Physica Medica 53 (2018) 72–79

Contents lists available at ScienceDirect

Physica Medica

journal homepage: www.elsevier.com/locate/ejmp

Original paper

Two-step validation of a Monte Carlo dosimetry framework for general radiology

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ARTICLE INFO

Keywords:
Monte Carlo
Dosimetry
General radiology

ABSTRACT

The Monte Carlo technique is considered gold standard when it comes to patient-specific dosimetry. Any newly developed Monte Carlo simulation framework, however, has to be carefully calibrated and validated prior to its use. For many researchers this is a tedious work. We propose a two-step validation procedure for our newly built Monte Carlo framework and provide all input data to make it feasible for future related application by the wider community. The validation was at first performed by benchmarking against simulation data available in literature. The American Association of Physicists in Medicine (AAPM) report of task group 195 (case 2) was considered most appropriate for our application. Secondly, the framework was calibrated and validated against experimental measurements for trunk X-ray imaging protocols using a water phantom. The dose results obtained from all simulations and measurements were compared. Our Monte Carlo framework proved to agree with literature data, by showing a maximal difference below 4% to the AAPM report. The mean difference with the water phantom measurements was around 7%. The statistical uncertainty for clinical applications of the dosimetry model is expected to be within 10%. This makes it reliable for clinical dose calculations in general radiology. Input data and the described procedure allow for the validation of other Monte Carlo frameworks.

1. Introduction

A survey from 2014 involving 36 countries from the European Union showed that projection imaging accounted for 87% of all medical diagnostic examinations [1]. The survey showed that the contribution to the total effective dose for diagnostic projection imaging of the population was 22% [1]. This highlights the need to track patient dose also in projection imaging, even if there is the tendency to mainly focus on computed tomography (CT) imaging.

A commonly used dosimetric index to estimate the radiation induced risk among the different modalities is the effective dose [2]. A frequently used method to obtain the effective dose from a projection exam is applying conversion factors to the dose-area product (*DAP*), entrance air kerma at a reference point or (a combination of) other parameters. These conversion factors are obtained for reference patients.

Our ultimate project is situated in the demand for patient specific dosimetry. The world health organization (WHO) estimated that in 2008 over 50% of both men and women in Europe will be overweight [3], which means that for dosimetric applications over 50% of the European population will not be appropriately represented by the actual reference body shape. Better dose estimates require new conversion factors based on anatomical models of different weight or body mass index (BMI) groups. Some commercial software packages are commonly used for dose assessment in projection imaging, but these approaches don't represent a realistic anatomical distribution of adipose tissue, soft tissues and bones for the obese or thin patients.

Monte Carlo techniques are often used in dosimetry because of the relative ease and flexibility to calculate or estimate quantities that are difficult to measure, such as absorbed organ doses [4]. Calculations are possible for different modalities, such as general radiology [5,6], multi slice or cone beam computed tomography (CT) [6–9] and image-guided radiotherapy [10]. Several Monte Carlo codes are available, such as EGSnrc [11,12], GEANT4 [13,14], MCNPX [15] or PENELOPE [16,17].

Any Monte Carlo framework however, has to be carefully calibrated and validated prior to its use. For many researchers the calibration and

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https://doi.org/10.1016/j.ejmp.2018.08.005
Received 6 February 2018; Received in revised form 27 July 2018; Accepted 6 August 2018

validation is a tedious work, especially if there is no easy access to a hospital setting. We propose a two-step validation and provide all input data to make it feasible for future applications by the wider community. First our framework was benchmarked through comparison with established codes and secondly we validated against measurements in clinical settings.

2. Materials and methods

An in-house developed Monte Carlo framework, which was originally built for cone beam CT [18,19], was adjusted for dosimetric purposes in projection imaging. The framework was based on EGSNRC (version 4-2.4.0) [12]. Both the Compton and the Rayleigh scattering were included in the photon-electron transport. The XCOM photon cross sections from the National Institute of Standards and Technology (NIST) were used [20]. The photon cut-off (PCUT) was set to 0.01 MeV. The Kerma Approximation was made on the electrons, i.e. secondary electrons were terminated on their place of birth and all their energy was deposited locally, given the short range of electrons in biological tissues at diagnostic energies [2].

Further input parameters for the Monte Carlo dosimetry framework were the spectrum (implemented as the number of photons per energy bin), the examination geometry and a voxel phantom. During a simulation, the deposited energy from the simulated particles to a predefined region of interest (ROI) is registered and the absorbed dose in the ROI (single voxel or voxel group) can be calculated. The Monte Carlo simulation uncertainty is estimated on a history-by-history basis [21] and quantified as the coefficient of variation (CV), which depends primarily on the number of simulated histories.

We followed a two-step procedure for calibration and validation of the Monte Carlo framework. First it was benchmarked against established data in the literature, by performing the calculations of case 2 of the American Association of Physicists in Medicine (AAPM) Task Group Report 195 [4]. Next, the framework was calibrated and validated for clinical protocols of a projection imaging system present at the University Hospitals Leuven.

2.1. Validation against established data

Case 2 for projection imaging of the AAPM Task Group Report 195 describes a soft tissue phantom with dimensions of $390 \times 390 \times 200\,mm^3$ (height × width × depth) placed at a source to surface distance of 155 cm. The detector dimensions are $39 \times 39\,cm^2$ at a distance of 180 cm from the source at the center of the field of view (0 degree case). The deposited energy per photon was scored in ten regions of interest in the phantom (Fig. 1). The applied spectra were a W/Al 120 kVp spectrum and a mono-energetic beam of 56.4 keV, which is the mean energy of the 120 kVp spectrum [4,22]. We repeated the simulation in our framework with exactly the same settings.

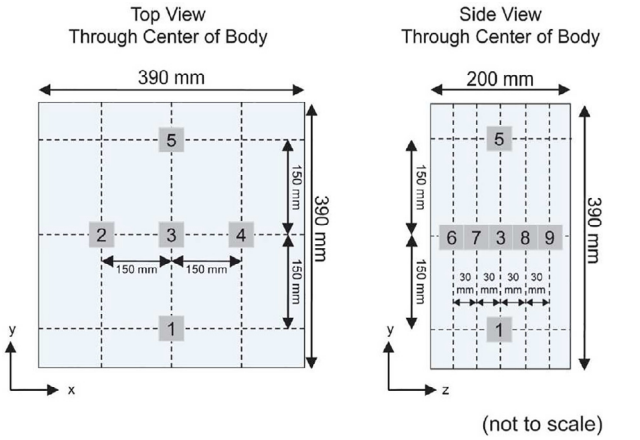

Fig. 1. Schematic of the soft tissue phantom and the scoring regions of interest (ROIs) described by [4]. The tenth ROI is the complete phantom.

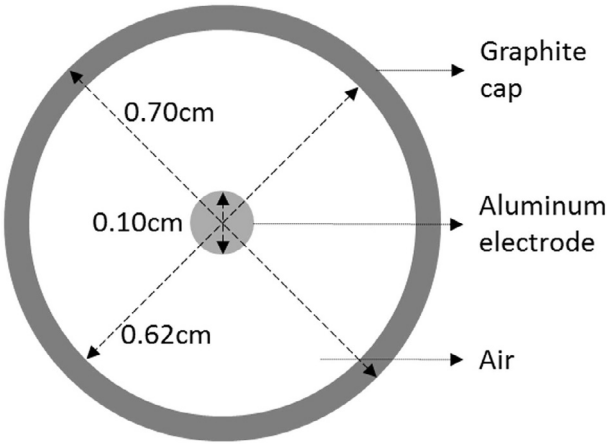

Fig. 2. Transversal view of the simulation model of the ionization chamber. The total length of the active part of the ionization chamber is 2.31 cm.

2.2. Calibration and validation in clinical setting

For the clinical evaluation, a Carestream DRX Evolution device with tube potential ranging from 40 to 150 kV with 1 kV steps was simulated. The detector has a maximum field size of $35 \times 42\,cm^2$ with a pixel resolution of $139 \times 139\,\mu m^2$.

The most common imaging protocols and their related examination settings were extracted with the dose monitoring software DOSE (Qaelum NV, Belgium). We focused on the trunk examinations (thorax, abdomen, pelvis, lumbar (L) spine and thoracic (T) spine).

2.2.1. Ionization chamber dosimetry

All experimental measurements were performed with a FC65-G farmer-type ionization chamber (SN:1698) coupled to an electrometer (IBA Dosimetry, Germany) (Fig. 2). The ionization chamber was calibrated by IBA's Secondary Standard Dosimetry Laboratory (IBA Dosimetry GmbH, Germany). The air kerma calibration was performed according to the IAEA TRS 277 Code of Practice resulting in an air kerma calibration coefficient ($N_{K,a}$) of 4.594×10^7 Gy/C. For the absorbed dose to water calibration, the procedure was performed according to the IAEA TRS 398 Code of Practice, which resulted in an absorbed dose to water calibration factor ($N_{D,w}$) of 4.724×10^7 Gy/C.

2.2.2. Input spectrum

No bowtie filter is involved in the plain X-ray imaging system. The spectrum is defined by its tube potential and the first half-value layer (HVL). The half-value layer can be determined experimentally with measurements. The ionization chamber was placed free in air at 60 cm from the tube in a collimated beam at the center of the field of view (FOV). The measurement was carried out with a fixed current-time product and by a step-wise addition of thin slabs (2 mm) of aluminum in the beam path until the measured air kerma was less than half of the initial air kerma. An exponential curve with parameters a and b was fitted to the measurements of the air kerma (K) as a function of the amount of added aluminum (x):

$$K = a\,e^{bx} \tag{1}$$

The half-value layer (HVL_m) was then calculated by:

$$HVL_m = \frac{1}{b}\ln\left(\frac{K_0}{2a}\right) \tag{2}$$

Next, a spectrum with the same half-value layer as the measured