Evaluation of PENFAST – A fast Monte Carlo code for dose calculations in photon and electron radiotherapy treatment planning

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Abstract

The aim of the present study is to demonstrate the potential of accelerated dose calculations, using the fast Monte Carlo (MC) code referred to as *PENFAST*, rather than the conventional MC code *PENELOPE*, without losing accuracy in the computed dose. For this purpose, experimental measurements of dose distributions in homogeneous and inhomogeneous phantoms were compared with simulated results using both PENELOPE and PENFAST. The simulations and experiments were performed using a Saturne 43 linac operated at 12 MV (photons), and at 18 MeV (electrons). Pre-calculated phase space files (PSF) were used as input data to both the PENELOPE and PENFAST dose simulations. Since depth-dose and dose profile comparisons between simulations and measurements in water were found to be in good agreement (within ± 1% to 1 mm), the PSF calculation is considered to have been validated. In addition, measured dose distributions were compared to simulated results in a set of clinically relevant, inhomogeneous phantoms, consisting of lung and bone heterogeneities in a water tank. In general, the PENFAST results agree to within a 1% to 1 mm difference with those produced by PENELOPE, and to within a 2% to 2 mm difference with measured values. Our study thus provides a pre-clinical validation of the PENFAST code. It also demonstrates that PENFAST provides accurate results for both photon and electron beams, equivalent to those obtained with PENELOPE. CPU time comparisons between both MC codes show that PENFAST is generally about 9-21 times faster than PENELOPE.

Key words: Fast Monte Carlo dose calculation, experimental verification, inhomogeneous phantoms

1. Introduction

Conventional treatment planning systems (TPS) are fast, but sometimes insufficiently accurate, especially in the vicinity of interfaces between materials, where the effects of electron transport cannot be accurately handled by conventional deterministic dose algorithms. Monte Carlo (MC) methods are considered to provide the best calculation engine today available in medical radiation physics [1,2]. However, conventional MC methods still require long calculation times. During the last few decades, a growing number of MC-based dose simulation engines (e.g. Macro MC [3], Super MC [4], Voxel MC [5], PEREGRINE [6], Dose Planning Method [7], MCDOSE/MCSIM [8] and others) have been developed, thanks to the optimization of dedicated radiation transport algorithms, and the implementation of variance-reduction techniques and suitable approximations. PEREGRINE was the first commercially available electron *and* photon beam treatment planning system, whereas the commercial implementation of other codes is applicable to electron beams only (Voxel MC incorporated into the Masterplan TPS of Nucletron, and Macro MC implemented in the Eclipse TPS of Varian). Recently, a new MC code called PENFAST, for the fast simulation of photon and electron dose distributions in computerized tomography (CT) structures, was developed by Salvat *et al.* [9-11] in the framework of the European Integrated Project referred to as MAESTRO (LSHC-CT-2004-503564). As PENFAST is a proprietary code, implemented as a new functionality in the TPS ISOgray™ of the French company DOSIsoft [12], we are restricted in our description to one of the physical models and transport

algorithms included in PENFAST. In the following paragraphs, a synthetic comparison between the general-purpose MC code PENELOPE [13-15] and its faster version PENFAST is presented.

PENELOPE [13-15] performs an accurate MC simulation of coupled electron-photon transport phenomena, in arbitrary materials over a wide range of energies (from a few hundred eV to 1 GeV). Photon transport is simulated by means of the conventional detailed simulation scheme. Electron and positron histories are generated on the basis of a mixed procedure, which combines a detailed simulation of hard events with a condensed simulation of soft interactions. A geometry package called PENGEOM [14,15] enables the generation of random electron-photon showers in material systems consisting of homogeneous bodies, limited by quadric surfaces (*i.e.*, planes, spheres, cylinders, etc), and allows the modelling of geometries of various forms and compositions. PENELOPE (as for all conventional MC codes) requires long computation times, which are impractical for the routine planning of clinical treatments.

PENFAST [9-11,16] simulates photons, electrons and positrons transport in voxelised geometries starting from pre-calculated PSFs with the general-purpose PENELOPE code. This algorithm is based on the same physical interaction models as those used in the conventional MC code PENELOPE, but it implements new transport mechanics for electrons and positrons tailored to optimize the simulation speed. One of the typical features of PENFAST is the consistent use of complete-grouping condensed simulation for electron and positron transport. In this approach the collective effect of all interactions undergone by a transported electron along a given path length is simulated, in an approximate manner, with a single computational step. The use of class I simulation for absorbed dose estimation in radiotherapy has been hindered by the inability of multiple-scattering theories to accurately account for energy straggling effects. PENFAST avoids this problem by using multiple-

scattering distributions (i.e. energy loss and angular distributions) generated with PENELOPE. This procedure freeds this class I algorithm from most of the limitations of conventional multiple-scattering theories. PENFAST also uses a simplified photon transport model adapted to CT geometries and which has been tailored to take full advantage of the peculiarities of the CT geometry and the limited variability of atomic numbers found in radiotherapy dose estimations. Apart from several small simplifications, the photon interaction models adopted in PENFAST are equivalent to those used in PENELOPE. Geometrical aspects are handled by using the delta-scattering method [17,18], which does not require control of the interface crossings, and simplifies the tracking of photons through the CT structure, combined with the variance-reduction techniques of interaction forcing and splitting (as implemented in PENELOPE). In practical dose calculations the CT scan which describes the morphology of the patient is used as input data to the PENFAST code. Since CT image information is expressed in Hounsfield units (HU), the use of such data involves converting HU to the chemical composition and the mass density required by PENFAST for each voxel. The conversion to chemical composition is performed on HU ranges. All voxels with a HU lying within a certain range are assigned to the same material. The chemical composition is defined using a material file read by the MC code. For the conversion to mass density (in $g.cm^{-3}$) the calibration curve of the CT scanner, that relates HUs to mass density, is used. PENFAST has in its database chemical composition and multiple-scattering distributions for air, water, compact bone, lung, titanium and lead (the two latter materials are used for specific applications: electron insert and prosthesis). Additionally, it gives the possibility of generating new materials for the database according to need. Therefore, the MC simulation in every part of the patient is performed in the appropriate medium. Additionally, it is possible to influence the voxelised patient matrix by varying its resolution. This is achieved by grouping together voxels from the original CT image. The resolution of the dose

estimation grid is limited to being an integer multiple of the resolution of the original CT image. In order to calculate the mass density corresponding to the grouped voxel, the HUs of the CT image voxels are averaged and then the conversion to mass density is applied to the average HU value via the considered CT calibration curve. The material assignment is determined by converting each of the CT image voxel to a material, and then the most abundant material present is chosen to represent the grouped voxels.

A large number of studies have already demonstrated the ability of the conventional MC code PENELOPE to provide a reliable description of the transport of photons and electrons in matter, as well as in the vicinity of material interfaces [19-22]. Recently, Blazy *et al.* [22] compared experimental measurements with simulated dose distributions obtained with the PENELOPE code (2003 release) in inhomogeneous phantoms (lung and bone slabs). The study was carried out for an 18 MeV electron beam, and a 12 MV photon beam from a Saturne 43 accelerator. Dose distributions were measured with Fricke dosimeters and with plane and cylindrical ionization chambers (IC). They showed a good agreement between the PENELOPE simulations and the experimental measurements, and demonstrated the ability of the PENELOPE code to account for fluence disturbances in the vicinity of the heterogeneity interfaces.

The goal of this paper is to assess the influence of the approximations used, in the physical models and transport algorithms included in PENFAST, for dose calculations. Experimental measurements of dose distributions in homogeneous and inhomogeneous phantoms were thus compared to simulated results obtained with PENELOPE (latest release 2006 [15]) and PENFAST. A comparison of the calculation times, for both MC codes, is also presented. This study provides a pre-clinical validation of the PENFAST code, using experimental data measured by Blazy *et al.* [22] at the LNHB (the French National Metrological Laboratory for ionizing radiation).

2. Materials and methods

2.1. Phantom configurations

Phantoms consist of parallelepipedic slabs of CIRS lung equivalent material (0.3 g/cm^3) and CIRS bone equivalent material (1.8 g/cm^3), placed in a water tank. The water tank is made of 1.5 cm thick plexiglas (PMMA) walls, with the exception of that in front of the beam which is only 0.4 cm thick. The phantom configurations used are presented in Figure 1. These cover a large number of heterogeneous interfaces encountered in radiotherapy: water/lung/water (figure 1a and 1d), water/bone/water (figure 1b) and water/lung/bone/water (figure 1c).

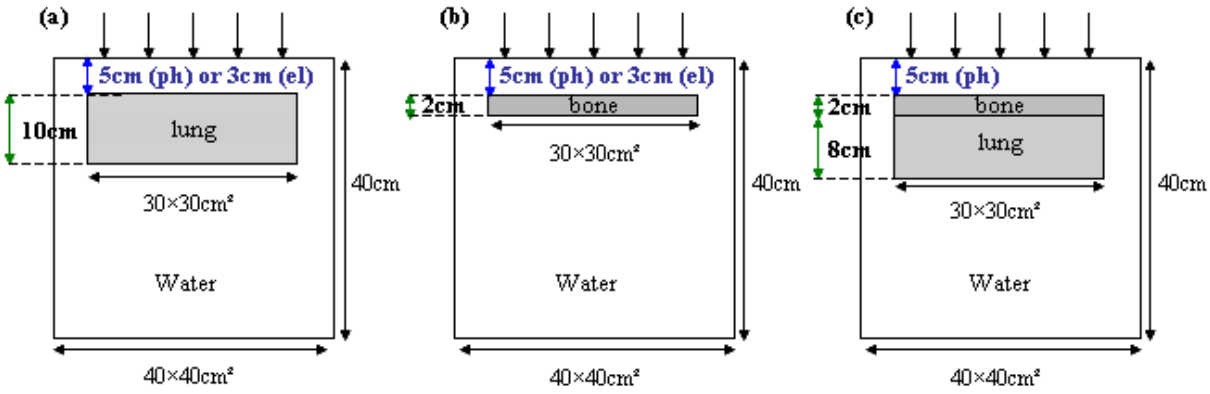

Figure 1. (a) and (b) show the inhomogeneous phantoms containing the lung and bone equivalent layers, respectively. The position of the heterogeneity layer depends on the beam quality; these configurations are common to the photon and electron beams. (c) shows the inhomogeneous phantom containing the bone and lung layers; this configuration is peculiar to the photon beam.

2.2. Dose measurements in inhomogeneous phantoms

Dose measurements have been carried out in inhomogeneous phantoms at the LNHB laboratory, using special precautions (*e.g.* accurate geometrical positioning, no drift, etc.) in order to achieve highly accurate results.

Depending on the experimental conditions, depth-dose distributions were measured upstream and downstream of the heterogeneity (figure 1a-c) using ICs (plane parallel: NACP-02, PTW-34001 and cylindrical: PTW-31002, NE-2571), and near to the interfaces using Fricke dosimeters. All measured central axis dose distributions were normalized relative to the dose at the reference depth (10 cm and 4.2 cm for photons and electrons respectively) measured in a homogeneous water phantom using IC and Frick dosimeters. The dosimeters are not included in the simulations. The measurements are therefore converted to an equivalent dose in the medium (water), for the purposes of comparison with the MC calculations, which directly provide the dose in the medium itself. The IC relative measurements in the electron beam are corrected by the water to air ratio of the stopping power calculated on the central axis. The Fricke dosimeter values are corrected for the wall perturbation calculated by PENELOPE. For further details concerning the determination of these correction factors and their values, see the paper of Blazy *et al.* [22].

Moreover, the dose profiles were measured using a cylindrical IC (PTW-31002) at depths 2.5 cm and 14.5 cm in the water-lung-water phantom (figure 1a) for the electron beam, and at depths 22 cm and 25 cm in the mediastinum configuration phantom (figure 1d) for the photon beam. All measured dose profiles were normalized relative to the dose on the central axis.

For IC relative measurements, the uncertainty due to the calibration factor of the dosimeter at a reference beam quality “disappears” when dealing with relative dose. The type A uncertainties (reproducibility) are negligible compared to type B uncertainties (experimental conditions) which are estimated to be 0.56% (1σ) for electron beams, and 0.36% (1σ) for photon beams. For the Frick dosimeters, the uncertainties are estimated to be 0.8%.

2.3. Monte Carlo simulations

The MC simulations were separated into two parts: the first involves the detailed simulation of particle transport through the accelerator head. An impact detector is located at the entry surface of the phantom; when crossing this scoring plane, particle characteristics (*e.g.* mainly type, position, energy, direction, weight) are stored in a PSF. These calculations were made using the 2006 release of the MC code PENELOPE [15]. Then, in the second part of the simulation, this PSF is used as input data to compute the three-dimensional dose distribution in phantoms, using both MC codes: PENELOPE and PENFAST.

2.3.1. PHASE SPACE FILE CALCULATIONS

Detailed modelling of the Saturne 43 linac head (used at the LNHB) was performed using the MC code PENELOPE. Both the photon (12 MV) and electron (18 MeV) beams are simulated according to the manufacturer's data. The chosen field size is 10 ×10 cm² at a distance of 100 cm from the source, in accordance with the IAEA protocol [23].

To determine the initial electron parameters (energy spectrum and beam spot size), several simulations were carried out with the PENELOPE code. The values of the parameters defining the source were then determined by least squares minimization of the difference between the measured and simulated dose distributions in water. This novel method for beam commissioning will be discussed in a further paper, since the main objective here is to validate the dose calculation with PENFAST, by starting from a pre-calculated PSF.

The PSF was calculated at 90 cm and 100 cm from the source, for the photon beam and the electron beam respectively. The number of simulated events was adjusted so as to achieve an average statistical uncertainty of 0.4%, in regions where the dose is greater than 50% of the maximum dose. This led to respectively 65 and 52 million particles scored in the PSF, for the photon beam and the electron beam. For the photon beam, four common

variance-reduction methods [15,18] were used to increase the efficiency of the simulation: bremsstrahlung splitting in the target, Russian roulette coupled with splitting, both outside and inside the beam, and circular splitting. When compared with analogue simulations (with no variance-reduction), these techniques allowed the simulation efficiency to be improved by a factor greater than 500 [24].

To validate the calculated PSF, depth-dose curves and dose profiles were measured in a 30 × 30 × 30 cm^3 water phantom, using a PTW-31002 cylindrical IC for the photon beam and a NACP-02 plane parallel IC for the electron beam. Dose profiles were measured at the reference depth: 10 cm for the 12 MV beam and 4.2 cm for the 18 MeV beam [22,23]. The measurement accuracy was better than 0.4% (1σ) for the photon beam and better than 0.6% (1σ) in the case of the electron beam. These measurements were compared to the simulated results, obtained with PENELOPE and PENFAST, in order to validate the PSF.

2.3.2. CALCULATION OF DOSE DISTRIBUTIONS IN INHOMOGENEOUS PHANTOMS

The PSF obtained during the beam modelling process serves as input data for both the PENELOPE and the PENFAST dose simulations. The same number of histories is thus used when comparing the results of both calculations.

For the PENELOPE dose simulations, the phantoms (figure 1a-d) were modelled using PENGEOM, assuming the chemical composition specified by the manufacturer. In all PENELOPE calculations, the cutoff energy was set to 100 keV for electrons and positrons, and 10 keV for photons. The parameter C_1, which determines the mean free path between hard elastic events, was set to 0.05, and the parameter C_2, which gives the maximum average fractional energy loss in a single step, was set to 0.05. The cutoff energy loss for inelastic collisions and bremsstrahlung emission were set to $W_{cc} = W_{cr} = 10$ keV.

For the PENFAST simulations, the voxelised phantoms (figure 1a-d) were generated numerically using a homemade Fortran program which assigns the correct chemical composition and mass density to each voxel. The CT scans were not used in order to avoid the accuracy limitations which arise when converting HUs to tissue parameters. In the following, we evaluate the influence of material composition on dose distributions. In fact, the phantom walls were made with PMMA, the data of which is not available in the original version of PENFAST. In the first step, for the PENFAST simulations, the PMMA voxels were assigned to have the same composition as water (the material with the closest density to that of PMMA), although with the correct density of PMMA (1.19 $g.cm^{-3}$). As a consequence, large discrepancies (> 3%) can be noticed, between the PENFAST and PENELOPE dose simulations, in the PMMA wall. Recently, Verhaegen and Devic [25] showed that medium and mass density mis-assignements can lead to dose errors of up to 10% for 6 and 15 MV photons, and 30% for 18 MeV electrons. Consequently, in the second step we generated look-up tables (angular multiple-scattering and energy loss distributions were generated by means of MC simulations using PENELOPE) for PMMA, which were then included in the PENFAST library. When the correct material composition and density are used, the differences previously seen in PMMA disappear. It must be emphasized that for accurate MC dose calculations, the material's chemical composition and mass density must be scrupulously respected.

3. Results and discussion

3.1. Homogeneous phantom comparisons

Figures 2(a-d) show the depth-dose curves and dose profiles, for IC measurements and simulated results computed with PENELOPE and PENFAST in water, for 12MV photon and

18 MeV electron beams. These simulations were made using cubic voxels of 5 mm wide for the photon beam and 4 mm wide for the electron beam.

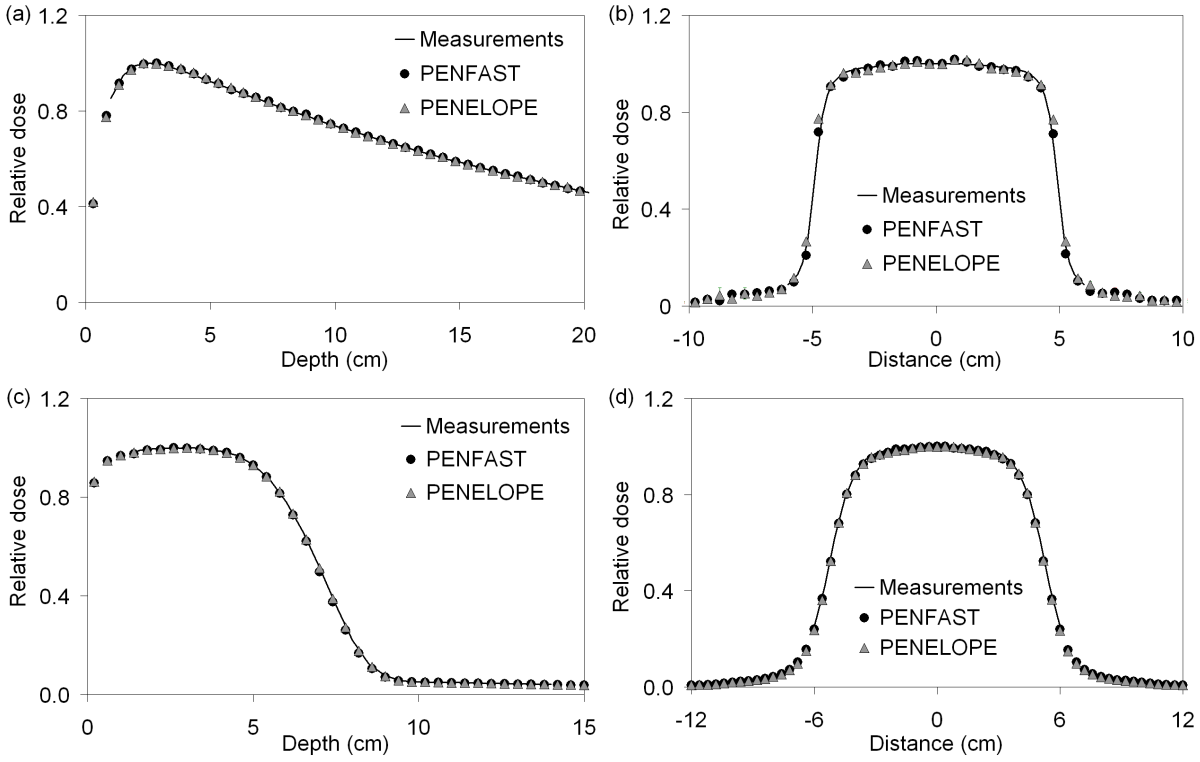

Figure 2. Depth-dose and profiles in water for a photon beam (a,b) and an electron beam (c,d).

All depth-dose curves are normalized to the maximum dose. The statistical uncertainties associated with the simulated depth-dose curves are less than 0.4% for the high dose regions and less than 1% elsewhere.

The dose profiles are normalized to the averaged dose of the three central points, in order to reduce fluctuations on either side of the central point. The statistical uncertainties of the simulated profiles are less than 0.6% (1σ) in the central region, and less than 2% (1σ) in the penumbra regions.

Overall, the PENFAST and PENELOPE simulations fall within 1% - 1 mm of the measured values. These results validate the PSFs used later, as input data for the inhomogeneous phantom dose calculations made with PENELOPE and PENFAST.

3.2. Inhomogeneous phantom comparisons

In the following analysis, experimental measurements of depth-dose curves and dose profiles are compared to simulated results obtained with both PENELOPE and PENFAST codes. The comparisons are made in inhomogeneous phantoms (figure 1) for both the electron beam and the photon beam. The measured and simulated central axis dose distributions are normalized relative to the dose at the reference depth in water, while all measured and simulated dose profiles are normalized relative to the dose on the central axis.

3.2.1. 12 MV PHOTON BEAM

Figures 3(a), (b) and (c) provide comparisons between measured and simulated depth-dose curves (for both PENELOPE and PENFAST simulations) in the water-lung-water (figure 1a), water-bone-water (figure 1b) and water-bone-lung-water (figure 1c) phantoms, respectively. Depth doses were tallied using cubic bins of 5 mm wide.

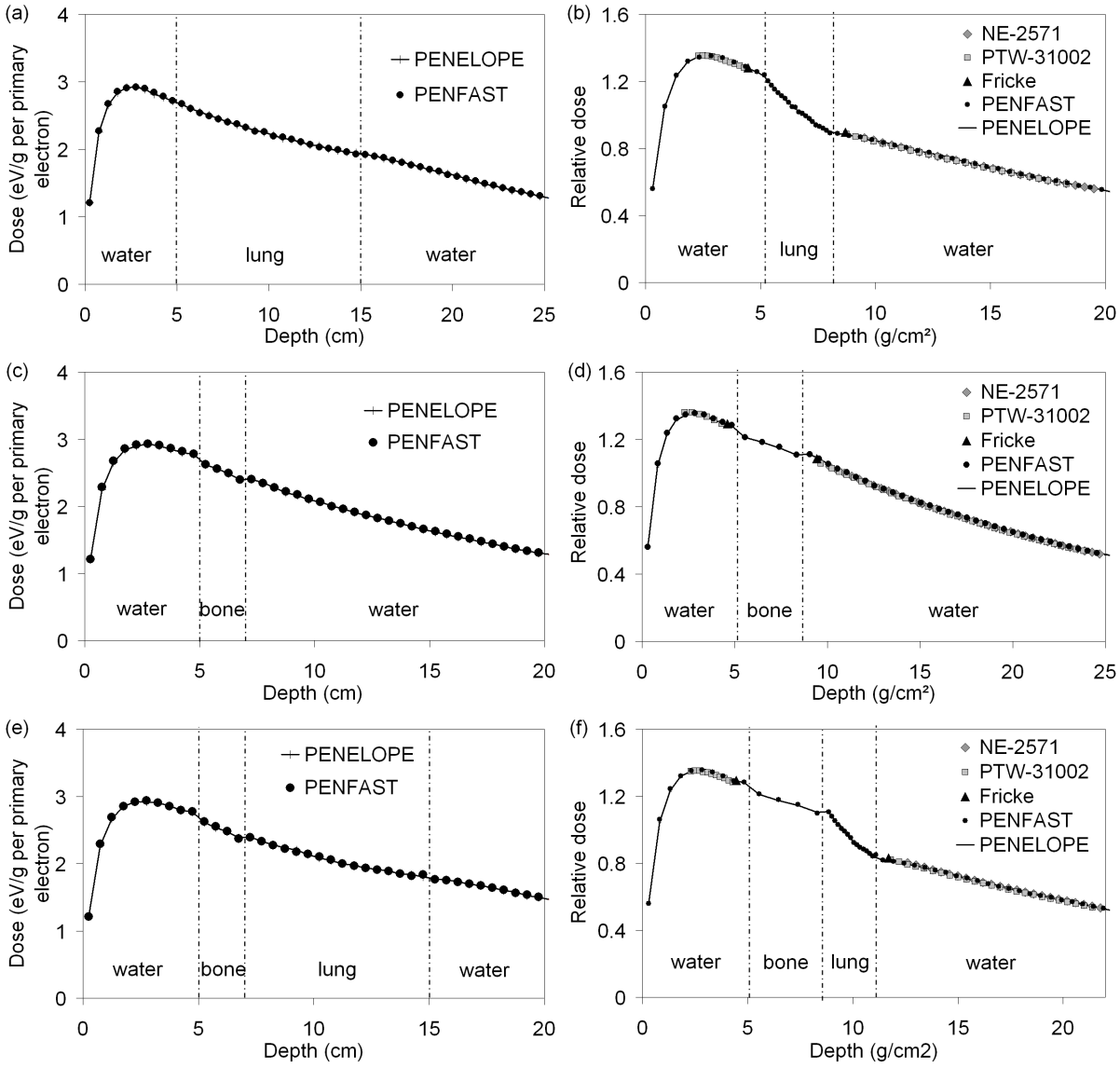

Figure 3. Photon 12 MV beam: (a), (b) and (c) central axis dose distributions; (d), (e) and (f) depth-dose curves for the water lung-water, water-bone-water and water-bone-lung-water phantoms respectively.

Figure 3(d) shows the comparison between measured and simulated dose profiles at depths 22 and 25 cm in the water-lung mediastinum phantom (figure 1d). Dose profiles were tallied using cubic voxels of 2 mm wide. The statistical uncertainty of the simulated results was of the order of 1% in the central region of the dose profiles and between 1.5-3% in the penumbra. In all regions, the discrepancies between measurements and PENFAST, as well as for PENELOPE do not exceed 2% - 2 mm.

Table 1 presents the statistical uncertainties (1σ) associated with the simulations, as well as the maximum relative difference between measurements and MC results in the three phantoms. For all phantoms, good agreement is found between the PENFAST and PENELOPE simulations in water and inside the heterogeneity layer. The largest discrepancy is less than 1% - 1 mm. Furthermore, the PENFAST calculations are found to be in good agreement with the experimental measurements. The maximum relative difference between the latter and the PENFAST depth dose calculations is observed in the vicinity of the heterogeneity, where it reaches 1.2% for the water-lung-water and the water-bone-water phantoms, and 1.7% for the water-bone-lung-water phantom. This difference decreases rapidly, when it is determined at greater distances from the heterogeneity. These discrepancies can be partly explained by the fact that the cylindrical IC tends to underestimate the dose, due to the attenuation of low energy electrons by its wall (see figure 3b, at depths of 7.3-8.5 cm near the bone/water interface). These electrons are however taken into account in the simulations, since the latter do not include the detectors.

Table 1. Maximum relative difference between measurements and both PENELOPE and PENFAST simulations in the three phantoms. The statistical uncertainty (1σ) range associated with MC simulations is also shown.

Phantom	Statistical uncertainty (1σ) range	Maximum relative difference (%)	
		PENELOPE	PENFAST
Lung	0.4 - 0.8	1.1	1.2
Bone	0.5 - 0.9	0.7	1.2
Bone-Lung	0.4 - 0.9	1.5	1.7

These results show good agreement between the measurements and the MC simulations made using PENFAST for the photon beam, since for TPS commissioning the suggested dose accuracy in inhomogeneous phantoms is usually 3% - 2 mm in the high dose and penumbral regions respectively [26-27].

3.2.2. 18 MEV ELECTRON BEAM

The comparisons between measured and simulated depth-dose curves with PENELOPE and PENFAST, for the water-lung-water and the water-bone-water phantoms, are shown in figures 4(a) and (b) respectively. The same voxel size was used for the PENELOPE and PENFAST simulations. This was set to 4 mm for the water-lung-water phantom and 2 mm for the water-bone-water phantom, in order to conserve the correct material composition and mass density at the centre of each voxel.

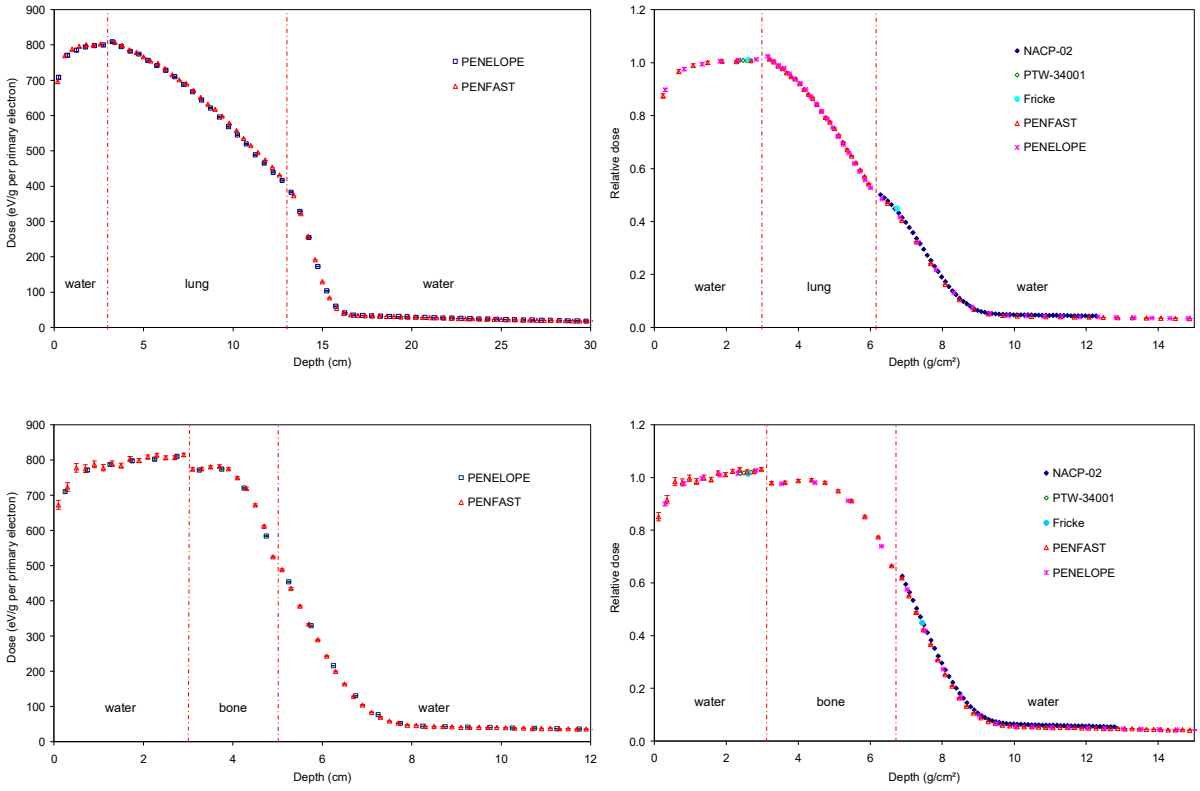

Figure 4. Electron 18 MeV beam: (a) and (b) central axis dose distributions; (c) and (d) depth-dose curves for the water-lung-water and water-bone-water phantoms respectively.

The statistical uncertainties (1σ) associated with the simulations, as well as the maximum relative difference between the measurements (IC and Fricke dosimeter) and the PENELOPE and PENFAST simulations, are presented in Table 2 for both phantoms. Overall,

to within the statistical uncertainties of the simulations, good agreement is found between PENFAST and PENELOPE, the largest discrepancy being less than 1% - 1 mm. Moreover, a good agreement is found between measured and simulated depth doses with PENFAST. The maximum relative difference between the measurements and the PENFAST simulations is 1.9% for the water-lung-water phantom, and 1.8% for the water-bone-water phantom.

Figure 4(c) shows the comparison between measured and simulated dose profiles at depths 2.5 cm (in front of the lung-equivalent heterogeneity) and 14.5 cm (behind the lung-equivalent heterogeneity) in the water-lung-water phantom. The statistical uncertainty of the simulated results was of the order of 1% in the central region of the dose profiles and between 1.5-3% in the penumbra. The agreement between the measurements and simulations is good, the differences being usually below the 2% - 2 mm level. The maximum differences are found especially at depth 14.5 cm and in regions which are outside the beam.

Table 2. Maximum relative difference between measurements and both PENELOPE and PENFAST simulations in the two phantoms. The statistical uncertainty (1σ) range associated with MC simulations is also shown.

Phantom	Statistical uncertainty (1σ) range	Maximum relative difference (%)	
		PENELOPE	PENFAST
Lung	0.3 - 1.7	1.8	1.9
Bone	0.4 - 3.9	1.7	1.8

3.3. Dose calculation time

A comparison between PENELOPE and PENFAST dose calculation times obtained in water and inhomogeneous phantoms is presented in Table 3. The PENELOPE and PENFAST calculations are performed under the same conditions, on a cluster including 22 processors (64 bits AMD Opteron), each equipped with 8 Go of RAM and operating at 2.4 GHz. Therefore, the same pre-calculated PSF is used for both PENELOPE and PENFAST dose simulations. Moreover, the same number of histories and voxels are used for both codes. The

simulation parameters used in PENELOPE are described above (Section 2.3.2), and the CPU times are those needed to obtain a 1% average statistical uncertainty, in regions where the dose exceeds 50% of the maximum dose.

Table 3. PENFAST computation times depending on the beam quality and number of voxels defining the phantom.

Phantom	Photon (12MV)		Electron (18 (MeV)	
	Number of voxels (millions)	CPU time (min)	Number of voxels (millions)	CPU time (min)
water	0.512	35	1	10
Lung	0.512	50	1	20
Bone	0.512	55	8	50
Bone/Lung	0.512	85	-	-

As noted in Table 3, the CPU time required by PENFAST is generally of the order of a few minutes, whereas PENELOPE requires several hours on the 22 PC network. Depending on the simulated configuration, PENFAST is estimated to be about 14-20.5 times faster than PENELOPE for photons, and 9-10.5 times faster than PENELOPE for electrons. The main reason for the largest execution speed gain in PENFAST face to the general-purpose PENELOPE code is the use of pre-calculated data in PENFAST (i.e. look-up tables of interaction properties are pre-generated for wide angles and various energy ranges, using some of the subroutines of PENELOPE), whereas with PENELOPE these data are determined during the particle transport simulation (as in any other conventional MC code). In that way a large part of calculation time is made before the use of PENFAST as a hidden execution time.

It is difficult to make an accurate comparison between the CPU times obtained using PENFAST and other available dose calculation engines, since the simulation time depends on the beam quality, the number of histories, the number of voxels used to define the patient anatomy, the number of histories per cm^2, the simulation parameters, the number of voxels defining the geometry, and CPU characteristics. However, to illustrate the typical differences in computation speed, the CPU time required for systems reported in several articles [1,3-8] is

generally of the order of a few minutes for an electron beam calculation, and a few hours for a photon beam calculation on a standard CPU. PENFAST is thus able to provide considerably shorter computing times for photons, and similar computing times for electrons, when compared to other existing systems. It must be noted that compared to the abovementioned fast simulation codes, PENFAST does not rely on virtual source models or MC kernels for performing MC treatment planning, but on a detailed description of the whole geometry of the accelerator together with a full MC simulation from the exit of the bending magnet down to the patient [9-11]. This approach allows the possibility of accurately estimating the absorbed dose in the patient.

4. Conclusions

The work presented in this paper demonstrates that, for photon and electron beams, PENFAST provides results which are equivalent, in terms of accuracy (with discrepancies of less than 1% - 1 mm), and which are substantially faster than those obtained using PENELOPE. The PENFAST code can run 9-21 times faster than the PENLEOPE code, depending on the beam quality and simulation geometry. In addition, the good agreement found between PENFAST simulations and measurements in inhomogeneous phantoms (generally within ± 2% - 2 mm relative agreement) demonstrates this code's ability to produce accurate photon and electron dose simulations, in situations with fluence disturbances in the regions of electronic equilibrium.

However, more testing will be needed before PENFAST can be used in a clinical environment. For example, the issue of accurate dose calculations in small field conditions is important for the evaluation of the MC code, in situations where lateral electron disequilibrium can be significant. Another important topic involves the conversion of HUs to the quantities required by the MC code (*i.e.* chemical composition and mass density), for each

voxel in the CT scan geometry. As this data is used as an input to the MC dose simulations, an accurate conversion is required in order to guarantee suitable accuracy in the dose calculations. These issues, along with others, are at the focus of current studies.

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