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ESTIMATION OF EXPOSURE DOSE OF HUMAN HEAD DURING CT SCANNING PROCEDURE USING MONTE CARLO SIMULATION

Abstract: *In this study we present method for determination of exposure dose on human head during computer tomography (CT) scanning procedure. Method is based on scan data obtained by CT. The gray level of image is converted to attenuation coefficient distribution in the head model. The exposure dose is calculated using this model data by Monte Carlo method simulation. This method is based on quantification of interaction between X-ray photon and head tissue. Obtained simulation results can be helpful to improve design, safety and quality of CT system for head imaging.*
Key words: *Monte Carlo, Computer Tomography, Exposure Dose*

1. INTRODUCTION

Recent studies in radiology suggest that computed tomography (CT) scans comprise only (3-11) % of all radiological exams, yet contribute to (35-45) % of the total radiation dose to the patient population [1,2].
Further research into the complex relationship between radiation exposure, image quality, and diagnostic accuracy should be encouraged, in order to establish the minimum radiation dose necessary to provide adequate diagnostic information [3].
According to this information protection of patients during scanning procedure is main requirement during imaging procedure and design of CT devices. The one way to minimize radiation dose is to better understand mechanisms of dose absorption and factor like construction of X-ray device, type of filter being used and characteristic of patient tissue. Dose reduction can be achieved using appropriate filterers [4,5] suitable reconstruction algorithms [6,7] or special mode of X-ray source operation [7].
In this study we try to calculate dose of CT scanner by Monte Carlo Simulation using multi slice image data.
The final goal is to develop tool for exposure dose calculation with the aim of enhancing quality of CT devices.

2. COMPUTER TOMOGRAPHY

2.1 Basic relation

Computer tomography (CT) is a non-destructive method for characterizing 3D objects by using X-ray radiation. This method is based on the differences in attenuation coefficient of X-ray beams for various materials and tissues. The final result is a grey level CT image where corresponding grey level is proportional to attenuation coefficient.
CT medical imaging includes exposure of the object of radiation at one side and detecting attenuated radiation at the other side of the object and this procedure is repeated from more than one direction. The next step is image reconstruction from the projection by using a number of techniques. All of these techniques are based on solving systems of integral equations which are formed as a result of total attenuation of the radiation beam from the source to the detector. Monochromatic X-ray reduction for homogenous materials is given by the relation

$$I = I_0 e^{-\mu d}, \qquad (1)$$

where I_0 is intensity of initial radiation, I is final intensity radiation after path length d in tissue with linear attenuation coefficient μ. If there are multiple materials, the equation becomes:

$$I = I_0 e^{-\sum_i \mu_i d_i}. \qquad (2)$$

Linear attenuation coefficient is very sensitive to energy variation of initial X-photon; that is to say the equations 1 and 2 are valid only for monochromatic beam.
If a polychromatic X-ray source is used, taking into account the fact that the attenuation coefficient is a strong function of X-ray energy, the complete solution would require solving the equation over the range of the X-ray energy (E) spectrum utilized:

$$I = \int I_0(E) e^{-\mu(E) d} dE \ . \qquad (3)$$

This equation largely complicates the reconstruction process and obtaining final results. For this reason the equation (1) is more applicable with a note that μ represents effective attenuation coefficient with spectral characteristics included.

2.2 CT number

CT numbers are also known as the Hounsfield units which are proportional to mean linear attenuation coefficient and they are formed at the end of reconstruction process.
The corresponding CT number is given by

$$CT = K \frac{\mu_t - \mu_w}{\mu_w}. \qquad (4)$$

Where μ_t, μ_w are linear attenuation coefficients of X – ray in tissue and water respectively and K is scale factor. The K is chosen to satisfy -1000 value of CT number for air and 1000 for some kind of cortical bone tissue. According to representation a Hounsfield unit in binary number system it is very useful take 1024 value for *K*. Then CT number ranges from -1024 to 1023 and 11-bit is used for representation. For 12-bit representation CT number takes values from -1024 to +3071.

3. MONTE CARLO METHOD

The Monte Carlo method is stochastic procedure for simulation of a finite number particle transport by using random number generation. In every iteration suitable random numbers determine direction and space angle of particle trajectory, energy of particle and type of their interaction with transported medium (tissue). The appropriate variables are obtained as a mean value of finite numbers of iteration. The accuracy of the estimated values is proportional to the number of iterations. Monte Carlo is stochastic method because the variable being sought can be obtained only in finite volume but not in one point unlike deterministic method.
If we consider motion of a particle, we should suppose a one point source with coordinates (0,0,0) as shown in Fig. 1.

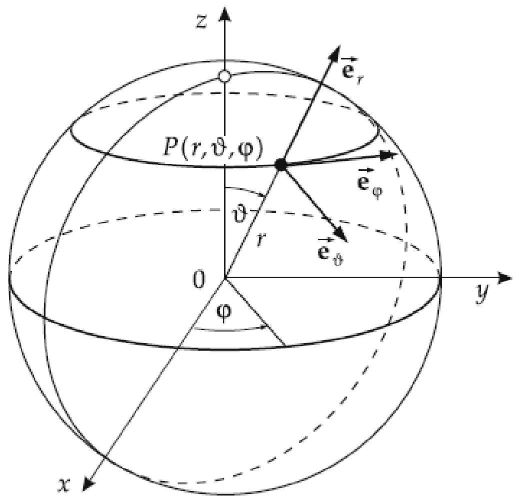

Fig. 1. Cylindrical coordinate for particle source

Random direction of particle motion is defined by equations

$$\theta = \arccos(1-2\xi_1)$$
$$\varphi = 2\pi\xi_2 \qquad (5)$$

where θ *is* polar angle, φ *is* azimuth angle and ξ_1, ξ_2 are random generated numbers with uniform density distribution in the interval (0-1).

Unit vector direction of particle motion is expressed by

$$\vec{\Omega} = \sin(\theta)\cdot\cos(\varphi)\cdot\vec{e}_x + \sin(\theta)\cdot\sin(\varphi)\cdot\vec{e}_y + \cos(\theta)\cdot\vec{e}_z \qquad (6)$$

Let's assume now that the particle is passing through a much more homogenous region as shown on Fig. 2.

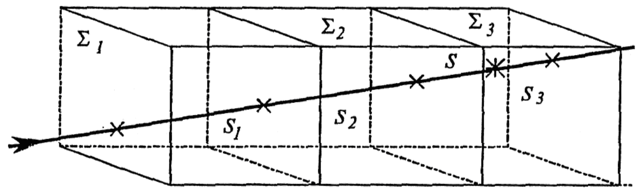

Fig. 2. Particle in non-homogenous media

The regions are characterized by macroscopic cross sections for absorption Σ_1, Σ_2, Σ_3, which are actually linear attenuation coefficients.
Tracking starts by generating a new random number ξ_3 and calculating the equation

$$t = -\ln\left(1 - x_3\right). \qquad (7)$$

If we find

$$\begin{aligned} &S_1 s_1 < t \\ &S_1 s_1 + S_2 s_2 < t \qquad (8) \\ &S_1 s_1 + S_2 s_2 + S_3 s_3 > t \end{aligned}$$

the interaction takes place in the region 3 after it travelled the distance *s* in the region 3,

$$s = \frac{\tau - \Sigma_1 s_1 + \Sigma_2 s_2}{\Sigma_3}. \qquad (9)$$

By analogy with this way we can find place of interaction and distance having been crossed for any region.
If the particle meets the requirement for interaction we will have increment of a total received dose of radiation for the amount of photon energy, otherwise there is no absorption and photon comes to detector without loss of energy.

4. MATERIAL AND METHODS

The first step in this study was to determine spectrum of X-ray source. We used low dose 3DCT scanner [8], X-ray tube Model XRS-125-7K-P for X-ray production with 4 mm aluminium filter and obtained spectrum as shown in Fig. 3.

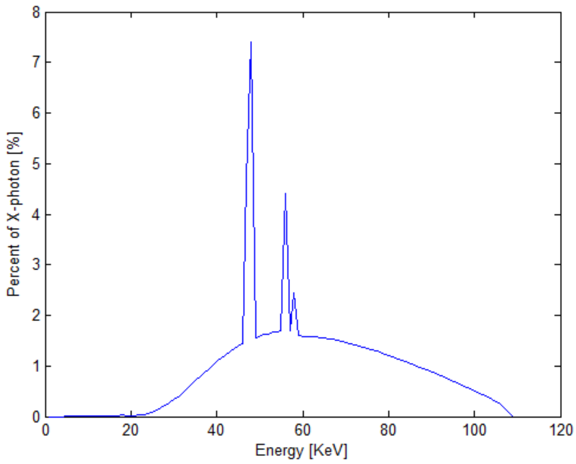

Fig. 3. Spectral distribution energy of X-photon

Calculation of the spectrum was performed by Spek-Calc [1] a software for X-photon spectrum estimation. Mean energy of X-photon of this spectrum is given by relation

$$\bar{E} = \frac{\int_E f(E) \cdot E dE}{\int_E f(E) dE} \qquad (10)$$

where E is X-photon energy and $f(E)$ represents a probability density function.
The mean energy is $\bar{E}$ =73.72 KeV. The next step is geometry determination of the problem and for this we used CT image as shown in Fig. 4.

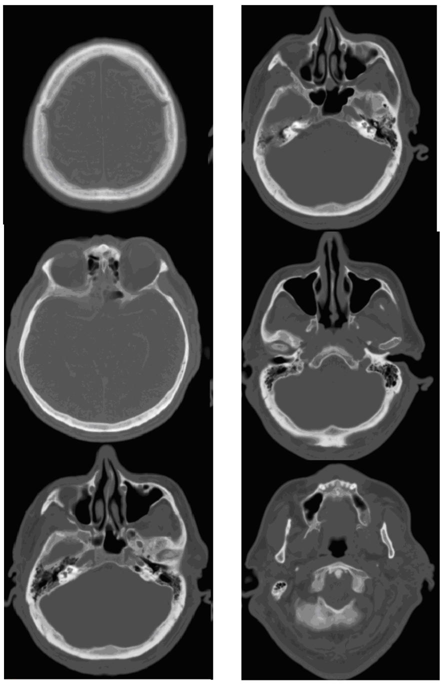

Fig. 4. Multi slice CT image

CT images are segmented and surface is produced for human head.
Model data consists of surface points shown in Fig. 5.

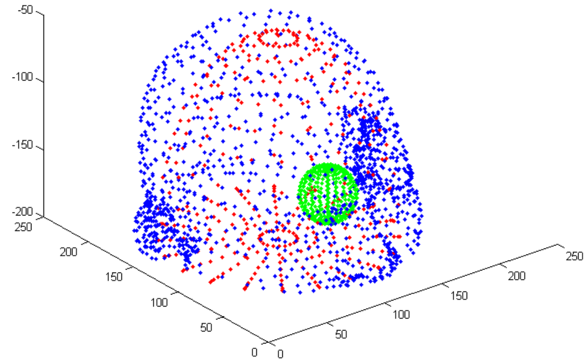

Fig. 5. Head model divided into three regions

The attenuation coefficient for water at effective energy $\bar{E}$ has value m_w = 0.1965 cm^{-1}. By substituting that in the equation (4) we can get a macroscopic cross section for any domain in the model. CT numbers for the structures having been considered are given respectively:

- CT=929 (Bone tissue external sphere) 170.5 mm diameter
- CT=-780 (Air shell internal) 27 mm diameter
- CT=85 (Internal sphere soft Tissue),

and they are obtained by an averaging on the domain.
During imaging procedure X-ray source rotates around head of patient, the procedure is finished when the source has closed a full circle. In our case we need to apply discretization by dividing full circle into ten segments. Each of these segments represents one direction source-detector between which there is object. For every direction we apply procedure described in section 3. The total amount of absorption energy is 8 % of initial radiation what corresponds to a received dose of 112 µS.

5. FINAL REMARKS

In this study we have briefly presented basic relation and technique of CT imaging method. Critical point of application of this diagnostic method is exposure dose of X-photon radiation.
The main goal of our research is estimation of the absorbed radiation dose using Monte Carlo simulation of X-photon transport in head tissue of the patient. For simulation we have created geometrical model from CT images. The model is divided into a number of regions with realistic dimensions and attenuation coefficients obtained from spectra data and grayscale level of images being used.
The final dose is around 112 µS and this result is comparable with value from 35 to 72 µS which was predicted for this scanner, according to 2007 ICRP [10].
The difference between the predicted and calculated

doses is the result of the approximations which were included in the calculation. Material characteristics of each voxel were not taken into account; instead, mean value in certain domain was attributed to every point of that domain. Also, the energy of X-photons was taken as the mean value and attenuation coefficients were calculated based on the unique energy value. In reality, the photons are polychromatic and this coefficient should be calculated for each energy value separately. Finally, the accurate initial radiation value can be obtained by measuring, but in this case, the estimation was done on the bases of scanner characteristics.

There are several goals of future research: to improve accuracy by using a more realistic geometrical model with a lot of sub-domains in model or even voxel representation of model, to do calculation which would include the attenuation coefficient dependence on energy to a higher degree and to compare the calculated values with the radiation dose values having been measured for the scanner being considered.

6. ACKNOWLEDGMENT

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