial response of the dosimeter used. For example, the broadening of the beam penumbra due to the detector size has been a well-known experimental fact since Johns and Darby1) first reported. Thus in clinical dosimetry, the detector size effect dealing with the spatial response of detectors has been extensively studied by many investigators.2-5) According to these studies, correction is necessary to the measurements because of the existence of the detector size effect. This correction is especially important on measurements6) of penumbra where dose gradient is large. In the stereotactic radiosurgery or tomographic IMRT where narrow beams are used, the detector size effect becomes even more critical since high accuracy is required in the dose calculation, delivery and verification to save the critical structure located on the edge of the field. The standard ionization chambers of volume of order 0.1 cm^3 or less are commonly used for the dose verification. But still the inaccuracy related to the detector size is unavoidable and results in the penumbral increase of several millimeters. In addition to the detector size effect, further correction to the measured dose is required if the detector and phantom materials are different. Since the charged particle fluence is perturbed due to the phantom material, the measured dose may contain inherent degradation to be corrected. This is known as in-scattering effect7) and becomes important when measurements are made with detectors embedded in a phantom and transient charged particle equilibrium is not attained.

One of the standard techniques to remove the detector size effect is by deconvolving the detector response artifacts from

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the measurements.[4,5,8,9] This technique is based on the consideration that the measured beam profile is the convolution of the inherent beam profile with artifacts specific to the physical characteristics of the used detector. Commonly a representative detector spatial response function, which is also known as detector kernel, represents the physical response artifacts of a detector. Then the inherent beam profile can be obtained by deconvolving the kernel from the measured data. Such mathematical models for detector kernel as step functions, parabolic functions and Gaussian functions have been proposed and studied.[4,5,8,9] Sibata *et al.*[4]used a parabolic kernel while Garcia-Vicente *et al.*[5] found that the Gaussian kernel gave the best fit to their data.

In this work, we propose a method to calculate the detector kernel of an ion chamber instead of taking an analytic model for granted. Using Monte Carlo simulation and the convolution theory, the detector kernel can be determined. This kernel, together with deconvolution technique, enables us to determine the inherent beam profile from the experimental data. For the simulation, we consider a cylindrical Farmer type chamber embedded in a water phantom at 5cm depth and being irradiated by 6MV photon beam.

MATERIALS AND METHODS

With a representative detector spatial response function or kernel K for a dosimeter, measured dose profile D_m can be written as the convolution of the inherent dose profile D with K as[4,5,8,9]

$$D_m(x) = \int_{-\infty}^{\infty} D(u) \cdot K(u-x) \cdot du \qquad (1)$$

Using the typical Fourier transformation, the above equation can be written as

$$F[D_m\ (x)] = F[D(x) \cdot F[K(x)] \qquad (2)$$

where

$$F[f(x)] = \frac{1}{\sqrt{2\pi}} \int_{-\infty}^{\infty} f(x) \cdot \exp(2\pi wix) \cdot dx \qquad (3)$$

From Eq. (2), the inherent dose profile can be obtained by inverse Fourier transformation as

$$D(x) = F^{-1}\ [F(D_m\ (x))/F(K(x))] \qquad (4)$$

In order to obtain $D(x)$, earlier works[4,8,9] used various form of analytic model functions for detector kernel such as step functions, parabolic functions and Gaussian functions. Some other works[5,10] tried to determine the best detector kernel by checking which model gave the best fit to their experimental data. In this work, the detector kernel is determined by using Monte Carlo simulation instead of artificially selected analytic function or experimental data fitting. For the purpose, we consider a Farmer type ion chamber embedded in a water phantom at 5 cm depth from its surface and being irradiated by 6MV photon pencil beam. Both the dose profile by ion chamber, $D_{m,cham}(x)$, and the inherent dose profile at 5 cm depth, $D_{wat}(x)$, in a water phantom being irradiated by 6MV pencil beam are not measured but calculated using Monte Carlo simulation. Then the kernel of the ion chamber, $K(x)$, can be derived from the convolution equation as the following

$$D_{m,\,cham}\ (x) = \int_{-\infty}^{\infty} D_{wat}\ (u) \cdot K(u-x) \cdot du \qquad (5)$$

In the parabolic model for the kernel, $K(x)dx$ represents a strip of detector of width dx in proportion to the total detector size.[9] Similarly, the detector kernel in Eq. (5) represents the response of the detector to the pencil beam localized in a strip of the detector volume. Using the ion chamber kernel obtained, the inherent dose profile can be deduced by deconvolution of the experimental data made by real beams.

Accurate calculation of ion-chamber response is an important topic in radiation dosimetry. Based on the Spencer-Attix formulation of Bragg-Gray cavity theory, various issues related with Monte Carlo simulation of chamber response have been discussed thoroughly.[11,12,13] In these earlier works, the measured ionization by a chamber was assumed to be proportional to the absorbed dose to the gas in the cavity if transient charged particle equilibrium exists. Following them, we also assume $D_{m,cham}(x)$ is proportional to the absorbed dose to the cavity gas (air).

The Farmer-type ion chamber has a cylindrical wall made of graphite with inner radius 3.12 mm, outer radius 3.5mm and length 24cm. The radius of aluminum central electrode and the outer radius of acrylic build up cap are 0.5 mm and 8mm respectively. A slim rectangular shaped pencil beam of size 0.02×24 cm^2 irradiates the surface of water phantom (SSD=100 cm) with the ion chamber embedded at 5 cm depth. The stem effect is ignored in this simplified cylindrical model.

In most of Monte Carlo simulations related to the ion chamber study, the frequently used code is CAVITY or CAVRZ in EGS4/EGSnrc code system due to the existence of cylindrical symmetry. Here, instead we use the code DOSXYZ because cylindrical symmetry of the system studied is broken: the system consists of a cylindrical shaped chamber located in the water phantom with plane surface. Thus the chamber has to be divided into small voxels of size $0.1\times0.1\times24$ cm^3 to calculate approximately the total absorbed dose to the cavity gas. The following parameters are used for the simulation: maximum fractional energy loss per step, ESTEPE=0.01, step size, SMAX=0.2 cm, transport control energy parameters, ECUT =AE= 0.521 MeV and PCUT=AP= 10KeV.

RESULTS AND DISCUSSION

The simulated results of $D_{m,cham}(x)$ and $D_{wat}(x)$, where x is the relative distance between the chamber center and the photon pencil beam center, are shown in Fig. 1 as solid and dashed line respectively. Even when the photon beam is off the chamber edge, i.e. the relative distance x is larger than the chamber radius, the ion chamber reading is finite. It is due to the secondary components of the radiation. Using these results and simulated annealing method, we can determine the detector kernel, $K(x)$, for the simulated ion chamber as shown in Fig. 2.

This kernel differs from any models proposed and studied previously. However, similarly to the parabolic model by Sibata *et al.*,[4] this kernel has maximum value with the beam at the detector center and decreases as the relative distance increases. At the boundary, the kernel has non-zero value, implying that the detector has response to the radiation before it touches the field boundary. It is due to the secondary components of the radiation coming from the water phantom.

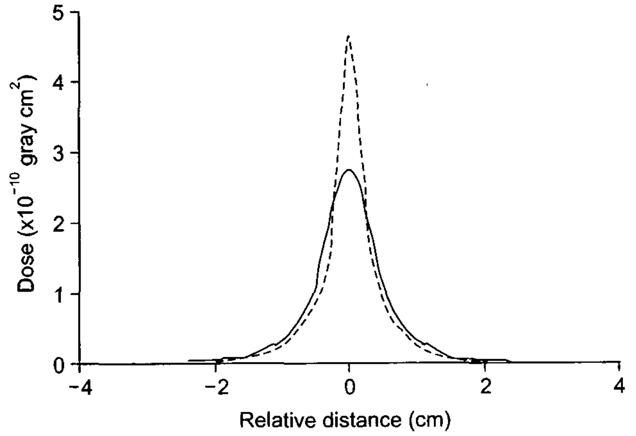

Fig. 1. The simulated dose profile by ion chamber, $D_{m,cham}(x)$, and the simulated inherent dose profile at 5 cm depth, $D_{wat}(x)$, are represented by the solid and dashed lines respectively. x is the relative distance between the chamber and beam centers.

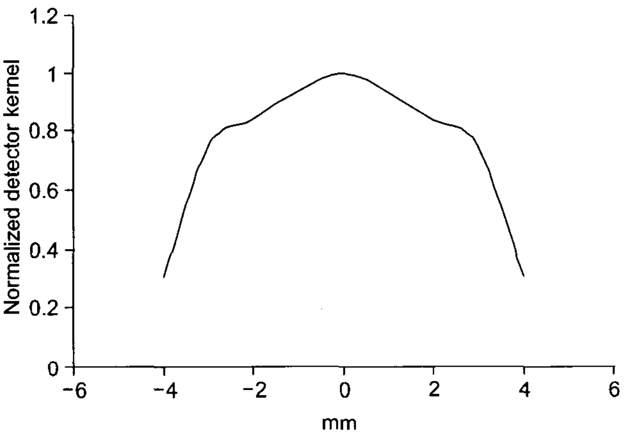

Fig. 2. Normalized detector kernel.

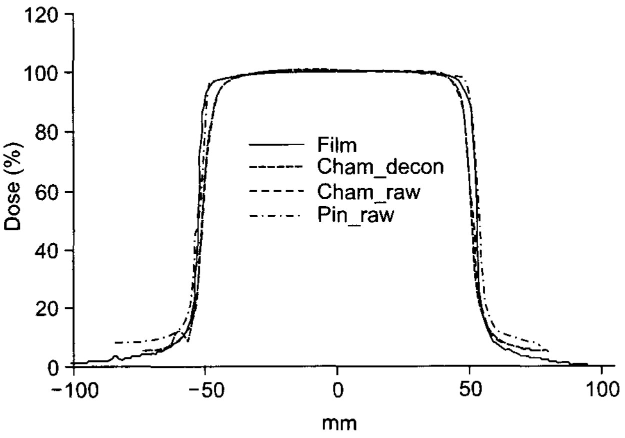

Fig. 3. Percentage dose profiles for a 10×10 cm^2 beam. Measurements by film, with an ion chamber of 0.6 cm^3 and pin-point chamber are plotted with a deconvolved profile.

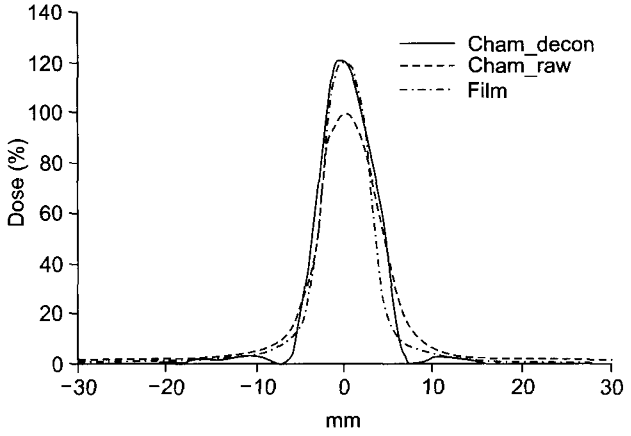

Fig. 4. Percentage dose profiles for a 0.5×10 cm^2 beam. Measurements by film, with an ion chamber of 0.6 cm^3 are plotted with a deconvolved profile.

Characteristically, this is similar to the Gaussian kernel obtained by Garcia-Vicente *et al.* via their data fitting method.[5)]

Fig. 3 shows the raw and deconvolved dose profile by ion chamber together with the film measurement for 6MV photon beam of size 10×10 cm^2. Also the data by pin-point ion chamber with supposedly negligible size effect is shown. To acquire the deconvolved data, we use the kernel in Fig. 2. Obviously, the deconvolved penumbra has higher shoulder height and narrower penumbral width. Even if we expected the tail to drop in the deconvolved profile, the existence of the ripple makes it not clear in the tail region. We have to investigate more this tail region and the physical/numerical origin of the ripple. The deconvolved beam profile is in good agreement with the measurements with Kodak XV-2 film and pinpoint ion chamber except in the tail region.

Fig. 4 shows the raw and deconvolved dose profile by ion chamber together with the film measurement for 6MV photon beam of size 0.5×10 cm^2. Compared with the deconvolved profile, the raw profile has lowerer peak. Since the chamber size is comparable to the field width, part of the chamber volume is out of the field even when the detector center and the beam center is in alignment. Therefore, the dose profile by the 0.6 cm^3 ion chamber is underestimated at the field center. The deconvolved data is in good agreement with the film data at the beam center but there exists discrepancy in the tail region. This may due to the well-known over-response of the film to the low energy charged particles or the lack of charged particle equilibrium.[4,14)]

CONCLUSION

We proposed a new method to determine the detector response kernel by using Monte Carlo simulation and convolution theory. Compared with artificially selected analytic models, the kernel obtained via this method may reflect the size effects for a detector in a phantom accurately. It differs from any models proposed and studied previously. After deconvolving the detector size effects from measurements, we find that the deconvolved beam profile is in good agreement with the measurements with film and pinpoint ion chamber except in the tail. This may show the validity of our numerically obtained detector kernel.

REFERENCES

1. **Johns HE, Darby EK**: Distribution of radiation near geometrical edge of x-ray beam. Br J Radiol 23:193-197 (1950)
2. **Brahme A**: Correction of easured distribution for the finite extension of the detector, Strahlentherapie 157:258-259 (1981)
3. **Rice RK, Hansen JL, Svensson GK, Siddon RL**: Measurements of dose distributions in small beams of 6 MV x-rays. Phys Med Biol 32:1087-1099 (1987)
4. **Sibata CH, Mota HC, Beddar AS, Higgins PD, Shin KH**: Influence of detector size in photon beam profile. Phys Med Biol 36:621-631 (1991)
5. **Garcia-Vicente F, Delgado JM Peraza C**: Experimental determination of the convolution kernel for the study of the spatial response of a detector. Med Phys 25:202-207 (1998)
6. **Dawson DJ, Schroeder NJ, Hoya JD**: Penumbral measurements in water for high energy x-rays, Med Phys 13: 1011-1014 (1986)
7. **Nilsson B, Montelius A**: Fluence perturbation in photon beams under nonequilibrium conditions, Med Phys 13: 191-195 (1986)
8. **Higgins PD, Sibata CH, Siskind L, Sohn JW**: Deconvolution of detector size effect for small field measurement. Med Phys 22:1663-1666 (1995)
9. **KaiSiung Chang, Fang Fang Yin, Kai-Wen Nie**: The effect of detector size to the broadening of the penumbra-a computer simulated study. Med Phys 23:1407-1411 (1996)
10. **Bednarz G, Saiful Huq M Rosenow UF**: Deconvolution of detector size effect for output factor measurement for narrow Gamma Knife radiosurgery beams. Phys Med Biol 47:3643-3649 (2002)
11. **Rogers DWO, Bielajew AF, Nahum AE**: Ion chamber response and A_{wall} correction factors in a ^{60}Co beam by Monte

Carlo simulation. Phys Med Biol 5:429-443 (1985)

12. **Rogers DWO**: How accurately can EGS4/PRESTA calculate ion-chamber response? Med Phys 20:319-323 (1993)
13. **Kawrakow**: Accurate condensed history Monte Carlo simulation of electron transport. II. Application to ion chamber response simulation. Med Phys 27:499-513 (2000)
14. **Laub WU, Wong T**: The volume effect of detectors in the dosimetry of small fields used in IMRT. Med Phys 30:341-347 (2003)

몬테카를로 시뮬레이션을 이용한 검출기의 크기효과 제거

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선량 측정기의 공간적인 반응특성 때문에 나타나는 detector의 크기효과는 임상적인 선량측정을 부정확하게 만드는 중요한 원인이기에 많은 연구의 대상이 되어왔다. 관례적으로 detector response kernel은 detector 자체의 크기가 측정한 방사선의 선량분포에 대해 미친 영향에 대한 정보를 포함하고 있다. 이 kernel에 대해 다양한 수학적 모델들이 제안되었고 실험적으로 이론적으로 연구되어왔다. 이 논문은 convolution이론과 Monte Carlo simulation만을 이용하여 detector의 kernel을 결정하는 방법을 제시한다. 이 수치해석적인 방법을 사용하여 물 phantom에 잠긴 Farmer형 ion chamber의 detector response kernel을 계산하였다. 계산된 kernel은 기존의 parabolic 모델의 특성과 Gaussian 모델의 특성을 동시에 나타내고 있다. 이 kernel과 deconvolution 방법을 사용하여 측정된 *6MV*, $10\times10\ cm^2$ $0.5\times10\ cm^2$ 광자선으로부터 크기효과를 제거하였다. 크기효과가 제거된 방사선의 선량분포는 꼬리부분을 제외하고는 film이나 pin-point ion chamber에 의해 측정된 결과와 유사한 선량분포를 나타냈다.

중심단어: 검출기 크기효과, Deconvolution, 몬테카를로 시뮬레이션