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Thorax organ dose estimation in computed tomography based on patient CT data using Monte Carlo simulation

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► ***Original article***

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ABSTRACT

Background: This study presents patient specific and organ dose estimation in computed tomography (CT) imaging of thorax directly from patient CT image using Monte Carlo simulation. Patient's CT image is considered as the patient specific phantom and the best representative of patient physical index in order to calculate specific organ dose. ***Materials and Methods:*** EGSnrc / BEAMnrc Monte Carlo (MC) System was used for CT scanner simulation and DOSXYZnrc was used in order to produce patient specific phantom and irradiation of photons to phantom in step and shoot mode (axial mode). In order to calculate patient thorax organ dose, patient CT image of thorax as voxelized phantom was divided to a 64x64x20 matrix and 6.25 x 6.25 x 6.25 mm^3 voxel size and this phantom was imported to DOSXYZnrc code. MC results in unit of Gy/particle were converted to absorbed dose in unit of mGy by a conversion factor (CF). We calculated patient thorax organ dose in MC simulation from all irradiated slices, in 120 kV and 80 kV photon energies. ***Results:*** Effective dose was obtained from organ dose and organ weighting factor. Esophagus and spinal cord received the lowest, and bone received the highest dose. In our study, effective dose in CT of thorax was 7.4 mSV and 1.8 mSv in 120 and 80 kV, respectively. ***Conclusion:*** The results of this study might be used to provide the actual patient organ dose in CT imaging and calculation of real effective dose based on organ dose.

Keywords: *Organ dose, patient specific dose; computed tomography; MC simulation; EGSnrc/BEAMnrc.*

INTRODUCTION

With the growth of technology in computed tomography imaging its use is significantly increased in diagnosis of disease [1]. But there are serious concerns on the harmful effects of radiation for adults and children [2,3]. Spiral CT imaging which is an imaging method with fast patient body data acquisition was introduced in the 1990s. Spiral CT has a high potential in cardiovascular studies with minimally invasive to patients. The number of CT scans imaging compared to all radiological procedures from 4.2% in 1996 increased to 7% in 2006. With improvements in CT scanner imaging technology and increase the number of CT scan studies, the cumulative dose arising from medical examination at the same time from 37% reached to 60% [4]. These data suggest that radiologists tend to replace two-dimensional study with volumetric method. Therefore patient dose assessment and radiation risk estimation in CT imaging is inevitable [5-8]. VII BEIR report about risk of low radiation dose in

2005 suggests that accurate estimation of risk is dependent on patient organs dose and sex [9]. 103 ICRP report shows that an accurate estimation of the risks of radiation dose would be possible by knowing the organ dose [10]. Direct measurement of organ dose is impractical in vivo therefore other methods are suggested such as measurement of radiation dose in physical phantom which is performed by TLD or MOSFET dosimeter [11, 12]. In computed tomography imaging, dose is generally measured using Ion chamber dosimeter with 100mm length on an acrylic cylinder with 32 cm diameter for trunk and 16 cm for head and CT dose Index (CTDI) is calculated. CTDI phantoms are uniform and differ with the human body so expression dose based on CTDI measurement does not seem to be logical. Therefore dose measurement in an environment similar to the real environment of the human body is essential. Several attempts were performed to calculate the patient's specific dose or organ dose in CT scan. AAPM Task Group 204 has proposed the size-specific dose estimate (SSDE) [13]. SSDE estimates patient dose with an anthropomorphic parameter. In this study a conversion factor is provided to convert the measured dose values on CTDI to SSDE. Another method of organ dose calculation is using the conversion factor of dose length product (DLP) to effective dose (E) [14-16]. In this study, phantoms designed by Oak Ridge National Laboratories (ORNL) were used. DLP to E conversion factor obtained based on direct measurement of the dose on CTDI phantom and Monte Carlo (MC) simulation calculations on anthropomorphic phantoms. Effective dose calculation is performed based on sum of organ dose weighted that is proposed by ICRP [10]. Patient and organ specific dose estimation, as POSDE, based on Monte Carlo simulation, was performed by Kalender *et al.* The patient CT data is loaded in software designed for that. Then the best match (best-fitting) between the CT image and voxel phantoms in the soft ware memory is achieved. Afterwards, it selects the closest anthropomorphic phantom in size to patient's CT image and Monte Carlo simulation runs to organ dose calculation [17]. In the present study, patient's CT image as the patient specific phantom and the best representative of patient physical index is used to calculate the specific organ dose.

MATERIALS AND METHODS

CT scan system

In order to obtain CT images and the patient specific organ dose calculation, GE- Light Speed CT system, 64 slices, manufactured by GE Healthcare Technologies (Waukesha, WI) was used. The source to isocenter distance on this CT scanner is 540mm, while the source to detector distance is 950mm. minimum slice thickness is 0.624 mm and fan beam angle is 56° [18].

Monte Carlo simulation of CT system

We used EGSnrc / BEAMnrc MC System for CT scanner simulation. All hardware components in CT system according to manufacture scanner geometry, such as X-ray tube, total X-ray tube filtration, collimator and bowtie filter were simulated using parameters as component module (CM) in BEAMnrc system. CMs as XTUBE, SLABS, JAWS, and PYRAMIDS were used to simulate X-ray tube, ALuminum filter, collimator and Bowtie filter, respectively. Bowtie filter were modeled based on the information provided by the manufacturer. Two SLABS of air were used as phase space (ph) file in after bowtie filter and gantry isocenter, respectively. Number of photons and particle produced in simulation and their histories were saved in phase space file. This file is used for X-ray beam thickness, X-ray spectrum and beam profile analysis. Ph file located after bowtie filter was used as radiation source in CT system simulation. In our study, in order to produce photon spectrum in simulation, 10^9 electron was emitted to anode X-ray tube. All options related to low energy particles transport in simulation as bound Compton scattering, photoelectron angular sampling, Rayleigh scattering, atomic relaxation and electron impact ionization were activated. Energy cut- off for electron (ECUT) and photon (PCUT) was selected to be 512 keV and 1 keV, respectively. For ECUT, rest mass energy of an electron (511 keV) was considered, so the electrons were tracked down to the kinetic energy of 1 keV - the same as that of the photons. Figure 1 shows the detailed geometry of CT X-ray tube components modeled in BEAMnrc simulations. For ph file data analysis and depiction of energy flounce, beam thickness in z axis and beam profile, we used BEAMDP (BEAM Data Processor) software.

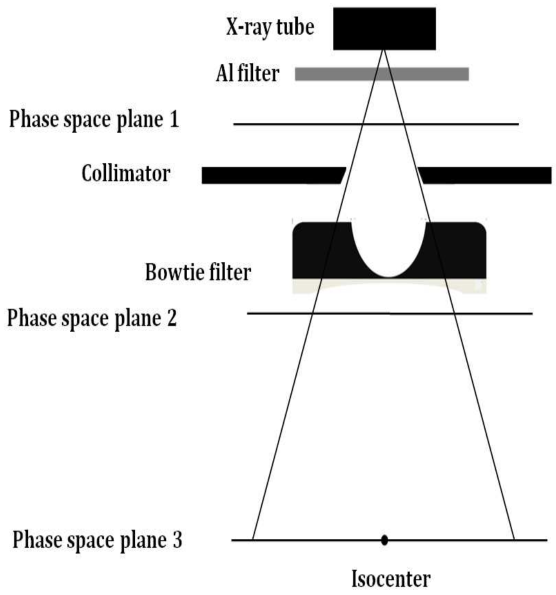

Figure 1. The detailed geometry of CT scanner modeled in BEAMnrc simulation, axial view.

Validation of CT scanner modeling in simulation

For evaluation of the X-ray beam data and X-ray tube component designed in simulation, energy spectra obtained by BEAMnrc were compared with the energy spectrum produced by Ipem78 as standard software for X-ray spectrum analysis [19]. For evaluation of the beam- shaping filter (bow tie filter), half beam profile has been measured using TLD crystal and computer radiography (CR) cassette. The CT scanner has a 56^{o} fan beam angle and 54.0 cm in source to isocenter distance. In this study, the TLD circular chips made up 6-Lif (Mg, Cu, P) with dimensions of 4.5 mm in diameter and 8mm thickness was used. In order to draw beam profile, TLDs were arranged in 20 points from center to peripheral of gantry isocenter and one exposure was performed in Scout mode (X-ray tube fix and table moves). This measurement processes was repeated two times with two TLDs series with the same model. The shape of the beam profile obtained from direct measurements was compared with the profile obtained by the MC code.

Patient specific phantom

DOSXYZnrc simulation code was used in order to produce patient specific phantom and irradiation of photons to the phantom. For accurate calculation of organ dose, patient CT image of thorax manually was voxelized in input file named INP file in DOSXYZnrc code. Thorax ct image was divided to a 64×64×20 matrix and 6.25×6.25× 6.25 mm3 voxel size. A pegs 4 file was developed with EGSnrcMP software in which specifications of the entire medium in phantom such as attenuation coefficient, interaction cross section and physical density are defined. DOSXYZnrc code uses this file for dose calculation in phantom voxels. Mediums in designed phantom included air, adipose (skin fat layer), blood, heart, muscle, soft tissue, bone cortex and bone marrow. Simulated area in thorax CT image, central slice and simulation phantom of central slice are shown in figure 2.

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Exposure model and source geometry in MC simulation

In the present study, the axial exposure model was used to irradiation of patient thorax and heart area. This mode of scan was simulated using DOSXYZnrc simulation code. In axial mode, X-ray tube rotates 360 degree around the patient body then table moves in new location based on user selection and initiates another scan. This method of scanning is named step and shoot or snap shot method. In MC simulation this mode of scan was modeled with moving the point of coordinate inside the phantom. Thorax phantom localized to heart area with 12.5 cm in length was irradiate in simulation with three axial mode without overlapping of adjacent slices. Mean dos to an organ was obtained based on dose to that organ from all slices. Twenty slices of phantom with 6.25 mm in thickness were irradiated. In our study, number of history for MC simulation run, was 10^9 photons per axial mode and beam width was 40mm. Figure 3 shows simulation set up in the present study.

In this study, phase space file after bowtie filter was used as radiation source in DOSXYZnrc MC System. Source model in DOSXYZnrc as ISOURCE=8 (Phase-Space Source Incident from Multiple Direction) was used. In this source model, number of projection, irradiation angle and weight of beam intensity per projection is selected. Here, 50 projections with the same beam intensity per projection were selected and organ dose in one rotation around the phantom was calculated.

Patient specific organ dose calculation

Dose in phantom voexls was stored in egslst and 3ddose files in DOSXYZnrc code. We used 3DDose Viewer [20] and MATLAB software for 3ddose file analysis and dose map depiction in phantom, respectively. Dose value in MC simulation is normalized per source history. In order to determine the absolute dose, MC results in unit of Gy/particle were converted to absorbed dose in unit of mGy by a conversion factor (CF). In our study, CF was calculated for 120kV and 80kV photon energy. In first step we measured the dose in air in isocenter of CT scanner gantry using Ion chamber dosimeter.

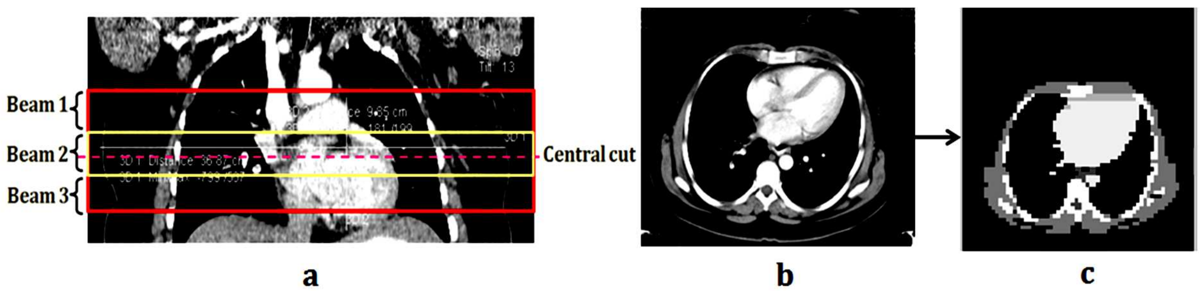

Figure 2. Coronal view of patient thorax ct image, simulated area and radiation beams (1, 2 and 3) (a). Axial view of central slice in which organ dose was calculated (b). Patient specific phantom simulated from central slices(c).

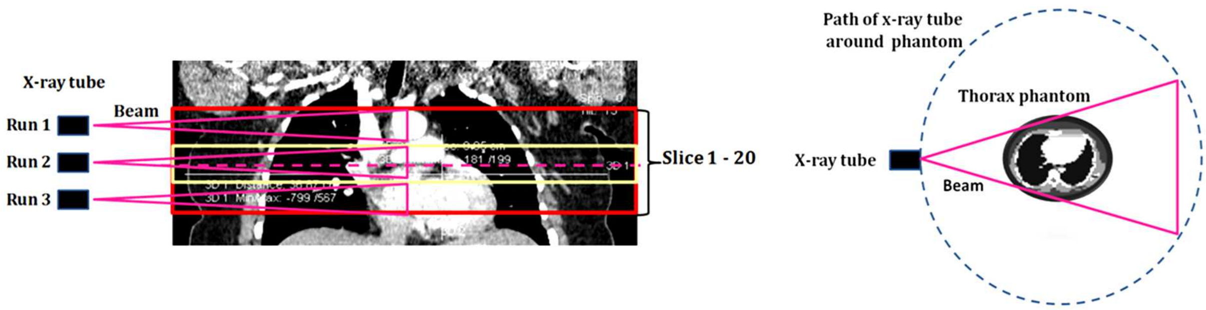

Figure 3. Three axial scan in thorax phantom (coronal view) in MC simulation with three MC code running (a). Seven slice with 6.25 mm in thickness included in every beam approximately. Axial scan modeled in simulation in transverse view for each run (b).

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Then, dose in air in the same point as measurement set up in simulation for two mentioned energy was calculated and CF was obtained based on equation 1.

Conversion factor (particle /mAs) = measurement dose (Gy/mAs)/simulation dose (Gy/particle) (1)

Absorbed dose in Gy was calculated based on equation 2.

Absolute dose (Gy) = Simulation dose (Gy/particle) x CF (particle /mAs) x mAs (2)

We calculated the organ dose for routine thorax imaging with 100 mAs beam intensity. We also calculated thorax organ dose in low energy X-ray photon with 80kV in comparison with 120kV. In order to verify the CF calculation, the body CTDI phantom with 32 cm in diameter was scanned in CT system as axial mode with 1 sec rotation time and 120 kV. Then, direct measurement was performed in central point and 4 peripheral points of CTDI phantom using Ion chamber. In simulation, a CTDI phantom was designed with 0.5 × 0.5 × 1 cm3 voxel size and irradiated with same set up as measurements. Results of simulation and measurement in same points inside the phantom were compared.

Effective dose was calculated using thorax organ dose and tissue weighting factor as equation 3.

Weighting factor reported in ICRP 103 was used for effective dose calculation [21].

$$E = \sum_T w_T H_T = \sum_T w_T \sum_R w_R D_{T,R} \qquad (3)$$

W_T = tissue weighting factor
W_R = radiation weighting coefficient (1 for photons)
$D_{T,R}$ = average absorbed dose to tissue T

RESULTS

X-ray beam spectrum results in BEAMnrc MC system and Ipem78 software are shown in figure 4. IPEM78 (Spectrum Processor of the Institute of Physics and Engineering in Medicine's Report 78) is a software that simulates X-ray spectra using a semi-empirical model based on the Birch and Marshall model [19]. In Ipem78 software, several variables that impact the X-ray spectrum are defined such as anode angle, flat filter material and thickness. Equality of characteristic spectrum in figure 4 a and b, resulted from BEAMnrc and Ipem78 software, shows that X-ray source in simulation was properly designed. Figure 5 shows beam profile from center to peripheral of CT scan gantry due to beam shaping filter (bow tie filter). X-ray photon profile was measured using TLD crystal and CR film and was calculated using MC simulation method. All beams profile resulted in simulation and measurements are in same shape; this means that bowtie filter was correctly simulated.

Conversion factor (CF) for 120kV and 80kV is shown in table 1. In our study, measurements and simulation calculation of absorbed dose in CTDI phantom was the same with 3% to 6% difference that is in the acceptable range and CF was correctly calculated. We did not use the CTCREATE soft ware [22] to convert the CT DICOM file to egsphant file (phantom file in DOSXYZnrc), due to the fact that, in our study, this soft ware had many errors and bugs in conversion processes and inaccuracy was possible in calculation of voxel material from CT number to electron density conversion curve. Thorax phantom (figure 3c) was irradiated as axial mode in MC simulation with 40mm field size, 120kV photon energy and 50 projections and organs dose were calculated. Khursheed *et al.* [23] indicated that 18 projections were sufficient in axial mode simulation without significantly affecting the calculated organ doses. In our simulation test with 50, 100 and 1000 projection irradiation model, no significant difference was found in dose transport to phantom voxel element. Imparted energy to the phantom is dependent on the rotation time (duration of one rotation around phantom) and beam intensity (mA). Gu *et al.* [24] used 16 projections for dose calculation in CTDI phantom. Figure 6 shows a case of the simulation set up and dose map in several slices and thorax organ dose calculation processes in this study. Figure 7 shows dose map in thorax

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phantom that it was irradiated using 120kV and 80kV photon energy in MC simulation. In thorax organ dose calculation, radiation dose due to scatter photons from adjacent beams was considered. In figure 6, simulation dose map in central slices is hot and slices far from center of X-ray field is cold. This is because central slice is exposed directly to primary photons and farther slices from center of field are exposed to scatter photons with low energy.

Thorax organ dose is listed in table 2. CF listed in table 1 was used for organ dose calculation. Our study showed that patient effective dose in thorax ct imaging is 7.4 mSv and 1.8 mSv in 120 kV and 80kV, respectively.

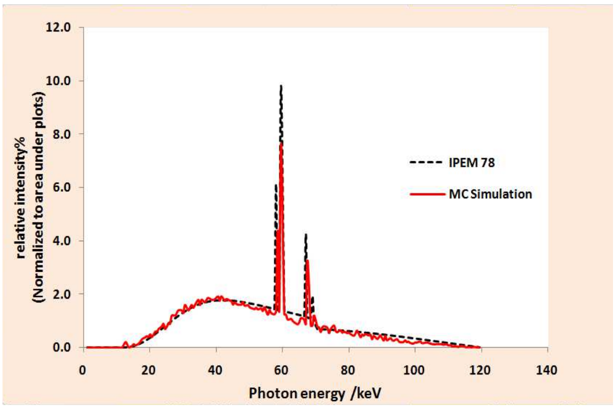

Figure 4. Comparison of X-ray spectra between IPEM report 78 data and MC simulation.

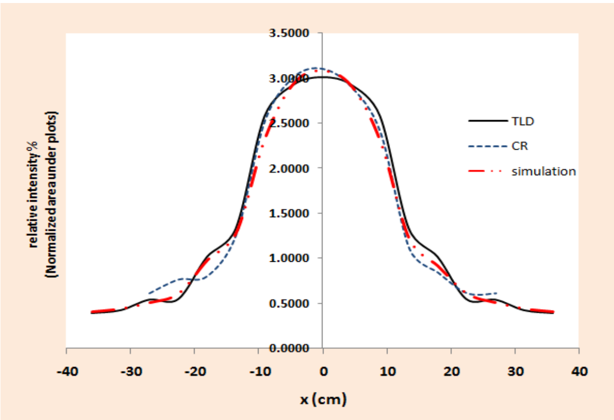

Figure 5. Comparison of the beam profile in air resulted in simulation, TLD crystal and CR cassette, from center to peripheral of CT scan gantry isocenter.

Table 1. Conversion factors for converting the MC simulation results from Gy/incident particle to mGy.

Photon energy (kV)	mAs	Slice thickness (mm)	Air dose measurement (mGy / mAs)	Simulation air dose calculation (mGy /particle)	Conversion factor (particle /mAs)
80	100	40	0.064	2.05E-18	3.12E+16
120	100	40	0.184	6.06E-18	3.04E+16

Table 2. Patient specific organ dose in patient thorax, irradiated with 80kV and 120kV photon energy, 100 mAs, 40 mm beam width and axial mode.

Photon Energy	Organ dose/ slice (mGy) (Mean ± SD)								
	esophagus	spinal cord	Heart	Lung	Spin		Adipose	Muscle	Rib
					Bone marrow	Bone surface			
80 kV	1.1±0.1	1.1±0.1	1.5±0.3	2.1±0.4	2.8±0.2	4.5±1.0	4.5±0.5	5.1±0.6	12.0±2.8
120 kV	5.6±0.7	6.2±0.8	7.2±1.1	9.1±1.2	12.3±3.5	20.0±3.3	17.2±1.7	18.6±2.2	43.6±8.2

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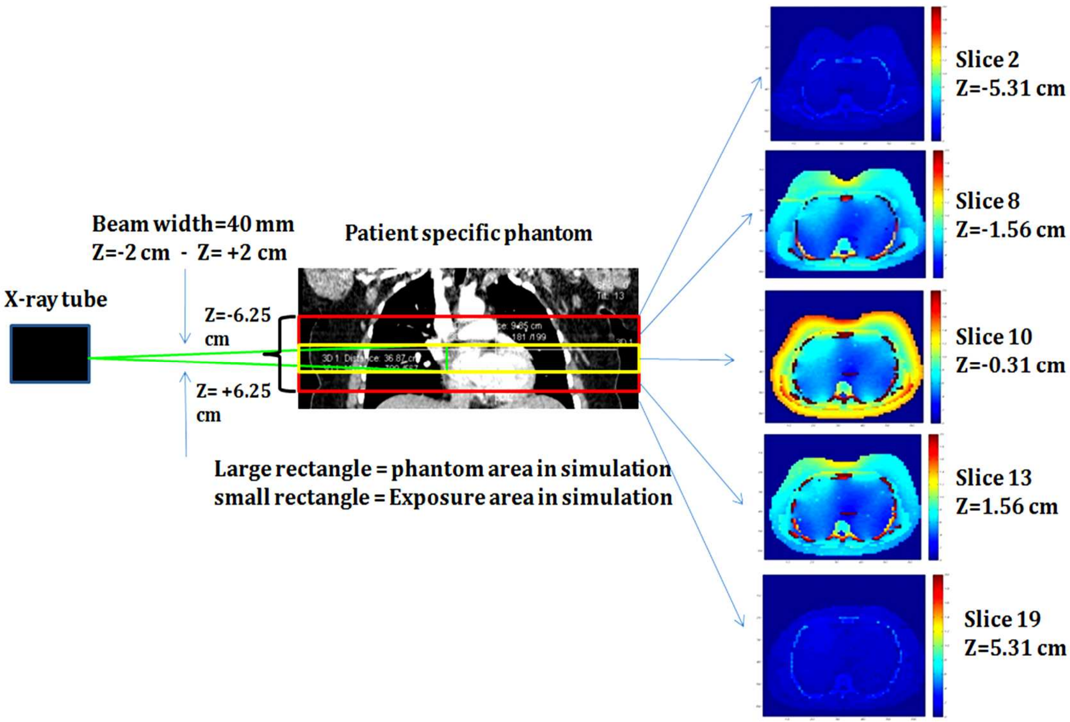

Figure 6. Simulation set up and dose map in slices. Slice 10 is central slice in phantom.

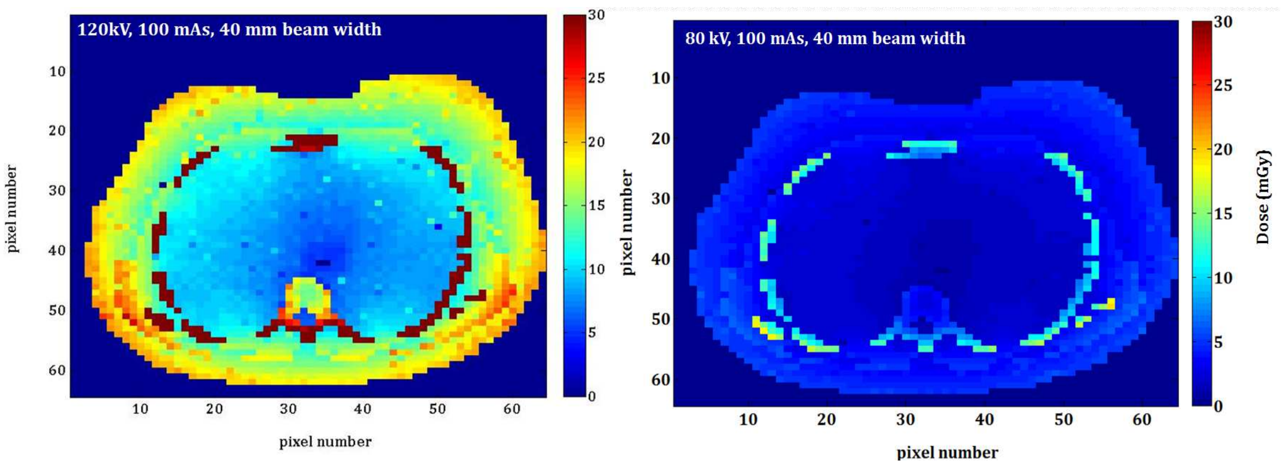

Figure 7. Dose map comparison in patient thorax phantom at 120 (left) and 80 kV (right) ,40mm X-ray beam width and 100 mAs.

DISCUSSION

According to figure 7, the ratio of the dose in the isocenter of patient thorax phantom to its surface in 120 kV energy photon was larger than that of 80kv at same points, because by increasing the X-ray photon energy, can also increase their penetration power. Chest wall (ribs and intercostal muscles) keeps the lung and mediastinum as a protective shield from receiving high radiation dose in two energy. Bones, especially the ribs, have received high dose due to high absorption power. The thoracic spine has received lower dose than ribs, because it is surrounded by ribs laterally and muscles posteriorly. Dose in surface bone (cortex) is more than bone marrow. Dose to transverse process and spinous process are more than body of spine. In most studies, organ dose was calculated based on dose measurements in CTDI phantom [25-27]. In these studies, organ dose level was reported for a reference man and was not patient specific. Several software have been introduced to calculate the organ dose in CT imaging such as ImPACT, CT Dose, CT Expo, ImpactDose, eXposure and WAZA-ARI. All of these software, tend to calculate organ dose based on several predefined body phantom imported to software memory. Whereas, in this study, patient CT data was directly used as patient specific phantom in order to calculate the organ dose. Organ dose values reported in articles vary extensively. For instance, lung dose in CT scan imaging of thorax, was reported to be 19.5 mGy with 16 slice, 37 mGy with 28 slice and 31.5 mGy with 60 slice, and there can be found no logical relationship between the number of slices and lung dose [28]. This is also the case for other organs. Our study shows that the lung dose in CT of thorax imaging is 9.1 and 2.1 mGy for 120 kV and 80 kV respectively. In a study by Huda *et al.* [25], organ dose in CT was reported based on dose measurements on CTDI phantom. In this study dose in lung and heart were 6.8 mGy and 10.3 mGy respectively with 120 kV, 100 mAs and 12 cm scan length. But in our study dose to heart with the same condition was 7.2 mGy. In another study, Effective dose in CT imaging of thorax was reported between 2.1 and 12.2 mSv [29]. Our study shows effective dose in CT of thorax to be 7.4 mSV and 1.8 mSv in 120 and 80 kV respectively. In general, disparity of organ dose values is due to differences in the measuring conditions. Since in the present study, patient CT image has been used to calculate the effective dose, it can be claimed that the calculated effective dose is the actual values, it is specific for each patient, and can be used for patient specific risk assessment in CT imaging. Radiation dose per slice in CT imaging could be a suitable indicator for patient dose assessment and comparing the technical procedures recommended in CT imaging for patient dose reduction. In this study we calculated the patient specific organ dose in CT of thorax based on CT image data using BEAMnrc MC system. The method presented in this research can be used to provide the actual patient dose in CT Imaging. Calculation procedures to patient specific dose per slice presented in this study can be used for all patient organs.

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Conflict of interest: *Declared none.*

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