Monte Carlo Simulation of the effects of tube current modulation on each organ doses in X-ray CT

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Purpose

The dosage of x-ray radiation from CT (Computed Tomography) scanner has been estimated recently by using a Monte Carlo simulation. The purpose of this study is to calculate accurate radiation dose in patient from CT examination using voxel phantom. In the calculation, scanner-specific descriptions that include filtration designs, and absorption of the CT table are necessary. For example, a special filter called "beam-shaping filter" which is positioned in front of the x-ray tube of the CT units causes less uniform x-ray fan beam. We took into account the effect of the filter based on measured data in the simulation.

MIRD (Medical Internal Radiation Dose) phantom based on westerner body shape has been used for estimation of exposure dose. The simulated exposure dose figures are different between westerner and Japanese, because the model size of Japanese is smaller than westerner. Accordingly, a standard Japanese Model phantom was developed by using the voxel phantom from CT DICOM (Digital Imaging and Communication in Medicine) format data. By using in-phantom dosimetry system which can be measure exposure dose of each organ, the calculation data was compared with measurements.
2.1 Overview

All simulations were performed using the EGS5 (Electron Gamma Shower ver.5) Monte Carlo code, which is a general purpose package for the Monte Carlo 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. For measurements and simulations, an X-ray CT unit Aquilion 64 (Toshiba Medical Systems, Tochigi, Japan) was used. Measurement and calculation were performed on thoracic scan protocol with a tube voltage of 120kV.

2.2 Effect of beam-shaping filter

A typical CT scanner is equipped with x-ray beam filtration that includes the beam-shaping filter. The attenuations across the fan beam were different due to the filter. The beam-shaping filter is used to adjust the beam quality of x-rays after passing through a patient. There is no such published data on the design of beam-shaping filter, which may vary considerably from scanner to scanner. For estimation of the beam-shaping filter, two types of measurements were obtained: Aluminum Half Value Layer (AL HVL) and dose profiles, namely, attenuation of beam-shaping filter. We measured aluminum Half Value Layer (AL HVL) and dose value after passing through the filter and incorporated change of energy spectrums and photon number based on measured values in the simulation. CT pencil ionization chamber was used for the measurements. In the measurement of the effect of beam-shaping filter, x-ray tube was positioned on the 0:00 direction on its orbital. Along a fan beam of x-ray CT, CT pencil ionization chamber was moved from 0 degree (downward vertical beam) up to 18 degrees (downward oblique beam) and measured at every three degree. It is assumed that the attenuation profile in the axial plane is symmetric about the central ray, so only measuring the half of the filter’s attenuation is sufficient. In the simulation, photon number at each beam angle along the fan beam was determined by dose profiles and energy spectrums were generated by Birch’s formula based on AL HVL measured on each angle.

2.3 Attenuation by the CT table

CT table top is usually made of carbon fiber due to its strength and low x-ray attenuation properties. The table was composed of density adjusted carbon in the simulation because the manufacture has not also published information on its actual density and composition. The density of the carbon was adjusted to correspond with the actual dose ratio table(+)/
The effect of attenuation of the table was developed in the simulation. These factors were also incorporated in our simulation code.

### 2.4 Measurement and calculation of CTDI

Computed Tomography Dose Index (CTDI) measured by using a CT acrylic cylindrical phantom was used for accuracy verification of our simulation. The CTDI phantom is 15 cm in length with a diameter of 30 cm. CTDI values are shown in air kerma (kinetic energy released in matter; mGy) in the phantom's center hole and four peripheral holes (0°, 90°, 180°, 270° position) when the x-ray tube rotates around the acrylic phantom.

Dose at peripheral four positions are CTDI peripheral (CTDI100,p), dose at center position is CTDI center (CTDI100,c). A relative value was given by:

\[
p-c\text{ ratio} = \frac{CTDI_{100,p}}{CTDI_{100,c}} (1)
\]

In order to determine the air kerma from CT examination, simulated deposit energies in units of MeV/gram/source particle were converted to air kerma in units of mGy by a conversion factor (C_f) [1]. An in-air normalization method that is based on pencil-chamber exposure reading for a single axial scan was taken at the center of the CT gantry. The conversion factor was defined as:

\[
C_f = \frac{CTDI_{100,air,\text{measured per mAs}}}{CTDI_{100,air,\text{measured per mAs}}} (2)
\]

where CTDI_{100, air, measured per mAs} is measured by an ion chamber free in air of the x-ray CT isocenter, and CTDI_{100, air, simulated per mAs} is obtained by simulation.

The absorbed dose \( D_{ab} \) is calculated by:

\[
D_{ab} = D_{sim} \times C_f (3)
\]

where \( D_{sim} \) is the calculated dose in the simulation.
2.5 In-phantom dosimetry system

In-phantom dosimetry system was used for verification of simulation using voxel phantom. In this system, small semiconductor dosimeters were implanted into phantom (THRA-1, Kyoto Kagaku Co., Kyoto, JAPAN)(Fig. 1) and these dosimeters were located at the centroid of the organ and tissue. As a result, a number of organ doses were obtained immediately undergoing x-ray CT examination. This system is called "in-phantom dosimetry system" and was used to compare measurement and simulation.

Fig.: 1. Overview of in-phantom dosimetry system

References: - Nagoya JP

2.6 Development of voxel phantoms
A voxel phantom was developed based on CT image. The resolution of axial slice of all CT image was 512×512. The resolution of the axial simulation matrix was decreased from 512 × 512 to 170 × 170 to decrease runtime of the simulation. The size of each voxel was 0.1875×0.1875×1.0 cm³. Three different types of phantoms were used for calculations. One of them (THRA-1) is only composed of lung, bone and soft tissue but can obtain organ doses using a number of internal semiconductor dosimeter installed at the position of tissues.

Another phantom called CTU-41 (Kyoto Kagaku Co., Kyoto, JAPAN) includes the shapes of organs with individual densities. The average radiation dose to each organ was estimated in CTU-41 phantom. Both "in-phantom dosimetry system" and CTU-41 phantom were scanned by x-ray CT, and image data was output as DICOM format. The organs and tissues were assigned in each voxels using digital value (Hounsfield Unit).

The other phantom was developed base on CT DICOM data of an actual patient. Differences between individuals could be evaluated by using actual patient's data. This phantom was defined as "patient phantom".

Original CT image and converted voxel image were shown in Figure 2. The simulations were performed using these three voxel phantoms.

![Figure 2: Compare original DICOM data (a) and Voxel phantom (b)](image)

**Fig.** 2. Compare original DICOM data (a) and Voxel phantom (b)

**References:** Nagoya/JP
2.7 Automatic tube current modulation

X-ray tube current modulation, a popular feature available in newer CT scanners, improves image quality but may result in a higher radiation dose. This function is essential for accurate simulation of organ dose from CT examination. Only x-ray tube current information of lateral view and an antero-posterior (AP) scanogram were outputted on the console of the CT device. Therefore, sinusoidal wave was applied to interpolate between two x-ray tube currents. Obtained current data was used as incident photon weight in the simulation. In this study, measurements and simulations were compared in case of without the current modulation (350 mA), and with the current modulation.
Images for this section:

Fig. 0: 2. Compare original DICOM data (a) and Voxel phantom (b)

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**Fig. 0:** 3. Comparison of measurement of in-phantom dosimetry system and simulation

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Fig. 0: 4. Comparison of measurement of in-phantom dosimetry system and simulation of the voxel phantoms

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Results

3.1 Comparison of CTDI between measurement and calculation

Table 1 shows comparison of measurements and calculations of p-c ratio. The difference between simulation and measurement result was calculated to evaluate the accuracy of the simulation. The agreement between measurements and calculations was within 3.6%. The p-c ratio was lowest at 180° position due to the absorption of the CT table. The simulated p-c ratios were approximately equal to measured p-c ratios. With this result, the x-ray source and the geometry of the x-ray CT could be incorporated in the simulation successfully.

Table 1 Measurement and Simulation of p-c ratio

<table>
<thead>
<tr>
<th>Location</th>
<th>Measurement</th>
<th>Simulation</th>
<th>Difference [%]</th>
</tr>
</thead>
<tbody>
<tr>
<td>Center</td>
<td>1.00</td>
<td>1.00</td>
<td>#</td>
</tr>
<tr>
<td>Top</td>
<td>1.81</td>
<td>1.75</td>
<td>3.61</td>
</tr>
<tr>
<td>Right</td>
<td>1.82</td>
<td>1.82</td>
<td>-0.52</td>
</tr>
<tr>
<td>Left</td>
<td>1.77</td>
<td>1.83</td>
<td>-3.17</td>
</tr>
<tr>
<td>Bottom</td>
<td>1.53</td>
<td>1.55</td>
<td>-1.39</td>
</tr>
</tbody>
</table>

3.2 Result of calculation using voxel phantom

First, measurements and simulations were compared in case of without current modulation (350mA constant current). Result of the comparison of measurement and calculation of the in-phantom dosimetry system is shown in Figure 3. Locations of the calculation point were accorded with that of semiconductor dosimeters in the in-phantom dosimetry system. Measurement and calculation showed the same tendency on the same location in the phantom. The differences between measured organ doses and calculated organ dose within 4.2 %.

Figure 4 shows comparison of the organ doses between the in-phantom dosimetry system and the CTU-41 phantom and patient phantom. Dose values of the in-phantom dosimetry system were measurement values of centroid of each organs, and those of
the CTU-41 phantom were calculation values of average dose of each whole organs converted from deposited energy. Calculated dose of the organ in the scan area such as lung was approximately equal to measurement dose, but the calculation dose of organ placed over the end of scan area such as thyroid resulted in significantly different from measured dose. In the calculation, the deposited energy in whole organ was divided by the organ mass and this value was presented as an average organ dose. When the organ was partially irradiated in the end of the scan area, average organ dose was smaller than the whole-organ irradiated case. In the measurement, the organ dose was presented maximum value as is the case in whole-organ irradiation if the evaluated point as a centroid of the organ fell within the scan area.

![Graph showing comparison of measurement and simulation](image)

**Fig.**: 3. Comparison of measurement of in-phantom dosimetry system and simulation

**References**: - Nagoya/JP
Fig.: 4. Comparison of measurement of in-phantom dosimetry system and simulation of the voxel phantoms

References: - Nagoya/JP

3.3 Effect of x-ray tube current modulation

Comparison of with current modulation and without current modulation (350mA constant current) using the simulation shows table2. Organ doses were decreased from approximately 1 to 11 mGy. Especially, organ dose in a scan area showed considerable decrease. Organ doses were decreased 3.66 mGy for lung, 5.20 mGy for esophagus, 7.43 mGy for heart when x-ray tube current modulation was taken into account in the simulation.

Table.2 Dose reduction with current modulation

<table>
<thead>
<tr>
<th></th>
<th>Without modulation</th>
<th>With modulation</th>
<th>Dose change [mGy]</th>
</tr>
</thead>
<tbody>
<tr>
<td>Thyroid</td>
<td>33.14</td>
<td>22.27</td>
<td>-10.88</td>
</tr>
</tbody>
</table>

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<table>
<thead>
<tr>
<th>Organ</th>
<th>First Measurement</th>
<th>Second Measurement</th>
<th>Difference</th>
</tr>
</thead>
<tbody>
<tr>
<td>Lung</td>
<td>28.86</td>
<td>25.20</td>
<td>-3.66</td>
</tr>
<tr>
<td>Esophagus</td>
<td>30.39</td>
<td>25.19</td>
<td>-5.20</td>
</tr>
<tr>
<td>Liver</td>
<td>25.63</td>
<td>21.17</td>
<td>-4.46</td>
</tr>
<tr>
<td>Heart</td>
<td>33.55</td>
<td>26.12</td>
<td>-7.43</td>
</tr>
<tr>
<td>Thymus</td>
<td>29.59</td>
<td>30.36</td>
<td>0.77</td>
</tr>
<tr>
<td>Spleen</td>
<td>25.97</td>
<td>25.37</td>
<td>-0.60</td>
</tr>
<tr>
<td>Pancreas</td>
<td>25.19</td>
<td>23.93</td>
<td>-1.26</td>
</tr>
<tr>
<td>Kidney</td>
<td>20.59</td>
<td>21.35</td>
<td>0.75</td>
</tr>
</tbody>
</table>
Conclusion

In this study the effect of the beam-shaping filter was successfully implemented in the simulation. The specific factor of CT unit such as the effect the attenuation of CT table was also incorporated in the simulation. Comparison of calculation and measurement by CTDI showed good agreement, and Measurement of in-phantom dosimetry system and calculation using voxel phantom showed good agreement at the same location in the phantom. In the simulation, dose distribution similar to the measurement was produced by using voxel phantom. The benefit of our method is that the effect of the beam-shaping filter can be incorporated easily if there isn’t information like material or shape of the actual beam-shaping filter. Detailed dose distribution in the voxel phantom was obtained using the simulation. The calculation dose of organ placed over the end of scan area such as thyroid resulted in significantly different from measured dose. In the measurement, care should be taken when organ dose is estimated in such organ.
1) J Gu et al, The development, validation and application of a multi-detector CT(MDCT) scanner model for assessing organ doses to the pregnant and the fetus using Monte Carlo simulations, Physics in medicine and biology

2) Radiation dose evaluation in 64-slice CT examinations with adult and paediatric anthropomorphic phantoms