MEDIRAD

Project title: Implications of Medical Low Dose Radiation Exposure

Grant Agreement Number: 755523

Call identifier: NFRP-2016-2017

Topic: NFRP-9

Deliverable D2.14

Report on a) the organ dose data representation and b) collected organ doses

Lead partner: University of Crete (UoC)

Author(s): Marios Myronakis, John Stratakis, John Damilakis

Work Package: WP2

Estimated delivery: 30/04/2020

Actual delivery: 29/05/2020

Type: Report

Dissemination level: Public

This project has received funding from the Euratom research and training programme 2014-2018 under grant agreement No 755523

Table of Contents

List of Figures

List of Tables

Abbreviations

API	Application Programming Interface
3D	Three-dimensional
ATCM	Automatic Tube Current Modulation
CT	Computed Tomography
CTDI	Computed Tomography Dose Index
CUDA	Compute Unified Device Architecture
DICOM	Digital Imaging and Communication in Medicine
FOV	Field of View
FTC	Fixed Tube Current
GUI	Graphical User Interface
HU	Hounsfield (scale) Unit
IRDBB	Image and Radiation Dose BioBank
MC	Monte Carlo
NAS	Network Attached Storage
PACS	Picture Archiving and Communications System
SSDE	Size-Specific Dose Estimates
WED	Water Equivalent Diameter

1. Development of a method to estimate patient organ dose from chest CT

Radiation risk to human health has been the subject of continuous research and debate for many years. Radiation exposure associated to computed tomography (CT) examinations has been identified as a major source of medical radiation burden for adult and pediatric subjects [1,2]. Attention has been emphasized on children as they are often considered to be more vulnerable to radiation due to their increased radio sensitivity and longer life span expectancy [3]. To assess stochastic radiogenic risks resulting from a CT examination, the dose to specific radiosensitive organs should be estimated [4].

The primary objective of the task 2.1.2 was to propose a method for the accurate estimation of dose to primarily irradiated organ structures (bones, esophagus, breast (female), heart, lungs, skin) in pediatric and adult CT examinations by considering patient size, automatic tube current modulation and specific CT-scanner characteristics with the use of patient-specific computational models.

2. Patient-based, scanner-specific organ dose estimation procedure

The proposed method for scanner-specific and patient-specific, organ-dose estimation and optimization, is based on a procedure that combines Monte Carlo computational techniques and patient CT scans. In general, the procedure is composed of five steps, detailed in subsequent paragraphs; (a) collection of suitable patient CT scans; (b) determination of scanner parameters; (c) Monte Carlo (MC) dosimetric computations; (d) three-dimensional (3D) patient-specific dose distribution output and (e) correlation between dose and patient characteristics. These five steps need to be performed only once for each CT scanner. Thereafter, the results obtained at step (e), i.e., the correlation between organ-dose and patient characteristics, is used to calculate organ doses for subsequent patients that undergo chest CT scans.

3. Collection of patient CT-scans

3.1 Pediatric Patients

Images from ninety-seven (97) pediatric patients were included in this task. All images were selected and extracted from the local Picture Archiving and Communication System (PACS) database following specific data protection guidelines as proposed by the MEDIRAD Data Management Board to ensure data privacy and security.

3.2 Adult Patients

In addition, standard thoracic CT data from two hundred and three (203) adult individuals with well-defined clinical indications (pulmonary-fibrosis, infectious diseases, metastatic) were also selected and extracted from hospital PACS and included in this task. Likewise, specific data protection scheme was followed as proposed by the MEDIRAD Data Management Board [5] for adult datasets.

3.3 Collection of patient CT-scans: Patient data privacy and security

Specifically, the following steps were taken: 1) Data was collected on-site, inside private-IP hospital network through PACS DICOM receive protocol (same IP-space) 2) initial data never left hospital infrastructure, 3) no external email/disk database was created, 4) two dedicated workstations were used for processing with Bitlocked® drives and were backed up to local network storage (NAS) using 1024-bit encryption. Before delineation and MC simulations, all CT images collected were anonymized and patient and exams id were kept in a different password protected archive, so as images' identifiers were all limited to alphanumeric characters with no direct connection to patient personal data.

3.4 Anonymization of patient sensitive data

The Digital Imaging and Communication in Medicine (DICOM) standard has been commonly used for storing, viewing, and transmitting information in medical imaging. DICOM was developed to ease the exchange of data between different manufacturers, but it also enables data sharing between institutions or enterprises for clinical research or clinical practice. A DICOM file not only

contains a viewable image, but it also encompasses a header with a large variety of data elements. Each data element is represented by a unique identifier (tag) with specific values and data types. Each tag is written with two hexadecimal numbers indicating its group and element number. These meta-data elements include information about the patient, the study, and the institution. Handling such sensitive data may demand proper handling to maintain patient privacy [6].

There are two methods to de-identify patient-related information in a DICOM header. The first method is anonymization which removes information carried by header elements or replaces the information with random data such that the remaining information cannot be used to reveal the patient identity at all. The other method, pseudo-anonymization, is implemented by replacing these fields using artificial identifiers that could be used by authorized personnel to track down the real identity of the patient in case of clinical analysis, processing, and research since good clinical practice requires that, should essential findings be encountered, it should be possible to track the patient to inform about these findings. An in-house developed Python application was used, incorporating toolboxes to access and edit DICOM Metadata. De-identified DICOM header tags are listed in Appendix 1. Patient age was a field that needs to be preserved, in order to calculate radiation induced risks.

3.5 Collection of patient CT-scans: Suitability of pediatric and adult scans for estimating patient organ dose from chest CT

Thorax CT data must have specific characteristics to be suitable for constructing patient-specific models for MC dosimetric computations. No contrast agent must be present in any of the reconstructed slices. Iodine contrast may notably affect the deposited radiation dose, as the application of iodinated contrast medium is accompanied by an increase in radiation absorption and attenuation through specific organ/tissues. In addition, for the accurate execution of all patient-model MC simulations, the reconstructed Field-of-View (FOV) of the examination must be suitable to maintain all volumetric information from the CT study. Parts of patient tissue not present within the boundaries of the CT-scan FOV will affect x-ray absorption and transmission through patient model resulting in dose inaccuracies. CT data with extreme patient anatomy/pathology was excluded; excessive buildup of fluid in thorax, lung necrosis or lung atelectasis/pneumothorax, situs inversus (a congenital condition in which the major visceral

organs are reversed/mirrored from their normal positions). The final number of thorax CT examinations that was used for constructing patient models was 300 i.e. 203 adult and 97 pediatric examinations.

4. Monte Carlo dosimetric computations

4.1 Monte Carlo software

The MC software selected for dosimetric computations was ImpactMC (version 1.6, CT Imaging© GMBH, Erlangen, Germany). ImpactMC is a software package providing fast calculation of 3D dose distributions for computed tomography (CT) scans using Monte Carlo algorithms. ImpactMC is a well validated MC software, specifically designed for 3D dosimetric evaluation on CT acquired images [7-10].

Based on the CT volumetric data as input, an individualized voxelized Monte Carlo simulation is performed in order to calculate the dose deposited. Dose is calculated on a per image voxel basis, considering all available physical interactions. ImpactMC's output comprises of parametric dose images in which every voxel carries the normalized dose (to CTDI) of the initial image grid. The generated 3D dose distributions are useful to estimate organ doses and to calculate effective dose of individual scans. Figure 1 shows a screenshot of the graphical user interface. The graphical user interface (GUI) consists of two main panels - one for parameter input and one for visualization. At the bottom of the image viewer a panel gives information on dose, density and material for the voxel selected with the mouse cursor.

ImpactMC utilizes any available Nvidia graphics processing unit (GPU) to accelerate computations. Specifically, it makes use of any CUDA-enabled GPU, employing NVIDIA's parallel computing platform and application programming interface (API) model. The CUDA platform is a software layer that gives direct access to the GPU's virtual instruction set and parallel computational elements (CUDA cores). In conjunction with high-speed parallel processing within the GPU, higher GPU memory bandwidth facilitates greater compute efficiency compared to CPU-only intensive calculations [11].

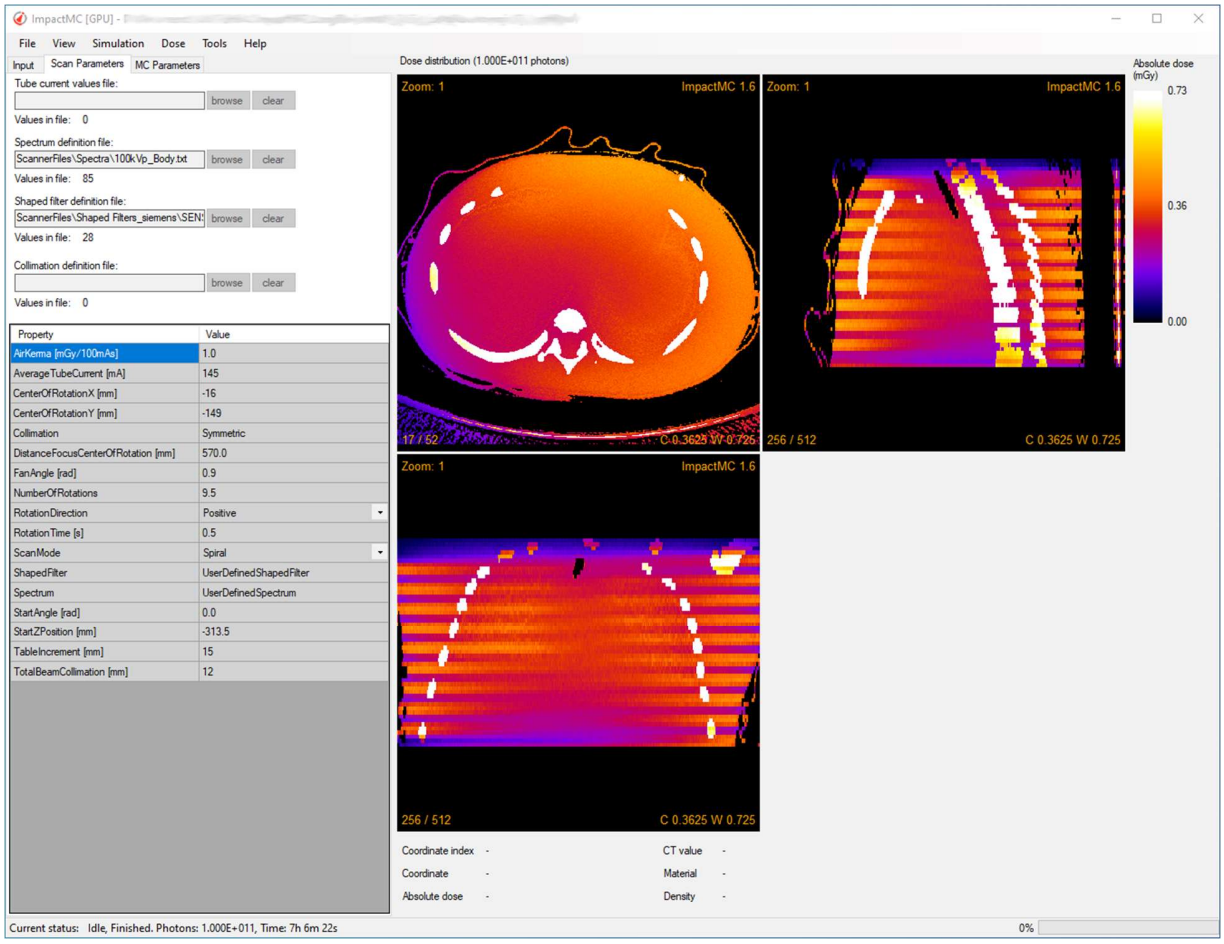


Figure 1:The graphical user interface of ImpactMC.

Each estimation of the dose distribution in a CT volume starts with setting up an ImpactMC project. Figure 2 provides an overview of the project structure.

The following input data were required by the software to start the dose computation procedure:

1. Input volume: The input volume represents the scanned patient or phantom; a set of CT reconstructed images from one examination, in DICOM format. In the current work the input volume is each patient (pediatric or adult) thorax CT.
2. Scan parameters: Data for beam spectrum, filtration and geometrical specifications (Table 1)
3. Simulation parameters: The number of x-rays depositing energy in the input volume was selected. Good statistical performance (<1% uncertainty) was obtained using a value in the order of 10^8 to 10^9 interacting x-rays.

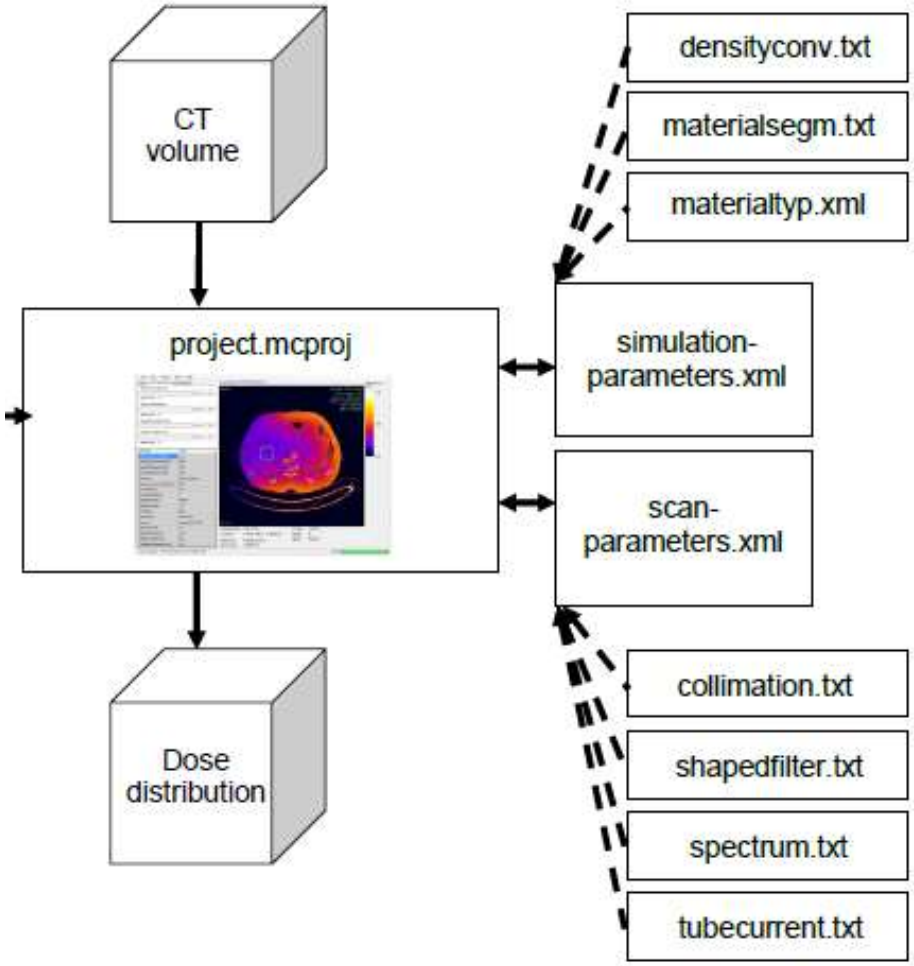


Figure 2: ImpactMC project structure.

4.2 Determination of scanner parameters – geometric characteristics

Scanner-specific dose computations required the parameterization of the Monte Carlo software according to operating and physical characteristics of each scanner considered. CT scanner models were based on data for x-ray beam spectra, beam shaping devices (bow-tie filters) and geometrical specifications. The required data were provided by the manufacturers and later compiled and converted to input parameters suitable for the Monte Carlo software. Geometrical specifications and beam shaping device modes used for scanner modelling are presented in Table 1.

Scanner	Focus to Isocenter (mm)	Fan Angle (rad)	Beam collimation (mm)	Beam shaping devices
Siemens SOMATOM 16	570	0.900	24.0	Small, Large
Siemens SOMATOM 64	570	0.900	19.2	Fixed
GE Revolution GSI 64	539	0.875	40.0	Small, Medium, Large

Table 1: Simulated CT Scanner geometrical characteristics.

4.3 Determination of scanner parameters –Spectral characteristics

Beam spectra for all modes of operation, i.e. 80kV, 100kV, 120kV and 140kV were produced for each scanner using both IMPACTMC integrated spectrum creator and SpekCalc spectrum calculation program. ImpactMC spectrum creator module is based on work of Tucker et al [12].

In addition, SpekCalc (Figure 3) allows the user to calculate and display the x-ray spectra emitted from tungsten-anode x-ray tubes. The underlying theoretical description for the bremsstrahlung and characteristic x-ray production was taken from a recently published model [13]. The user can select tube potential in kVp, the take-off angle and the amount of filtration. The resulting spectrum can be calculated, displayed, and saved/exported for later use. Beam quality parameters such as the half-value-layer, in mm of aluminum and copper, and the mean beam energy, in keV, can also be considered. Filtration can be selected in mm, for 7 materials (eg, Al, Co, Zn, etc)

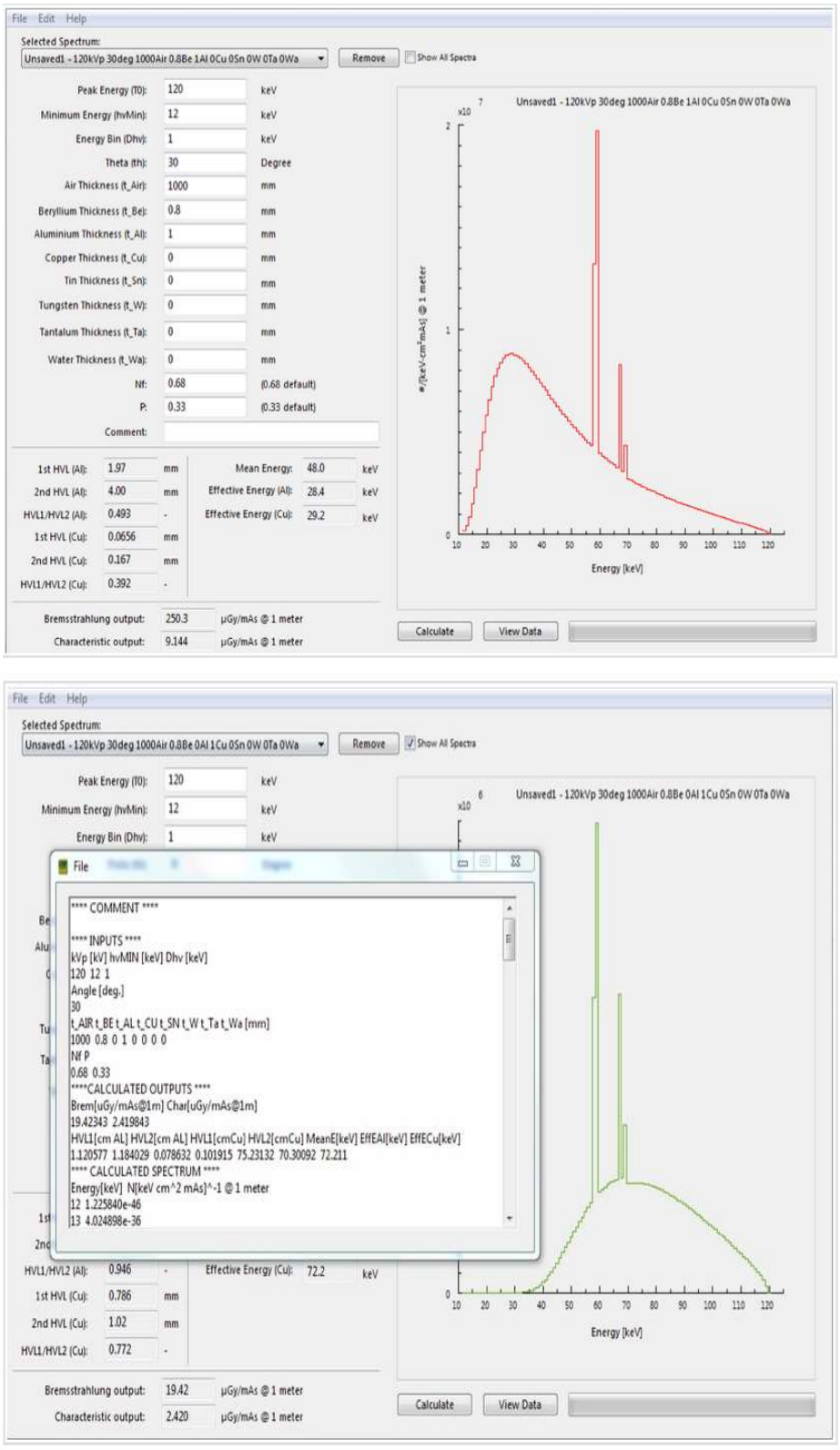
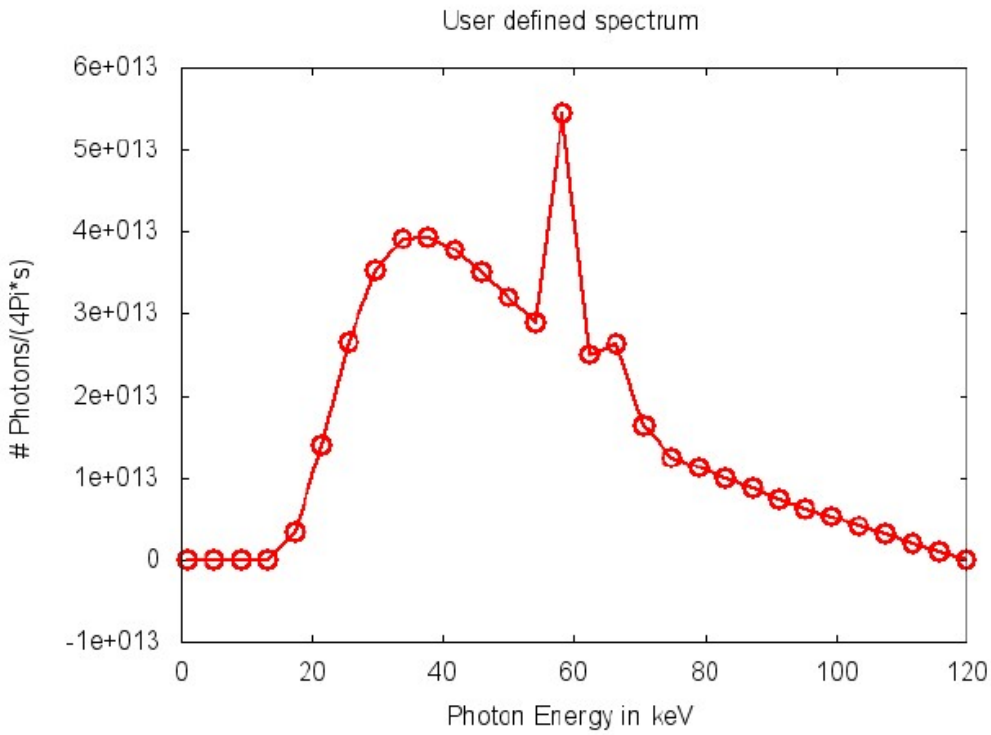


Figure 3: SpekCalc spectrum calculation software GUI.

The structure of IMPACTMC 's spectrum text file consists of several entries, each defining the number of photons at a certain energy level (Figure 4). Each line has the following syntax:

energy e (in keV), number of photons in the spectrum in energy bin e

The values for e must be positive, monotonically increasing and equidistant. The number of photons did not have to be normalized because normalization was done by the software, if necessary. This text file was imported using the GUI.




```
# Spectral Data File
# Energy, Photons/(4Pi*s)
1.000, 0.0
5.103, 9.1194e-030
9.207, 1.7386e+006
13.310,9.3135e+010
17.414,3.3558e+012
21.517,1.3956e+013
25.621,2.6502e+013
29.724,3.5191e+013
33.828,3.9030e+013
```

Figure 4: A user defined spectrum and spectral data file.

4.4 Determination of scanner parameters – Tube current modulation

All modeled scanners were capable for fixed current and modulated current acquisitions. The DICOM header of each reconstructed image (z) included a unique mAz value along with the corresponding table location. Each mAz value is the average of the angularly and longitudinally modulated values applied over the gantry rotation used to reconstruct this z-th image. mAz profile values were retrieved from the DICOM header of the image data using ImageJ manipulation software and used as a custom tube current file for the MC simulation software (Figure 5).

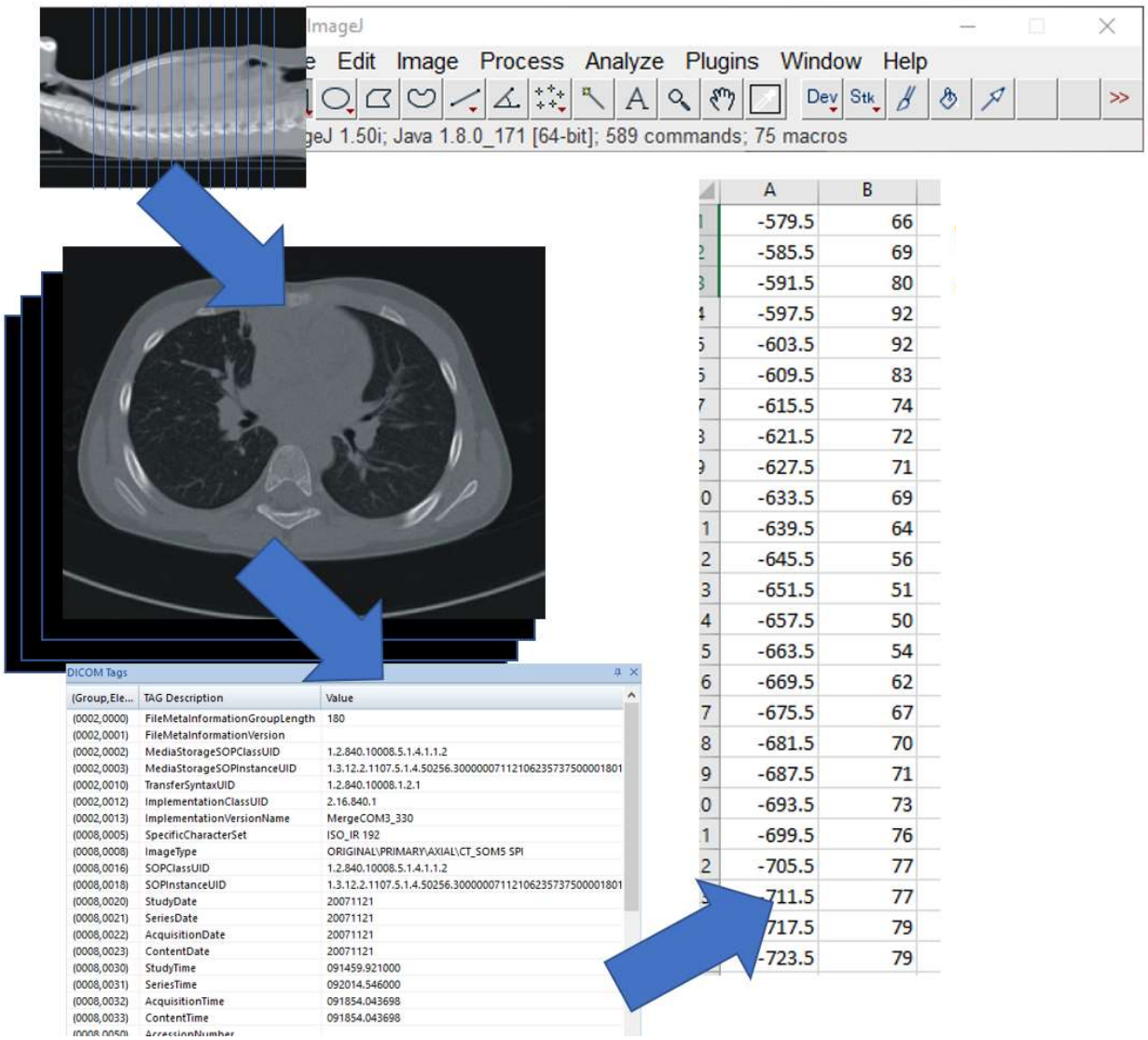


Figure 5: Graphical representation of extraction of longitudinally modulated tube current from DICOM header. Columns A and B represent z-axis position and tube current, respectively.

4.5 Determination of scanner parameters – CT shaping filters modelling

In ImpactMC software, the user has the option to include a shaped filter where the thickness of the filter is varying with the fan angle (Figure 6) at the x-ray source in the MC simulations. The shaped filter can be switched off and on from the GUI. The user can add a shaped filter from a specification file. In this case the values for filter material and thickness are read from the file according to the following file structure:

line 1: Filter (keyword)
line 2: # Name of the material (mandatory comment line)
line 3: Name of the material, e.g. Al. (software specific name)
line 4: # Density of the material (g/cm^3) (mandatory comment line)
line 5: Density of the material in g per cm3,
line 6: # Angular increment (°) (mandatory comment line)
line 7: Angular increment between the given thickness values in degrees.
line 8: # Thickness array (mm) (mandatory comment line)
line 9: thickness of the shaped filter at angle = 0°, i.e. in the middle of the fan beam
line 10: thickness of the shaped filter at angle +/- angular increment
following lines: thickness of the shaped filter at n angle +/- angular increment

```
Filter
# Name of the material
Al
# Density of the material (g/cm^3)
2.7
# Angular increment (°)
5.0
# Thickness array (mm)
0.49014
0.74040
1.30470
2.15470
3.26210
4.89555


Filter
# Name of the material
C
# Density of the material (g/cm^3)
1.88
# Angular increment (°)
10.0
# Thickness array (mm)
0.49014
0.74040
1.30470
2.15470
```

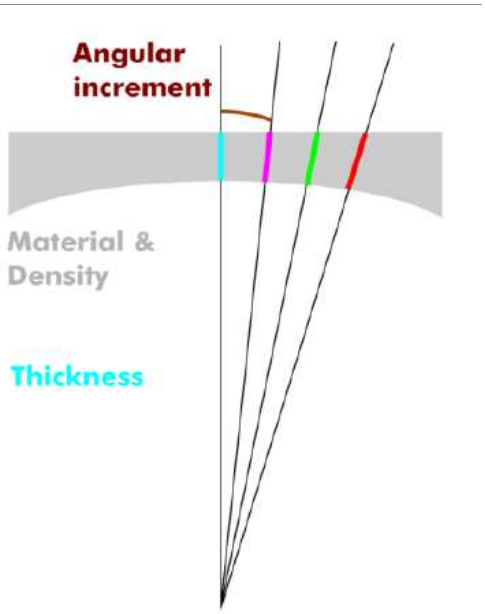


Figure 6: Definition of the shaped filter material and density, angular increment and thickness (Adapted from ImpactMC user-manual).

4.6 Acquisition protocols parameters

CT exposure for pediatric and adult cases were performed with predefined protocols suitable for thoracic imaging in all scanners. These protocols have specific settings that were translated to corresponding parameter values in the Monte Carlo software. For each voxelized model, the number of rotations per simulation was calculated based on the length of the imaged volume, beam collimation and pitch. Table 2 contains a list of CT settings and their equivalent value in the Monte Carlo software.

CT Setting	**Monte Carlo Parameter**
kV	User defined file with spectrum
Beam Collimation (mm)	Total Beam Collimation (mm)
mA modulation	Tube Current Values File
mA fixed	Average Tube Current
Pitch	Table Feed (*)
Rotation time (s)	Rotation Time (s)

Table 2: CT exposure settings and corresponding Monte Carlo parameters.

(*) Pitch is converted to table feed using the total beam collimation and pitch value

4.7 Monte Carlo simulations

Patient CT scans were used as input volumes to perform dosimetric computations in ImpactMC. Pediatric cases were simulated for 80, 100, and 120 kV for each scanner. Adult cases were simulated for 80, 100, 120, and 140 kV. A subset of the collected CT scans was acquired using tube current modulation (Figure 7).

Two simulations were performed for each model: (i) a simulation with a Fixed Tube Current (FTC) value and (ii) a simulation with the ATCM system activated. In the first simulation, the mA was kept constant in all tube rotations throughout the entire examination length. In the second simulation, a different mA value was applied for each individual tube rotation. The ATCM system available on the simulated Siemens CT scanner applies combined angular and longitudinal mA modulation.

The average tube current value was set equal to 100 mA over one second gantry rotation. Tube current normalization to 100 mA simplifies dose output processing without loss of accuracy. In total, the number of dosimetric simulations performed were over 6000 for pediatric and adult cases.

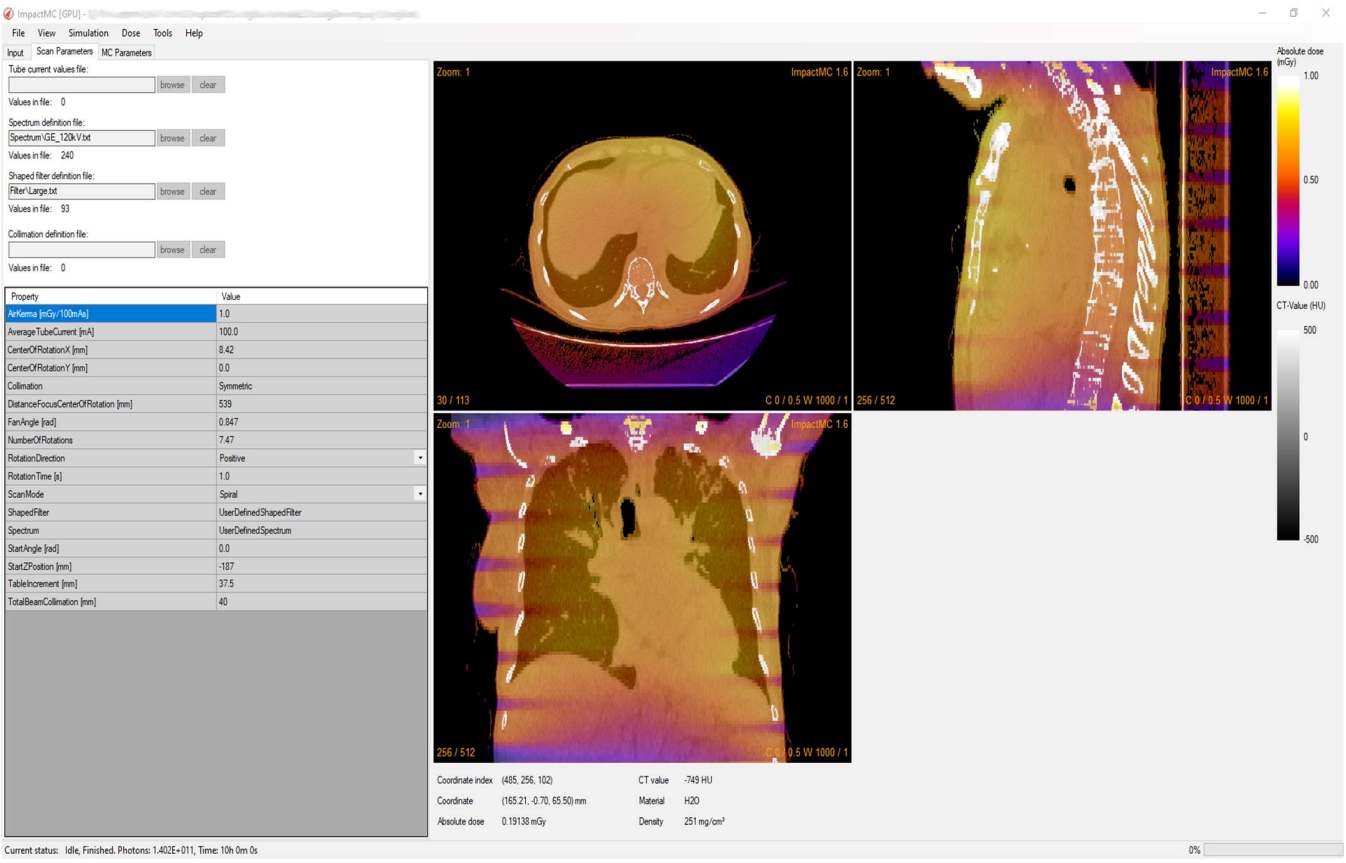


Figure 7: Example output of ImpactMC software after a successful simulation run.

5. Delineation of radiosensitive organs within the primary exposed volume

ImageJ and open source tools were utilized to delineate radiosensitive organs within the primary exposed volume of all pediatric and adult CT scans. Structures that were automatically delineated were bones, lungs and the outline of the patient. In order to segment these structures, thresholding techniques were utilized (Figure 8, Figure 9). Thresholding is a technique for dividing an image into two (or more) classes of pixels, which are typically called "foreground" and "background." Global thresholding was used within ImageJ which works by choosing a value cutoff, such that every pixel less than that value is considered one class, while every pixel greater than that value is considered the other class. For automatic segmentation of bones and lungs, iterative procedure based on the isodata clustering algorithm of Ridler & Calvard [14]. The procedure divides the image into object and background by taking an initial threshold, then the averages of the pixels at or below the threshold and pixels above are computed. The averages of those two values are computed, the threshold is incremented, and the process is repeated until the threshold is larger than the composite average. That is, threshold = (average background + average objects)/2. Lung entities were manually selected, and bone structures were automatically assembled using the analyze particles ImageJ tool, which extracts and measures objects in binary or thresholded images. Size and circularity were used to trace particles:

Size: Particles with size (area) outside the range specified in this field are ignored. Values may range between 0 and 'Infinity'. For DICOM images pixel units were used.

Circularity: Particles with size circularity values outside the range specified in this field are also ignored. Circularity (4π × [Area][Perimeter]2), ranges from 0 (infinitely elongated polygon) to 1 (perfect circle).

Esophagus, breast and heart were manually delineated by a radiologist using a Wacom tablet for added accuracy and by adjusting the brightness and contrast of the image to optimum levels (Figure 10). ImageJ ROI (Region of Interest) Manager was used for working with multiple selections. Multiple organ selections for each structure were saved as a ZIP file, containing multiple ROIs incorporating complete CT-sequence's organ assortment.

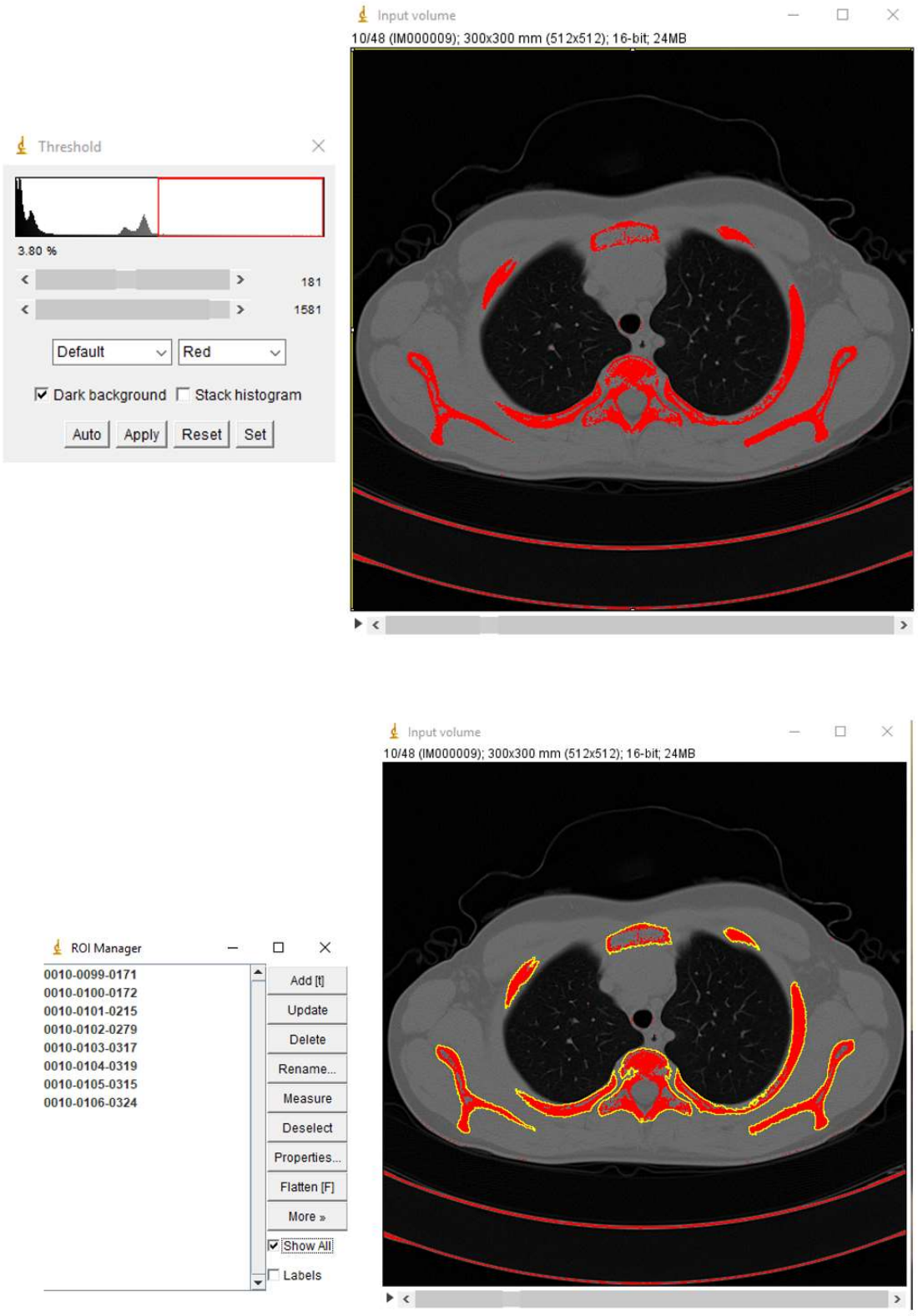


Figure 8: Automatic segmentation of bone structures using thresholding techniques.

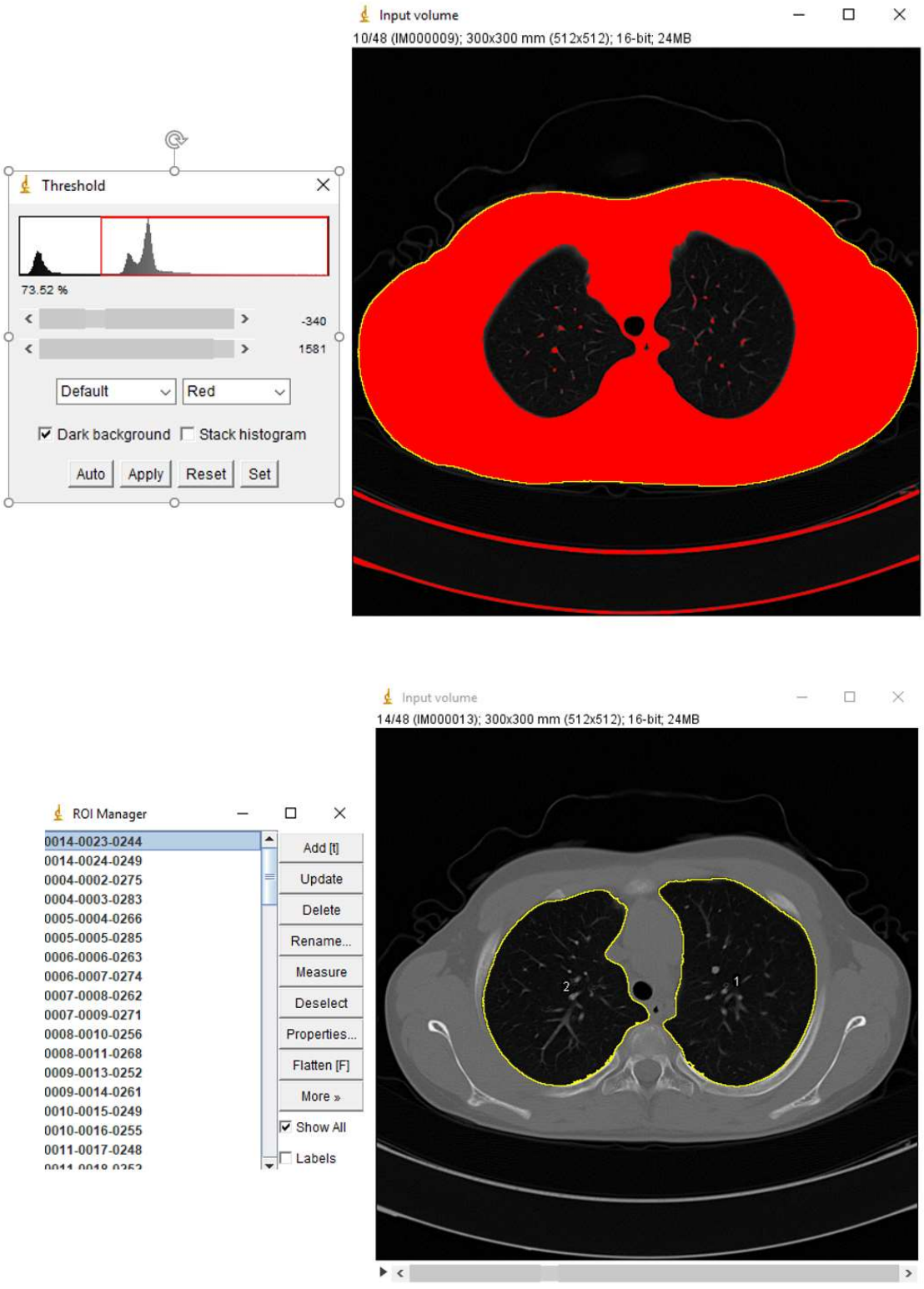


Figure 9:Automatic segmentation of skin contour and lungs using thresholding techniques.

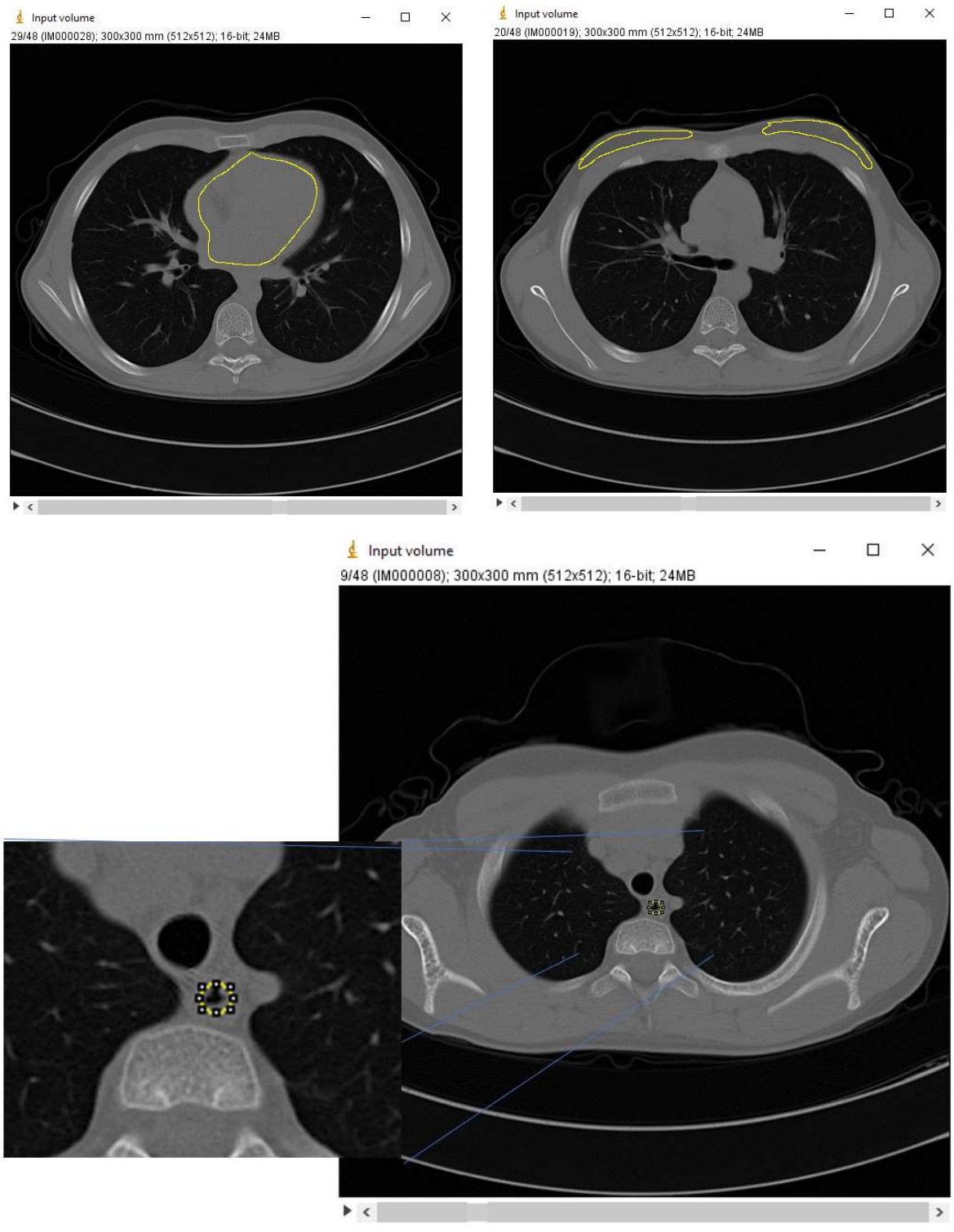
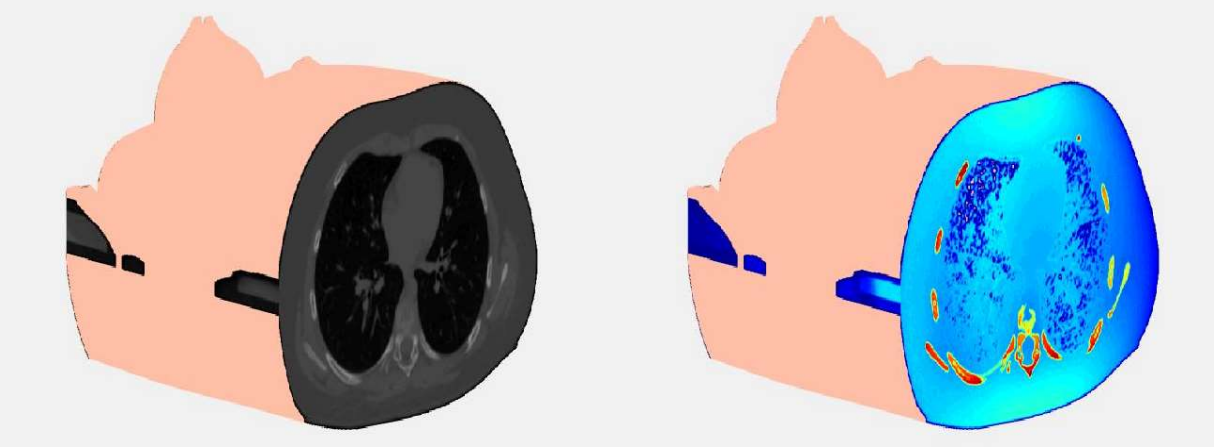


Figure 10:Manual delineation of esophagus, heart and breast contour using ImageJ drawing tools.

6. Patient-specific 3D dose distribution

The Monte Carlo software output after each computation was in the form of a 3D dose distribution (fig. 3), based on the physical properties (i.e. attenuation, composition and size) obtained from the input CT scan. Each slice in the dose volume corresponds to the same slice in the CT scan. Each pixel in a specific slice of the CT volume has a corresponding dose value in the 3D dose distribution output. The dose distribution was exported in DICOM file format. To facilitate dose data processing, the output dose was normalized to CT dose index in free air (CTDIair) and CTDIvol. The unit of dose values in the 3D volume was mGy/mGy per 100mAs.

Organ dose information was extracted from 3D dose distributions through appropriate delineation. The tissues of interest in this work were bones, lungs, esophagus, breast, heart and the skin.



(a) (b)

Figure 11: (a) CT image slices and (b) corresponding dose distribution in the thorax region.

The contours of an organ were overlaid on the corresponding slices of the dose distribution (Figure 12) and the respective dose was extracted. The dose over the whole organ was computed as:

$$D = \sum_{i}^{N} D_i \qquad (1)$$

where, D_i is the dose within the contour at slice i, and N the total number of slices that contain contours of a specific organ. The calculated dose (D) was normalized to 100 mA, pitch factor (p) equal to one, and rotation time (t_{rot}) equal to one second using the following formula:

$$D_n = D \times \frac{p \times 100}{t_{rot} \times mA} \qquad (2)$$

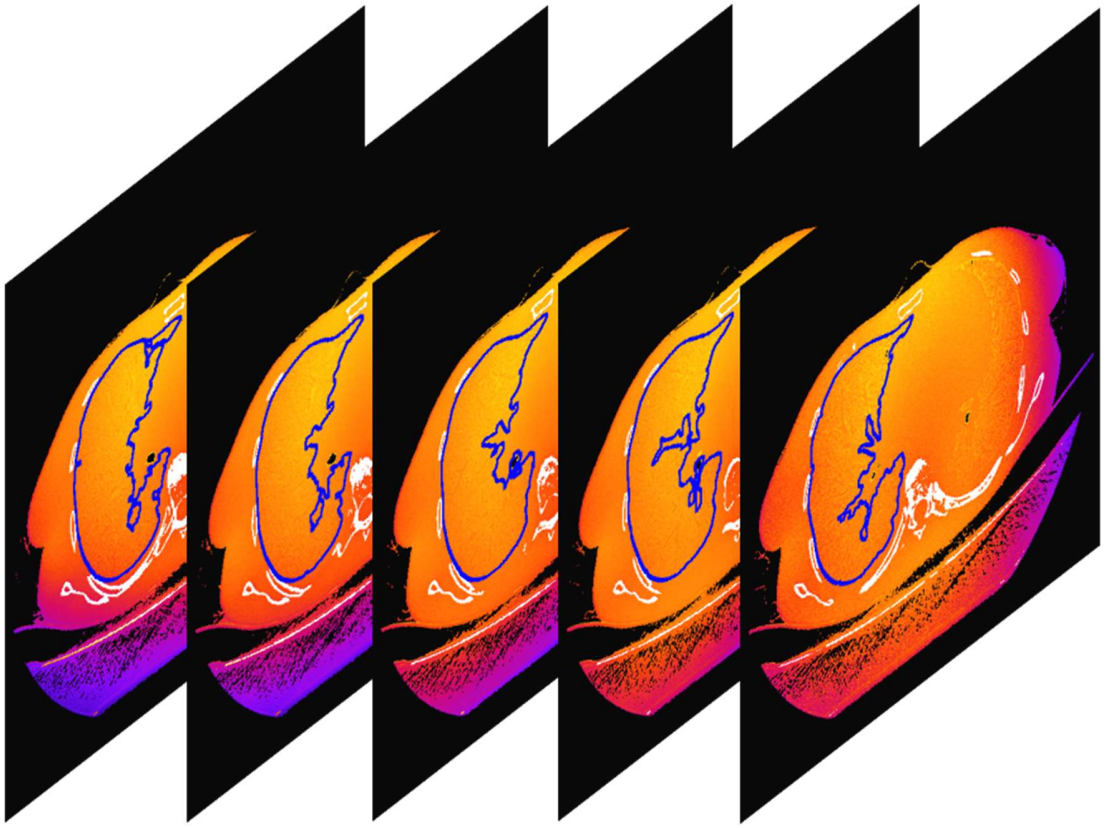
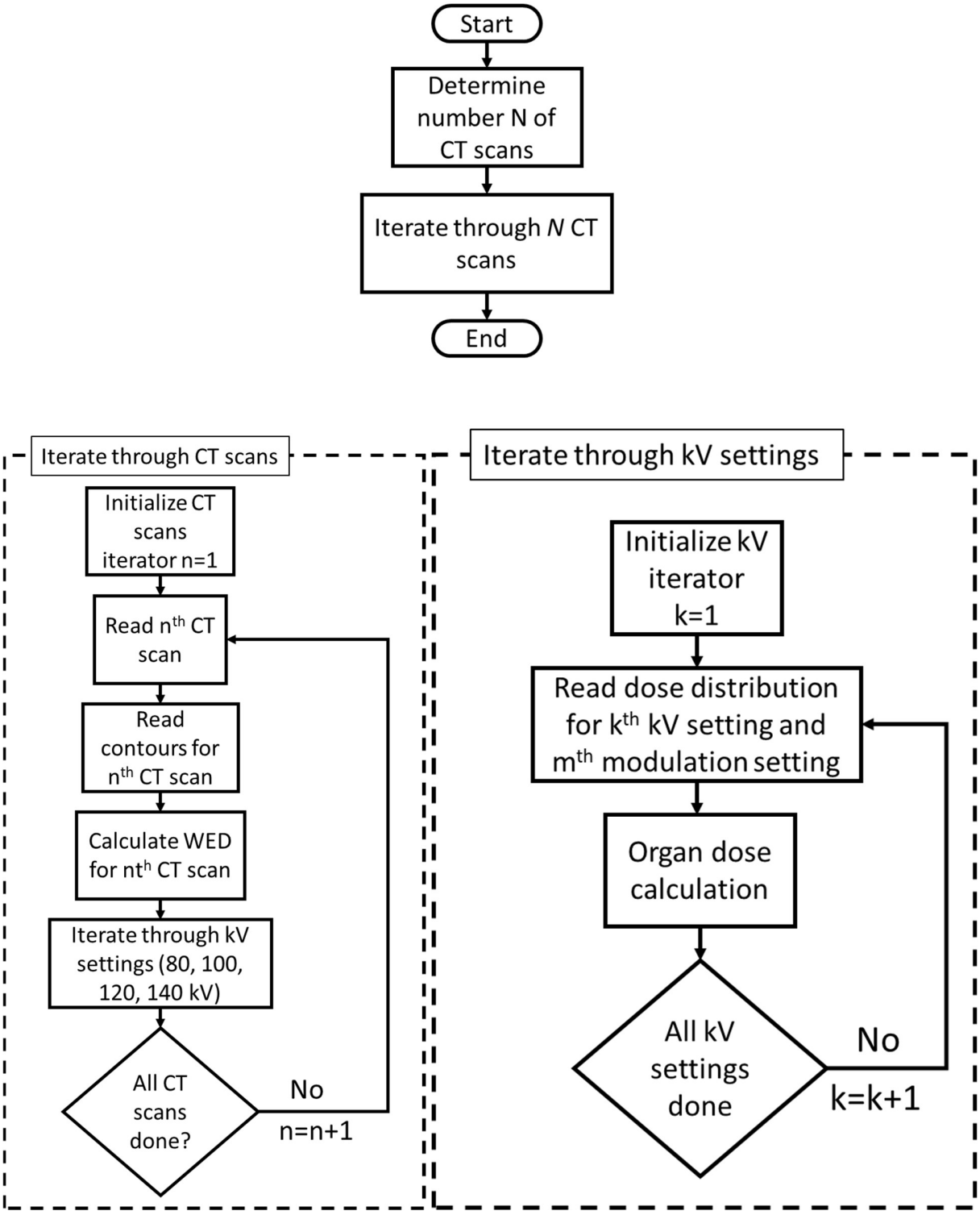


Figure 12: Graphical representation of right lung contours overlaid on corresponding dose slices.

Due to the large number of simulated results (over 6000 dose distributions), organ dose calculation procedure was automated. Relevant algorithms were implemented in ImageJ to provide unsupervised organ dose calculation, WED values and corresponding linear regression outcomes. The implemented procedure's flowchart is displayed in Figure 13.


Start
Determine number N of CT scans
Iterate through N CT scans
End
Iterate through CT scans
Initialize CT scans iterator n=1
Read nth CT scan
Read contours for nth CT scan
Calculate WED for nth CT scan
Iterate through kV settings (80, 100, 120, 140 kV)
All CT scans done?
No
n=n+1
Iterate through kV settings
Initialize kV iterator k=1
Read dose distribution for kth kV setting and mth modulation setting
Organ dose calculation
All kV settings done
No
k=k+1

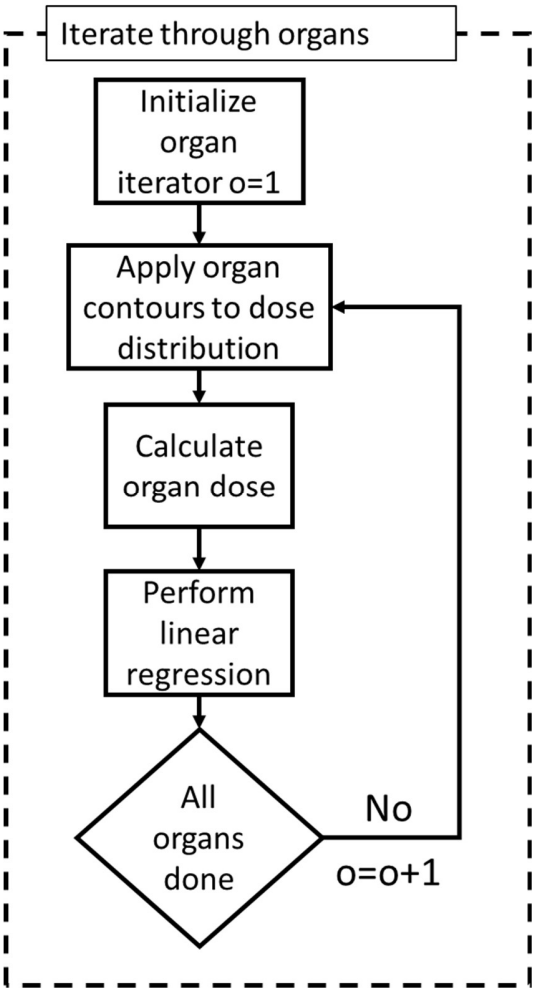


Figure 13: Flow-chart of automated patient-specific, organ-dose calculation algorithm.

7. Water-equivalent diameter (WED)

Water equivalent diameter (WED) is an attenuation-based size metric that may be used to quantify patient size in CT [15]. The WED size metric proposes the diameter of a hypothetical cylindrical water phantom that has the same average attenuation as the body region being examined (Figure 14).

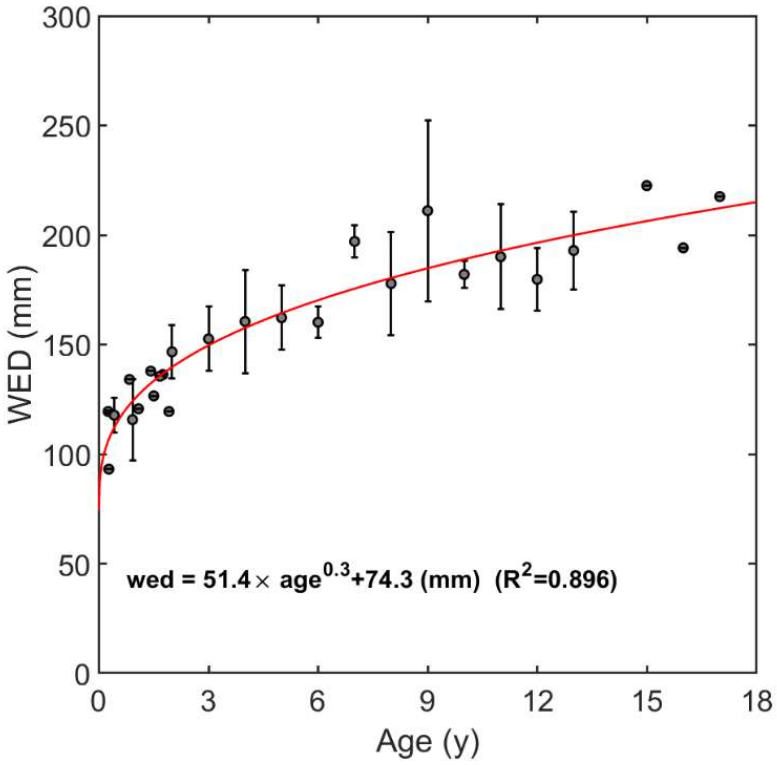


Figure 14: Pediatric WED versus age.

8. Water-equivalent diameter (WED) automatic calculation

AAPM TG220 recommended the adoption of water equivalent diameter (WED or Dw) as the preferred patient size metric for accurate size-specific dose estimates (SSDE) from CT scanning. WED calculation is performed on axial CT image slices and is based on the CT number, i.e. Hounsfield Units (HU), of the imaged patients using the following formula:

$$WED = 2 \times \sqrt{\left[\frac{1}{100}\overline{CT(x,y)_{ROI}} + 1\right] \times \frac{A_{ROI}}{\pi}} \qquad (3)$$

where $\overline{CT(x,y)_{ROI}}$ is the average HU value over each x, y location in the ROI that contains the imaged patient in one slice. A_{ROI} is the area of the ROI. WED can also be calculated on localizer (scout) images but yields less accurate results [15].

Accurate WED determination requires detailed delineation of the patient body outline on the CT slice without including the CT couch/table. In radiation therapy, body, organ and tissue delineation is part of the treatment planning routine, hence, body outline is readily available. In diagnostic radiology, such delineations are not usually provided. Although this task can be performed manually by experts, it is practically cumbersome and time-consuming over a large number of patient CT scans.

Within the scope of this project, an algorithm to automatically delineate the patient body and remove the couch from each CT slice was developed (body segmentation). The algorithm is based on morphological operations applied on each image slice (Figure 15a). A square structuring element with heuristically determined size equal to 12 pixels was used to perform morphological opening. This step effectively removed small structures in the vicinity of the body and preserved the body shape and size (Fig. 15b). The next step was conversion of the image to binary using Otsu's method to determine the single HU threshold between background and any other structure [16, 17]. During this step, the couch is usually removed from the image without affecting the body size or shape (Fig. 15c). Lungs and other low-density tissues are likely to produce empty space (holes) in the binarized image within the region defined by the body shape. This empty space was filled by detection of zero value locations that were not connected with the image borders (Fig. 15d). In the final step, non-zero areas that were not connected were identified and the largest area was kept (i.e. the body) – this step was rarely required but nevertheless performed (Fig. 15e).

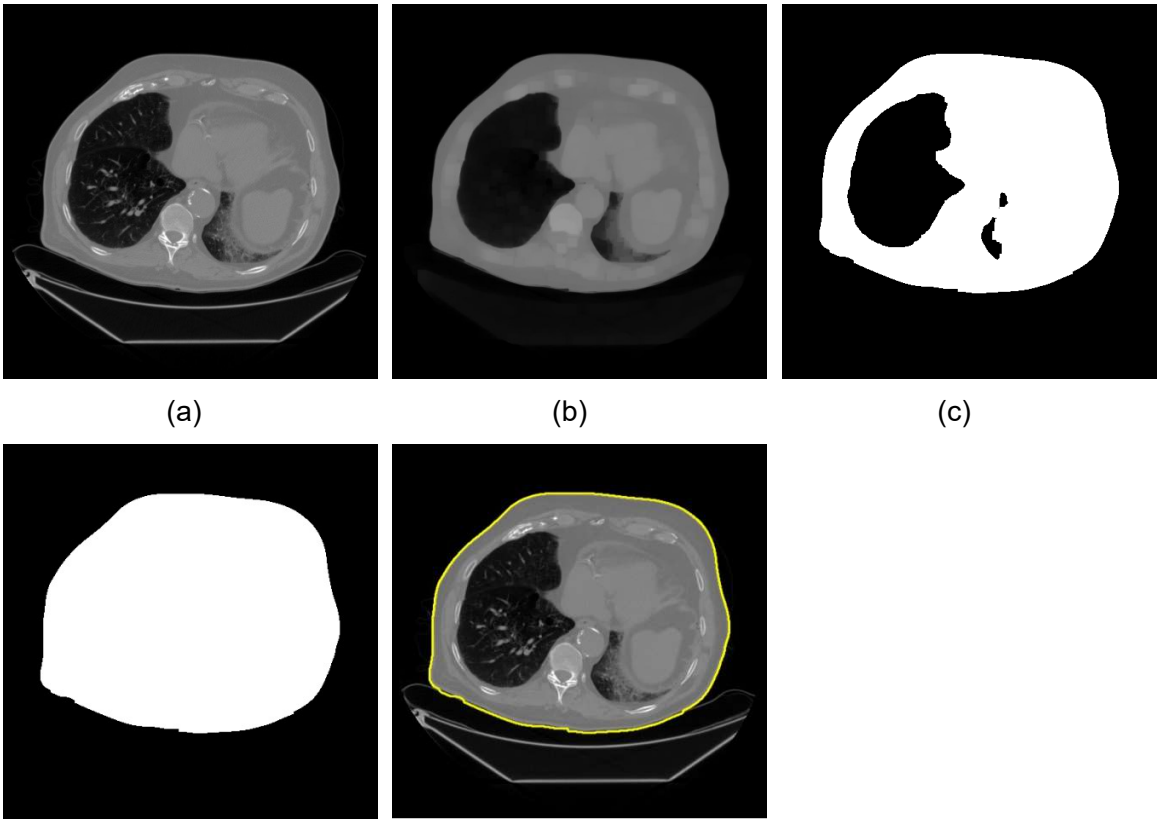


Figure 15: (a) original image slice of patient with pulmonary fibrosis; (b) image after morphological opening; (c) binarization – notice the couch has been already removed; (d) filling the empty space; (e) outline overlaid on original image.

9. Correlation between dose and patient characteristics

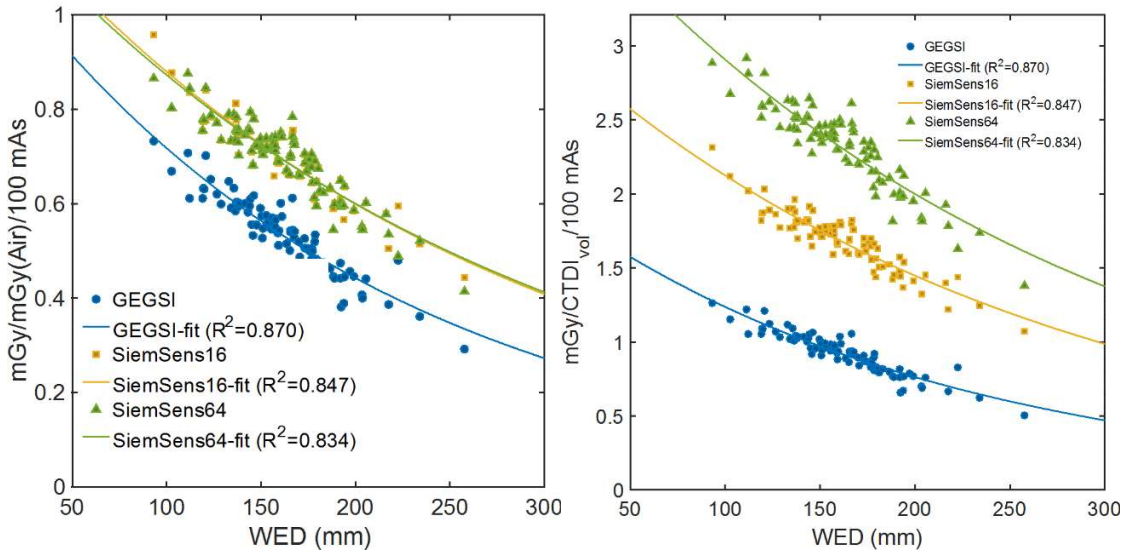
WED and organ dose correlation was determined through univariate regression analysis. The estimated regression function was of the exponential form

$$y = a \cdot e^{b \cdot x} \qquad (2)$$

were the independent variable, x, corresponds to the WED value in millimeter (mm) units, estimated per patient as described previously. The WED was measured at the central axial plane crossing the geometrical center of the heart. The variable y, corresponds to the dependent variable, organ dose in mGy/mGy/100mAs (or similar normalized unit).

Parameters a and b were estimated for each patient-specific WED at each organ dose computed by the Monte Carlo software. The coefficient of determination r^2, was used as a measure to assess the strength of correlation between WED and organ dose.

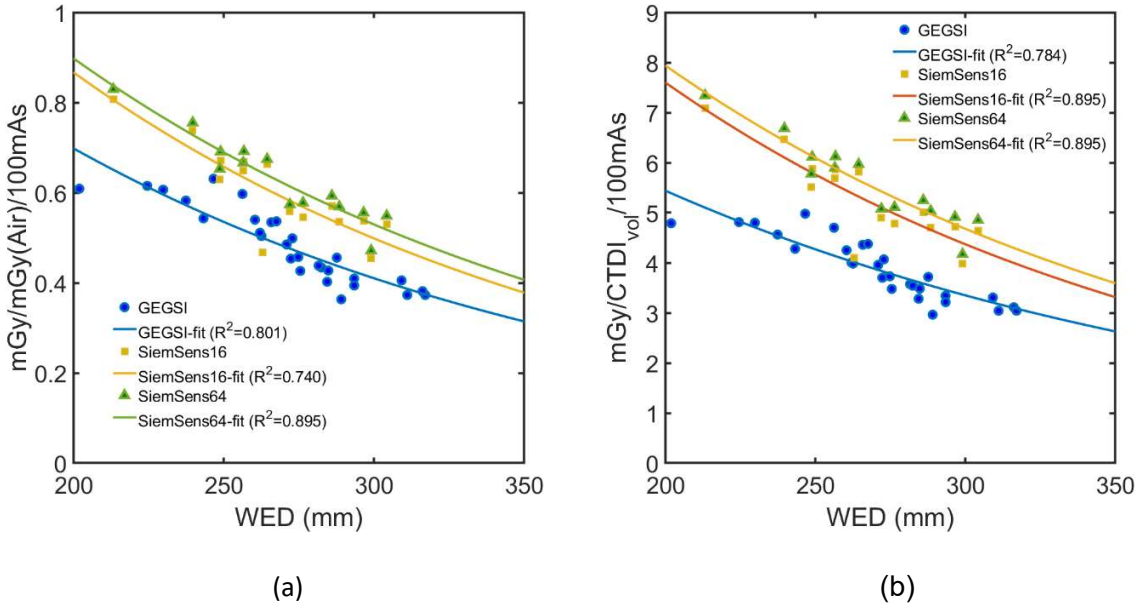
In following figures (Figure 16 to 20), examples of regression analysis for heart dose in pediatric cases are demonstrated. In this example, the heart dose was obtained using 80 kV beam spectra, fixed tube current, and the results were normalized to CTDIair (fig. 4a) and CTDIvol (fig. 4b). The blue line corresponds to regression associated with dose values computed using the GE Revolution GSI model. The green and yellow lines correspond to Siemens SOMATOM Sensation 16 and 64 respectively.



(a) (b)

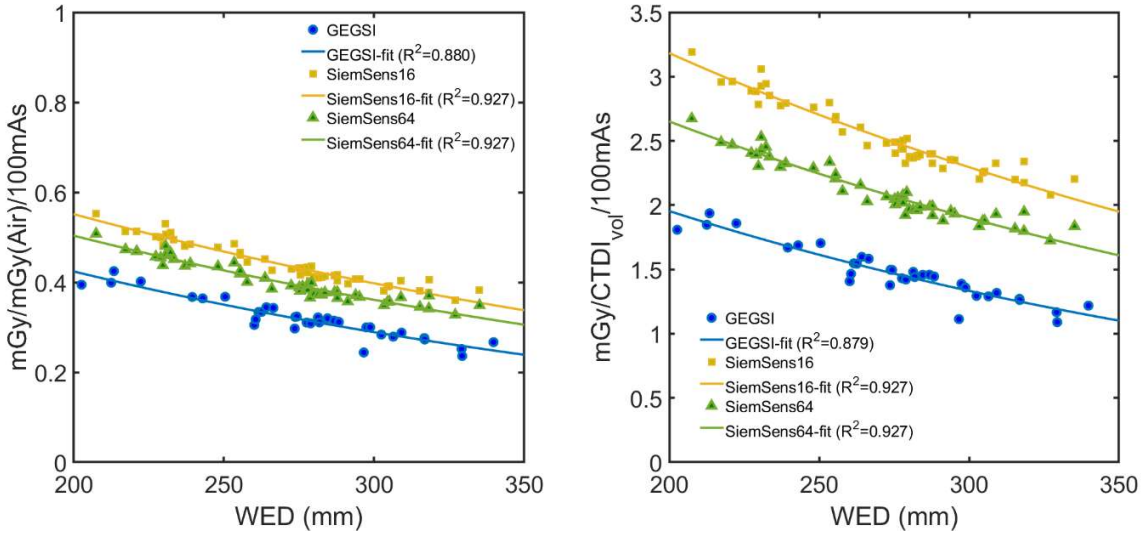
Figure 16: WED versus heart dose and associated regression lines for pediatric cases normalized to (a) CTDIair, and (b) CTDIvol.

Similar regression analysis was performed for adult cases. Figures 5 to 8 demonstrate examples of regression curves for selected organs/tissues for various adult cases at different kV settings. The doses were normalized to CTDIair and CTDIvol.



(a) (b)

Figure 17: WED versus heart dose at 120 kV and associated regression lines for adult cases with infectious disease as clinical indication normalized to (a) CTDIair, and (b) CTDIvol.



(a) (b)

Figure 18: WED versus skin dose at 100kV and associated regression lines for adult cases with pulmonary fibrosis as clinical indication normalized to (a) CTDIair, and (b) CTDIvol.

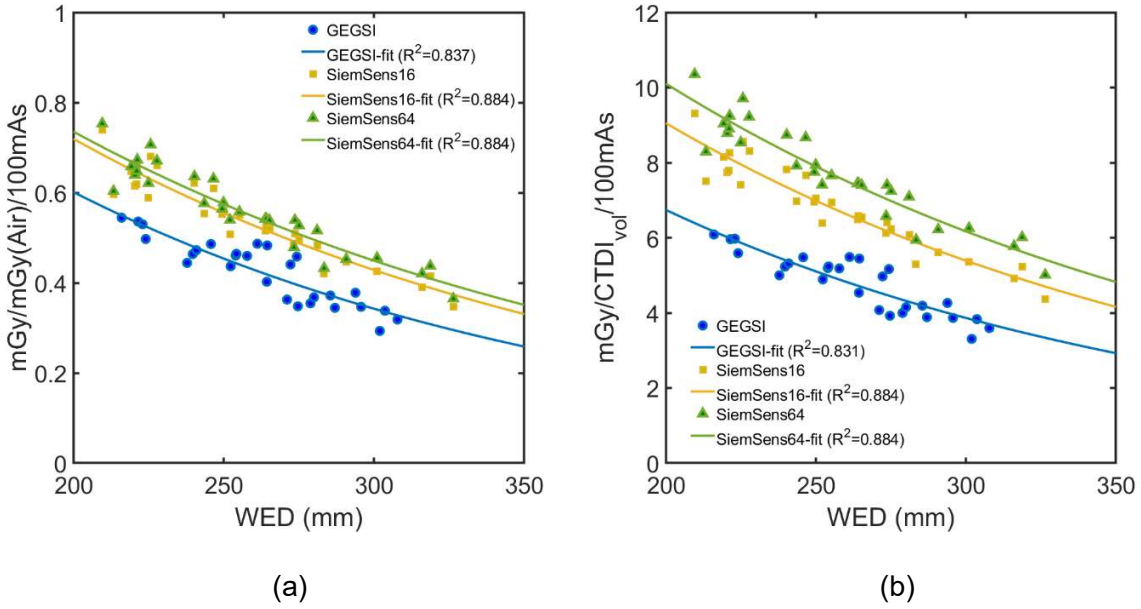


Figure 19: WED versus esophagus dose at 140kV and associated regression lines for adult cases with metastatic disease as clinical indication normalized to (a) CTDIair, and (b) CTDIvol

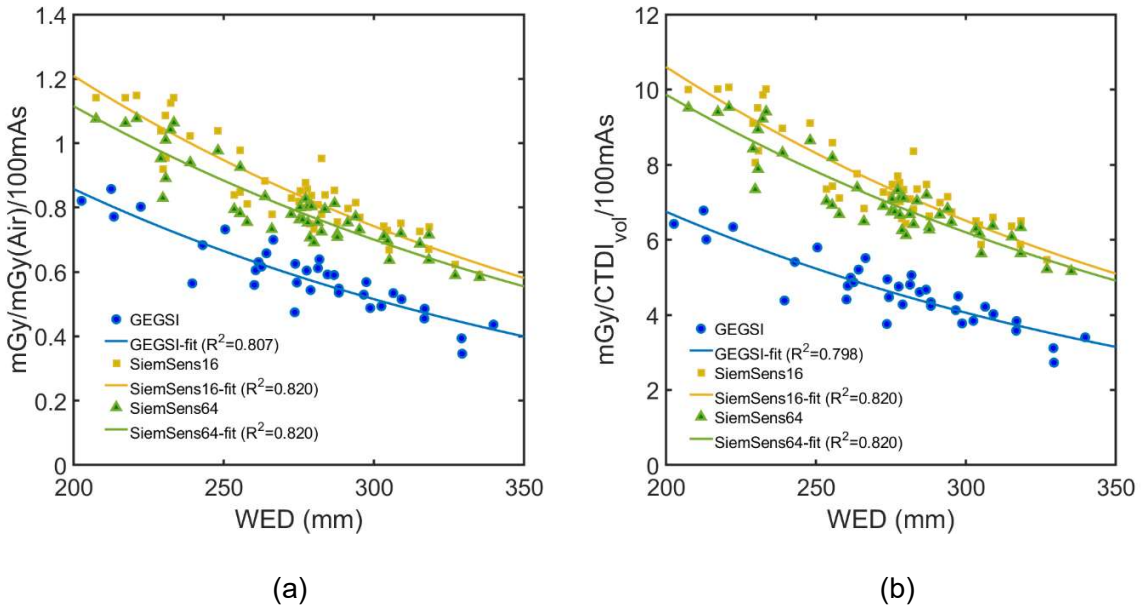


Figure 20: WED versus bone dose at 120kV and associated regression lines for adult cases with pulmonary fibrosis as clinical indication normalized to (a) CTDIair, and (b) CTDIvol.

10. Conclusions

A scanner- and patient-specific Monte Carlo (MC) method is proposed to accurately estimate absorbed doses to irradiated organs and tissues during thoracic computed tomography (CT) examinations. The proposed method for scanner-specific and patient-specific, organ-dose estimation is based on a procedure that combines Monte Carlo (MC) computational techniques and patient CT scans. The procedure is composed of five steps: (a) collection of suitable patient CT scans; (b) determination of scanner parameters; (c) Monte Carlo dosimetric computations; (d) three-dimensional (3D) patient-specific dose distribution output and (e) correlation between dose and patient characteristics.

Patient-specific voxelized phantoms were constructed using volumetric data acquired from pediatric and adult thorax CT scans. The voxelized phantoms were used as input in a GPU-enabled MC dosimetry software for CT exposure simulations. Geometric specifications, x-ray beam characteristics, and transient current modulation (where applicable) for each CT scanner were considered in the simulations. Organ structures were segmented with image analysis software using automated and freehand techniques and dosimetric calculations were performed. The derived organ doses were normalized and correlated with patient's WED -a size-specific metric- using non-linear regression analysis. Normalized organ doses demonstrated good correlation with WED (R^2=0.7-0.9, $P<0.001$). Non-linear regression models of size-specific normalized doses can be used to estimate organ dose in patients subjected to modern thoracic MDCT protocols.

The proposed method has several advantages. First, patient models were used instead of phantoms and, therefore, results are based on true patient-specific dosimetry. Second, strong correlations between normalized doses and patient WED ensure highly accurate organ dose estimations. Third, the method is applicable to both adult and pediatric patients undergoing chest CT examinations. Finally, data produced allows the development of a web-based tool (CTRAD) capable of estimating organ radiation doses quickly and accurately.

11. References

1. Brenner DJ, Hall EJ. Computed tomography: an increasing source of radiation exposure. N Engl J Med 2007; 357:2277–2284.

2. Mettler FA, Bhargavan M, Faulkner K, et al. Radiologic and nuclear medicine studies in the United States and worldwide: frequency, radiation dose, and comparison with other radiation sources-1950-2007. Radiology 2009; 253:520–531.

3. International Commission on Radiological Protection, "The 2007 recommendations of the International Commission on Radiological Protection," ICRP Publication 103, 2007.

4. Committee to Assess Health Risks from Exposure to Low levels of Ionizing Radiation, Nuclear and Radiation Studies Board, Division on Earth and Life Studies, National Research Council of the National Academies, Health Risks from Exposure to Low Levels of Ionizing Radiation: BEIR VII Phase 2 (The National Academy, Washington, DC, 2006).

5. MEDIRAD Data Management Board questionnaire concerning data management and data protection aspects of the clinical studies in MEDIRAD on detailed information on the procedures that will be implemented for data collection, storage, protection, retention and destruction and confirmation that they comply with national and EU legislation.

6. Aryanto, K.Y.E., Oudkerk, M. & van Ooijen, P.M.A. Free DICOM de-identification tools in clinical research: functioning and safety of patient privacy. Eur Radiol 25, 3685–3695 (2015).

7. Schmidt, B & Kalender, WA (2002), "A fast voxel-based Monte Carlo method for scanner- and patient-specific dose calculations in computed tomography", Phys. Med. 18(2): 43-53 .

8. Deak, P, et al (2008), "Validation of a Monte Carlo tool for patient-specific dose simulations in multi-slice computed tomography", Eur. Radiol. 18:759-772.

9. Myronakis, M, et al (2009), "Evaluation of a patient-specific Monte Carlo software for CT dosimetry", Rad. Prot. Dosim.133(4): 248-255.

10. Chen, W, et al, (2012), "Fast on-site Monte Carlo tool for dose calculations in CT applications", Med. Phys. 39:2985-2996.

11. Bert, Julien & Perez-Ponce, Hector & El Bitar, Ziad & Jan, Sébastien & Boursier, Y. & Vintache, Damien & Bonissent, Alain & Morel, Christian & Brasse, David & Visvikis, Dimitris. (2013), GEANT4-based Monte Carlo simulations on GPU for medical applications. Physics in medicine and biology. 58. 5593-5611.

12. Tucker DM, Barnes GT, Chakraborty DP (1991), Semiempirical model for generating tungsten target x-ray spectra. Med Phys; 18 (2): 211–218.

13. Poludniowski G et al (2009), Calculation of x-ray spectra emerging from an x-ray tube. Part II. X-ray production and filtration in x-ray targetsPhys Med Biol. Oct 7;54(19):N433-8.

14. Ridler, TW & Calvard, S (1978), "Picture thresholding using an iterative selection method", IEEE Transactions on Systems, Man and Cybernetics 8: 630-632.
15. American Association of Physicists in Medicine, “Use of water equivalent diameter for calculating patient size and size-specific dose estimates (SSDE) in CT,” AAPM Report No. 220, 2014.
16. Nobuyuki Otsu (1979). "A threshold selection method from gray-level histograms". IEEE Trans. Sys. Man. Cyber. 9 (1): 62–66.
17. Yousefi, Jamileh. (2015). Image Binarization using Otsu Thresholding Algorithm. 10.13140/RG.2.1.4758.9284.

12. Appendix

List of DICOM elements anonymized (de-identified) before patient CT data were stored outside hospital infrastructure (PACS) or uploaded to IRDBB database.

Tag ID	Tag Name
0008,0020	StudyDate
0008,0021	SeriesDate
0008,0022	AcquisitionDate
0008,0023	ContentDate
0008,0024	OverlayDate
0008,0025	CurveDate
0008,002A	AcquisitionDatetime
0008,0030	StudyTime
0008,0031	SeriesTime
0008,0032	AcquisitionTime
0008,0033	ContentTime
0008,0034	OverlayTime
0008,0035	CurveTime
0008,0050	AccessionNumber

0008,0080	InstitutionName
0008,0081	InstitutionAddress
0008,0090	ReferringPhysiciansName
0008,0092	ReferringPhysiciansAddress
0008,0094	ReferringPhysiciansTelephoneNumber
0008,0096	ReferringPhysicianIDSequence
0008,1040	InstitutionalDepartmentName
0008,1048	PhysicianOfRecord
0008,1049	PhysicianOfRecordIDSequence
0008,1050	PerformingPhysiciansName
0008,1052	PerformingPhysicianIDSequence
0008,1060	NameOfPhysicianReadingStudy
0008,1062	PhysicianReadingStudyIDSequence
0008,1070	OperatorsName
0010,0010	PatientsName
0010,0020	PatientID
0010,0021	IssuerOfPatientID
0010,0030	PatientsBirthDate
0010,0032	PatientsBirthTime
0010,0040	PatientsSex
0010,1000	OtherPatientIDs
0010,1001	OtherPatientNames
0010,1005	PatientsBirthName
0010,1010	PatientsAge
0010,1040	PatientsAddress
0010,1060	PatientsMothersBirthName
0010,2150	CountryOfResidence
0010,2152	RegionOfResidence

0010,2154	PatientsTelephoneNumbers
0020,0010	StudyID
0038,0300	CurrentPatientLocation
0038,0400	PatientsInstitutionResidence
0040,A120	DateTime
0040,A121	Date
0040,A122	Time
0040,A123	PersonName