Progress in NUCLEAR SCIENCE and TECHNOLOGY, Vol. 3, p.72-75 (2012)

TECHNICAL MATERIAL

Evaluation of Maximum Dose Position Shifting in Phantom in X-ray CT

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In this study, the maximum dose position shifting in cone beam CT (CBCT) was analyzed in detail using Monte Carlo simulation method. The simulation was carried out on the assumption of 320 detector row CBCT. Our results show that maximum dose position is 18 mm depth from the phantom surface, and peak deposit energy is 11 % higher than the phantom surface dose. In the CBCT, peripheral position dose is underestimated using conventional weighted CTDI estimation method.

KEYWORDS: Monte Carlo simulation, cone beam CT, CTDI, percentage depth dose

I. Introduction

X-ray CT dose estimation is performed using computed tomography dose index (CTDI) which is weighted average of doses of center position and peripheral positions (10 mm depth) in cylindrical PMMA (Poly Methyl Meta Acrylate) phantom placed in a CT gantry for a single axial scan.

When Percentage Depth Dose (PDD) was simulated in plain radiography using Monte Carlo simulation (MC simulation), we found that maximum dose position was slightly shifted from phantom surface to deeper part. Wider x-ray field and higher x-ray tube voltage shifted maximum dose position from surface part to deeper part in the phantom and it seemed to be because of Compton scattering. Furthermore, we think that this effect will also occur in CT, especially in cone beam CT (CBCT), because it has wide x-ray field and high x-ray tube voltage. If maximum dose depth is more than 10 mm from phantom surface, conventional weighted CTDI ($CTDI_w$) might be underestimated because the peripheral dose positions disagree with maximum dose position. However, the measurement of dose distribution in phantom for kilovoltage x-ray beams is difficult because there are problems related to large gradients in the dose distribution and the large energy dependence of most dosimetry systems.[1)]

Therefore, in this study, the maximum dose position shifting in CBCT was analyzed in detail using MC simulation method.

II. Materials & Methods

1. Checking the conformity of MC simulation with measurement

It is needed to consider the conformity of MC simulation with measurement. Therefore, we simulated and measured $CTDI_{100}$ in a Non-helical X-ray CT unit TCT-300 (Toshiba Medical Systems, Tochigi, Japan). The x-ray tube voltage was 120 kV, and Source Center Distance (SCD) was 600 mm. PMMA cylindrical phantom of 300 mm diameter and 150 mm length and CT ionization chamber of 100 mm length were used. In the phantom, there were five cylindrical cavities in different depth along the cylinder axis to insert the CT chamber; The depths were 11, 59, 81, 115, and 150 mm from the phantom surface.

Figure 1 shows measurement geometry. The center of the phantom was placed in the isocenter of the CT unit. By the way, it is difficult to incorporate the effect of x-ray scattering and attenuation of CT patient bed in MC simulation because the composition of the bed has not been disclosed. Therefore, the phantom was supported using wooden blocks and the bed was removed from scan area to reduce this effect. In addition, it is also difficult to incorporate the effect of x-ray tube rotation overlapping in MC simulation because the overlapping angle has not been also disclosed. Therefore, a scanogram mode (non-rotating mode) which had 2 mm slice thickness in TCT-300 was performed. To be the same as a full scan, the phantom was rotated by 10 degrees up to full circle in a counterclockwise direction and each dose was integrated into a cylindrical cavity; In the measurement of the cavity, cavities at other depths were filled with PMMA rods.

In MC simulation, the simulation code which we used was Electron Gamma Shower ver.5 (EGS5)[2)]. EGS5 was a general purpose package for the MC simulation of the coupled transport of electrons and photons in an arbitrary geometry for particles with energies above a few keV up to several hundred GeV. The simulation geometry was the same as the above measurement geometry. The number of photons was 1.44×10^9. Fractional Standard Deviation (FSD) in the center of the phantom was less than 3.0 %. Typical CT scanner is equipped with the beam-shaping filter which is used to adjust the beam quality of x-ray after passing through a patient. This effect of the beam shaping filter was incorporated into MC simulation. Aluminum Half Value Layer (AL HVL) and dose value after passing through the beam shaping filter were measured using Rapidose (Radcal, Monrovia, CA) and these measured values were incorporated in MC simulation.

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Rapidose is a semiconductor detector which can measure dose and AL HVL at the same time and it is suitable for measurement of relatively high energy x-ray such as x-ray CT. When the effect of the beam shaping filter was measured, x-ray tube was fixed on the 0:00 position on its orbital. Rapidose was moved at every one degree along a fan beam of x-ray up to 19 degrees. Energy spectra as x-ray souce along the fan beam of the CT were generated by Tucker's method based on AL HVL measured at each angle.[3)]

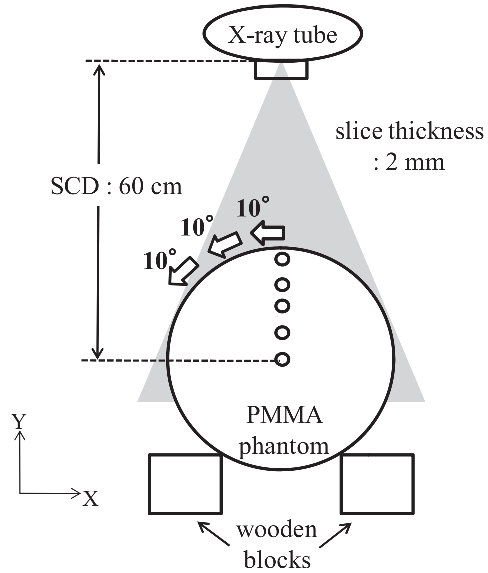

Fig. 1 Measuremet geometry.

2. PDD calculated by MC simulation in CBCT

PDD along the axis passed through the isocenter in a direction perpendicular to cylinder axis on the phantom cross section in CBCT was calculated using EGS5 to analyze the maximum dose position shifting. The simulation was carried out on the assumption of 320 detector row CBCT. The 320 detector row CBCT's geometry was applied to previous TCT-300 geometry (The x-ray tube voltage was 120 kV, and SCD was 600 mm), but the slice thickness was only changed from 2 mm to 160 mm **Figure 2** shows MC simulation geometry in the CBCT. The x-y coordinate plane is parallel to the transverse axis, and the z coordinate axis (z axis) is parallel to the longitudinal axis of cylindrical phantom. The x coordinate axis (x axis) is horizontal to the floor, and the y coordinate axis (y axis) is perpendicular to the floor. Phantom was PMMA cylinder of 320 mm diameter and 600 mm length. The number of photons was 3.6×10^9. FSD in the center of the phantom was less than 1.0 %.

We calculated PDD curve along the y axis passed through the isocenter on the phantom cross section. PDD was calculated per 1.0 mm depth. The size of calculation region of PDD was 10.0 mm (x), 1.0 mm (y), and 100 mm or 300 mm (z). The size of z direction was based on the assumption of 100 mm or 300 mm CT chamber. Energy deposition by primary x-ray and scatter x-ray was obtained separately.

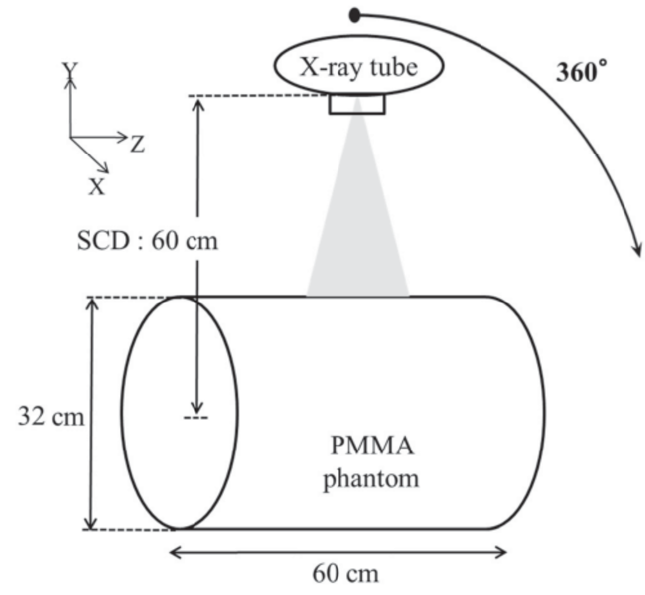

Fig. 2 MC simulation geometry in CBCT.

III. Results

1. Checking the conformity of MC simulation with measurement

We checked the conformity of MC simulation with measurement (**Table 1**). The depth showed the distance from the phantom surface to the center of CT chamber. Each depth dose was normalized to 11 mm depth dose in the simulation and the measurement.

We calculated Percent Average Error (PAE). PAE is given as follows:

$$PAE = \frac{D_m - D_s}{D_m} \times 100 \quad [\%] \tag{1}$$

where D_m was the normalized measured dose, D_s was the normalized calculated dose.

In **Table 1**, PAE was within ±2.4 % in all depths.

Table 1 The conformity of simulation with measurement.

depth from the phantom surface [mm]	normalized dose		PAE [%]
	simulation (Ds)	measurement (Dm)	
11	1.0	1.0	0.0
59	0.94	0.93	-1.5
81	0.84	0.85	2.2
115	0.77	0.77	-0.3
150	0.71	0.73	2.4

2. PDD calculated by MC simulation in CBCT

We calculated PDD along the y axis passing through the isocenter and obtained deposition energy by primary x-ray and scatter x-ray separately.

Figure 3 shows that PDD curve which was calculated on the assumption of 100 mm CT chamber. **Figure 4** shows that PDD curve which was calculated on the assumption of 300 mm CT chamber. In both **Figure 3** and **Figure 4**, solid line indicates total deposition energy; broken line indicates deposition energy which only comes from Compton scattering component of x-ray; and dotted line indicates deposition energy which comes from primary component of x-ray. The depth indicates the distance from the upper surface to the lower surface of the phantom. Each deposition energy was normalized to the deposition energy in the surface region.

In **Figure 3**, peak dose of total deposit energy was 11.0 % higher than phantom surface dose. The position of maximum dose was 18.0 mm depth from the phantom surface. In scatter component of x-ray, maximum dose was 17.0 % higher than the phantom surface dose. The position of maximum dose was 25.0 mm from the surface. In primary component of x-ray, the position of peak dose was on the phantom surface.

In **Figure 4**, peak dose of total deposit energy was 22.0 % higher than phantom surface dose. The position of maximum dose was 30.0 mm depth from the phantom surface. In scatter component of x-ray, maximum dose was 33.0 % higher than the phantom surface dose. The position of maximum dose was 40.0 mm from the surface. In primary component of x-ray, the position of peak dose was on the phantom surface.

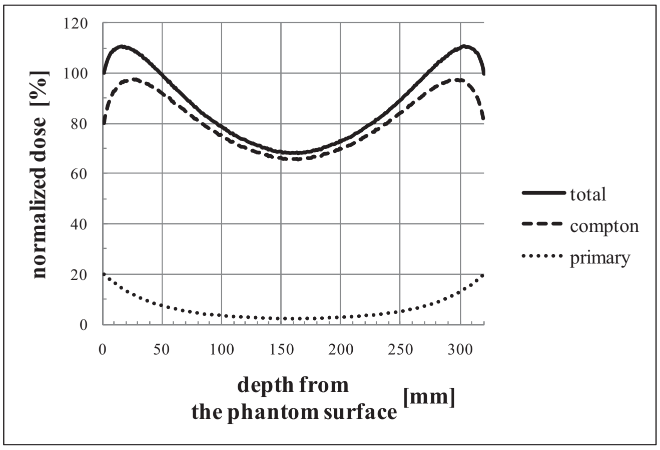

Fig. 3 PDD curve which was calculated on the assumption of 100 mm CT chamber.

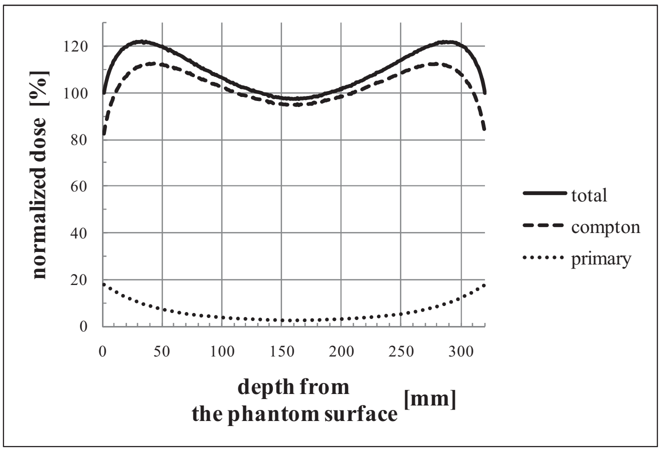

Fig. 4 PDD curve which was calculated on the assumption of 300 mm CT chamber.

IV. Discussion

We first discuss the conformity of MC simulation with measurement. **Table 1** shows PAE which represents the error between simulation and measurement. PAE is within ±2.4 % in all depths from the phantom surface. This result indicates that there is a good agreement between MC simulation and measurement.

In **Figure 3** and **Figure 4**, detailed analysis of energy deposition finds that Compton scattering is contributed largely to maximum dose position shifting. In the International Electrotechnical Commission (IEC) standard, dose estimation in CT scanner is now performing using $CTDI_{100}$ which is derived from measurements of 320 mm diameter and 150 mm long CTDI phantom and 100 mm CT chamber. However, in 320 detector row CBCT, Geleijns et al. said that $CTDI_{100}$ underestimated $CTDI_{300}$ ($CTDI_{300}$ is derived from measurements of 350 mm long CTDI phantom and 300 mm CT chamber.) [4-5]and Mori et al. said that the length of both the CTDI phantom and CT chamber needed to be more than 300 mm for dosimetry of CBCT[6].

From our simulation of 100 mm or 300 mm CT chamber, the effect of maximum dose position shifting is greater with the 300 mm CT chamber than with the 100 mm CT chamber. This is because the 300 mm length along the direction of z axis covers wider x-ray beam width in 320 detector row CBCT and the energy deposition of Compton scatter component of x-ray is higher.

Our results show that, in the existing CTDI estimation of CBCT, the dose of peripheral position of 10 mm depth is underestimated because maximum dose position shifts to the deeper part (more than 10 mm depth), and the effect of maximum dose position shifting is remarkable in the 300 mm CT chamber. We consider that this effect is very important for CTDI estimation. Therefore, we think that it is better to measure CTDI in the condition that the 300 mm CT chamber is used and the peripheral position is not 10 mm depth but maximum dose position depth.

V. Conclusion

In this research, we calculated PDD in the phantom using MC simulation method in order to analyze the maximum dose position shifting in CBCT. In previous research, the need of long length CT chamber and CTDI phantom was suggested in CBCT. We think that it is also important to consider about the measurement depth of peripheral positions in CTDI phantom in CBCT.

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