*Transactions of the Korean Nuclear Society Spring Meeting*
*Jeju, Korea, May 17-18, 2018*

# Performance evaluation of phantom dosimeter with organic scintillator array for measuring the penumbra of Co-60 tele-therapy unit using a CMOS camera

Hyun Young Shin [a], Shin Sang Hun [a], Jae Hyung Park [a], Hyungi Byun[a], Si Won Song[a], Ji Ye Kim[a], Young Beom Song [a], Bongsoo Lee [a*]
[a] *School of Energy Systems Engineering, Chung-Ang University, Heukseok-dong, Dongjak-gu, Seoul, 156-756, Republic of Korea*
[*] *Corresponding author: bslee@cau.ac.kr*

## 1. Introduction

The most important thing to perform radiotherapy is to deliver accurate dose to targeted treatment area. Till now, although advances in care equipment and dosimeters have made radiotherapy much more accurate, there are some disadvantages of conventional dosimeters such as a large sensor size, radiation damage, complicated calibration and not real-time measurement. And a high energy radiation is typically used for dose uniformity in the targeted treatment area, this can cause a loss of field flatness and increased penumbra width [1]. Normally, a considerable amount of radiation can be delivered to normal and cancer cells in the penumbra region. For the accuracy of radiation treatment, the precise dose distribution must be obtained in the penumbra region according to the depth.

The phantom dosimeter has many advantages to measure dose distribution in radiotherapy. In this dosimetric system, scintillating lights generated from organic scintillators are measured by a CMOS camera and it makes real-time measurement possible with ease calibration. Because it also has high spatial resolution due to the small size of organic scintillators, the penumbra area can be measured more precisely [2,3].

In this study, we fabricated a phantom dosimeter using organic scintillators to measure dose distribution for Co-60 tele-therapy radiation source. The array of scintillating lights was measured by CMOS camera and analyzed using MATLAB program. To evaluate the performance of the phantom dosimeter, the widths of penumbra were measured, and the results were compared to those obtained using a Gafchromic EBT3 film. Although the penumbra width is affected by many factors in radiotherapy, we focused on the beam size and the depth of the phantom in Co-60 tele-therapy dosimetry.

## 2. Materials and experimental setup

An organic scintillator (BCF-12, Saint-Gobain Ceramic & Plastics) with a diameter of 3.0 mm, was used as a sensing part to generate scintillating lights. The refractive index of the core and cladding are 1.60 and 1.49 and the numerical aperture (NA) is 0.58. The decay time and the emission peak are 3.2 ns and 435 nm, respectively. The structure of the phantom dosimeter is shown in Fig. 1. The total size of this dosimeter is 20 × 20 × 1 $cm^3$ and the interval between the organic scintillators is approximately 8.0 mm.

Scintillating light signals generated from the organic scintillators in the phantom dosimeter were simultaneously measured by the CMOS camera (UI-3580CP Rev. 2, IDS Imaging Development Systems GmbH). The image sensor of the camera consists of a number of square pixels with dimensions of 2.2 × 2.2 $\mu m^2$; its sensor size is 5.63 × 4.22 $mm^2$

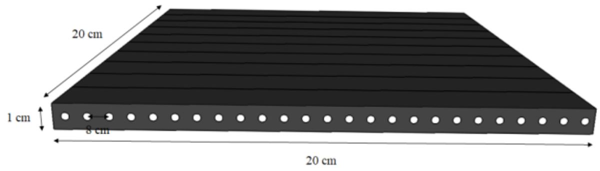


Fig. 1. Arrangement of the organic scintillators in the phantom dosimeter.

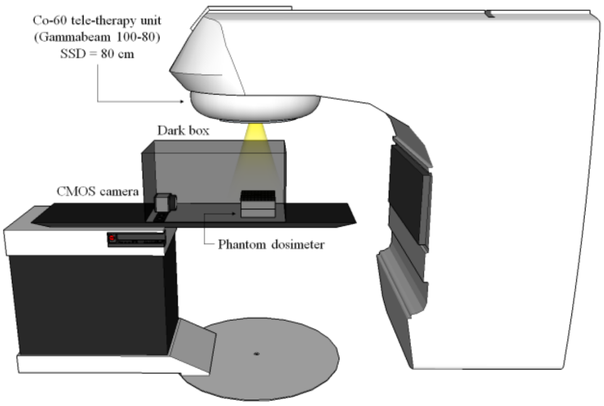


Fig. 2. Experimental setup.

The phantom dosimeter was placed at a source-to-surface distance (SSD) of 80 cm and the beam angle was set at 0°. EBT3 films were also placed at the same position with same angle.

## 3. Results

Figure 3 shows the dose distributions and the beam edges measured using the phantom dosimeter and the EBT3 film according to the field size. To evaluate the performance of the phantom dosimeter, gamma-rays generated by Co-60 were measured as a function of field sizes of 5 × 5, 10 × 10 and 15 × 15 $cm^2$. The PMMA stack phantom with 1 mm thickness was placed on the phantom dosimeter, scintillating lights measured with the CMOS camera were analyzed. Measured all data were normalized to 100% of the maximum and compared to those of EBT3 film.

The distributions of measured scintillating lights in Fig. 3 shows a sharp slope but not a perfectly perpendicular beam edge, because they are affected by the two kinds of penumbra such as transmission and physical, the scattered radiation, and the size of the scintillator in the phantom dosimeter.

Generally, to measure penumbra at the beam edge exactly the resolution of a dosimeter should be high. Therefore, the size and interval of the scintillator are very important factor for accurate dose measurements in the region of penumbra. If the size of scintillator is smaller than that of the proposed phantom dosimeter, the beam edge is measured more exactly.

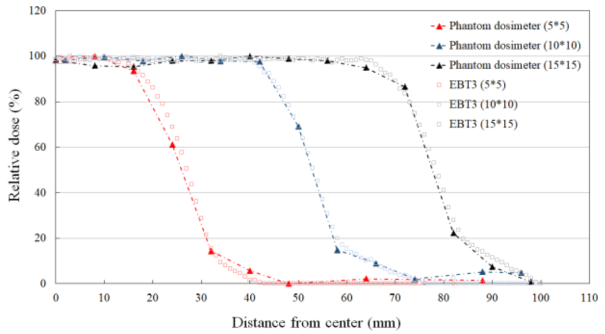


Fig. 3. Measured dose distributions at the beam edge according to the beam field size in the depth of 5 mm.

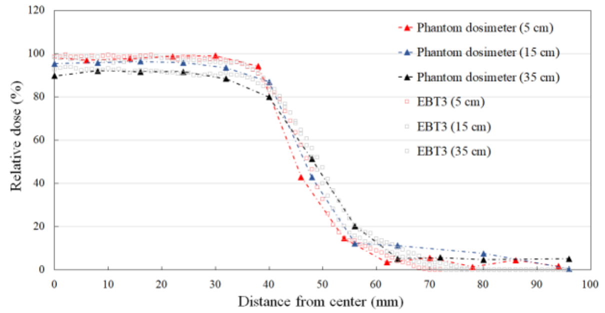


Fig. 4. Measured dose distributions with 10 × 10 $cm^2$ beam field size according to the depth of a phantom.

Figure 4 shows the measured dose distributions with 10 × 10 $cm^2$ beam field size according to the depth of a phantom. Dose distribution becomes flat with decreasing depth of the PMMA phantom. This result is attributed to an increase in scattered radiation in accordance with increasing depth of the phantom and decreasing photon energy [4].

In this result, the measured scintillating lights at the beam edge have a gentle slope with increasing depth because the penumbra region is also increasing. Actually, the penumbra width can be measured as the distance from 20% to 80% of relative dose [1].

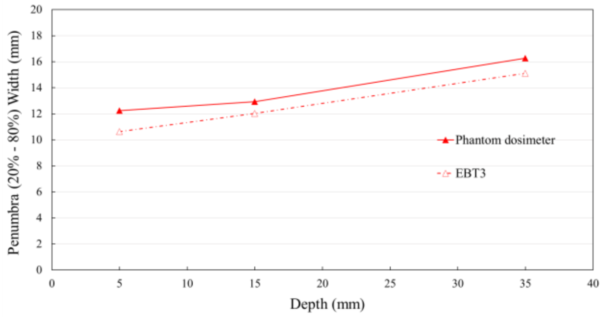


Fig. 5. Measured penumbra (20% – 80%) width at the different depths of a phantom.

Figure 5 shows the measured penumbra width at the different depths of a phantom. By using of the phantom dosimeter the penumbra widths were measured to be 12.3 mm, 12.9 mm and 16.3 mm at the depths of 5, 15 and 35 cm of a phantom, respectively. In Fig. 5, the penumbra width increases as the depth of a phantom increases due to increasing scattered radiations. The widths of penumbra measured using the phantom dosimeter are slightly overestimated compared to those of EBT3 film because the diameter of a scintillator is larger than the thickness of a film.

## 4. Conclusions

In this study, we fabricated a phantom dosimeter using organic scintillators to measure dose distribution of Co-60 tele-therapy source. The cylindrical type scintillators were used to fabricate the organic scintillator array. In order to evaluate the performance of the fabricated phantom dosimeter system, the beam profiles and the beam edge were measured using the transverse scintillator array with various kinds of beam field sizes and depths of the phantom. As experimental results, dose distribution becomes flat and the beam edge have a shaper slope with decreasing depth of the phantom.

Based on the results of this study, it is expected that the phantom dosimeter can be developed to accurately measure the dose distributions and the width of the penumbra in the radiotherapy dosimetry by reducing the sizes of scintillators and the intervals between them in the phantom dosimeter.

## REFERENCES

[1] S. A. Oh, M. K. Kang, J. Y. Yea, S. K. Kim, Y. K. Oh, Study of the penumbra for high-energy photon beams with Gafchromic EBT2 films, Journal of the Korean Physical Society, Vol. 60, p.1973, 2012.
[2] K. W. Jang, D. H. Cho, W. J. Yoo, J. Ki Seo, J. Y. Heo and B. Lee, A scintillating fiber-optic dosimeter for Co-60

radiotherapy, Proceedings of the IEEE SENSORS Conference, Oct.25-28, 2009, Christchurch, New Zealand.

[3] W. J. Yoo, S. H. Shin, D. Jeon, S. Hong, H. I. Sim, S. G. Kim, K. W. Jang, S. Cho, W. S. Youn, B. Lee, Measurement of entrance surface dose on an anthropomorphic thorax phantom using a miniature fiber-optic dosimeter, Sensors, Vol.14, p.6305, 2014.

[4] W. J. Yoo, J. Moon, K. W. Jang, K. T. Han, S. H. Shin, D. Jeon, J. Y. Park, B. G. Park, B. Lee, Integral T-shaped phantom-dosimeter system to measure transverse and longitudinal dose distributions simultaneously for stereotactic radiosurgery dosimetry, Sensors, Vol.12, p.6404, 2012.

## ACKNOWLEDGEMENT

This research was supported by Basic Science Research Program through the National Research Foundation of Korea (NRF) funded by the Ministry of Science, ICT and future Planning (No. 2017R1A2B2009480) and This research was supported by National Nuclear R&D Program through the National Research Foundation of Korea (NRF) funded by the Ministry of Science, ICT and future Planning (No. 2016M2B2B1945255).