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Development and implementation of a Monte Carlo frame work for evaluation of patient specific out- of - field organ equivalent dose

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ABSTRACT

Background: The aim of this study was to develop and implement a Monte Carlo framework for evaluation of patient specific out-of-field organ equivalent dose (OED). ***Materials and Methods***: Dose calculations were performed using a Monte Carlo-based model of Oncor linac and tomographic phantoms. Monte Carlo simulations were performed using EGSnrc user codes. Dose measurements were performed using radiochromic films. Furthermore, the applicability of this framework was examined for a 3D conformal radiotherapy of breast. ***Results:*** Commissioning of the beam model was done by comparing the measured and calculated out-of-field dose values of several points in the physical and tomographic phantoms, respectively. The maximum percentage difference was 17%, which was smaller than 30% acceptance criteria for Monte Carlo modeling. The maximum statistical uncertainty in out -of-field dose calculation was 23%. Organ equivalent doses for out of field organs in 3D conformal radiotherapy of left breast varied from 2.4 cGy for right kidney to 134.6 cGy for the left humeral head. ***Conclusion:*** The framework developed in this research is a valuable tool for calculating peripheral dose and out-of field patient specific OEDs, the quantities needed for calculating risk of secondary cancer induction as a result of radiotherapy. This code can be used as a patient specific treatment plan optimization tool in order to select a treatment plan with the lowest risk of secondary cancer induction.

Keywords: *Out-of-field organ doses, secondary cancer risk, tomographic models, Monte Carlo simulations.*

INTRODUCTION

In radiotherapy, the dose absorbed outside a radiation treatment field, i.e. the peripheral dose (PD), is believed to increase the risk of stochastic effects and the severity of deterministic effects [1]. The sources of PD are scattered radiation inside the irradiated volume in patient's body and scattered and leakage radiation originated from accelerator head [2]. For each beam energy, PD at each point outside the field depends on the field size and the distance from the field edge [2, 3]. Several studies investigated out-of-field organ equivalent dose (OED). Moreover, the risk of secondary cancer incidence may also differ for different treatment methods, such as conventional, 3D conformal (3D-CRT) and intensity-modulated radiation therapy (IMRT) [2, 4-8]. Some studies have reported higher PD values for IMRT plans, due to higher number of fields and longer treatment times which result in higher leakage radiation levels [2, 9], while others

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have shown comparable PD values for IMRT and 3D-CRT plans using multi leaf collimators (MLC) [6, 7]. Monte Carlo (MC) studies have also indicated that linac makes and models beam energy and collimator rotation, moreover, the utilized patient models may affect the results of calculating out of field OED [2]. Therefore, an ideal framework for selecting an optimized treatment plan with minimum risk of secondary cancer induction must include out-of-field head scatter and leakage radiation for planned energy, field size and collimator/MLC specifications as well as out-of-field organ specifications of patients. The accuracy of treatment planning system (TPS) calculations, beyond the treatment border, is inadequate and out-of-field dose must be evaluated by either direct measurements or MC calculations [10]. The aim of this study was to describe the required technical details for developing a MC framework to estimate PD. The developed framework was implemented to calculate OED for out-of-field organs in a 3D-CRT of breast.

MATERIALS AND METHODS

To develop a Monte Carlo-based framework for calculating out-of-field organ dose the followings were employed: Monte Carlo model of a linac head developed using BEAMnrc user code of EGSnrc [11], patient's tomographic phantoms developed using patient's CT images and CTCREATE[12] user code of EGSnrc, and patient's treatment plan data exported from TPS.

MC modeling of treatment radiation fields

The contribution of scattered and leakage radiation to PD was calculated using a previously developed and validated MC model of a Siemens Oncor linac [3]. This model includes a 6 MV photon beam-line, x-ray jaws and multileaf collimator (MLC) and the shielding components of the linac head. The phase space data is scored at the level of Y jaws, before beam enters the collimator system, using BEAMnrc's source module no. 19 [11]. This data was then used repeatedly as the source term to simulate planned treatment radiation fields, BEAMnrc's source module no. 21, using patient's treatment field data exported from TPS. While using phase space file as a source, the specific number of each component module in linac model or the LATCH value was passed on and the number of LATCH bits was identical for both simulations [11]. Simulation of treatment field was done using the JAWS and MLC component modules. The position of each leaf in the MLC bank (X_{MLC})was calculated using the following minification factor: $X_{MLC} = X_{isocenter} \times \frac{SCD}{SAD}$, where $X_{isocenter}$ is the leaf position projected at the isocenter, SCD is source collimator distance and SAD is source axis distance.

Construction of computed tomographic phantom

For each patient, a tomographic phantom was constructed using CTCREATE user code of EGSnrc by converting patient's CT images from DICOM format into voxelized geometry. The input parameters of CTCREATE were 2mm phantom voxel size, 4 material for the CT ramp and CT ramp parameters (for converting CT number into material densities (ρ)).

CT number-ρ calibration curve was constructed for our CT scanner (Siemens, SOMATOM Sensation) using an anthropomorphic torso phantom (QDV phantom, LinaTech) made of bone (ρ=1.4g/cm^3), lung (ρ=0.29g/cm^3), and soft tissue equivalent material (ρ=1.04g/cm^3).

Integrating linac models and computational phantoms

MC dose calculations in patient tomographic model were done using DOSXYZnrc user code of EGSnrc by incorporating the simulated patient treatment plan model. In order to integrate the two models, the coordinate system used in CT DICOM data was converted to the system used in DOSXYZnrc code using equations 1 to 3 [13]. Beam arrangements in DOSXYZnrc are described by θ, φ, and φ_{col}, relative to X, Y, and Z axes according to the gantry, couch and collimator angles [12]. The θ_G, θ_T, and θ_C are the angles of gantry, table and collimator, respectively.

$$\theta = \cos^{-1}(-\sin\theta_T \sin\theta_G) \quad (1)$$

$$\varphi = \tan^{-1}\left(\frac{-\cos\theta_G}{\cos\theta_T \sin\theta_C}\right) \quad (2)$$

$$\varphi_{col} = \frac{3\pi}{2} - \theta_c - \tan^{-1}\left(\frac{-\sin\theta_T \cos\theta_G}{\cos\theta_T}\right) \quad (3)$$

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Commissioning the computational framework

Commissioning of the beam model was done by comparing the measured and calculated dose distributions along beam central axis and in out-of-field regions [14]. The developed framework was also commissioned for clinical applications in a tomographic phantom designed using CT images of the phantom. The phantom head was irradiated by a 6MV 10×10 cm^2 photon field and dose calculations were performed in out-of-field lung and bone regions. To verify the accuracy of calculated doses, film dosimetry was performed using radiochromic films (GafChromic, EBT2, International Specialty Products, Inc., Wayne, New Jersey), calibrated according to Devic film calibration procedure [15]. A Microteck scanner with scan resolution of 127 dpi (0.2 mm / pixel), and an in-house MATLAB routine was used for image processing. The percentage difference values

$$\frac{|calculated\ dose - measured\ dose|}{measured\ dose} \times 100$$

Were calculated to compare with the acceptance criteria of 30% for model validation [18].

Calibration of the linac model

In order to calculate the total dose for a multiple field treatment plan, contribution of each field is considered by MC absolute dosimetry or virtual linac calibration which determines the relationship between number of incident particle fluence and MU [16]. A calibration run was performed to calculate the incident particle fluence per MU for a 10×10$_{cm^2}$ field size,

i.e. $\left(\frac{D^{cal}_{abs,(10\times10)}\left(\frac{cGy}{MU}\right)}{D^{cal}_{xyz,(10\times10)}\left(\frac{cGy}{incident\ particles}\right)}\right)$, using the physical linac output calibration conditions: SAD setup, 10$_{cm}$ depth and 10×10 cm^2field size.

$D^{cal}_{abs,(10\times10)}\left(\frac{cGy}{MU}\right)$ is the absolute dose calibration factor and $D^{cal}_{xyz,(10\times10)}\frac{cGy}{incident\ particles}$ is dose per particle fluence deposited in each voxel located at x,y and z in the calibration condition. The "cal" and "abs" labels, refer to values acquired in calibration conditions and absolute dose values respectively. Normalized dose values were reported in DOSXYZnrc output file (.3ddose).

For other field sizes, (x × y), relative output factor (ROF) was calculated in BEAMnrc user code: the monitor chamber was labeled as a dose scoring region and the dose delivered to the chamber was calculated in two separate simulations, first due to the beam entering from above $D^{forward}_{ch}$ and then due to the particles backscattered from the collimators (D^{back}_{ch})

The values with "ch" subscript, were scored in monitor chamber of model. For each field size, ROF was calculated using the equation (4):

$$ROF_{(x\times y)} = \frac{D_{xyz(x\times y)}}{D_{xyz(10\times10)}} \frac{D^{forward}_{ch(10\times10)} + D^{back}_{ch(10\times10)}}{D^{forward}_{ch(x\times y)} + D^{back}_{ch(x\times y)}} \quad (4)$$

Finally, for each field size, the absolute dose in cGy in each voxel located at x, y and z was calculated using the equation (5):

$$D_{xyz,abs,(x\times y)}(cGy) = ROF \times D^{cal}_{abs,(10\times10)}\left(\frac{cGy}{MU}\right) \times MU_{(x\times y)} \quad (5)$$

Implementation of Monte Carlo framework

Patient treatment plan specifications for the medial and lateral tangential breast fields including gantry angles (124 and 299 degrees), Y jaws and MLCs positions, monitor unit (MU) (116 and 90), and normalization depth (7.8 cm) were exported from TPS to the developed MC code. Then the same plan was simulated using the tomographic model of patient. Figure 1 shows the simulated MLC configuration for the medial and lateral tangential breast fields using the exported plan data from TPS.

The calculated dose matrix was used to calculate the OEDs using the equation (6):

$$OED_{org} = \frac{1}{V}\sum_{i=1}^{N} D_i V_i \quad (6)$$

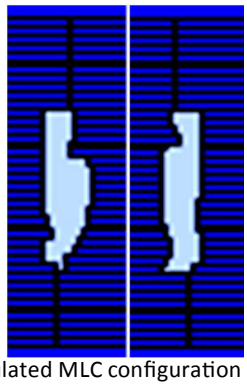

Figure 1. The simulated MLC configuration for the medial and lateral tangential breast fields using the exported plan data from TPS.

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D_i is the calculated absorbed dose in each volume element V_i of the organ taken from organ's dose volume data (DVH), and the sum is taken over N dose calculation points. DVH data for each organ was calculated by importing the MC calculated dose matrix to a research TPS (computational environment for radiotherapy research (CERR)) [17].

RESULTS

The first step in constructing a tomographic phantom was to define a conversion ramp. Table 1 shows the conversion ramp used to convert patient's CT images from DICOM into voxelized format.

The result of model validation, shown in figure 2, compares calculated and measured relative PD values as a function of distance from the edge of the field. For both sets of data, PD values were obtained using the same treatment plan specifications and were normalized to the dose at the depth of maximum dose (d_{max}) along central axis of the beam. The comparisons were made between calculated and measured out-of-field dose values in several points in tomographic and physical phantoms, respectively. The maximum percentage difference calculated for out-of-field was17%.

TPS derived coordinates for medial and lateral breast field were as follows: collimator angle (θ_C)= 0, table angle (θ_T) =0 for both fields and gantry angles (θ_G) were 124 and 299 degrees, respectively. This data which were in DICOM format was converted to DOSXYXnrc coordinates using equations 1-3 and the following results were derived for θ, $\varphi_{col}$and φ: 90, 270 and 34 degrees for the medial and 90, 270 and 209 degrees for the lateral fields.

In order to calculate the incident particle fluence per MU for a 10×10 cm² field size, i.e. $\left(\frac{D^{cal}_{abs,(10\times10)}\left(\frac{cGy}{MU}\right)}{D^{cal}_{xyz,(10\times10)}\left(\frac{cGy}{incident\ particles}\right)}\right)$, a calibration run was performed. Scored dose values in the monitor chamber of the model ($D^{forward}_{ch}+D^{back}_{ch}$), were obtained from BEAMnrc output file (.egslst) and dose per particle fluence matrix was acquired from DOSXYZnrc output file (.3ddose). Results of the calibration run for a 10×10 cm² field were: D^{cal}_{xyz} =8.55 ×10^{-16} $\left(\frac{cGy}{incident\ particles}\right)$, $D^{forward}_{ch(10\times10)}$ $D^{back}_{ch(10\times10)}$ = 2.672×10^{-18} $\left(\frac{cGy}{incident\ particles}\right)$,and $D^{cal}_{xyz,(10\times10)}$ =1.4145×10^{-16} $\left(\frac{cGy}{incident\ particles}\right)$ and $D^{cal}_{abs,(10\times10)}$ =0.789 $\left(\frac{cGy}{MU}\right)$. A calibration run was also performed for the breast treatment fields using beam configuration data exported from TPS and the following results were derived: $D^{forward}_{ch(22\times11)}$ = 8.55 × 10^{-16} $\left(\frac{cGy}{incident\ particles}\right)$, $D^{back}_{ch\ (22\times11)}$ = 2.608 × 10^{-18} $\left(\frac{cGy}{incident\ particles}\right)$ and $D^{cal}_{xyz,(22\times11)}$ =1.4145×10-16$\left(\frac{cGy}{incident\ particles}\right)$The maximum

Table 1. The conversion ramp used to convert patient's CT images into voxelized format.

Material name (ICRU700/ pegs4)	Material CT number upper bound	Material density upper/lower bound (g/cm3)
Air	-750	0.001-0.044
Lung	30	0.05-0.27
Tissue	1500	0.27-1.101
Bone	>1500	1.101-1.68

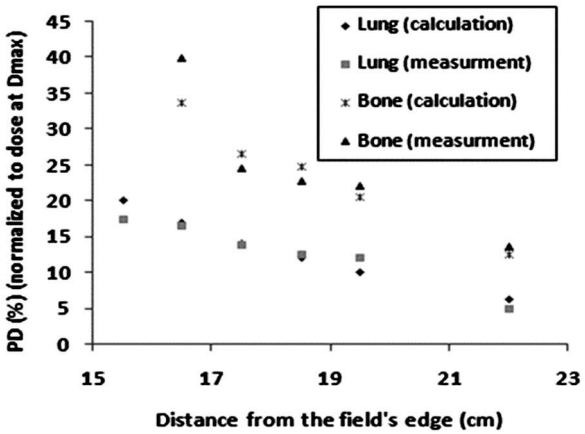

Figure 2. The measured and calculated percent peripheral dose (PD) values in lung and bone regions as a function of distance from the edge of the field.

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statistical uncertainty in dose calculations reported by DOSXYZnrc was 23%.

Finally, for each breast planned field (medial/ lateral), the absolute dose matrix (D_{xyz} (cGy)) for the patient tomographic phantom, was calculated by substituting the above values in equations 4 and 5.$D_{xyz(20\times11)}$ matrix was obtained from DOSXYZnrc output file and was acquired $MU_{(20\times11)}$ from TPS planned data.

$$D_{xyz,abs,(20\times11)med/lat}(cGy) = D_{xyz(20\times11)med/lat} \times \frac{8.55\times10^{-16}+2.672\times10^{-18}}{8.55\times10^{-16}+2.608\times10^{-18}} \times \frac{0.789\left(\frac{cGy}{MU}\right)}{1.4145\times10^{-16}} \times MU_{(20\times11)med/lat}$$

The results of calculated OED for different out of field organs in 3D-CRT of breast are illustrated in figure 3, using DVH data calculated by importing the MC calculated dose matrix to CERR.

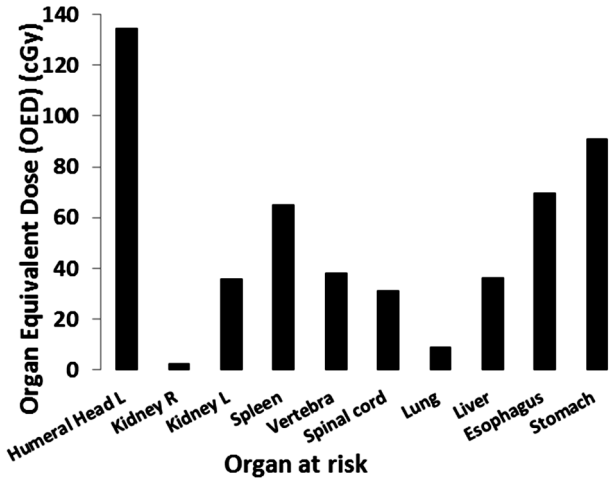

Figure 3. The results of calculated OED for different out of field organs in 3D-CRT of left breast (50 Gy prescribed dose).

DISCUSSION

The goal of this study was to describe the required technical details in developing a MC frame work for estimating the out of field dose in radiotherapy. MC dosimetry method is considered as the most accurate method for quantifying the absorbed dose of out of field organs [2]. In order for a MC framework to accurately model all variables that influence out of field organ specific doses, treatment unit and patient specific parameters need to be addressed [2]. In this regard, most developed models have limited accuracy due to application of specific standard phantoms, specific treatment plans and treatment units [2, 4, 19]. Hence, the diversity in reported results is evident [2]. The framework developed in this research is superior to previous models due to inclusion of patient specific peripheral dose calculation features such as: 1- calibration of fluence and MU, 2- utilization of patient's tomographic phantom, 3- transfer of patient's planed beam configuration data to create the patient specific MC beam model, 4- conversion of TPS derived beam coordinates to MC beam model coordinates and 5- multiple conformal beam dose calculations.

A MC framework including a 6 MV beam model and a tomographic phantom was developed for patient's specific OED calculations. The model was implemented and validated for a 3DCRT breast treatment. The results of model validation showed an acceptable agreement between calculated and measured data. The maximum percentage difference for out-of-field calculation was 17%. The acceptance criterion for percentage difference is 30% for out-of-field region [18]. The uncertainty of the radiochromic dosimetry, error in MC calculation and differences in the composition of the phantom and corresponding model contributed to the overall uncertainty. The overall one-sigma dose measurement uncertainty in film dosimetry, using Devic protocol for a uniform field of above 0.4 Gy, is up to 2% [15]. The maximum statistical uncertainty in out-of-field dose calculation was 23%.

Our results are comparable to those of a similar study in which the mean difference between calculated and measured out-of-field doses was reported as 11.4%. Calculated doses for out of field organs were comparable to our results as well. The dose values for stomach, esophagus, liver, lung, spleen and kidney were reported as 109.54, 85.45, 43.15, 21.82, 71.24 and 24.36cGy, respectively [19]. Comparing these values to our results shows the maximum difference of 18.7 cGy for stomach. This difference can be the result of inherent error in low dose out-of-field dose calculations,

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difference in linac and phantom models and setup differences (source-to-surface distance setup verses SAD). It has been reported that SAD setup may result in lower out-of-field doses [19].

The framework developed in this research can be a valuable tool for calculating peripheral dose and out-of field patient specific OEDs, the quantities which are needed when calculating risk of secondary cancer induction as the result of radiotherapy [20]. In other words, this code can be used as a patient specific treatment plan optimization tool in order to select a treatment plan with the lowest risk of secondary cancer induction.

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