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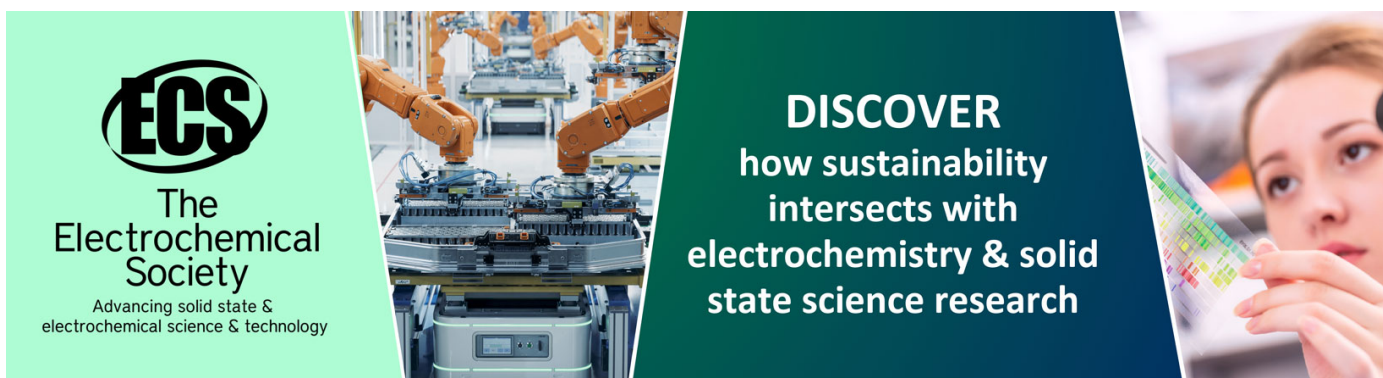
Basic study on evaluation of X-ray dose distribution using plastic scintillator plate and digital CMOS camera

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Basic study on evaluation of X-ray dose distribution using plastic scintillator plate and digital CMOS camera

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Abstract. Evaluation of dose distribution in a phantom is effective for quality control of radiation diagnostic equipment and evaluation of exposure. However, measurements at multiple points using a dosimeter is complicated. An alternative is to combine a plastic scintillator (PS) plate and digital camera to obtain the dose distribution of the entire phantom. In this study, the basic characteristics of this system were examined. A PS was sandwiched between Polymethyl Methacrylate (PMMA) disk phantoms, and irradiation was performed while the tube current was changed from one direction. The tube voltage was set to two values, specifically 60 and 120 kV. During irradiation, the entire phantom was recorded at 0.1 s intervals using a digital Complementary Metal Oxide Semiconductor (CMOS) camera. The lowest dose rate where luminescence could be detected was approximately 1 mGy per 0.1 s. The amount of luminescence increased linearly with the dose rate. The proportional relationship that was confirmed between dose rate and luminescence suggests the possibility of using the system for dose estimation. This proposed dose-evaluation method can enable the dose distribution to be evaluated with a time resolution of 0.1 s and a high spatial resolution.

1. Introduction

To provide medical care safely, it is important to control the quality of radiation diagnostic equipment, and to regularly check whether there is any abnormality and whether the equipment conforms to standards. For example, computed tomography dose index (CTDI) measurement is performed for the quality control of CT equipment [1, 2, 3]. Through CTDI measurement, the doses at the centre and periphery of a phantom are measured, enabling the dose distribution to be determined in a simple method [4]. To judge whether the quality control of CT equipment for radiotherapy should be corrected, calibration measurement is performed approximately once a month [5].

The exposure dose of CT examination is relatively high among radiological examinations [6, 7]. Thus, patient dose estimation is performed using thermoluminescent dosimeter (TLD) measurements and Monte Carlo simulations [8]. In addition, with the development of CT technology, the size-specific dose estimate (SSDE), which is determined through the application of CTDI, is used as one index [9].



Therefore, evaluating the dose distribution in a phantom is effective for the quality control of radiation diagnostic equipment and estimation of exposure dose.

In CTDI measurement, a 100-mm pencil ionisation chamber dosimeter [10] and a semiconductor dosimeter [11] are used to derive the CTDI₁₀₀ [12, 13]. However, this method can be complicated because it requires repeated measurements at multiple points, and because there are few measurement points for the entire phantom, the spatial resolution of the dose distribution is poor. The radiation dose can also be evaluated via a method that uses a film [14]; however, this method is expensive because the film cannot be used repeatedly.

Therefore, a method that combines a plastic scintillator and a digital camera was devised to obtain the dose distribution of an entire phantom in real time. The method involves inserting a plastic scintillator between phantoms and photographing the cross section with a camera to grasp the dose distribution in all areas of the cross section. In addition, the number of irradiation times is small because there is no part in the procedure where the dosimeter is moved depending on the measurement position, as in past methods. To accurately evaluate this dose-distribution measurement system, it is necessary to examine its basic characteristics, such as its feasibility of use, the range of dose rates that can be measured, and its energy dependence, if any. In this study, these basic characteristics were investigated.

2. Materials and methods

The general X-ray equipment used in this study was UD 150L—30 (SHIMADZU, Kyoto, Japan), and the digital CMOS camera used was ORCA-spark C11440-36U (Hamamatsu Photonics, Shizuoka, Japan). In the recording mode of this camera, continuous shooting is performed in time intervals equal to the sum of the exposure time and image reading time to obtain multiple images.

The image acquisition software used was a high-speed recording software (Hamamatsu Photonics, Shizuoka, Japan) corresponding to the camera. For the scintillator experiment, a 10-cm-thick PMMA phantom, which was assumed to be a CTDI phantom, was prepared using 10 PMMA disk phantoms (16 cm in diameter × 1 cm). The scintillator used was made of plastic and was of Luminade brand (TOKYO PRINTING INK MFG CO., LTD., Tokyo, Japan, 150×150×1 mm). The square scintillator was shaped into a circle to match the size of the disk phantom, and thus part of the original scintillator was missing in the experiment. A small solid detector, CT dose profiler (RTI Electronic, Sweden), and a CTDI phantom for the head were used to measure the air kerma.

2.1. Luminescence measurement using CMOS camera and scintillator

A scintillator was sandwiched between five PMMA disc phantoms. A phantom-sized box was prepared, and the phantom was fixed in position. The distance between the lens and phantom was 10 cm. During the study, the scintillator was covered with cloth to shield it from light. The phantom was irradiated with X-rays from the side, and the luminescence state was photographed in the recording mode (figure 1).

In this experiment, the measured characteristics were sensitivity, energy dependence, linearity of luminescence amount with dose rate. In the first part of this experiment, the difference in the luminescence amounts between one scintillator and two scintillators was determined. The amount of light emission was measured using a digital CMOS camera and a plastic scintillator. The irradiation conditions were as follows: tube voltage = 120 kV, X-ray tube focus-phantom distance = 50 cm, and irradiation time = 0.5 s. The dose rate was adjusted through changes in the tube current (10, 20, 50, 100, 160, 200, and 320 mA).



Figure 1. Irradiation of scintillator using general X-ray equipment. The scintillator is sandwiched between PMMA disk phantoms, and X-rays are emitted from the left side.

In the second part, differences in the luminescence amount due to differences in the tube voltage were evaluated. The luminescence amounts were compared under two tube voltages, specifically 120 kV and 60 kV, and two scintillators were used. The irradiation conditions were the same as those for the first part of the experiment. Similarly, the tube current was changed to adjust the dose rate.

In the third part, the linearity with respect to the dose rate was evaluated via measurement of the amount of luminescence at three points: on both ends and at the centre of the horizontal disk phantom for X-ray irradiation (A, B, and C were 15 mm, 80 mm, and 145 mm, respectively, from the phantom entrance surface). Two 0.5 mm thick plastic scintillators were stacked and used as a luminescent material. The tube voltage was 120 kV, and the tube current was 100 mA.

For each test, images were obtained three times, and the average value was calculated. Digital CMOS cameras were set up with a gain of 240, binning of 2×2 , and exposure time of 0.1 s. The pixel number was set to 960×600 , and the image was output in 12 bits.

2.2. Dose rate measurement

The air kerma was measured under the same conditions in which the luminescence was measured in subsection 2.1, using a small solid detector and a CTDI phantom. The air kerma at point A was measured to determine changes in the amount of light emission due to differences in the number of scintillators and differences in the tube voltage. The dose rate was calculated from the air kerma and irradiation time (0.5 s). For each condition, three measurements were obtained, and the average value was determined.

2.3. Image analysis

The image analysis software ImageJ 1.52a (Wayne Rasband, National Institutes of Health, USA) was used to measure the amount of luminescence. There were three or four images wherein the scintillator emitted light for one irradiation. The image was smoothed using a mean filter to suppress the effects of camera noise, after which the increase in the pixel value from the background was measured for each image, and the luminescence amount in 0.1 s was calculated. The pixel number per centimetre was calculated from the image based on the assumption that the cross-sectional diameter of the scintillator was 16 cm, resulting in 28 pixels/cm. The measurement range of the pixel value was 1 cm (28 pixels) in diameter, to measure the same size as that of a small solid detector. The pixel value measurement point was the same as the air kerma measurement point. Furthermore, the image acquisition time is the sum of the exposure time and reading time, where no light is received during the reading time because

no exposure is performed. Therefore, for the correction assuming that the exposure was performed during the reading time, the correction was performed by a coefficient obtained by dividing the total time of the exposure time and the reading time by the exposure time. Finally, the relational expression between the luminescence amount (pixel value) and dose rate was plotted.

3. Results

Figure 2 depicts the actual luminescence. This image was obtained from the left with X-rays and was captured with an exposure time of 0.1 s. Figure 3 visualizes the difference in luminescence depending on the number of scintillators. The proportionality coefficient of the linear approximation curve when there were two scintillators was 31.15, and the coefficient of determination was 0.9993. Meanwhile, the proportionality coefficient of the approximate curve for one sheet was 15.25, and the coefficient of determination was 0.9998. Moreover, the lower limit of the dose rate at which the increase in pixel value could be confirmed was approximately 1 mGy/0.1 s.

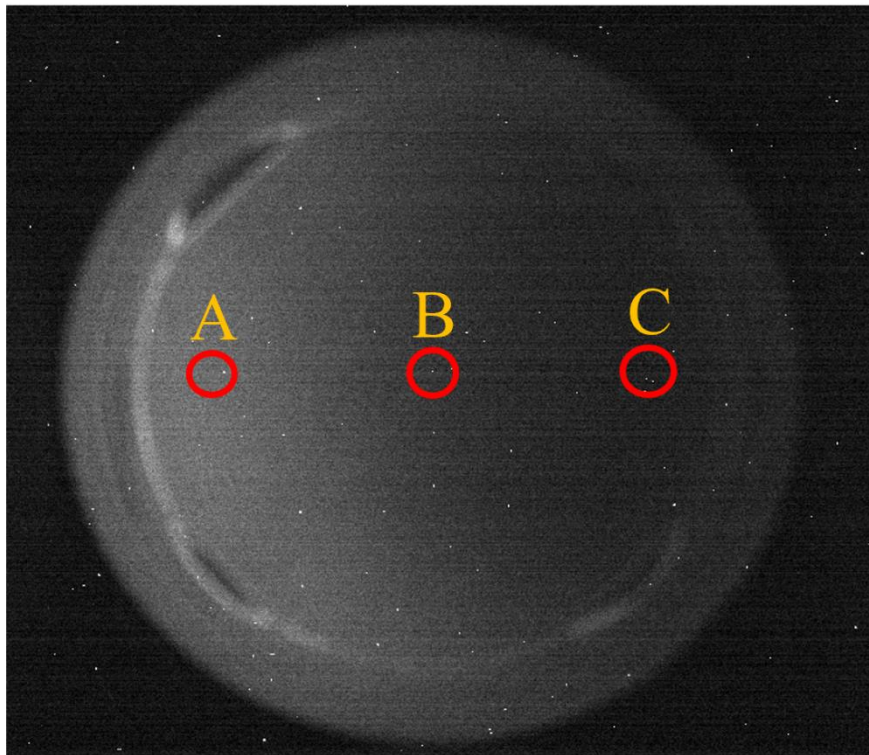


Figure 2. Luminescence of scintillator (120 kV, 100 mA, exposure time: 0.1 s). Irradiated from left side. Therefore, A has strong luminescence, and C has weak intensity. Circular luminescence is a cross section of the scintillator. Because a square scintillator is formed into a circular shape and its size is not complete, a portion of the luminescence is irregular at the peripheral portion.

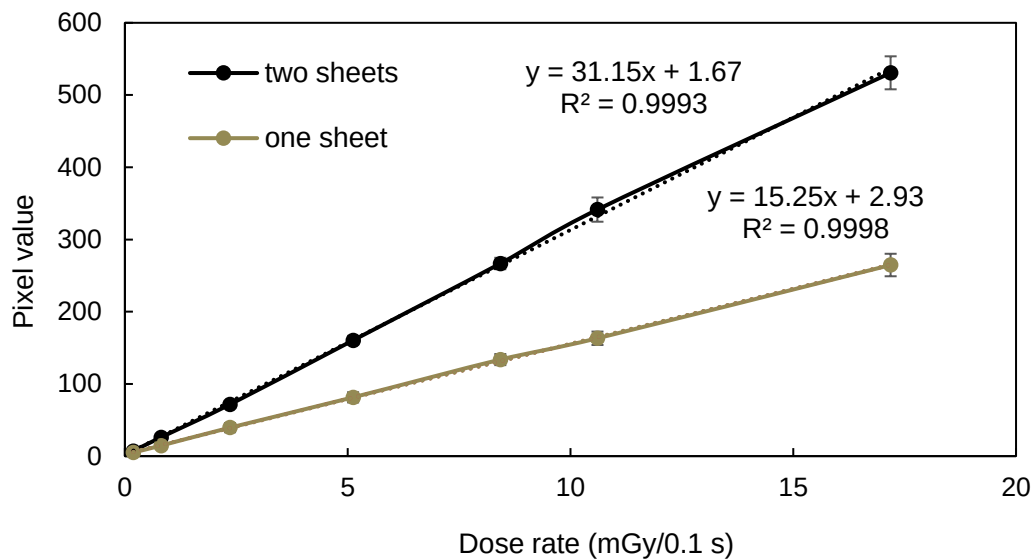


Figure 3. Increase in pixel value with respect to dose rate and the number of scintillators at point A .

Figure 4 shows the relationship between the increase in pixel value with respect to tube voltage and dose rate. In the case of 60 kV, the increase in pixel value was reduced by approximately 20% relative to that in the case of 120 kV at the same dose rate.

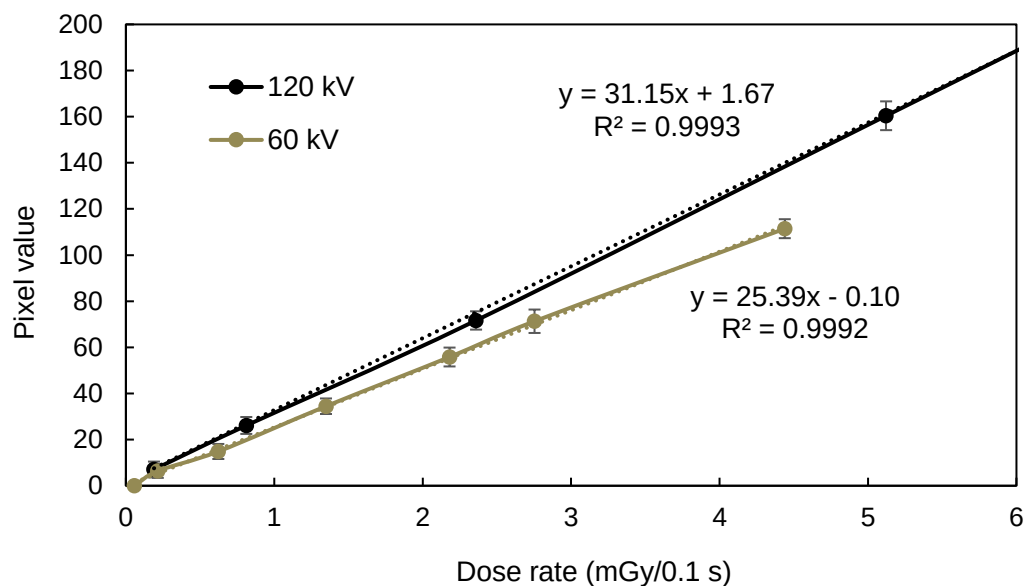


Figure 4. Increase in pixel value with respect to dose rate and tube voltage at point A .

Figure 5 shows the horizontal pixel value profile at 100 mA and the dose rates at points A, B, and C. The pixel value was higher for the dose rate in the central area than in the peripheral area.

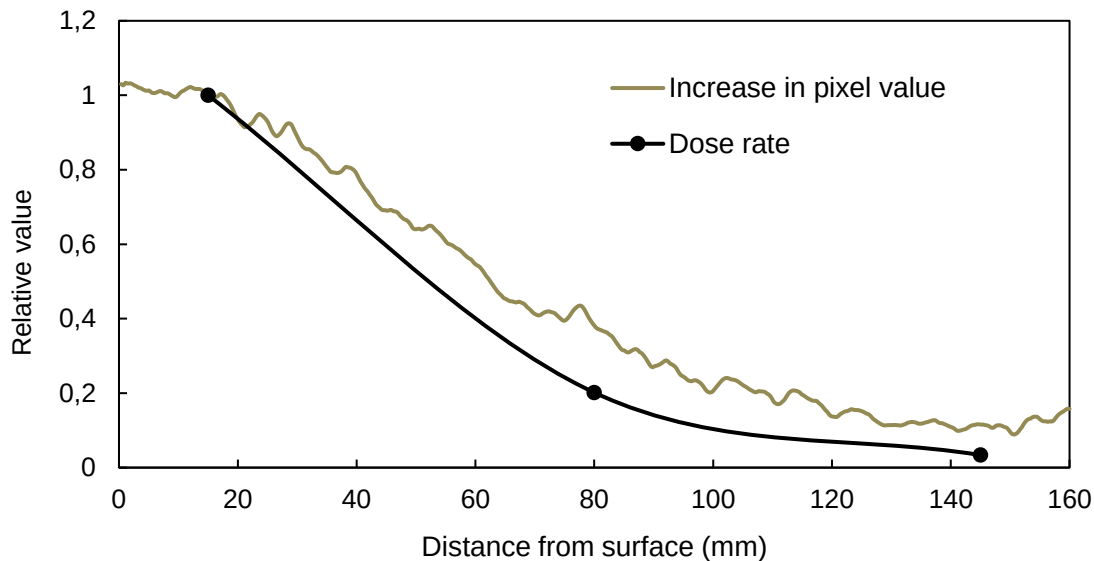


Figure 5. Profile of pixel value with respect to depth direction and dose rate at points A, B, and C. The value is standardized based on point A.

4. Discussion

As depicted in figure 3, the pixel value of this system also increases in proportion to the dose rate. Therefore, this result suggests that the dose rate can be measured from the luminescence of the scintillator. When the number of scintillators was increased to two, the amount of luminescence emitted doubled. Therefore, the sensitivity is increased when the number of scintillators is increased. However, if the number of scintillators is increased exceedingly, the scintillator itself will absorb the emitted light, and the pixel value will decrease [15]. Aside from the scintillator, the phantom may also absorb light. Therefore, the transparency and placement of the phantom must be considered in future experiments. Yamamoto of Nagoya University reported that PMMA emits radiation even when the energy is lower than the threshold of Cherenkov light [16]. Therefore, it is necessary to verify whether the phantom itself emits light.

Ideally, if there is no energy dependence, then there is only one straight line between the pixel value and dose rate for any tube voltage. However, as shown in figure 4, when the energy was low, the amount of luminescence decreased, and when the tube voltage was actually decreased from 120 kV to 60 kV, the increase in pixel value decreased by approximately 20%. This difference in the amount of emitted light due to energy may be caused by the different compositions of the scintillator and of the small solid detector. Therefore, this energy dependence may be solved through changes in the type of scintillator.

According to figure 5, there was a difference in the profile of the pixel value and the distribution of the dose rate in the deep direction for one image, especially in the central part. One of the causes may be the energy dependence described in the previous paragraph. Beam hardening may have also occurred as the depth from the incident surface increased. Another possible cause is that the phantom itself moved when the detector was moved, impairing reproducibility.

The advantage of this dosimetry device is that the entire dose distribution can be visually grasped with a single imaging. Furthermore, in the recording mode, multiple images at short time intervals are saved as a single series. Therefore, this system may be able to estimate the cumulative value of the air kerma at any given time. This system could also be used for exposure dose estimation.

In future research, improving the sensitivity will be important. Furthermore, in this measurement system, the phantom is transparent for light to not be blocked, and the scintillator must be a plastic scintillator with a composition close to that of a human body phantom. Given these limitations, a system with better performance must be developed.

5. Conclusion

The imaging system developed in this study, composed of a scintillator and digital CMOS camera, could confirm the proportional relationship between pixel value and dose rate, and that the pixel value was doubled through the use of two scintillators. The lower limit of the dose rate at which luminescence could be detected was approximately 1 mGy/0.1 s. Energy dependence was confirmed, and when the tube voltage was changed from 120 kV to 60 kV, the pixel value decreased by approximately 20% for the same dose rate.

References

- [1] AAPM 2008 The measurement, reporting and management of radiation dose in CT
AAPM report No. 96.(www.aapm.org/pubs/reports/RPT_96.pdf) (Accessed: 12 January 2021)
- [2] McNitt-Gray M F 2002 AAPM/RSNA physics tutorial for residents: topics in CT–radiation dose in CT *RadioGraphics* **22** 1541
- [3] AAPM Task Group 23 2008 The measurement, reporting, and management of radiation dose in CT *AAPM Report* No. 96
- [4] John A. Bauhs, Thomas J. Vrieze, Andrew N. Primak, Michael R. Bruesewitz and Cynthia H. McCollough 2008 *RadioGraphics* **28** 245
- [5] Kanematsu N 2003 *Japanese journal of medical physics* **23**(2) 140
- [6] Ruan C, Yukihiro E G, Clouse W J, Gasparian P B R, Ahmad S 2010 *Medical Physics* **37**(7) 3560
- [7] National Council on Radiation Protection and Measurements 2009 Ionizing radiation exposure of the population of the United States *NCRP Report* No. 160,
- [8] Lee C, Yoon J, Han S-S, Na J-Y, Lee J-H, Kim YH and Hwang JJ 2020 *PLOS ONE* **15**(3)
- [9] AAPM 2011 Size-specific dose estimates (SSDE) in pediatric and adult body CT examinations
AAPM Report No. 204 (www.aapm.org/pubs/reports/rpt_204.pdf) (Accessed: 4 February 2015)
- [10] Suzuki A and Suzuki M N 1978 *Med. Phys* **5** 536
- [11] Anam C, Fujibuchi T, Haryanto F, Widita R, Arif I and Geoff Dougherty 2019 *J Radiol Prot* **39** 112
- [12] Kalender W A 2014 *Phys Med Biol* **59** R129
- [13] Goldman L W 2007 *J. Nucl. Med. Technol* **35** 213
- [14] Sipila P, Ojala J, Kaijaluoto S, Jokelainen I, Kosunen A 2016 *J Appl Clin Med Phys* **17**(1) 360
- [15] Inada K 1971 *Environment Control in Biology* **8**(2) 111
- [16] Yamamoto S, Kato K, Abe S 2020 *Applied Radiation and Isotopes* **157** 109037