

# Interpretation of measured dose data in X-ray imaging

Toroi P.<sup>1,2</sup>, Kellaranta A.<sup>1</sup>, Vock P.<sup>2</sup>, Siiskonen T.<sup>1</sup>, Tapiovaara M.<sup>1</sup> and Kosunen A.

<sup>1</sup>STUK – Radiation and Nuclear Safety Authority, Laippatie 4, P.O.Box 14, FI-00881 Helsinki, Finland

<sup>2</sup>Inselspital University Hospital, Freiburgstrasse 10, CH-3010 Bern, Switzerland

## Abstract

The main part of the uncertainty in patient dose estimation is related to the conversion from measurable dose quantities to organ doses and effective doses. In this study, methods available for these dose conversions were reviewed. In this study, the thorax radiograph (posterior-anterior projection) was taken as an example, and the uncertainties related to the properties of the beam and of the patient were examined. Radiation quality has a large effect on the conversion, and conversion factors differing more than by a factor of two can be found in the clinical range of X-ray spectra. The conversion factor can be chosen reasonably accurately based merely on the half-value layer of the X-ray beam. Even a small adjustment in the beam position can have a marked impact on organ doses, but at least in this case the effects on doses in various organs partly compensate each other and the impact on the effective dose is not large (here of the order of 10%). Usually, dose conversion is based on a standard patient model, and patient-specific conversion is not performed. Dose conversion factor differences between a real patient and a standard-sized patient can be up to  $\pm 50\%$  for some organs and  $\pm 20\%$  for the effective dose, and even higher if paediatric patients are considered.

Key words: dose conversion factor, organ dose, effective dose, uncertainty

## 1 Introduction

Patient dose determination is an essential part of the process of optimizing between adequate image quality and radiation detriment in diagnostic radiology. The radiation dose from an X-ray examination is needed to estimate the radiation-induced risk of cancer to the patient. Risk estimates are normally based on organ doses or the effective dose, which cannot be measured directly. Therefore, air kerma-based, measurable quantities are generally used for clinical dosimetry. IAEA Publication TRS No. 457 (IAEA, 2007) provides a unified approach to dosimetry in diagnostic radiology and describes the measurable quantities. These quantities can be used for quality control and as diagnostic reference levels, but they are not directly related to the patient's cancer risk. Thus, great care should be taken when these quantities are used for optimization. Additional interpretation is needed to estimate the radiation risk.

Several methods are available to estimate risk-related dose quantities for an X-ray examination, but the accuracy of the doses depends on the methods used for this estimation. The coarsest level of estimation is to use the average or typical dose value of the given examination. Such values generally relate to the mean value of a limited set of X-ray systems and standard-sized patients, and they do not account for variations in X-ray beam quality or beam positioning. Moreover, as they represent typical dose values, the actual amount of radiation used is not accounted for. More accurate data can be obtained if measured dosimetric data is used together with tabulated conversion factors from measured data to organ doses. Nowadays, these factors are typically based on Monte Carlo simulations. The more information is used to select the correct conversion factor, the better the accuracy will be. However, the human models used for conversion factor calculation are typically of an average size and do not take into account differences in patient size and anatomy. A recent AAPM report provides guidance for patient size corrections in computed tomography (CT) imaging (AAPM 2011). However, this report does not include size correction factors for projection imaging. The best accuracy can be achieved when individual and examination-specific simulation is performed based on real 3D-image data of the individual.

In this study, one of the most common X-ray examinations, the posterior-anterior (PA) thorax projection, was taken as an example. Patient doses in this specific projection have been investigated, for example, by Schultz et al. (1994) and Campos de Oliveira et al. (2011). In this article, available methods for the interpretation of measured dose data are reviewed and discussed together with the uncertainties. Uncertainty related to differences in beam positioning and properties and patient size is estimated.

## 2 Material and methods

### 2.1 Available methods

Information on available methods for organ and effective dose estimation was collected based on a literature survey. Many published papers include tabulated effective doses for certain diagnostic procedures and examinations (e.g. Hart et al. 2002, Mettler 2008, EC 2008). However, the use of such values is not sufficient for accurate dose calculations, because the real situation can differ greatly from the one assumed.

Dose conversion factors together with measured dose values can be used to offer more accurate estimates of organ dose levels in certain examinations. Conversion coefficients have been published to give organ and effective doses from the incident air kerma, the entrance surface dose or dose-area product. They are presented as a function of tube voltage and filtration for different X-projections modelled in Monte Carlo runs. Conversion factors for projection X-ray diagnostics have been published by different research groups, such as NRPB Reports from the UK (Hart et al. 1994 a and b and 2002), GSF reports from Germany (Drexler et al. 1990, Zankl et al. 2002), CDRH reports from the USA (Rosenstein 1988) and reports of a group from the University of Delft in the Netherlands (Schultz et al. 2001).

Spreadsheet calculation programs are also available that are based on the above simulation results, e.g. NRPB data (Hart et al. 1994 b). Kramer et al. (2008) developed the CALDose\_X program, which provides an improved patient model by using male and female voxel phantoms instead of a standard MIRD-type geometrical phantom model. Normally, these calculation programs include assumptions about the beam positioning, their radiation quality range does not cover Cu filtrations and they use a standard-sized phantom.

Some commercial computer simulation programs allow the user to modify the scanning parameters and calculate doses for different patient sizes and radiation qualities. For example, the Monte Carlo simulation programs PCXMC from STUK, Finland (Tapiovaara and Siiskonen 2008) and ImpactMC from CT Imaging, Erlangen, Germany (Deak et al. 2008) are available for such purposes. There are also other groups performing patient-specific dose calculations with their own simulation set-ups (e.g. Jarry et al. 2003, Angel et al. 2009, Li et al. 2008, 2011 a and b). ICRP (2009) has published reference male and female phantoms intended to be used for the calculation of radiation protection quantities, including the effective dose. Although these phantoms are not primarily intended for estimating the organ doses in X-ray diagnostics, they can be introduced to widely-used general-purpose Monte Carlo codes such as EGSnrc, MCNPX or FLUKA, allowing flexible adjustments of exposure geometry and beam properties.

Information on the methods used for organ dose estimation was collected as part of a coordinated research project of the International Atomic Energy Agency (CRP E2.10.08, Development of advanced dosimetry techniques for diagnostic and interventional radiology). This activity was implemented with questionnaires that were distributed by the participants (Table 1). More than one response from a country was accepted. Respondents provided information on the methods in use for the conversion, and they estimated organ and effective doses from a given set of clinical scenario data (Table 2). The participants were instructed to use the methods and references that would be used in their institutes.

**Table 1.** Institutes participating in the activity

<b>Participant</b>	<b>Institute</b>
Australia	Australian Radiation Protection & Nuclear Safety Agency
IAEA	International Atomic Energy Agency
China	Hospital of Capital Medical University, Beijing
Czech Republic	National Radiation Protection Institute (NRPI)
Finland (Activity coordinator)	Radiation and Nuclear Safety Authority (STUK)
Greece	Greek Atomic Energy Commission (GAEC)
Kenya	Kenyatta National Hospital
Serbia	Vinca Institute of Nuclear Sciences
United Republic of Tanzania	Tanzania Atomic Energy Commission
United Kingdom of Great Britain and Northern Ireland	Newcastle upon Tyne Hospitals Foundation Trust

**Table 2.** Clinical scenario used in the questionnaire

<b>Case</b>	<b>1</b>
<b>Sex</b>	Male
<b>Nationality</b>	Danish
<b>Age (years)</b>	40
<b>Height (cm)</b>	183
<b>Weight (kg)</b>	85
<b>Modality</b>	General radiography
<b>Examination</b>	Thorax

<b>X-ray system anode</b>	W-anode, target angle 20°
<b>Projection</b>	PA
<b>Tube voltage (kV)</b>	130
<b>Total filtration (including inherent filtration)</b>	3mmAl+0.2mmCu
<b>HVL (mmAl)</b>	8,3
<b>Tube current-time product (mAs)</b>	3
<b>KAP (<math>\mu\text{Gy}\cdot\text{m}^2</math>)</b>	6
<b>Focus-image receptor distance (cm)</b>	180
<b>Beam size at the image receptor (cm×cm)</b>	35×40

## 2.2 Beam properties

The effect of beam properties on the uncertainty of conversion was studied by using the PCXMC 2.0 program (Tapiovaara and Siiskonen 2008) for the case given in Table 2. Different radiation qualities and geometrical parameters were used to simulate possible variations in clinical situations and determine the potential effect on organ doses and the effective dose. The reference parameters for the simulations were adopted from the IAEA CRP project and the ranges used for this study are presented in Table 3.

The effect of radiation quality on the conversion factors was examined with three different filtration combinations together with a range of tube voltages (Table 3). Variations were studied as a function of the half-value layer (HVL) and the tube voltage. The HVL values of these radiation qualities were calculated using a computer program (Tapiovaara and Tapiovaara, 2008) based on the semiempirical spectrum model of Birch and Marshall (1979).

In our questionnaire, the clinical situation was described by giving the beam size and focus-image distance (FID), but not the focus-skin distance (FSD), which is normally not readily known. Therefore, each participant had to estimate the patient thickness and the phantom exit-image distance. The effect of this uncertainty was assessed by modifying the phantom exit-image receptor distance from 0.0 to 20.0 cm, and the FSD and beam size were calculated by the program according to this value. The patient thickness of the phantom model used in PCXMC was constant. The effect of beam positioning was examined by placing the fixed-area beam at different cranio-caudal positions. In our reference situation, the z-value of 55 was chosen. To evaluate the uncertainty of this parameter, it was varied between 50 and 60 cm while keeping all the other parameters fixed at their reference values.

**Table 3.** X-ray beam parameters and patient properties in PCXMC simulations

<b>Parameter</b>	<b>Reference value</b>	<b>Range</b>
Phantom height (cm)	183	
Phantom weight (kg)	85	
Phantom thickness (cm)	21.3	
Incident air kerma (mGy)	1.0	
Focus-image distance (cm)	180	
Image size (cm×cm)	35×40	
Reference point x (cm)	0	
Reference point y (cm)	0	
Reference point z (cm)	55	50 – 60
Filtration (mmAl/mmCu)	3/0.2	3/0.0, 3/0.1, 3/0.2
Tube voltage (kV)	130	80 – 140
Phantom exit-image distance (cm)	3.0	0.0 – 20.0
Focus-to-skin distance (cm)	155.71	138.71 – 158.71
Beam size (cm×cm)	30.28×34.60	28.92×33.05 – 30.86×35.27

## 2.3 Patient properties

The effect of patient size and shape was investigated by using the Monte Carlo simulation program ImpactMC from CT Imaging, Erlangen, Germany (Schmidt and Kalender 2002, Deak et al. 2008). CT images of three male patients of different sizes were selected for the simulations (Table 4). The beam properties were those given in Table 2, except that the beam size was enlarged so that it covered the whole chest area and the lungs of all patients. The lungs and breasts were chosen as organs of interest, because they were entirely inside the radiation beam and the effect of differences in beam positioning could be avoided. Although a male patient was considered in our

example, he was also assumed to have a small amount of breast tissue. These organs were segmented from patient images by using the ITK-SNAP program (Yushkevich et al. 2006). Organ doses were calculated from dose distributions by using the ImageJ program (Abramoff et al. 2004) and 3D ROI Manager plug-in (Iannucelli et al. 2010).

**Table 4.** Patient sizes

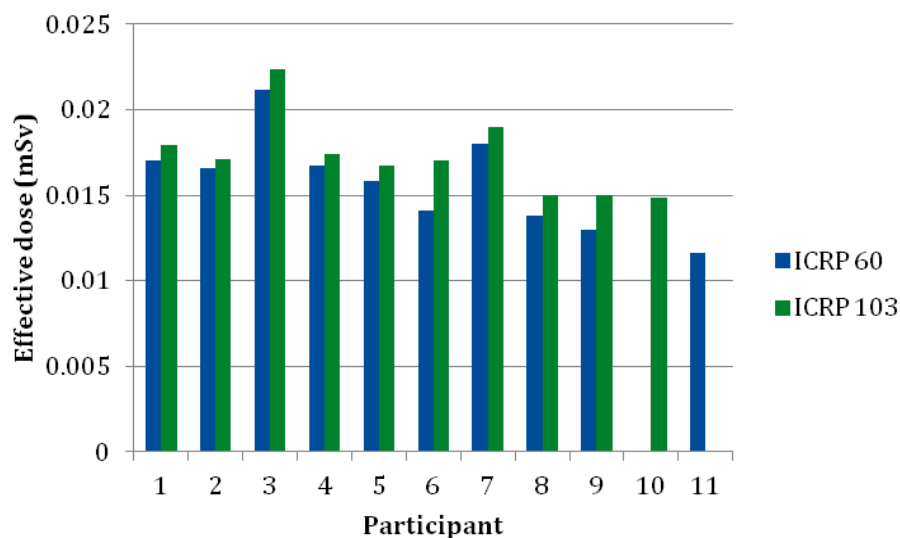
Patient code	Weight (kg)	Height (cm)	Thickness (cm)	Width (cm)	Effective diameter (cm)	Thickness of PCXMC phantom (cm)
MXS	45	172	20.3	29.5	24.5	16.0
MM	72	175	26.3	34.0	29.9	20.0
MXL	108	172	34.0	40.0	36.9	24.8

Simulations for these patients were also performed with the PCXMC program by using the actual weight and height of the patients. PCXMC scales the phantom model based on these patient size parameters. The thickness of the phantom in these cases is given in Table 4. It can be seen that the scaled phantoms used in PCXMC do not necessarily accurately represent the anatomy of the patients, since in PCXMC the sizes of all organs and tissues are simply increased or decreased according to horizontal and vertical scaling factors. For example, no extra inter-organ fat is inserted. Moreover, PCXMC uses a hermaphrodite phantom in simulations. Hence, the amount of breast tissue is greater in PCXMC simulations than in our ImpactMC simulations. Moreover, the thickness of the breast is larger, potentially causing a different attenuation and scattering environment at the location of the breast tissue.

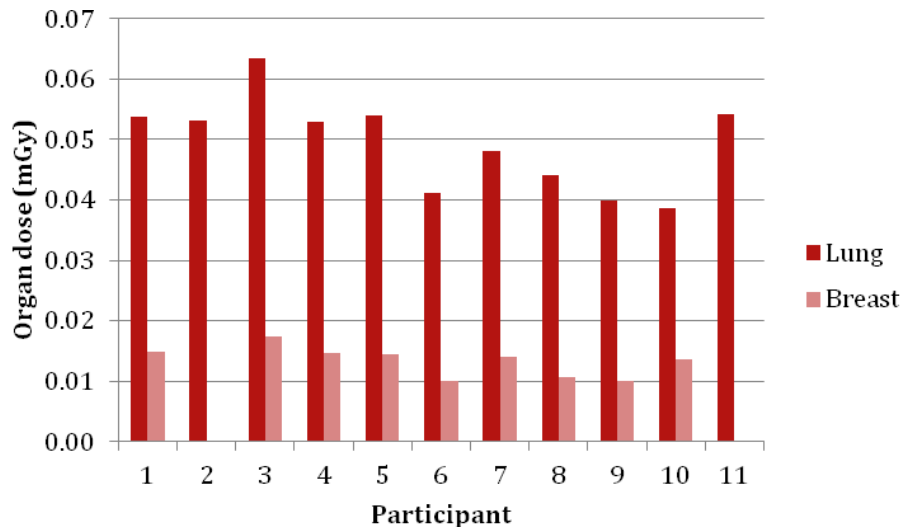
### 3 Results

#### 3.1 Available methods

In the CRP activity, information was collected on the methods that are used for organ dose estimations. Fourteen questionnaire responses were received. Three responses were not included in the analysis because they were clearly erroneous. Variation among participants in the effective doses can be seen from the Figure 1. Participants from 1 to 7 used the PCXMC program, participants 8 and 9 used data based on NRPB publications and participants 10 and 11 used other methods (Rosenstein 1988, Schultz et al. 2001). The average value for the effective dose (ICRP 103) was 0.017 mSv, the standard deviation was 13% and the maximum variation compared to the average was 44%. Breast and lung doses are presented in Figure 2. Participant 2 did not provide the breast dose, while participant 11 used conversion factors from simulations carried out with the ADAM phantom (Kramer 1982), which does not have breast tissue, and the breast doses were therefore zero.



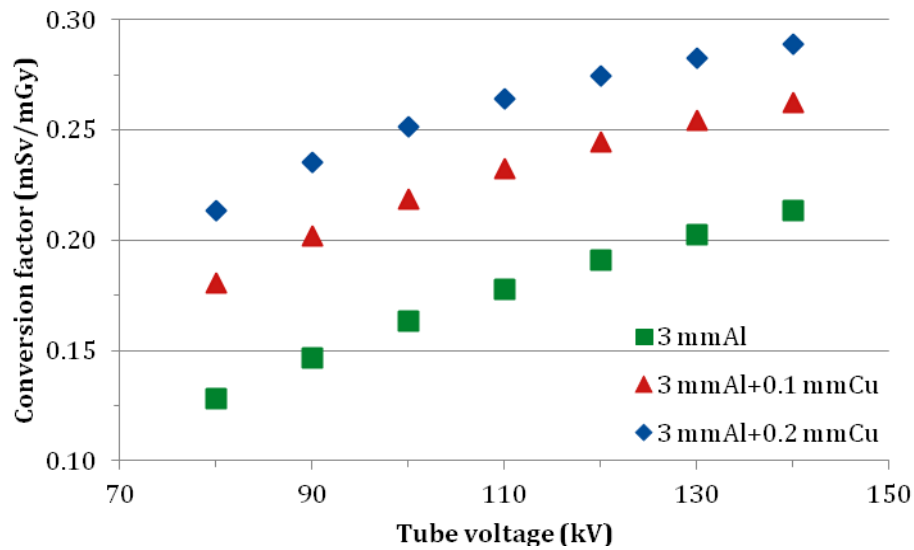
**Figure 1.** Effective doses in the example thorax PA examination of Table 2.



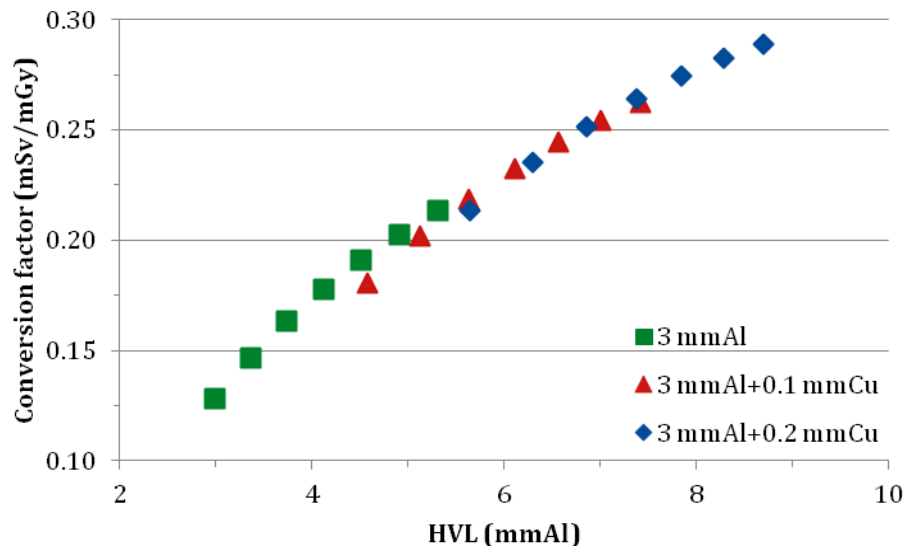
**Figure 2.** Breast and lung organ doses in the example thorax PA examination of Table 2.

### 3.2 Effect of beam properties

Conversion factors from incident air kerma to effective doses for the thorax PA examination with different radiation qualities are represented as a function of the tube voltage and HVL in Figures 3 and 4, respectively. Radiation quality had a strong effect on the conversion factors, which differed by a factor of more than two in the studied range. The tube voltage was not a good parameter for specifying the radiation quality alone. However, the HVL could be used reasonably accurately (see Figure 4).

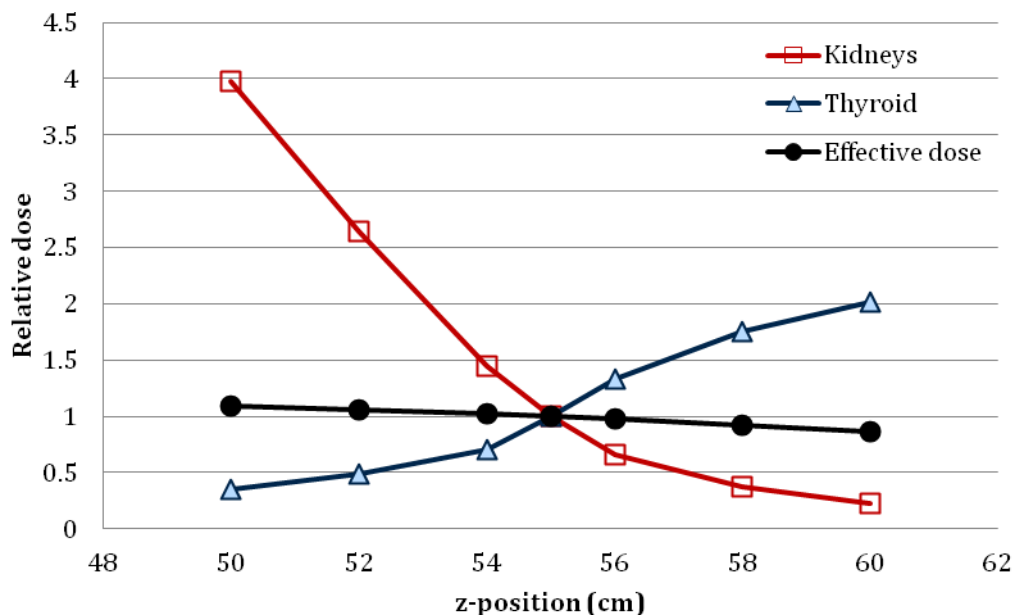


**Figure 3.** Conversion factor from incident air kerma to effective dose for the example thorax PA examination of Table 2 for different total filtrations as a function of tube voltages.



**Figure 4.** Conversion factor from incident air kerma to effective dose for the example thorax PA examination of Table 2 for different total filtrations as a function of the half-value layer (HVL).

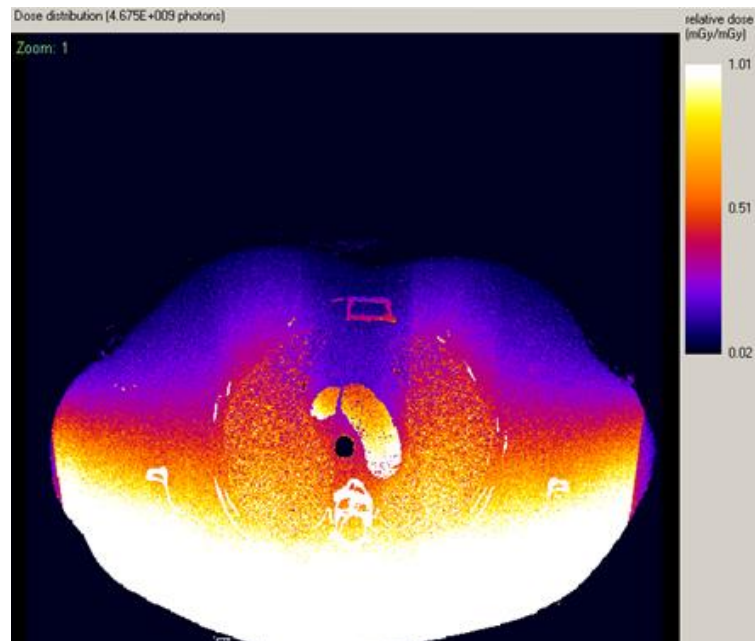
The focus-skin distance (FSD) depends on patient thickness and is not normally readily known. In our study, we calculated conversion factors and used a fixed incident air kerma for the calculations, and changing the FSD therefore only changed the area of the incident beam. In the studied FSD range, the maximum variation compared to the reference situation was 17%. As the beam area changes, organs are positioned differently in relation to the beam. Some organs (such as kidneys or thyroid, Figure 5) that were originally outside the primary beam became partially in the beam when the FSD increased. The effect of beam positioning was also examined by changing the position of the X-ray beam in the vertical ( $z$ ) direction (Figure 5). In PCXMC, the position of the beam is adjusted by changing the  $z$ -coordinate of the reference point; this point describes the location of the X-ray beam central axis with respect to the patient. In the case of the thorax examination, the total variation of the effective dose was 23% compared to the reference situation in the studied range ( $50 \text{ cm} \leq z_{\text{ref}} \leq 60 \text{ cm}$ ). However, the effect on a specific organ can be very large. For example, in Figure 5 the dose to the kidneys rapidly decreases as the reference point of the beam is moved upwards. At the lowest beam position, the kidneys are partially in the primary beam, while they are entirely outside the primary beam for  $z_{\text{ref}}$  values higher than 60 cm. Contrary to this, the thyroid is outside the primary beam until  $z_{\text{ref}}$  is 53.8 cm. As the effective dose combines the doses to various organs, the effect on the effective dose is much lower. These beam positioning effects are examination specific and depend on the anatomical area.



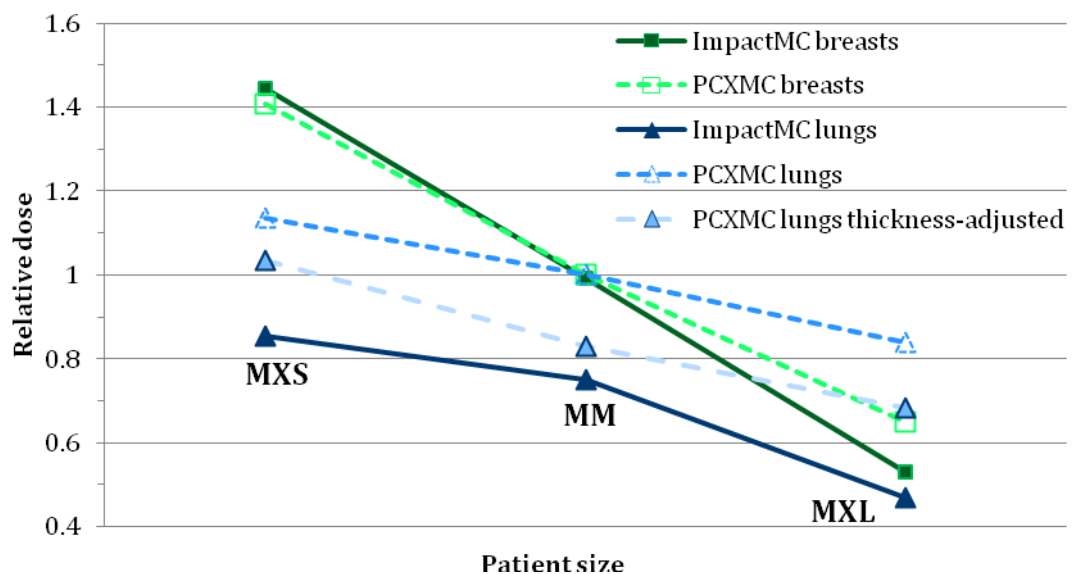
**Figure 5.** Relative organ doses and effective dose for thorax PA examination as a function of the vertical position of the X-ray beam.

### 3.3 Effect of patient size

An example of the dose distribution in a patient-specific simulation using ImpactMC is provided in Figure 6. Organ dose results from all patient simulations of Table 4 were compared with the values calculated using PCXMC (Figure 7). Breast doses agreed well, but there was a large difference between the lung doses. This was assumed to result from the different thickness of the phantom and the real patient (see the values from Table 4). Therefore, simulations with PCXMC were repeated by using the real patient thickness as the phantom thickness by changing the weight of the patient, and thickness-adjusted lung doses are also presented in Figure 7. The remaining differences can be explained by other differences between the real patients and the phantom model. When the values of small and large patients are compared with those of medium-sized patients, the total variations are up to 92% in breast doses, 38% in lung doses and 38% in effective doses.



**Figure 6.** ImpactMC (CT Imaging, Erlangen, Germany) simulated dose distribution for MXL patient.



**Figure 7.** Comparison of relative organ dose conversion factors calculated using simulations with ImpactMC and PCXMC for different patient sizes of Table 4.

### 3.4 Uncertainties

The international recommendation for the accuracy of dose measurement in diagnostic radiology is 7% ( $k = 2$ ) if the value is used for the optimization of patient doses (Wagner et al. 1992, ICRU 2005, IAEA 2007). This is related to the statement that even a 10% reduction in the patient dose is a worthwhile objective for optimization (IAEA 2007). If optimization tasks are close to the 10% level, this sets high requirements for accuracy in patient dose

estimations. In this study, uncertainties were only estimated for one clinical case. However, similar variation can be expected for other clinical cases. Table 5 roughly shows the maximum differences compared to the smallest value in effective dose estimates that may be encountered in real clinical settings. Clearly, the accuracy requirement for a measurable quantity is very high compared to the uncertainties related to conversions. If dose estimation in the optimization process is only based on a standard-sized patient model, the outcome of the study can be wrong. This emphasizes the need for careful conversion of measurable doses. However, it should be kept in the mind that the next step, risk estimation, may have large uncertainties of up to an order of magnitude or more (BEIR 2006).

**Table 5.** Maximum differences related to conversion factor in estimating the effective dose in thorax PA examination for a large range of exposure conditions and patient properties.

Property	Maximum difference (%)
Beam positioning	30
Radiation quality	130
Patient size	50

## 4 Discussion

Different levels of accuracy are needed for organ dose and effective dose estimations depending on the purpose of use. In many cases, for example in the case of an anxious patient, a typical effective dose value for the examination in question is usually enough. Such estimates give, by nature, the order of magnitude of the exposure (and subsequent cancer risk). However, more accurate dose conversion factors from measured values to organ doses and the effective dose are needed for research and the optimization of patient doses. Tabulated conversion factors have been published, but they are normally average values for some specific situation and certain assumptions about the beam positioning, radiation quality and patient model are made. The most accurate organ dose assessment is based on patient-specific calculations, where the individual properties of the patient are taken into account. This may be accomplished by Monte Carlo radiation transport simulations in phantoms based, for example, on a CT scan of the particular patient. However, this approach is very time consuming and cannot be routinely used in clinical work.

Based on our study, most of the respondents used the PCXMC program for organ dose calculations in thorax examinations. In this program, the radiation quality, X-ray beam size and positioning can be accurately simulated if the information is available. However, the patient model is based on a standard MIRD-type phantom that does not correspond to a real patient model in all cases. In PCXMC it is possible to scale the phantom size to represent the real patient weight and height so that the user is able to fit the correct beam size for patients of different sizes. However, this property does not always enable accurate estimation of the effect of patient size on organ doses.

## 5 Conclusion

The effect of patient size on dose conversion factors should be studied more in projection imaging. An adequate accuracy in organ and effective dose estimations could be achieved by using measured dose data together with a simulation or radiation quality-specific conversion factor and a patient size-related correction factor.

## Acknowledgements

This study is related to a coordinated research project (CRP E2.10.08: Development of advanced dosimetry techniques for diagnostic and interventional radiology) of the International Atomic Energy Agency. All the participants of this project are thanked for their cooperation.

## References

American Association of Physicist in Medicine (AAPM) 2011 Size-specific Dose estimates (SSDE) in Pediatric and Adult Body CT Examinations *AAPM Report No 204*

Angel E, Yaghmai N, Jude C M, DeMarco J J, Cagnon C H, Goldin J G, Primak A N, Stevens D M, Cody D D, McCollough C H and McNitt-Gray M F 2009 Monte Carlo simulations to assess the effects of tube current modulation on breast dose for multidetector CT *Phys. Med. Biol.* **54** 497–511

Abramoff M D, Magalhaes P J, Ram S J 2004 Image Processing with ImageJ *Biophotonics International* **11** (7) 36-42



BEIR- Committee to Assess Health Risks from Exposure to Low Levels of Ionizing Radiation 2006 Health risks from exposure to low level of ionizing radiation BEIR VII Washington D.C: National Academy of Sciences

Birch R and Marshall M 1979 Computation of bremsstrahlung x-ray spectra and comparison with spectra measured with a Ge(Li) detector *Phys. Med. Biol.* **24** 505–17

Campos de Oliveira C P M, Squair P L, Lacerda M A and Da Silva T A 2011 Assessment of organ absorbed doses in patients undergoing chest X-ray examinations by Monte Carlo based softwares and phantom dosimetry *Radiation Measurements* **46** 2073 – 2076

Deak P, Straten M, Shrimpton P C, Zankl M, Kalender W A 2008 Validation of a Monte Carlo tool for patient-specific dose simulations in multi-slice computed tomography *Eur. Radiol.* **18** 759–72

Drexler G, Panzer W, Widenmann L, Williams G and Zankl M 1990 The calculation of dose from external photon exposures using reference human phantoms and Monte Carlo methods Part 3 Organ doses in X-ray diagnosis *GSF-Bericht* **11/90** ISSN 0721-1694

European Commission (EC) 2008 Radiation Protection N°154 European Guidance on Estimating Population Doses from Medical X-Ray Procedures, Luxemburg

Hart, D, Jones, D G, Wall B F 1994 a Estimation of Effective Dose in Diagnostic Radiology from Entrance Surface Dose and Dose-Area Product Measurements *Rep. NRPB-R262* National Radiological Protection Board, Chilton

Hart, D, Jones, D G, Wall B F 1994 b Normalized Organ Doses for Medical X-Ray Examinations Calculated Using Monte Carlo Techniques *Rep. NRPB-SR262* National Radiological Protection Board, Chilton

Hart, D and Wall B F 2002 Radiation Exposure of the UK Population from Medical and Dental X-Ray Examinations National Radiological Protection Board NRPB-W4, Chilton

Iannuccelli E, Mompert F, Gellin J, Lahbib Y, Yerle M and Boudier T 2010 NEMO: a tool for analyzing gene and chromosome territory distributions from 3D-FISH experiments *Bioinformatics*

International Commission on Radiation Units and Measurements (ICRU) 2005 Patient dosimetry for x-rays used in medical imaging. ICRU Report 74 *J. ICRU* **5**(2) 1–113

International Atomic Energy Agency (IAEA) 2007 Dosimetry in diagnostic radiology: An international code of practice *Technical report series no. 457* Vienna, IAEA

International commission on radiological protection (ICRP) 2009 Adult Reference Computational Phantoms ICRP Publication 110 *Ann. ICRP* **39** (2).

International commission on radiological protection (ICRP) 2007 Recommendations of the International Commission on Radiological Protection, Publication 103, Pergamon Press, Oxford and New York

International commission on radiological protection (ICRP) 1991 1990 Recommendations of the International Commission on Radiological Protection, Publication 60, Pergamon Press, Oxford and New York

Jarry G, DeMarco J J, Beifuss U, Cagnon C H and McNitt-Gray M F 2003 A Monte Carlo-based method to estimate radiation dose from spiral CT: from phantom testing to patient-specific models *Phys. Med. Biol.* **48** 2645–63

Kramer R, Zankl M, Williams G and Drexler G 1982 The calculation of dose from external photon exposures using reference human phantoms and Monte Carlo methods. Part 1: the male (ADAM) and female (EVA) adult mathematical phantoms *GSF-Bericht* **S-885** ISSN 0721-1694

Kramer R, Khoury H J and Vieira J W 2008 CALDose\_X—a software tool for the assessment of organ and tissue doses, effective dose and cancer risk in diagnostic radiology *Phys. Med. Biol.* **53** 6437–59

- Li X, Samei E, Segars W P, Sturgeon G M, Colsher J G, Frush D P 2008 Patient-specific dose estimation for pediatric chest CT *Med. Phys.* **35** 5821-8
- Li X, Samei E, Segars W P, Sturgeon G M, Colsher J G, Frush D P 2011a Patient-specific Radiation Dose and Cancer Risk for Pediatric Chest CT *Radiology* Volume **259**: 862-74
- Li X, Samei E, Segars W P, Sturgeon G M, Colsher J G, Toncheva G, Yoshizumi T T, Frush D P 2011b Patient-specific radiation dose and cancer risk estimation in CT: Part II. Application to patients *Med. Phys.* **38** 408-19
- Mettler F, Huda W, Yoshizumi T and Mahesh M 2008 Effective Doses in Radiology and Diagnostic Nuclear Medicine: A Catalog *Radiology* Volume **248**: 254 – 263
- Rosenstein M 1988 Handbook of Selected Tissue Doses for Projections Common in Diagnostic Radiology, HHS Publication FDA 89-8031, Center for Devices and Radiological Health, Rockville, MD
- Schmidt B, Kalender WA 2002 A fast voxel-based Monte Carlo method for scanner- and patient-specific dose calculations in computed tomography *Physica Medica* **18**, N. 2
- Schultz FW, Geleijins J and Zoetelief J 1994 Calculation of dose conversion factors for posterior-anterior chest radiography of adults with a relatively high x-ray spectrum *Br J Radiol* **67** 775 - 785
- Schultz F W, Jansen J Zh M, Zoetelief J 2001 Organ and Effective Dose Conversion Factors for various Types of Examination in Diagnostic Radiology *Stratechreport: ST - MSF-2001-012*
- Tapiovaara M and Siiskonen T 2008 PCXMC. A Monte Carlo program for calculating patient doses in medical x-ray examinations (2<sup>nd</sup> Ed.). Report STUK-A231, Radiation and Nuclear Safety Authority, Helsinki
- Wagner L K, Fontenla D P, Kimme-Smith C, Rothenberg L N, Shepard J and Boone J M 1992 Recommendations on performance characteristics of diagnostic exposure meters: report of AAPM Diagnostic X-Ray Imaging Task Group No. 6 *Med. Phys.* **19** 231–41
- Yushkevich P A, Piven J, Hazlett H C, Smith R G, Ho S, Gee J C and Gerig G 2006 User-guided 3D active contour segmentation of anatomical structures: Significantly improved efficiency and reliability. *Neuroimage* **31(3)** 1116-28