Fast skin dose estimation system for interventional radiology

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ABSTRACT

To minimise the radiation dermatitis related to interventional radiology (IR), rapid and accurate dose estimation has been sought for all procedures. We propose a technique for estimating the patient skin dose rapidly and accurately using Monte Carlo (MC) simulation with a graphical processing unit (GPU, GTX 1080; Nvidia Corp.). The skin dose distribution is simulated based on an individual patient’s computed tomography (CT) dataset for fluoroscopic conditions after the CT dataset has been segmented into air, water and bone based on pixel values. The skin is assumed to be one layer at the outer surface of the body. Fluoroscopic conditions are obtained from a log file of a fluoroscopic examination. Estimating the absorbed skin dose distribution requires calibration of the dose simulated by our system. For this purpose, a linear function was used to approximate the relation between the simulated dose and the measured dose using radiophotoluminescence (RPL) glass dosimeters in a water-equivalent phantom. Differences of maximum skin dose between our system and the Particle and Heavy Ion Transport code System (PHITS) were as high as 6.1%. The relative statistical error (2σ) for the simulated dose obtained using our system was ≤3.5%. Using a GPU, the simulation on the chest CT dataset aiming at the heart was within 3.49 s on average: the GPU is 122 times faster than a CPU (Core i7–7700K; Intel Corp.). Our system (using the GPU, the log file, and the CT dataset) estimated the skin dose more rapidly and more accurately than conventional methods.

Keywords: Monte Carlo simulation; MC-GPU; radiation dose; IR

INTRODUCTION

Interventional radiology (IR) produces minimally invasive surgery for patients, but the number of cases accompanied by radiation dermatitis has increased recently. Generally, IR operations require a long examination time. Furthermore, complex procedures have increased in number. High doses have led to increased radiation injury to patients and staff members [1–3]. Many reports in the relevant literature claim that the patient exposure dose depends on the type of procedure [3–5]. Therefore, monitoring the exposure dose as accurately as possible, preferably in real time, is extremely important to avoid injury from radiation.

Nevertheless, conventional methods of dose evaluation in IR are difficult. Moreover, they provide uncertain results. For example, radiophotoluminescence (RPL) glass dosimetry is well known for its excellent precision and good properties such as stability (negligible fading) and uniformity of products [6]. The use of these dosimeters, however, requires great effort that entails manual handling, annealing, preheating, and readout. Particularly, annealing takes 1 h at 400°C. Preheating takes 1 h at 70°C.

A dose area product (DAP) meter can measure DAP, which is a product of the irradiated dose and the area of the exposure fields. The International Electrotechnical Commission (IEC)
specifies 25% tolerance for this device [7]. This uncertainty value is insufficient for dose control. In addition to this shortcoming, and unfortunately by its definition, DAP loses information related to exposed positions.

Recently, several real-time dosimeters have become commercially available [8–12]. However, they measure doses only at one or at a few points on a patient.

The Monte Carlo (MC) method is well known for use as a dose calculation algorithm. Nevertheless, it has also been notorious for its requirement of long durations to produce highly precise estimates.

Recently, one means of reducing the calculation time was proposed by Badal et al. The method uses graphical processing units (GPUs). Actually, MC-GPU v1.3 [13] is a Monte Carlo simulation package for use with GPUs to generate synthetic radiographic images in the public domain.

A few real-time skin dose estimation methods for IR have been reported [14, 15]. They estimate skin dose on one standard human model using a theoretical mathematical model. No report of the relevant literature has described a study using an individual patient’s body along with the MC method for estimating the detailed skin dose distribution for IR. This paper presents a proposal of a novel fast dose estimation system for IR (FDEIR) using an individual patient’s computed tomography (CT) dataset. Fluoroscopic conditions extracted from examination records were fed to the simulation system automatically. In this way, the patient skin dose is calculable immediately after fluoroscopic examination, with accuracy of 6.1%.

**MATERIALS AND METHODS**

**MC-GPU–based system**

FDEIR is based on MC-GPU, a public domain MC simulation package that can simulate synthetic radiographs and computed tomography (CT) scans using the computational power of commercially available GPU cards [13]. The package uses the PENELOE 2006 subroutine [16] for photon interaction models, photon transport models, and material cross sections. The Woodcock tracking algorithm [17] for photon transport and suppression of electron transport is also implemented to accelerate calculation in the package. For calculations, the package requires the x-ray source position, the source trajectory for the CT scan, the x-ray energy spectrum, a random seed, the number of simulated histories and GPU that threads use, a voxelized trajectory for the CT scan, the x-ray energy spectrum, a random seed, is also implemented to accelerate calculation in the package. For calculation algorithm. Nevertheless, it has also been notorious for its requirement of long durations to produce highly precise estimates.

For MC simulation, FDEIR requires fluoroscopic conditions that include tube voltage, milliampere–second (mAs) values, source–to-image-receptor distance (SID), field of view (FOV) size, source position, and imaging table position, which are all included in a fluoroscopic log file. For every exposure, FDEIR reads them out.

The energy spectra are indispensable for MC simulation. They were computed in an earlier study using spectrum computation software with the Birch and Marshall model [18, 19].

In FDEIR, a CT phantom is constructed from the patient CT dataset before dose calculation. Then, FDEIR reads in the phantom and material data, along with the calendar time (seconds since 1 Jan 1970) (as a random seed to prevent production of the same result) and the fluoroscopic condition from the log file in random access memory (RAM). Subsequently, these data are copied to the GPU. Next, FDEIR executes the MC simulation. Outputs simulate the skin dose distribution. After simulation with each fluoroscopic condition, the simulated dose distribution is converted into the absorbed dose distribution using a linear regression equation, as described in the *Computing scaling factor section* herein.

To validate FDEIR, a comparison with Monte Carlo code Particle and Heavy Ion Transport code System (PHITS) ver. 2.88 [21] and the measurements was made. Electron Gamma Shower Version 5 (EGSS) [22] is incorporated for photon and electron transport in PHITS. Their transport in EGSS was confirmed [23, 24]. This validation ensured that FDEIR has a plausible transport model and no need for electron transport. For comparison, the percentage depth dose (PDD) curves for a 76 kV x-ray beam were evaluated in water. The electron transport was simulated in PHITS. The cut-off energy was 5 keV in each case. An RPL glass dosimeter (GD-352M; Chiyoda Technol Corp.) was used for measurements. An annealing process was applied before exposing the RPL glass dosimeter to x-rays. The dose was measured every 2.5 cm along the x-ray beam axis (Fig. 1).

Additionally, we compared the dose at the center of the surface of the water-equivalent phantom as determined by FDEIR, PHITS and measurement. We used the same geometry presented in Fig. 1.

Similarly, we compared the relative skin dose distribution between FDEIR and PHITS. PHITS constructs a voxelized phantom using the DICOM2PHITS program, which is a part of PHITS for inputting the CT DICOM dataset [21, 25]. Actually, FDEIR constructs it using a technique described in the *Voxelization from a CT dataset section* herein. For this comparison, we assumed for cardiac fluoroscopy that the chest was exposed to the x-ray beam in the posterior–anterior (PA) direction (Fig. 2).

**Computing scaling factor**

The original MC-GPU returns the absorbed dose divided by the number of incident photons. Determination of the number of
incident photons from a fluoroscopic apparatus makes it possible to estimate the x-ray beam dose that is absorbed. We therefore defined the number as a scaling factor. The number of incident photons depends on the energy spectrum and the irradiation field size. Therefore, the scaling factor must be defined for every energy spectrum and irradiation field size. For this study, we adopted a 76 kV spectrum, 20 cm FOV and 100 cm SID. The number of incident photons is proportional to the mAs value. The scaling factor was also associated with the mAs value.

The measured absorbed dose was proportional to the simulated relative dose at a reference point. The reference point is the center of a boundary between a water-equivalent phantom and an imaging table. The geometry is presented in Fig. 1.

The absolute doses at the reference point were measured using a RPL glass dosimeter (GD-352M) with application of the annealing process. The dose figures represent the average of two RPL glass dosimeters placed at the reference point.

For the simulation, a voxelized geometry was constructed with similar geometry of the measurement (Fig. 1). The fluoroscopy conditions were extracted from the log file. Then, the relation between the simulated dose and that measured using RPL glass dosimeters in a water-equivalent phantom was approximated using a linear function with a zero intercept. The linear regression equation between them was $y = 0.203x$.

Voxelization from a CT dataset
For this study, a voxelized phantom was constructed from an anonymized chest CT dataset (1.074 × 1.074 × 5 mm, Fig. 3) provided by Teikyo University Hospital. To reduce statistical error and loading time of the phantom, the CT dataset was downsampled to 5.37 × 1.074 × 5 mm by taking the average of pixel values.

Air, water and bone were distinguished within the CT dataset, based on pixel values, to allocate the tissue type for each pixel. The values of air, water and bone are mutually distinct, which simplifies their categorization. In contrast, categorizing organs and soft tissues is difficult because their values are very similar. Organs and soft tissues have values that are approximately equal to that of water. We treated organs and soft tissues, except for skin, as water for construction of the voxelized phantom.

The phantom skin was assumed to be one voxel layer at the outer surface of the body. The voxel thickness is 1.074 mm, which is approximately equal to the reference value of the ICRP: 1.3 mm [26]. Material data for air, water, bone and skin were obtained from the MC-GPU package.

Attenuation and scattering at an imaging table are important factors for calculating the skin dose. Therefore, we added a graphite board (3.8 cm Innova 4100-IQ; GE Healthcare) to the phantom as the imaging table at Teikyo University Hospital. Material data were based on the database from PENELOPE. A slice of the voxelized phantom is portrayed in Fig. 3.

Rapid dose calculation
At the start of each simulation, the MC-GPU reads the voxelized phantom and material data from a hard drive to memory. When using the chest CT phantom, the reading time is ~1 s. It is a large fraction of GPU simulation time. FDEIR reads these data only once at the beginning of the process.

Fig. 1. Geometry arrangement for measurements and simulations to evaluate dose in the water-equivalent phantom. In the simulation, the imaging table is assumed to be a graphite board. Glass dosimeters are sandwiched between MIX-R and the imaging table. Source-to-surface distances (SSDs) were 60 cm and 50–70 cm, respectively, for PDD and the scaling factor.

Fig. 2. Geometry arrangement for skin dose simulation in FDEIR and PHITS. The phantom size is 43.0 × 25.8 × 36.0 cm$^3$. The beam was incident on the center of the phantom, aimed at the heart.

Fig. 3. Volume rendered 3D chest CT dataset (left) and a slice of the voxelized phantom at the heart level (right). The voxelized phantom is segmented into five regions: air, water, bone, skin and graphite.
To measure the calculation time, we simulated 10 million and 100 million histories with Case A (Table 1) in FDEIR on the computer. Similarly, these simulations were executed in PHITS. We compared the maximum skin doses estimated using FDEIR, PHITS, and the dose recorded in the log file. The log file shows values calculated using a commercial apparatus (Innova 4100-IQ; GE Healthcare) at Teikyo University Hospital. The values also include maximum absorbed dose records in mGy units for each exposure. However, the value was rounded to the nearest integer. The recorded doses were extracted from the log file with each fluoroscopy condition. In FDEIR and PHITS, the maximum skin dose was estimated using these conditions. The conditions are presented in Table 1. The geometry is depicted in Fig. 2.

### RESULTS

#### System validation

Validation of FDEIR was performed using a comparison with PHITS. Figure 4 shows PDD curves evaluated using FDEIR, PHITS, and measurement in the water phantom. The difference between them, i.e. 

\[
\frac{(\text{our FDEIR} - \text{PHITS dose})}{\text{PHITS dose}}
\]

was 4.3% on average. That result implies that FDEIR has plausible transport of photons, with no necessity of electron transport in the diagnostic energy range.

#### Compute absorbed dose

Figure 5 presents measured and simulated doses with the least-squares fitting straight line for computing the scaling factor. The linear regression equation between them was

\[
y = 0.203x
\]

The coefficient of determination, the square of Pearson’s correlation coefficient \(r \approx 0.998\), was 0.995. That result shows that measured doses and simulated ones are proportional and that linear regression is plausible. For this study, we defined the linear regression equation as the scaling factor. We provide a couple of examples of the absolute skin dose distribution in Figs 6 and 7. They show the absorbed skin dose distribution for the part of the skin in the chest CT phantom. The maximum dose was estimated as 0.573 mGy/mAs. The error in the irradiation field was 5.8% on average. The difference derives from statistical error or the construction methods for the phantom, which was made using image thresholding in FDEIR; in PHITS, the phantom was constructed using DICOM2PHITS.

#### Comparison of the absorbed doses

The absorbed doses at the center of the surface of the water-equivalent phantom were evaluated. The measured dose was 3.66 \(\pm 0.18\) mGy when the phantom was exposed to the x-ray beam for 30 s with Case B (Table 1). The dose was calculated with FDEIR as 3.71 mGy under the same conditions. This result and the others are

![Fig. 4. PDD curves evaluated using FDEIR, PHITS, and measurement in a water phantom for 76 kV x-ray beam at 60 cm SSD. Error bars show variations in the reading reproducibility (5%).](image)

\[y = 0.203x\]

![Fig. 5. The scaling factor to convert from relative dose to absolute dose is the slope of the line. Points on this graph correspond to the measurements and simulation dose on the geometry shown in Fig. 1. Error bars show variations in the reading reproducibility (5%).](image)

Table 1. Fluoroscopic conditions of X-ray beams aimed at the heart

| Case | SSD | kV | FOV | SID | Direction |
|------|-----|----|-----|-----|-----------|
| A    | 50  | 76 | 20  | 100 | PA        |
| B    | 60  | 76 | 20  | 100 | PA        |
| C    | 70  | 76 | 20  | 100 | PA        |
| D    | 60  | 76 | 20  | 100 | RAO30     |
presented in Fig. 8. The difference between the absorbed dose from FDEIR and that from measurement was 2.9% on average.

Regarding the skin dose, the log file recorded the absorbed dose as 3 mGy and the mAs value as 0.16234 per second when the phantom was exposed to the x-ray beam for 30 s with Case B (Table 1). We calculated the absorbed dose using FDEIR under the same conditions. The simulated absorbed dose was 2.91 mGy in the voxel exposed maximum dose. When the x-ray beam was aimed in the RAO 30 direction (Table 1), the recorded dose was 6 mGy. The simulated absorbed dose was 5.22 mGy. Figure 9 presents these results and other results for some conditions, where the PHITS dose was normalized by the simulated absorbed dose at SSD60. The relative statistical error (2 σ) was 3.5% at the maximum. Differences between the maximum skin dose between FDEIR and PHITS were as great as 6.1%.

Calculation time
Simulation of 10 million and 100 million incident particles on the chest CT phantom with Case A (Table 1) was executed on the computer. Table 2 presents a summary of the calculation times of the simulations. For the simulation of 100 million incident photons, the GPU (GTX 1080) was ~122 times as fast as the single-threaded CPU (Core i7–7700K).

DISCUSSION
The simulation results for maximum skin dose showed agreement between FDEIR and the conventional method (Figs 7 and 9). The dose from the log file was rounded to the nearest integer. The electron transport was simulated in PHITS.

Results show two salient benefits of FDEIR. First, the exposed location on the skin is identifiable in the calculated 3D-dose grid. Although an exposed position is indispensable for estimation of the skin dose distribution, the log file shows not the dose distribution, but the maximum absorbed dose, excluding exposed positions. Accurate knowledge related to the skin dose distribution facilitates prediction of the extent of radiation dermatitis.

Second, the simulation can be performed rapidly using the GPU. Estimation of the skin dose distribution can be done immediately after an examination. Moreover, with extraction of fluoroscopic examinations from the log file during an examination, FDEIR is useful for real-time estimation. Cognition of the exposed position and dose help to suppress high-dose exposure at any one specific point. The occurrence of radiation dermatitis can therefore be controlled.

Comparison of the simulation times for 10 million incident particles and 100 million incident particles on the same hardware revealed that the GPU (GTX 1080) took about three times longer, although the incident particles were 10 times more numerous. Results suggest that reading the patient CT phantom and material data from RAM to

Fig. 6. Absorbed skin dose distribution on the front (left) and the back (right) of the chest in the chest CT phantom. There were 100 million incident photons. The x-ray beam used for fluoroscopy of the heart had the following specifications: 76 kV, 20 cm FOV, 100 cm SID, and 60 cm SSD in the PA direction.

Fig. 7. Absorbed skin dose distribution in 2D for an area of the skin on the back of the thorax of the CT phantom (left) and axial dose distributions indicated in a line shown in the left panel (right). The PHITS dose was normalized to the simulated maximum dose from FDEIR. The X mark shows the maximum dose position. It is 0.573 mGy/mAs.
GPU global memory accounts for a large fraction of the execution time. For further acceleration of the simulation, these data should be retained in GPU global memory during simulations.

For this study, FDEIR was used for skin dose estimation. It provides a detailed dose distribution for internal organs if the patient CT phantom is constructed elaborately, with inclusion of internal organs. Furthermore, by altering the patient CT phantom to an IR room phantom, the air dose distribution in the IR room can be estimated rapidly. This makes it easier to identify points with a low air dose rate.

In conclusion, this paper presents FDEIR based on the MC-GPU package. Use of individual CT datasets and various fluoroscopic conditions strongly affects accurate dose estimation. Simulation using the GPU supports rapid calculation. FDEIR is expected to be helpful for ensuring radiation safety for IR.

**CONFLICT OF INTEREST**
The authors declare that there are no conflicts of interest.

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