

X-RAY SOURCE MODEL FOR MODERN DIAGNOSTIC X-RAY SETS

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Abstract

This paper presents the model of a static diagnostic X-ray set. The model takes into account such parameters as a high anode voltage, total beam filtration, anode voltage pulsations, field dimensions and the type of a diagnostic investigation. X-ray spectra of a tube were generated using the "TASMIP". The ICRP voxel phantoms were adapted for modeling one of the most frequent diagnostic procedures of a chest in order to obtain the profiles of absorbed dose distributions from the developed X-ray source model.

Key words: X-ray radiation, X-ray tube, voxel phantom, Monte-Carlo modeling, dosimetry

1. INTRODUCTION

The number of X-ray diagnostic procedures increases in many clinics all over the world. According to UNSCEAR (2008) the annual number of diagnostic X-ray examinations exceeds 3.6 billion. As a result, the radiation burden per capita also shows a significant growth. In order to minimize risks of oncological diseases and hereditary pathologies from these examinations the reliable methods of a dose estimation and optimization of the radiation exposure of patients should be developed.

Monte-Carlo modeling of X-ray transport within virtual human phantoms allows assessing the absorbed dose distribution in organs and tissues. Using this data one can easily estimate the effective dose obtained by a patient in a specific X-ray examination.

Such a modeling requires an accurate model of the source of radiation. A diagnostic X-ray set can be considered as an X-ray tube and a system that forms the field of necessary dimensions and shape. During modeling it is necessary to take into account the dependence of final X-ray spectra on such factors as a high voltage, a current, a total filtration, voltage pulsations (ripple), parameters of a forming system, etc.

Forming systems of various shapes can be used depending on type of X-ray examination. In this paper only rectangular collimation system was considered. The developed X-ray set model was adjusted for simulating one of the most frequent X-ray examinations – investigations of a chest. The transport of X-rays from model source within the reference voxel phantoms was simulated using the Monte-Carlo method implemented in the MCNP. The profiles of the absorbed energy within the phantoms were obtained and analyzed in order to optimize the number of voxels included into simulation and minimize the time required for modeling.

2. X-RAY TUBE AS AN IMPORTANT ELEMENT OF A STATIC X-RAY SET

Factually, X-ray tube is a powerful diode where the beam of electron, emitted from the cathode filament, is accelerated by the electric field and hits the anode. The X-ray radiation arises in the anode material as the result of slowdown of electrons that have a certain velocity determined by electric voltage between the cathode and the anode. The range of anode voltages from 20 kV to 250 kV is used in most medical diagnostic units.

The anode surface is located under a certain angle towards the electron beam direction (Frame n.d.). Thus, X-ray quanta generated in the anode are emitted at a necessary angle (see figure 1). In order to obtain a sharp image, electron beam must be focused on the anode at a very small area called focal

spot. The larger the tube focus width is, the larger the geometrical blur is and, conversely, the sharper the focus is, the less the blurriness is. Modern X-ray diagnostic tubes have the following dimensions of the focal spot: micro focus is $0.3 \times 0.3 \text{ mm}^2$, small focus is from $0.6 \times 0.6 \text{ mm}^2$ to $1.2 \times 1.2 \text{ mm}^2$, and large focus is from $1.3 \times 1.3 \text{ mm}^2$ and more (Kishkovsky, Tutin & Esinovskaya 1987).

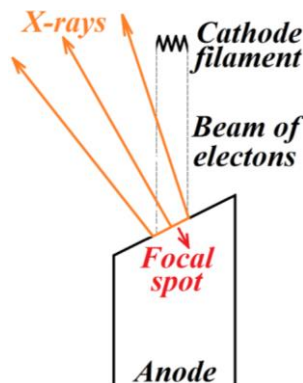


Fig. 1. Cathode and anode of an X-ray tube (Frame n.d.)

Diagnostic tubes create intensive pulses of X-ray radiation during the short period of time so that the irradiated object will not displace. Since the main part of electron beam energy is spent not on radiation but ionizing losses (i.e., is converted into heat), then there are special precautions for heat removal. Liquid cooling or rotating anode is used to decrease the intensity of anode heating. Also, the anode material should be of a high atomic number as it increases the intensity of X-rays. Usually, a tungsten anode is used in medical X-ray sets.

Bremsstrahlung generated as a result of electron radiation losses is continuous and depends only on a tube voltage and a beam filtration. Besides bremsstrahlung, characteristic radiation is presented at X-ray tube spectra. This type of radiation depends on the anode material and appears as a result of an electron conversion from outer electron shells to inner. The spectrum of characteristic radiation consists of separate lines.

In diagnostic apparatus the role of characteristic radiation is small in comparison with bremsstrahlung. Furthermore, final radiation spectrum can be corrected using filtration, e.g., the low-energy part of the spectrum can be cut off. As the filter thickness increases, the absorption of low-energy quanta also increases, and the spectrum becomes “harder”.

3. X-RAY SOURCE MODEL FOR STATIC X-RAY SETS

The reliability of a computer simulating an X-ray set is primarily determined by using a realistic X-ray spectrum. There are three main directions of obtaining X-ray spectra: experimental, theoretical (Monte-Carlo modeling) and semi-empirical methods (Lankipalli et al. 2001). Each of these methods has its advantages and disadvantages but the Monte-Carlo method provides more appropriate results because of using the realistic representation of radiation interaction with matter and complete description of the geometry of a task.

Unfortunately, it is difficult to implement the full-scale Monte-Carlo model of an X-ray set with the detailed description of an X-ray tube because of the lack of manufacturer’s information about tube components and their material composition. Moreover, simulating such a model takes a lot of computer time because it takes into account all processes connected with generating X-ray radiation, although some of them will not be used for final spectrum. That is why in this work an X-ray tube was modeled as a point source with a predetermined radiation spectrum. The influence of focal spot size on an absorbed dose distribution within a human body will be studied in the next parts of the research.

Although our model of an X-tube does not require simulating its inner structure and mode of operation but it takes into account such parameters as the anode voltage, the total filtration and the anode voltage pulsations.

The “TASMIP”-model was used for generating energy X-ray spectra. This model is considered to be the most appropriate for simulating spectra of an X-ray tube with tungsten anode working at the range of anode voltage from 30 kV to 140 kV (Boone & Seibert 1997).

In the model the photon flux at the distance of 100 cm from the X-ray tube focus is given as a tabulated function $\Phi(E, d, V, \zeta)$ normalized by 1 mA·s. The number of photons emitted by the source at the energy interval $(E, E + dE)$, where $dE = 1$ keV, is given for certain voltage V , its pulsation ζ and the total filtration expressed in the equivalent thickness d mm of aluminum (Boone & Seibert 1997). Changes of each of these parameters result in variations of a final X-ray spectrum. On the figure 2 one can see how adding an aluminum filtration changes the spectrum. The number of low-energy quanta for the spectrum with the filtration of 3 mm Al decreases in comparison with the no-filtered spectrum and the average energy of the filtered spectrum shifts to a higher energy range.

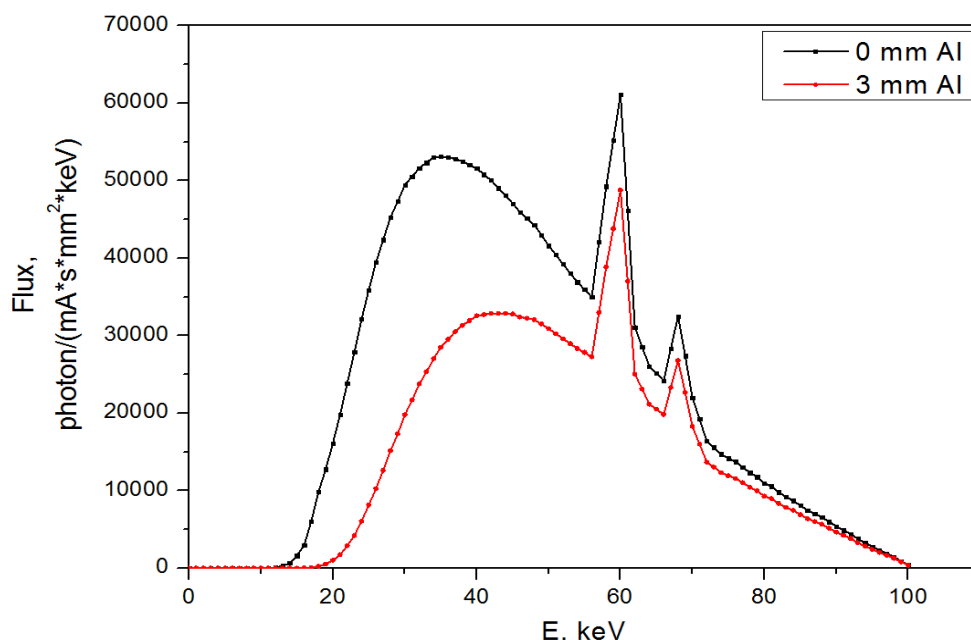


Fig. 2. Energy spectra of an X-ray tube in the “TASMIP”-model for the anode voltage $V = 100$ kV, $\zeta = 0$, $d = 0$ mm Al and $d = 3$ mm Al

Taking into account the anode voltage ripple in the “TASMIP”-model allows realistic modeling of the X-ray beam formation. Pulsations of the anode voltage can be quantitatively characterized by a ripple coefficient (ζ)

$$\zeta = \frac{V_{max} - V_{min}}{V_{max}} \cdot 100\%$$

where V_{max} and V_{min} are maximum and minimum anode voltage values respectively.

Typical anode voltage ripple values are $\zeta \sim (5 - 10)\%$ (Ay et al. 2004). The most important effect of the voltage ripple is not in the energy redistribution of the emitted radiation but in the decrease of the source radiation output. X-ray spectra with different values of the ripple are shown on the figure 3. The ripple does not change the position of the spectrum maximum but noticeably decreases its amplitude and, as a consequence, the radiation intensity that is proportional to the area under the spectrum. Quantitatively, the decrease of the area under the spectrum depends on the total filtration.

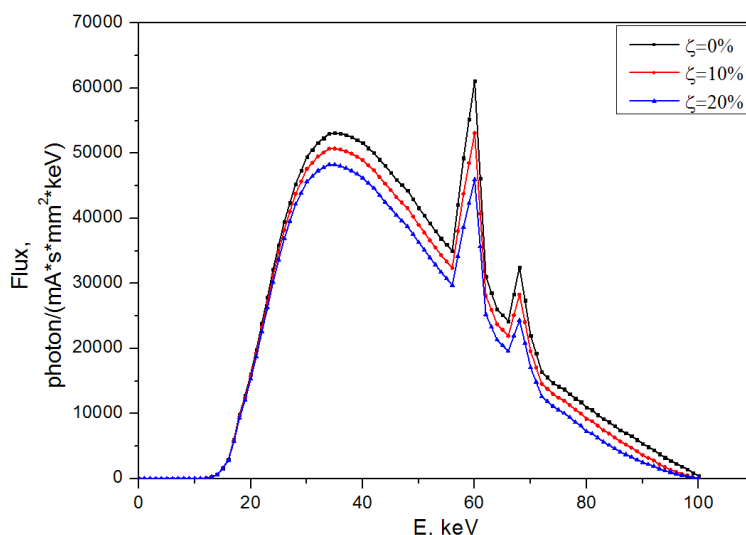


Fig. 3. Energy spectra of an X-ray tube in the “TASMIP”-model for the anode voltage $V = 100$ kV, $\zeta = 0\%$, $\zeta = 10\%$ and $\zeta = 20\%$, $d = 0$ mm Al

By now 150 X-ray spectra have been calculated using the “TASMIP”-model for the following parameters: the high voltage range from 60 kV to 140 kV in increments of 20 kV; the total filtration range from 0 mm Al to 5 mm Al in increments of 1 mm; the ripple range from 0% to 20% in increments of 5%. The choice of these parameters is conditioned by the fact that they completely cover the range of the voltage values, filtrations and pulsations that is used for conducting the X-ray examinations of a chest.

In order to simulate the operation of an X-ray set it is necessary to consider the system that forms a required X-ray field configuration on a film. The rectangular field is used for most X-ray examinations; modern fluorography sets use slotted collimation. The round field is often used for dental X-ray examinations.

In our model the forming system is pairs of lead shutters located at different distances from a tube focus (see figure 4). The point X-ray source radiates into a cone with a base circumscribed around the X-ray film. The lead shutters truncate the unnecessary part of this cone to form the rectangular field of the required size. Upper pair of the shutters provides the expansion of the irradiation field along the X axis, and the lower pair corrects the field size along the Z axis. Input parameters for the model are the distance from focal spot to the irradiated object, X-ray field dimensions, the distance from the focal spot to the forming system, characteristics of the forming system.

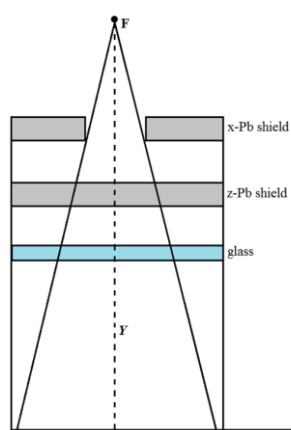


Fig. 4. Model of the X-ray set with two pairs of lead shutters

4. THE RADIATION OUTPUT OF THE MODEL SOURCE

The created model of an X-ray set will be used for modeling typical X-ray examinations in order to estimate radiation doses to patients. Thus, it is necessary to figure out how to apply the results of a simulation to a specific X-ray unit used by a radiologist at a clinic.

As the “TASMIP”-model was used for generating X-ray spectra the results obtained from a simulation will be suitable only for the X-ray tube that was used to create that model. Different X-ray tubes operating at the same values of high voltage and added filtration have various radiation spectra and, as a result, different doses will be obtained during the same examinations. This is due to the fact that X-ray tubes are characterized by different anode angles, focal spot size, inherent filtration and anode aging. As the “TASMIP”-model does not consider these parameters it is important somehow to take them into account.

Each X-ray tube can be characterized by a radiation output. The radiation output is determined as a free-air kerma measured on the primary X-rays beam axis at a distance of 1 m from the X-ray tube focus for a given anode voltage (Guidelines 2.6.1.2944-11 2011). If the developed model of an X-ray set is correct it is possible to determine which of the “TASMIP”-spectrum of our model corresponds to the spectrum of a real X-ray tube by comparing the radiation output of our model tube and the radiation output of a real tube at a given anode voltage and ripple.

In order to verify the created model we calculated the X-ray output of our source model for $10 \times 10 \text{ cm}^2$ field at a distance of 1 m for the abovementioned voltage and filtration ranges and the ripple of 0%. The MCNP was used to simulate coupled photon-electron transport and estimate free-air kerma from F6 tally (Briestmeister 1997). After that we compared our results with the experimental data for the same parameters presented in the article about the “TASMIP” (Boone & Seibert 1997). Some results of comparison are shown in table 1.

Table 1. Radiation output (*mR/mAs*) calculated after simulation and experimental values

<i>d</i> , mm Al	Simulation	Experiment (Boone & Seibert 1997)	Deviation, %
60 kV			
0	5,2842	5,3161	0,60
1	3,4644	3,4731	0,25
2	2,4778	2,4951	0,70
3	1,8621	1,8686	0,34
4	1,4474	1,4320	1,08
5	1,1504	1,1260	2,17
100 kV			
0	13,4017	13,4444	0,32
1	10,0404	10,0486	0,08
2	8,0155	8,0598	0,55
3	6,6364	6,6680	0,47
4	5,6166	5,6250	0,15
5	4,8359	4,8270	0,19
140 kV			
0	23,1608	23,2902	0,56
1	18,5142	18,5349	0,11
2	15,5147	15,5568	0,27
3	13,3703	13,4966	0,94
4	11,7312	11,8001	0,58
5	10,4167	10,5354	1,13

Good agreement of the comparable results allowed us using our X-ray set model as a source of radiation for simulating X-ray diagnostic procedures.

After that we calculated the radiation output of our model source for other values of the anode voltage ripple and derived formulas of dependencies of the radiation output on the anode voltage, ripple and total beam filtration. They will be used to determine which values of “TASMIP”-spectrum parameters should be used as input values for modeling of the X-ray examination with a real X-ray tube that has the certain value of a radiation yield.

5. HUMAN BODY MODEL FOR X-RAY EXAMINATIONS OF A CHEST

As mentioned before, the developed X-ray model will be used a source of radiation for simulating the radiation transport within a human body in order to evaluate doses to organs and tissues. The estimation of doses obtained by patients during X-ray investigations is based on applying an absorbed dose distribution in the anthropomorphic computational model (phantom) of a human body. Such phantoms have to include the description of the human internal structure, organs positions, density and composition of tissues.

Until recently, so-called mathematical phantoms were used for simulating processes of radiation transport within a human body (ICRU Report 48 1998; Cristy 1980). In these phantoms human body was represented as a set of three dimensional bodies bounded by second order surfaces (planes, spheres, ellipsoids, cylinders, cones). Intersections and unions of these bodies were used for modeling organs of complicated shapes. This rather rough model of geometry allowed solving many tasks of radiation protection.

Modern techniques of obtaining human body images (computed tomography, magnetic resonance imaging) provide sufficiently accurate anatomical information for constructing voxels (volumetric cells) of phantoms necessary for simulation. Such models can be developed for a specific person as well as for a referent person (Zubal et al. 1994; Zankl, Petoussi & Wittmann 1996).

The ICRP suggested using conventional voxel phantoms of an adult male and female with body parameters close to the reference person for dosimetric calculations (ICRP Publication 110 2009). These phantoms correspond to the anatomical data of the reference person (ICRP Publication 89 2003).

Each reference ICRP phantom is the model of the human body placed into rectangular parallelepiped filled with air. This parallelepiped is divided into cells (voxels) of the same size. Each cell contains the number of organ to which it belongs. Each organ is aligned to a specific material (tissue) which describes the content of the cell. Cells that do not belong to the phantom are filled with air. The phantoms contain more than 140 different structures that consist of 50 tissues types. It was made in order to maximally correspond to the reference individual.

The ICRP phantom of female consists of more than 14 million of voxels ($299 \times 137 \times 348$). Voxel dimensions are $1.78 \text{ mm} \times 1.78 \text{ mm} \times 4.8 \text{ mm}$. The ICRP phantom of male contains more than 7 million of voxels ($254 \times 127 \times 222$). Voxel dimensions are $2.14 \text{ mm} \times 2.14 \text{ mm} \times 8 \text{ mm}$. Because of such a large number of voxels in the phantoms it is impossible to create an input file for the MCNP manually. Thus, special programmers CreatePhantomMale.nb and CreatePhantomFemale.nb were worked out in Mathematica to write automatically the input files with the descriptions of the phantoms and the X-ray set model for the MCNP.

The trial simulation of the transport of X-rays from the model source within the phantoms was made. The examination of a chest was chosen because it is one of the most frequently made X-ray examination in our country. The X-ray set model was adjusted to form radiation field with dimensions $35 \text{ cm} \times 35 \text{ cm}$ on the film. Focus-film distance was 100 cm. The F8* tally was used to estimate the absorbed dose in each voxel of the phantom (Briestmeister 1997). The results of simulation for the male phantom are shown on the figure 5. Evidently, that absorbed dose in voxels directly not got into irradiation field is rather small.

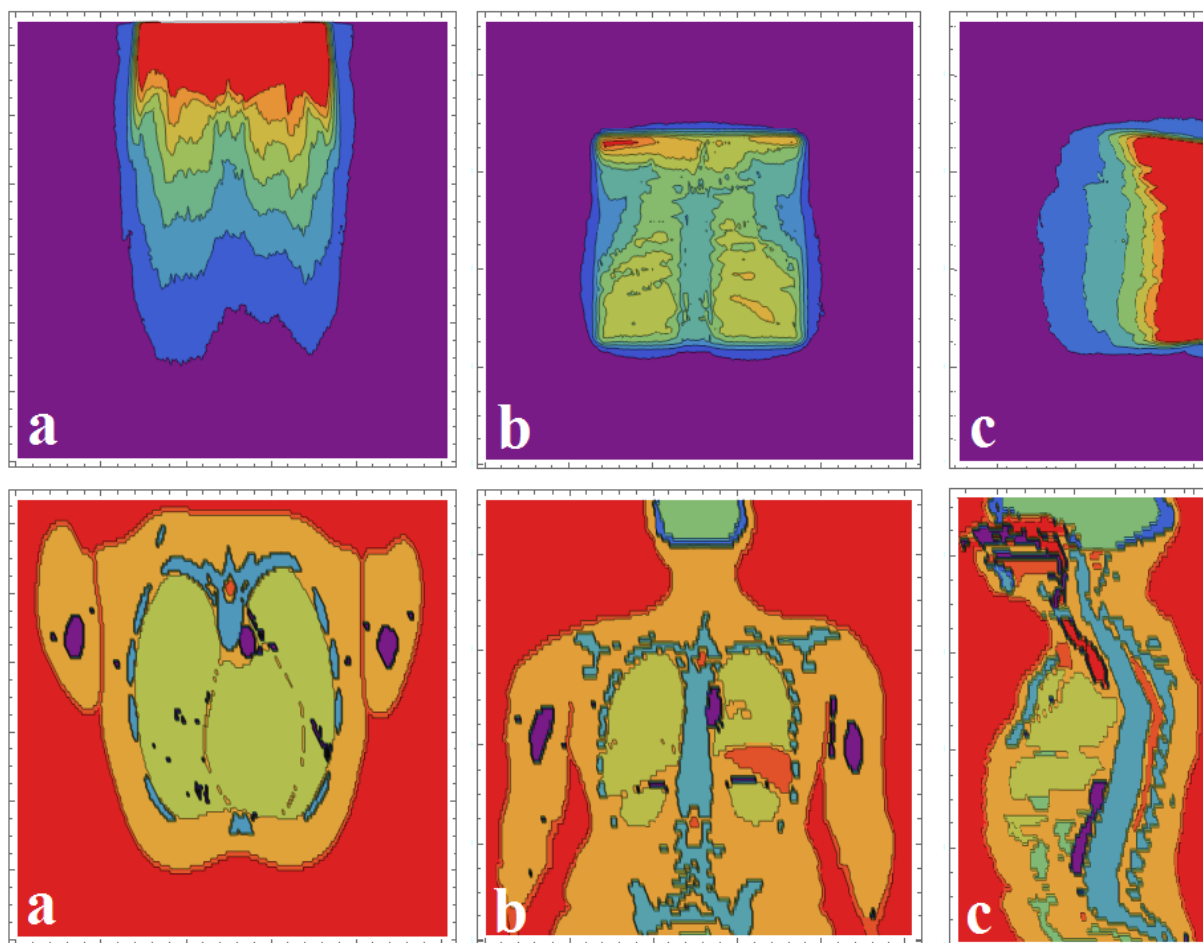


Figure 5. Absorbed dose distribution within the male phantom during chest irradiation where a) – transverse section; b) – coronal section; c) – sagittal section

The analysis made it possible to optimize the number of voxels that should be included into a simulation. Thus, the programs *CreatePhantomMale.nb* and *CreatePhantomFemale.nb* depending on the type of X-ray examination, recommended field size, focus-film distance and other parameters will select from the entire phantom only those layers that are directly irradiated. Also the programs will add layers of 20 cm to upper and lower edges of the selected part of the phantom in order to take into account the effects of penumbral regions.

6. CONCLUSIONS

Modeling of different medical exposure aspects is widely used because of the involvement of various computer programs applying the Monte-Carlo method for simulating the interaction of radiation with matter. The correct model of an X-ray source plays an important role for such simulations.

The “TASMIP”-model of an X-ray tube with a tungsten anode was used for generating X-ray spectra. 150 spectra were calculated for wide ranges of the anode voltage, total filtration and the voltage ripple. The X-ray set model includes a point source with a predetermined spectrum and the collimating system that forms the rectangular field of required dimensions.

The verification of the model was made based on the results of comparison of the calculated values of the model source radiation output with the experimental data presented by Boone & Seibert (1997). All data showed good agreement.

The ICRP reference voxel phantoms were used for estimating the absorbed dose distribution from the model X-ray source. These phantoms were adapted for modeling X-ray examinations of a chest. It was found that it is reasonable to use for simulation only that part of the phantom that is directly irradiated and additional layers of 20 cm to take into consideration the penumbral regions effects. This modification allows minimizing the computer time necessary for simulating without increasing uncertainties in estimations of doses in irradiated organs and tissues.

The results obtained will be used for carrying out next stages of the research dedicated to Monte-Carlo modeling of human irradiation during X-ray examinations and estimating the spatial distribution of the absorbed dose within the patient's body depending on different characteristics of medical X-ray sets and irradiation procedures.

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