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Validation of a dose deposited by low-energy photons using GATE/GEANT4

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Abstract
The GATE Monte Carlo simulation platform based on the Geant4 toolkit has now become a diffused tool for simulating PET and SPECT imaging devices. In this paper, we explore its relevance for dosimetry of low-energy $^{125}$I photon brachytherapy sources used to treat prostate cancers. To that end, three $^{125}$-iodine sources widely used in prostate cancer brachytherapy treatment have been modelled. GATE simulations reproducing dosimetric reference observables such as radial dose function $g(r)$, anisotropy function $F(r, \theta)$ and dose-rate constant ($\Lambda$) were performed in liquid water. The calculations were splitted on the EGEE grid infrastructure to reduce the computing time of the simulations. The results were compared to other relevant Monte Carlo results and to measurements published and fixed as recommended values by the AAPM Task Group 43. GATE results agree with consensus values published by AAPM Task Group 43 with an accuracy better than 2%, demonstrating that GATE is a relevant tool for the study of the dose induced by low-energy photons.

(Some figures in this article are in colour only in the electronic version)

1. Introduction
Use of Monte Carlo simulations is becoming more and more diffused to compute dose deposits where complex geometries are involved. As the transport of ionizing particles is a complex process, the accuracy of currently available dose computation models for the planning of radiation treatments is limited. Also, it has been shown that some discrepancies compared to true dose distributions may be clinically significant in specific cases. Monte Carlo tools enable a universal accuracy: all materials, modalities, anatomic geometries and devices can be modelled so that they are used to obtain an accurate estimation of quantities difficult or impossible to measure.
As a consequence, Monte Carlo tools are progressively integrated into treatment planning software (TPS) used in radiotherapy to improve their accuracy. PENELOPE tools are being integrated into the ISOGRAY TPS, MCRTV (Monte Carlo for Radiotherapy Treatment Plan Verification) consist of the EGS4/PRESTA MC codes interface with the TPS Eclipse from Varian Medical Systems, Palo Alto, CA, USA (Yamamoto et al 2007). XVMC (Kawrakow et al 1996, Kawrakow 1997, Fippel 1999, Kawrakow and Fippel 2000) is being interfaced to CMS, Elekta and Brainlab TPS, Peregrine (Hartmann-Siantar et al 1997) is proposed by North American Scientific. VMC++ (Kawrakow and Fippel 2000) is used by Nucletron and MMC (Neuenschwander and Born 1992, Neuenschwander et al 1995) with VMC++ are used by Varian Medical Systems.

Recently, a new platform has received a lot of interest in the medical physics research community because of a number of attractive features. The GATE platform (Jan et al 2004, Lazaro et al 2004, Cullen et al 1997) was designed from the beginning to meet specific needs of the simulations for nuclear medical imaging. In addition to the many potentialities provided by GEANT4 (Agostinelli et al 2003, GEANT4 user’s guide For Application Developers web site), GATE includes specific modules necessary to perform realistic simulations including modules managing time and time-dependent processes (detector and source movements, radioactive decay, dynamic acquisitions), complex source distributions and easy description of geometry.

GATE is built on Geant4, software developed by the high energy physics community to design very large detectors. While GATE and its underlying library Geant4 have been extensively validated for photon emitters used in PET and SPECT (Jan et al 2004, Assie et al 2005, Staelens et al 2003), its potential use for dosimetry has not been properly explored and documented.

In this paper, we study the relevance and accuracy of GATE for dosimetric applications with low-energy photon sources commonly used to treat prostate cancers. Geant4 brachytherapy example (Agostinelli et al) has been proposed for the simulation of one of the $^{125}$I sources (BEBIG source) studied in this paper but is limited to the study of the energy deposited around the source. Therefore, this paper can be defined as a Monte Carlo based study recommended to unambiguously define all dosimetric quantities required by the AAPM Task Group 43; combined also to measurements, it represents a complete work that extends from the modelling of sources emitting photons with mean energies below 50 keV ($^{125}$I) to comparisons with published and validated values. This paper provides therefore the first validation of the GATE platform at energies below 50 keV.

In section 2, we specify the $^{125}$I sources studied, the corresponding GATE simulations that have been performed, and the relevant dosimetric quantities that have been calculated using a grid infrastructure. Section 3 presents a comparison of the results obtained with GATE to the existing body of data in the literature, while section 4 proposes a discussion on the results obtained and exciting perspectives for the use of GATE in dosimetry.

2. Material and methods

2.1. Brachytherapy sources

2.1.1. The Best model 2301 source. The Best model 2301 $^{125}$I source (see figure 1) is made of a double-encapsulated titanium source (density of 4.54 g cm$^{-3}$) surrounding a tungsten x-ray marker coated with an organic carbon layer impregnated with $^{125}$I. Figure 1 shows a schematic diagram of this source (courtesy of BMI web site) and a screenshot of the source simulated with GATE.
Figure 1. Schematic drawing and OpenGL visualization of $^{125}$I Best model 2301 source simulated with GATE.

Figure 2. Schematic drawing and OpenGL visualization of $^{125}$I Symmetra model I25.S06 source simulated with GATE.

The source has a physical length of 5 mm and an outer diameter of 0.8 mm. The internal cavity of the source capsule has a diameter of 0.64 mm.

The cylindrical tungsten x-ray marker has a physical length of 3.7 mm and a diameter of 0.25 mm and is coated with a 0.1 mm thick organic matrix. The thickness of the organic matrix at each of the two ends of the marker is 0.15 mm. The titanium wall thickness is 0.08 mm (0.04 mm thick for each capsule). The thickness of the two ends of the capsule is 0.08 mm.

2.1.2. The Symmetra model I25.S06 source. The source capsule consists of a 0.05 mm thick titanium tube, density of 4.54 g cm$^{-3}$, that is laser welded at both ends. The welds are spherical with an average thickness ranging from 0.44 to 0.48 mm depending on the batch. On average, the weld is slightly thinner on the axis than near the cylindrical wall. The radioactive seed core consists of a cylindrical ceramic shell with outer and inner diameters of 0.60 and 0.22 mm, respectively, and a length of 3.50 mm. The ceramic has a density of 2.88 g cm$^{-3}$ and consists of alumina (Al$_2$O$_3$) within which a radioactive iodine $^{125}$I is uniformly distributed. A gold marker, density of 19.32 g cm$^{-3}$, 0.17 mm diameter and 3.5 mm long inside the ceramic core, permits radiographic localization of the seed (see figure 2). The Symmetra model is produced by BEBIG (BEBIG web site).
2.1.3. The Amersham model 6711 source. This source consists of a 4.5 mm welded titanium capsule, 0.05 mm thick, with welded end caps. The capsule contains a 3.0 mm long silver rod onto which $^{125}\text{I}$ is adsorbed (see figure 3) (Amersham web site).

2.2. Dose calculation formalism

Following the recommendations of the AAPM Radiation Therapy Committee Task Group No.43 (Nath et al. 1995) updated in March 2004 (Rivard et al. 2004), in order to evaluate the 2D dose distribution around cylindrically symmetric sources, the dose-rate at point $(r, \theta)$ can be written as in (1):

$$D(r, \theta) = S_k \wedge \left[ \frac{G(r, \theta)}{G(r_0, \theta_0)} \right] g(r) F(r, \theta).$$

We have adopted the line source model.

$S_k$ is the air kerma strength of the source at 1.0 cm from the source (in cGy cm$^2$ h$^{-1}$).

$\Lambda$ is the dose-rate constant at 1.0 cm from the source (in cGy h$^{-1}$ U$^{-1}$) (see equation (2))

$$\Lambda = \frac{D(1, \pi/2)}{S_k}.$$

$\theta$ is the angle between the source axis and the radial vector from the source centre at point $(r, \theta)$.

The geometry factor in equations (3) and (4) accounts for the variation of relative dose due only to the spatial distribution of activity within the source, ignoring photon absorption and scattering in the source structure. For a line source approximation, it is defined as

$$G_L(r, \theta) = \frac{\beta}{L r \sin \theta} \quad \text{if} \quad \theta \neq 0$$

$$G_L(r, \theta) = \left( r^2 - \frac{L^2}{4} \right)^{-1} \quad \text{if} \quad \theta = 0^\circ.$$

The radial dose function $g(r)$ in (5) accounts for the effects of absorption and scatter in the medium along the transverse axis of the source, it is defined as

$$g_L(r) = \frac{D(r, \theta_0)}{D(r_0, \theta_0)} \frac{G_L(r_0, \theta_0)}{G_L(r, \theta_0)}.$$
The 2D anisotropy function in (6) describes the variation in dose as a function of polar angle relative to the transverse plane and is defined as

\[ F(r, \theta) = \frac{D(r, \theta)}{D(r, \theta_0)} \frac{G_L(r, \theta)}{G_L(r, \theta_0)}. \]  

(6)

2.3. **GATE Monte Carlo simulations**

In this study, the results presented are calculated with the version 3.0.0 of the GATE generic Monte Carlo platform. This version uses the GEANT4 version 4.8.0.p01. Compton scattering, photoelectric absorption and Rayleigh scattering for photons were considered. Electron processes (ionizations, multiple scattering and Bremsstrahlung) were simulated but not followed in the simulation. In particular the two electromagnetic physics models standard and low-energy were used and tested. GATE inherits the Geant4 capability to set thresholds for the production of secondary electrons, x-rays and delta rays (Agostinelli et al 2003), when particles created are tracked at the end of their range. A cut allows the user to suppress particles whose range would be less than a user-defined value that we name the range cut. Therefore, the cut applied on the secondary electrons is chosen high enough (1 m) because the maximum range of secondary electrons is small compared to the recovering ring dimensions. The cut on x-rays is fixed to 5 keV. The ROOT System analysis tool was used to calculate the relevant dosimetric quantities. \( 3 \times 10^8 \) events for air kerma strength calculation to avoid statistical fluctuations in the dose calculation in air (see the dose-rate constant section). The simulations of anisotropy functions and radial dose functions were performed in liquid water. The energies of the gamma rays generated by a 125I sources were 27.202 keV (40.6%), 27.472 keV (75.7%), 30.98 keV (20.2%), 31.71 keV (4.39%) and 35.492 keV (6.68%) as recommended by the TG-43; therefore 14 760 000 gamma particles were generated for each simulation in order to calculate dosimetric quantities.

2.3.1. **Air kerma strength and dose-rate constant calculation.** The dose-rate constant was calculated by dividing the dose deposited at 1 cm from the source and \( \theta = 90^\circ \) in liquid water by the air kerma strength \( S_k \).

By taking into account the isotropic emission around the source, \( D(1, \pi/2) \) was calculated by recovering the energy deposited in a ring around the source with a width \( \Delta r = 0, 2 \) cm and a height \( \Delta z = 0, 2 \) cm following figures 4 and 5, the total energy was then divided by the volume of the ring. The source is positioned at the centre of a water spherical phantom of 30 cm in diameter. We tested the fact that dose gradient over the size of the scoring ring is almost linear and therefore is well adapted for the calculations to follow the recommendations of the AAPM-TG43 requiring a volume averaging less than 1%.

The in-air calculations were performed in a ring located at 5 cm from the source with \( \Delta r = 0.8 \) cm and \( \Delta z = 0.8 \) cm and then corrected using an inverse square law in order to reproduce the NIST calibration procedure (see (7)):

\[ S_k = \frac{K_{\text{air}}(1 \text{ cm})}{K_{\text{air}}(5 \text{ cm})} 5^2. \]  

(7)

Results obtained were corrected for photon attenuation and scattering in air.

In order to avoid random fluctuations in the scoring volume at 5 cm, we ensure that the statistics of events were sufficient for the calculations as demonstrated in section 3.4.

Cut-off on low-energy photons as titanium characteristic x-ray production has been fixed to 5 keV in order to exclude it in the calculated \( S_k \) value.
2.3.2. Anisotropy function and radial dose function calculations. As for the dose-rate constant calculation, the system used to calculate the radial dose function $g(r)$ and the anisotropy function $F(r, \theta)$ by scoring the energy and positions of interactions is described in figures 4 and 5, the total energy is then divided by the volume of the ring. Table 1 indicates the values for the $\Delta r$ and $\Delta z$, respectively radial and longitudinal distances. For each simulation, the source is positioned at the centre of a spherical phantom of 30 cm in diameter.

2.4. Deployment of GATE calculations on a grid infrastructure

A computational grid is, as described by Foster and Kesselman (1999), a hardware and software infrastructure that provides dependable, consistent, pervasive and inexpensive access to high-
end computational capabilities; we talk about an infrastructure because a computational grid is concerned, above all, with large-scale pooling of resources. We use a computational grid to run our Monte Carlo simulations in a faster way.

The computing time of a Monte Carlo simulation depends on different parameters, for instance: the number of particles generated during a simulation, the medium where particles interactions occur (typically voxelized or homogeneous phantoms). Depending on the type of material filling the medium and the type of particles generated, the number of physical interactions can vary. The number of calls to the Random Number Generator (RNG) is consequently dependent on these parameters (Perkins et al. 1997).

Each Monte Carlo simulation uses a sequence of random numbers to reproduce the probability of the physical interactions in matter. The more numerous the interactions in a medium are, the longer the sequence of random numbers generated for the simulation is. A simple way to reduce the execution time of a Monte Carlo simulation used in physical experiments, is to sub-divide a long or a very long simulation into little ones by indexing to each simulation a sub-sequence of random numbers obtained by partitioning a long sequence of random numbers. Sub-sequences obtained have to be independent.

To perform such parallelization of the code, we used the EGEEII (Enabling Grids for E-sciencE) grid infrastructure. The EGEE grid is built on the EU Research network GEANT and exploit grid expertise generated by many EU, national and international grid projects to date. Funded by the European Commission, the EGEE project community has been divided into 12 partner federations, consisting of over 70 contractors and over 30 unfunded participants covering a wide range of scientific and industrial applications. The biomedical applications are supported by a large number of sites in EGEE: 150 sites representing more than 5000 CPUs and 21 TBs are accessible for the storage of medical data. The OpenGATE collaboration is involved with the EGEE-II project to produce and develop some tools dedicated to medical staff so they can deploy, in a very transparent and secure way, Monte Carlo simulations of medical experiments and treatments, reducing therefore significantly the computing time of the simulations.

3. Results

3.1. Uncertainty evaluation

Uncertainties associated with the dose-rate constant and radial dose functions $g(r)$ have been calculated by doing a quadrature sum of the statistical variations, geometric uncertainties and cross-section uncertainties.

For the moment, as stated in the update version of the AAPM Task Group No.43 Report (Nath et al. 1995, Rivard et al. 2004), a similar uncertainty analysis for anisotropy function is not provided.

The geometry seed uncertainties, as explained in this report, have been taken into account by taking the generic uncertainty assessment recommended for Monte Carlo calculations.

The results presented in this paper are affected by a total uncertainty $\sigma_Y$ calculated as the quadrature sum of the cross-section uncertainty $\%\sigma_{Y|\mu}$, the uncertainty of the seed geometry $\%\sigma_{Y|\text{geo}}$, and the statistical uncertainty $\%\sigma_{Y|S}$.

$$\%\sigma_Y = \sqrt{\%\sigma_{Y|\mu}^2 + \%\sigma_{Y|\text{geo}}^2 + \%\sigma_{Y|S}^2}$$

$$\%\sigma_Y = \sqrt{\left(\frac{\partial Y}{\partial \mu}\right)^2 \%\sigma_{\mu}^2 + \left(\frac{\partial Y}{\partial \text{geo}}\right)^2 \%\sigma_{\text{geo}}^2 + \%\sigma_{Y|S}^2},$$

(8)
where the relative uncertainty propagation factor has been calculated for the cross section value \( \mu \) as defined in (9) with a coverage factor of unity, approximating a 68% level of confidence:

\[
\frac{\partial Y}{\partial \mu} \equiv \frac{\mu \partial Y}{Y \partial \mu}.
\]

(9)

3.1.1. Dose-rate constant uncertainties. Total cross sections implying photons (photoelectric effect, Compton scattering, pair creation and Rayleigh scattering processes implied) obtained with the low-energy package of Geant4 have been compared to the XCOM database given by NIST (Berger et al 1999) for an energy spectrum going from 1 eV to 100 keV. The relative deviation between NIST and Geant4 values is below 2.0%. The estimation of \( \% \partial \lambda / \partial \mu \) has been calculated by comparing the calculation of \( \lambda \) with the cross section libraries of Geant4 and XCOM and is of 0.9%. As a consequence, the cross-section uncertainty \( \% \sigma_{\lambda/\mu} \) has been evaluated to be of 1.8%. Geometric uncertainties \( \% \sigma_{\lambda_{geo}} \) is taken as 2.0% following the AAPM TG-43 report. Statistical uncertainty \( \% \sigma_{\lambda_{S}} \) is below 0.2%. As a consequence, the total uncertainty \( \% \sigma_{\lambda} \) concerning dose-rate constant is below 2.7%.

3.1.2. Radial dose function uncertainties. Cross-section uncertainty \( \% \sigma_{g/\mu} \) is below 1.8% following the same calculations as in Section 3.1.1. Geometric uncertainties \( \% \sigma_{g_{geo}} \) is taken as 2.0%. Statistical uncertainty \( \% \sigma_{g_{S}} \) in all regions of interest for radial dose function calculations is below 0.2%. As a consequence, the total uncertainty \( \% \sigma_{g} \) concerning radial dose functions is below 2.7%.

3.2. Grid impact to reduce computing time

Figure 6 illustrates the computing time in minutes for the simulation of a 125-iodine brachytherapy source in water. Comparisons are presented for a simulation running on a single Intel Xeon CPU locally and for the same simulation split in 10, 20, 50 and 100 jobs on 1320 CPU on the grid. In this case, the lowest computing time is obtained for 50 jobs running in parallel.

Such an example shows that the computing time is not proportional to the number of jobs running in parallel. The main reasons for this behaviour are the following:
• the launching time of the jobs,
• the building of the geometry at the beginning of each simulation which does not depend on the random generation of the particles,
• the batch queue system managing the jobs: even if the splitting of the simulation is very high, the number of jobs waiting in the batch queue depends on the load on the processors.

Finally, the use of the grid allowed reducing the time needed to simulate the 125-iodine source in a water phantom from 16 h on a local Intel Xeon processor to 40 min on 1320 CPU of the EGEE grid infrastructure. By splitting a simulation of sources corresponding to a prostate brachytherapy treatment on the whole grid infrastructure for biomedical applications (∼7500 CPU), we obtain a dosimetry in about 12 h.

3.3. Comparison of EM physics models

GEANT4 provides a large set of physics processes for electromagnetic (EM) interactions. For the EM interactions of photons and electrons, three models are available: standard, low-energy and Penelope. The three EM physics models employ different cross-sectional data sets and final state-sampling algorithms (Poon and Verhaegen 2005). We have compared the standard and low-energy models from GEANT4, available for photon interactions in the GATE platform, with the Task Group 43 consensus values in figure 7 in order to quantify the differences obtained using these two models at the characteristic low energy of 125-iodine sources.

The mean relative differences obtained between the two models do not exceed 7%. The maximum relative difference between TG43 consensus values and the low-energy EM model is 1%. As the low-energy model allows particle simulations down to 250 eV and because the atomic relaxation subsequent to photoelectric and ionization interactions is included, we decided to use in our simulations the low-energy model.

3.4. Dose-rate constant

The in-air simulations for the calculation of the dose-rate constant following equations (2) and (7) are difficult to achieve with a low number of events in the simulations. Statistical fluctuations on the dose-rate constant calculation are shown in figure 8. The calculation of the
dose-rate constant has to be performed with at least $3 \times 10^8$ events in an air medium in order to avoid erroneous results due to insufficient statistics.

For Best medical model 2301 source, the simulated $\Lambda$ in liquid water was found to be $1.012 \text{ cGy h}^{-1} \text{ U}^{-1}$. Calculations for the Symmetra model I25.S06 source lead to a value of $1.018 \text{ cGy h}^{-1} \text{ U}^{-1}$.

For the model 6711 source, the simulated $\Lambda$ in liquid water was found to be $0.990 \text{ cGy h}^{-1} \text{ U}^{-1}$.

All the uncertainties affecting $\Lambda$ take into account statistical and cross-section components.

Table 2 shows a comparison of our results with recommended values of the AAPM TG-43 report and with other Monte Carlo codes cited as reference Monte Carlo results by the TG-43.

In table 3, we list the mean relative deviation between GATE results and the referenced Monte Carlo simulations of the TG-43 report. The Best medical model 2301 dose-rate constant differs by 0.2% from the corresponding Monte Carlo result of Sowards and Meigooni and differs by less than 1% from the TG-43 consensus value.
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Figure 9. Anisotropy function for ¹²⁵I Best Medical model 2301 as a function of angle at 2 cm.

Table 3. Mean relative deviation between GATE and Monte Carlo Consensus values.

<table>
<thead>
<tr>
<th>Comparisons</th>
<th>BMI model 2301</th>
<th>Symmetra model</th>
<th>Model 6711</th>
</tr>
</thead>
<tbody>
<tr>
<td>GATE (geant4.8.0)/TG43 Consensus value</td>
<td>0.6</td>
<td>0.1</td>
<td>2.6</td>
</tr>
<tr>
<td>GATE (geant4.8.0)/MCPT-DLC99 Hedtjärn et al (2000)</td>
<td>1.6</td>
<td></td>
<td></td>
</tr>
<tr>
<td>GATE (geant4.8.0)/MCPT-DLC146 Hedtjärn et al (2000)</td>
<td></td>
<td>0.1</td>
<td></td>
</tr>
<tr>
<td>GATE (geant4.8.0)/PTRAN Sowards and Meigooni (2002)</td>
<td>0.2</td>
<td></td>
<td></td>
</tr>
<tr>
<td>GATE (geant4.8.0)/PTRAN Williamson (1991)</td>
<td></td>
<td></td>
<td>1.2</td>
</tr>
</tbody>
</table>

Concerning Symmetra model I25.S06, our result deviates by less than 2.0% from the result obtained by Hedtjärn et al (2000) deviation from TG-43 consensus value is less than 1%.

The dose-rate constant obtained for the model 6711 deviates by 1.2% from the value obtained by Williamson and by 2.6% from the TG-43 consensus value.

3.5. Anisotropy function

Anisotropy function values have been calculated from the Monte Carlo simulations in liquid water phantom. We compared our results with consensus values from the AAPM Radiation therapy Committee Task Group 43 report in liquid water. Results of Sowards and Meigooni (2002) are recommended as consensus values for the Best medical model 2301 source. The results of Hedtjärn (Hedtjärn et al 2000) are recommended as consensus values for the Symmetra model I25.S06 and results of Weaver (Weaver et al 1989) are recommended for the model 6711.

3.5.1. Best Medical model 2301. Figures 9 and 10 present anisotropy values as a function of polar angle $\theta$ at 2 and 7 cm from the source centre. These results are also compared with the consensus values $F(r, \theta)$ of the AAPM TG-43. Close to the source, internal geometrical features are strongly reflected in the anisotropy function. Far from the source, contributions from scattered radiation in water medium increase, washing out the geometrical effects from the source.
Anisotropy values greater than unity can be observed beyond 50°. A very good agreement with Sowards and Meigooni calculations is found: the mean relative deviation is less than 3.2% at 2 cm and 7 cm. Our results show a more anisotropic dose distribution close to the source near the longitudinal axis.

3.5.2. Symmetra model I25.S06. We represent in figures 11 and 12 the anisotropy as a function of the angle at two different distances. Our data show a less anisotropic dose distribution close to the source near the longitudinal axis: mean relative deviation is less than 1.5% at 2 cm and less than 2.2% at 7 cm. A very good agreement can be observed beyond 50°.

3.5.3. Model 6711. Figures 13 and 14 present anisotropy values as a function of angle at two different distances. Our results are in very good agreement with those of Weaver (below 2.0%) beyond 20° at all distances. Mean relative deviation is around 2.8% at 2 cm and 3.2% at 5 cm.
3.6. Radial dose function

Radial dose functions $g(r)$ have been calculated for the three types of sources in liquid water phantoms. The $g(r)$ data are presented in figures 15–17.

We compared our calculations with published Monte Carlo data; those of Sowards and Meigooni (2002) for the Best model 2301 source, those of Hedtjärn et al (2000) concerning the Bebig I25.S06 source (BEBIG web site) and those of Williamson (Williamson 1991, Daskalov et al 1998) for the model 6711 as all those results are the recommended values of the AAPM TG-43. It has to be noted that the consensus values of $g(r)$ below 10 mm were obtained using an extrapolation fit.

A good agreement is obtained between Best model 2301 GATE simulation and TG-43 recommended values (mean relative difference is 3.5%, less than 2% before 6 cm). Concerning Bebig I25.S06 source differences obtained are lower (mean relative difference is 1.0%, less than 2.0% for all distances). For the Amersham model 6711 however, discrepancies with TG43 consensus values are higher (mean relative difference is 15.0%). Some explanations for this result are proposed in the next section.
4. Discussion

GATE and therefore GEANT4 are already recognized tools in the field of medical physics. Thanks to their powerful geometry construction capabilities and thanks to the fact that GEANT4 provides many alternative physics models and functionalities, it is now possible to achieve very precise calculations by keeping a high-level control on the tracking of the particles.

We chose to compare the standard and the low-energy models available in the GEANT4 release in order to see which of the two models were the more suitable to the simulation of such sources by comparing our results with TG-43 consensus values. It has been shown that the low-energy model fit better the TG-43 results and is therefore the model chosen to simulate dosimetric quantities characteristic of the sources.

GATE gives results in good agreement with the referenced values already published (Monte Carlo calculations and measurements). The dose-rate constant $\Lambda$, the anisotropy function $F(r, \theta)$ and the radial dose function $g(r)$ have been obtained for the three different source models with a relative mean difference with the referenced published results less than...
Figure 16. Radial dose function $g(r)$ for $^{125}$I Symmetra I25.S06 from 0 to 10 cm in liquid water compared to the calculated values reported in the literature.

Figure 17. Radial dose functions $g(r)$ for $^{125}$I Amersham model 6711 from 0 to 10 cm in liquid water compared to the calculated values reported in the literature.

3%. However, for the simulation of the radial dose function $g(r)$ of the $^{125}$I Amersham model 6711 quite important discrepancies have been observed reaching 15% with the TG-43 results. This relative high difference could be explained by a dose-scoring volume not well adapted. Some little modifications in the shape or in the radioactive layer of the source could also be explained the results. Therefore, the dose-scoring volume, the length of the radioactive layer and the thickness of the capsule have been tested to study their influence on $g(r)$ calculation. For the moment, we did not succeed to find which parameter is responsible for the differences observed.

The computing time is a bottleneck to make GATE compete with other codes for dosimetry application, particularly when using voxel phantoms to describe the human body. In addition to the variance reduction techniques actually in deployment for the GATE platform, we proposed to use the EGEE Grid infrastructure to compute the simulation in a parallel mode. The gain in computing time obtained by splitting the simulations is very encouraging. Thereafter, the development of a convivial tool to split, launch and retrieve GATE Monte Carlo simulations on a grid environment using a secured web portal is underway in our research group.
5. Conclusion

This paper provides the first validation of the GATE platform using GEANT4 low-energy model at energies below 50 keV. It shows that the present level of accuracy reached by GATE is adequate for its use for dosimetry using photons.

These results open very interesting perspectives because GATE is a fast Monte Carlo simulation platform thanks to its deployment on grid infrastructures. There is a clear opportunity to develop treatment planning in clinical routine using GATE Monte Carlo simulation deployed on the grid.

Further improvement in GATE performances is under study to ensure that the current GATE platform be consolidated and made more efficient to produce a new platform (fGATE project 2007) that would become a worldwide resource standard for Monte Carlo simulations in emission tomography and radiotherapy.

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