Dosimetric verification of PENSSART, a new Monte Carlo dose calculation system based on PENELOPE and dedicated to patient-specific treatment quality control in radiotherapy

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I. Introduction

Beam delivery in radiotherapy has become increasingly complex since the advance of 3D conformational radiotherapy (3DCRT), intensity modulated radiotherapy (IMRT) and stereotactic body radiotherapy (SBRT). With these high precision modalities, accurate dose calculations are essential for radiotherapy planning since the accuracy of the absorbed dose as prescribed determines the clinical outcome.

However, in presence of air cavities or inhomogeneities in the patient anatomy, commercial treatment planning systems (TPSs), in particular those based on the pencil beam algorithm, often fail to predict accurate dose distributions [10]. Systematic errors exceeding 10 % in the thoracic area have been reported.

To further spatially optimize dose distribution, dose calculation algorithms have thus to be improved. Monte Carlo (MC) simulation is the only method that explicitly transports photons and electrons within a material and is therefore likely to provide more accurate results at material interfaces and within inhomogeneities. However, MC methods remain time consuming and their integration into commercial TPS and their use in clinical settings[11, 4] are the result of several improvements, among which implementation of variance reduction techniques and assumptions in the physical models. As a consequence, MC-based calculations engines embedded within TPS could be less accurate than dose calculation engines based on MC general-purpose codes. This has recently motivated the development of several in-house MC based verification tools for routine quality assurance in radiotherapy [17, 14, 9, 7].

To obtain accurate dose calculations in MC simulations, tissues have to be correctly characterized. Indeed, it has been reported that to overcome significant dose artifacts [3, 16], it is of prime importance to distinguish between the various densities of lung [1] and also between the various compositions and densities of bone [15].

In this work, we present a new MC dose calculation system called PENSSART (PENELOPE Simulation for the Safety in Radiotherapy). Prior to dose calculation, we have developed an accurate characterization of biological tissues thanks to a conversion method similar to the one proposed by Vanderstraeten *et al* [15], based on a stoichiometric calibration method and making used of dosimetrically equivalent tissues subsets. A special attention has been given to bone and lung materials. The schemes obtained for a GE LightSpeed16 scanner have been introduced in PENSSART and validated with a slab phantom.

II. Materials and Methods

II.1 Description of the PENSSART system

The PENSSART system is divided into three modules. The dose calculation module is based on the 2006 release of the general-purpose MC code PENELOPE [13] and was designed to perform MC dose calculations within voxelized geometries. It requires as input data a description of the geometry as well as a description of the radiation source, provided by a geometry module and a radiation source module, respectively. The radiation source module is also based on the 2006 release of PENELOPE and allows the simulation of different kinds of radiation sources, going from simple sources such as monoenergetic beams to more complex sources like the one resulting from the complete modeling of a linear accelerator (linac) treatment head. To generate phase space files (PSF) resulting from the complete description of a linac, a new version of the main program PENMAIN was developed, in which several conventional variance reduction techniques (Russian roulette combined with angular splitting, selective bremsstrahlung splitting and rotational splitting) were implemented. Finally, the geometry module was implemented to assist the user in the description of complex geometries (phantoms, patients).

In this work, we will focus on the geometry module and especially on the conversion of DICOM images into MC data.

II.2 Description of the geometry module

The geometry module has been developed to prepare voxelized geometries in which dose computations will be performed. Two kinds of geometries are supported by the system: *i)* numerical phantoms created using a routine provided in the PENELOPE package which allows the conversion of volumes limited by quadric volumes into voxelized objects and *ii)* voxel-based patient models obtained by converting DICOM images into data usable by the MC dose calculation module. This enables to make direct comparisons between PENSSART and PENELOPE computations and will ease the validation process between the two codes.

A two step procedure has been developed to convert DICOM images into MC data. First, a stoichiometric calibration[8] was used to obtain CT number to density curves. Images of the electron density phantom CIRS Model 062 were used to perform the calibration of a GE LightSpeed16 scanner. In a second step, we used a CT conversion scheme [15] to extract the

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elemental composition data and divide the CT number scale into dosimetrically equivalent tissue subsets (or bins). PENELOPE dose calculations were performed with a 10×10 cm² 12 MV photon beam produced by a Saturne 43 linac, within homogeneous phantoms with dimensions $31 \times 31 \times 20$ cm^3. The dimensions of the scoring grid voxels were 1 cm in the X and Y directions and 0.2 cm in the Z direction. The statistical uncertainty (1σ) of the MC calculations was less than 0.5 % within each voxel of the central axis. For materials between lung and bone, the goal is to keep a deviation less than 1 % for each voxel of the central axis within two materials representing adjacent tissue subset. To eliminate the influence of mass density, it was set to 1 g.cm^{-3} for all these materials. For materials between air and lung, no method was proposed to sample the bins. For these materials, mass density is the only factor that influences dose computation as the chemical composition keeps the same between adjacent tissue subsets. Under this condition two lung bins have been first defined to allow the discrimination between air, inflated lung and deflated lung.

II.3 Validations

II.3.1 Dosimetric validation of PENSSART

A first validation benchmark has been conducted to ensure that the PENSSART system gives similar dose results as PENELOPE. PENELOPE was already comprehensively validated against dose measurements in previous studies for several homogeneous and heterogeneous phantoms [5, 6] and is therefore considered as the reference code in this study. A PSF file corresponding to a 10×10 cm² field size for a 12 MV photon beam of a Saturne 43 linac was created at a distance of 100 cm from the source, using a linac model previously validated [5]. Dose computations were then performed using this PSF file with PENSSART and PENELOPE in a homogeneous water phantom and a layered inhomogeneous phantom.

To assess the accuracy of the dose computation, we also performed the ICCR00 benchmark [12] for PENELOPE and PENSSART. A realistic clinical 1.5×1.5 cm^2 photon beam is used to irradiate a multi-slab phantom made of the following materials: water (from 0 to 3 cm depth), aluminum (from 3 to 5 cm depth), lung (from 5 to 12 cm depth) and water (from 12 to 30 cm depth). The source is uniform and placed at 100 cm upon the phantom. The 18 MV photon spectrum used is given in the benchmark. The PENELOPE energetic parameters used for the dose calculation are in agreement with the EGS4 parameters given in the benchmark [12]. The depth dose distribution is scored in $0.5 \times 0.5 \times 0.2$ cm^3 voxels.

II.3.2 Conversion of DICOM images into MC data

Conversion schemes proposed for the GE LightSpeed16 CT scanner were validated for a slab heterogeneous phantom described on Figure 1. Four conversion schemes were performed for 140 kV, 120 kV, 100 kV and 80 kV. The slab phantom was acquired at 120 kV. To assess the influence of the scanner potential, the four conversion schemes were tested. Dose computation was then performed on the 120 kV converted images. The slab phantom was also directly described as a numerical phantom in the geometry module, using the exact chemical composition for each material it is made of. The comparison between the 3D dose maps computed using the exact geometry of the slab phantom and using the geometry obtained after applying the conversion scheme was used as a validation test.

Material	*Density (g.cm^{-3})*	*Elemental composition (percentage in mass)*
Water	1.00	O (88.89) ; H(11.11)
Lung	0.30	C (60.08) ; O (23.04) ; H (8.33) Mg (4.8) ; N (2.73) ; Cl (1.02)
Bone	1.92	O (43.5) ; Ca (22.5) ; C (15.5) ; P (10.3) ; N (4.2) H (3.4) ; S (0.3) ; Mg (0.2) ; Na (0.1)

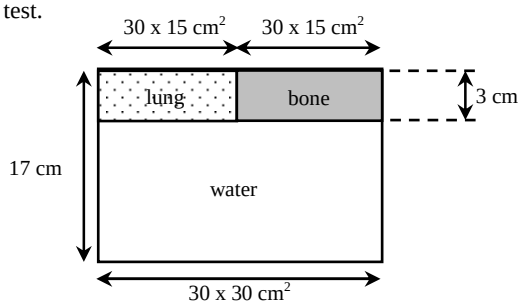

Figure 1: Slab phantom and materials used to validate the scheme of conversion of CT numbers into MC data.

III. Results

III.1 Conversion of DICOM images into MC data

The bin conversion obtained with the method proposed by Vanderstraeten *et al* [15] and with the PENELOPE MC code are reported in Table 1. Thirteen bins corresponding to thirteen biological materials were finally determined: among them, seven different materials are necessary to accurately describe bony structures for MC dose calculations and two different materials are necessary to differentiate lung exhale tissues from lung inhale tissues. This is in contradiction with conversion schemes already proposed by other groups who developed quality assurance tools for TPS verification. For example, the EGSnrc/DOSXYZnrc default CTCREATE subroutine used in some in-house MC calculations systems [9, 4] assigns only 4 types of tissues (air, lung, soft tissues and bone). It has been demonstrated that such a coarse segmentation can lead to serious errors in the dose computation, up to 10 % for 6 MV photon beams [2].

We have reported on Figure 2 the evolution of the Hounsfield number with the potential for the GE LightSpeed 16 CT. We note a large influence of the scanner potential used to make the calibration on the segmentation, especially for bony tissues. Indeed, as reported by Verhaegen and Devic[16], for bony tissues the Hounsfield numbers decrease rapidly with kVp

(25 % difference between 80 kVp and 140 kVp) whereas for lung and soft tissues they slightly increase with kVp (5 % difference between 80 kVp and 140 kVp).

We have then performed a dose computation on the 120 kV converted image. The same dose computation has also been performed for a numerical phantom describing the slab phantom with the exact materials. The matching between the two dose maps is excellent for the materials considered here (thus compact bone, lung and water). On Figure 3, we have reported the profiles obtained at 5 cm depth, thus after the interface between the two inserts (lung and bone) and water.

Table 1: Conversion of CT numbers into material composition.

	Bin	*Composition (masse percentage)*												*Density*
		H	*C*	*N*	*O*	*Na*	*Mg*	*P*	*S*	*Cl*	*K*	*Ca*	*Ar*	*(g.cm^{-3})*
1	*AIR*	-	0.16	78.44	21.08	-	-	-	-	-	-	-	0.33	0.001
2	*Lung tissues*	10.29	10.16	3.08	75.28	0.20	-	0.20	0.30	0.30	0.20	-	-	0.328
3		10.30	10.18	2.94	75.38	0.20	-	0.20	0.30	0.30	0.20	-	-	0.654
4		10.76	32.38	2.05	54.04	0.11	-	0.11	0.21	0.21	0.11	-	-	0.816
5	*Soft tissues*	11.07	47.23	1.46	39.78	0.05	-	0.05	0.15	0.15	0.05	-	-	0.978
6		9.98	16.29	4.28	68.38	0.36	-	0.05	0.36	0.25	0.05	-	-	1.073
7		8.78	21.17	3.89	58.52	0.28	0.04	2.15	0.32	0.18	0.03	4.62	-	1.167
8		7.77	25.32	3.56	50.13	0.21	0.08	3.94	0.29	0.13	0.02	8.56	-	1.261
9		6.90	28.89	3.27	42.91	0.15	0.11	5.48	0.26	0.08	0.01	11.95	-	1.355
10	*Bony tissues*	6.13	32.01	3.02	36.61	0.10	0.13	6.82	0.23	0.03	-	14.91	-	1.449
11		4.58	22.60	3.69	40.54	0.10	0.17	8.80	0.27	0.01	-	19.23	-	1.68
12		3.95	18.82	3.96	42.12	0.10	0.19	9.60	0.29	-	-	20.97	-	1.80
13		3.40	15.50	4.20	43.50	0.10	0.20	10.30	0.30	-	-	22.50	-	1.92

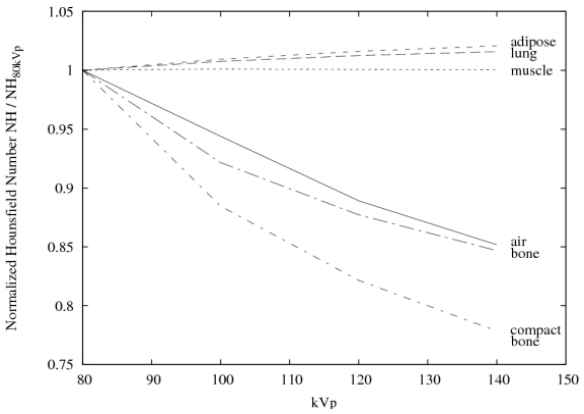

Figure 2: Evolution of Hounsfield numbers with the scanner potential for the GE LightSpeed16 CT.

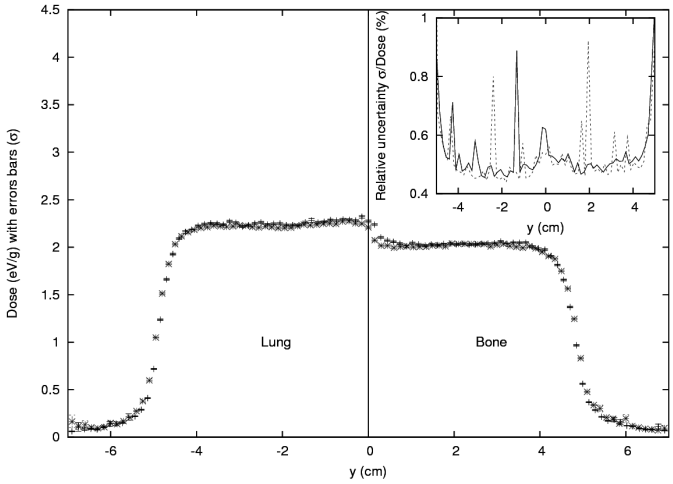

Figure 3: Profiles obtained at 5.0 cm depth. Solid line and "+" crosses correspond to the dicom image and dash line and "x" crosses correspond to the exact slab geometry.

III.2 Dosimetric validation of PENSSART

For the three benchmarks performed, the standard deviation between PENELOPE and PENSSART is less than 2 % except at the interface between lung and water and also for the depths where an uncertainty higher than 2 % is observed for one of the two codes. The build-up computed by PENSSART at the interface between lung and water is moved compared to the one computed by PENELOPE. This difference is due to the fact that dose computation is performed in a voxelized geometry in PENSSART whereas it is performed in a quadratic geometry in PENELOPE.

Relative depth dose distributions computed with PENELOPE, PENSSART and EGS4 for the ICCR00 have been also performed for 100 millions of particles in the conditions specified by the benchmark [12]. A good agreement between the three codes is observed for the depth dose distributions. Nevertheless, some discrepancies appear at the interfaces between heterogeneities: it was observed that PENELOPE and PENSSART both underestimate the lung dose compared to EGS4. At each interfaces, large deviations comprised between 5 and 10 % can be noted. Out of these interfaces, the deviation between the codes is comprised between ± 2 %.

IV. Conclusions

Development of the PENSSART system was successful for performing MC simulations in heterogeneous media. We have presented here the key features of the system. A CT conversion scheme has been developed whereby a material can be assigned to each voxel of a DICOM image. Materials have been chosen is such way that the doses obtained by MC calculations are not expected to differ by more than one percent from the doses obtained with the real tissues. Investigations are currently driven to precise this conversion scheme in the case of pulmonary SBRT treatments.

The next step will be to adapt the PENSSART system for quality control of IMRT treatment plans in radiotherapy. To this end, developments are already on-going: models of the Siemens ARTISTE linac and its 160 MLC^{TM} were already validated and clinical cases will be tested in the future months with the PENSSART system.

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