Development and application of a dose verification tool using a small field model for TomoTherapy

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Abstract

Helical TomoTherapy® is a radiation delivery technique that uses the superposition of many small fields to precisely deliver the prescribed dose to the patient. This work presents a dose verification tool that can be used as part of a quality assurance program for a tomotherapy system. This tool is based on a small field model that takes into account the two main effects that influence the dose distribution in small fields: the extended shape of the radiation source and the loss of lateral charged particle equilibrium (CPE) within the field. The dose verification tool was implemented for simple beam configurations and used to study the influence of temporal beam parameter variations on the delivered dose. After comparing measured and calculated output factors (OFs) and dose profiles for different field configurations, it was found that they agree well to within the globally-defined gamma acceptance criteria of 2%/2mm. The study demonstrated that none of the studied systematic and random variations applied resulted in failed gamma scores using gamma acceptance criteria of 3%/3mm. The developed model implemented in the verification tool allows to evaluate the performance of devices applying narrow photon beams in the treatment delivery and, in particular, to evaluate the delivery performance of a tomotherapy unit.

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1 Introduction

Modern radiation therapy techniques make use of different beam modifiers (e.g., MLCs, micro-MLCs and stereotactic collimators) in order to achieve the desired geometrical accuracy in the delivered dose to the target. One of the drawbacks in the use of such techniques, is that the treatment is typically based on the superposition of many small fields that define very sharp dose gradients. This carries a number of issues that need to be considered from the dosimetric and modelling points of view. The most important effects to consider under these conditions are: the partial occlusion of the radiation source [1] and the loss of charged particle equilibrium within the field [2].

Under small field conditions the dosimetric methods used to acquire the input data for a particular TPS must be carefully selected in order to reduce the uncertainties in dose calculation. The dosimeter used for this purpose should have an adequate lateral resolution and tissue equivalence [3]. Moreover, it is recommended that this data should be measured with more than one dosimetric system [4]. The use of small fields is also demanding in the modelling aspect. The dose calculation model needs to be designed to account for the extended source, the finite size of the focal spot, the shape of collimating devices, as well as secondary particle transport.

Nowadays, the optimal patient treatment is obtained in many cases using inverse-planning. This technique makes the planning process much more efficient, but the optimized parameters obtained for the treatment are no longer intuitive. This entails the requirement of a better understanding of the beam delivery and quality assurance (QA) processes associated, underlining the importance of independent dose calculation methods to verify the output of the planned treatment. International organizations, such as the ESTRO [5] and the AAPM [6,7], have recommended the inclusion of this practice as part of a comprehensive clinical QA program. An independent dose calculation should be performed prior to the treatment in order to detect and prevent possible flaws in it. Ideally, these calculations should rely on input data that is independent from the planning system.

The aforementioned issues motivated the development of a dose verification tool, based on a beam model applicable to small fields [8], that can be used to provide a supplemental procedure for the QA program of a Hi-ART TomoTherapy System. This tool allows one to calculate distributions delivered by tomotherapy to a homogeneous phantom, using input data that is independent of the TPS, to verify the overall performance of the system. Within this context, the dosimetric challenges and demands in dose modelling and QA, posed by the advent of non-standard beams in radiotherapy, are discussed; with particular interest in tomotherapy.

The dose verification tool can also be used to study the uncertainties related to the single processes involved in the helical delivery technique, as it allows variation of individual treatment parameters. This feature is particularly useful to study the influence on the overall treatment due to the variation of parameters such as output, gantry speed or couch velocity. The importance of such independent calculations in complex delivery techniques used in modern radiotherapy should be underlined, as the success of the treatment in these cases depends on the performance and synchrony of many individual components.

2 Materials and Methods

2.1 Dose calculation

The dose calculation engine used for this tool can accurately model small field distributions, as it explicitly takes into account the main effects that influence the dose distributions for narrow beams: the spatial extension of the source and the lack of CPE within the field [8]. To determine the relative 2-D intensity distribution of the extended source, a “slit-method” [9], based on the measurement of strip integrals of the source...
and the use of a CT image reconstruction technique, together with a fitting procedure for measured collimator factors, are used. The polyenergetic pencil beam kernel (PBK), which explicitly simulates the energy transport for narrow beam conditions, is calculated from mono-energetic dose deposition kernels (DDKs) and the weights of the energy spectrum of the beam. DDKs are obtained directly from Monte Carlo simulations using the EDSnrc calculation tool from the EGSnrc distribution, and the weights of the spectrum by fitting the calculated depth dose profile obtained for a reference field (5x10 cm²) to a percentage depth dose (PDD) curve measured under the same conditions. Taking into account these two effects and the known geometry of the radiation system, lateral fluence profiles, dose distributions and output factors can be modelled by convolution methods.

Three-dimensional dose distributions were obtained in a homogeneous unit density virtual cylindrical phantom, coaxially aligned with the gantry rotation axis. A schematic representation of this phantom, indicating the main evaluation planes, is shown in Figure 1(a). The axis of the “cheese”-phantom (seen in Figure 1(b)) has a length of 18 cm and a radius of 15 cm. The calculation voxels were defined to be of 1x1x1 mm³.

In the reference frame used on the Helical TomoTherapy system, the X-axis corresponds to the transversal direction, the Y-axis to the longitudinal direction, along the gantry rotation axis, and the Z-axis to the posterior-anterior direction. The gantry rotation angle \( \theta \) is defined with respect to the z-axis in the clockwise direction.

Three dynamic modes of the system are available. These are:

- **Topographic**: the couch moves longitudinally while the gantry angle is fixed during the complete beam-on time.
- **Rotational**: the couch is fixed while the gantry continuously rotates around it until the procedure is finished.
- **Helical**: both couch and gantry move simultaneously during the treatment.

The final dose distributions were obtained as follows:

1. **Setting the initial conditions**: Before starting the calculation process, the initial longitudinal position of the center of the phantom (\( Y_i \)) and the gantry angle (\( \Theta_i \)) need to be set.
2. **Input the treatment parameters**:
   
   (a) The opening of the jaws (\( Y \)), with possible values of 1, 2.5 or 5 cm at the isocenter plane
   
   (b) The number of projections per gantry rotation (\( P_r \)). The default value is 51.
   
   (c) The gantry period (\( T \)), from 10 to 60 s
   
   (d) The displacement of the couch per gantry rotation (\( Y_r \)) or alternatively the pitch (\( P \)), with \( P = Y_r / Y \).
3. **Enter MLC Sinogram**: The sinogram \( S(l,p) \) contains all the information related to the modulation of the beam in the transverse direction and indicates the fraction of time that each leaf (\( l \)) of the MLC remains open in every projection (\( p \)). Then the opening time \( O_{l,p} \) can be calculated as \( O_{l,p} = S(l,p) \times T/P \).
4. **Model the beamlet dose rate**: The beamlet dose rate \( b(l,y,z) \) in the virtual cylindrical phantom, for each one of the leaves opened individually and for each one of the three possible openings of the jaws, was calculated using the static mode of the system (with the central axis perpendicular to the couch plane and the phantom centred). Each “beamlet” distribution was stored for later use. After this step, the small beam model is no longer needed.
5. **Calculate the projection dose**: The dose deposited in the phantom by the delivery of a single projection (\( D_p \)) is calculated in the following way: The pre-calculated beamlet
\((b_i^j(x, y, z))\) corresponding to the first leaf \((l = 1)\) and to the nominal opening of the jaws \((Y)\) is rotated (interpolation is needed) and shifted; according to the gantry angle \((\theta_p = \theta_l + 2\pi \cdot (p - 1)/P_r)\) and the longitudinal position of the phantom \((y_p = Y_l + Y_r \cdot (p - 1)/P_r)\) corresponding to the projection and then weighted by its opening time \((O_1 \cdot \theta_l)\). Then the process is repeated for the other 63 leaves and the distributions are added.

\[
D_p(x, y, z) = \sum_{l=1}^{64} O_{1,p} \cdot b_i^j(x', y', z'),
\]

where \(b_i^j(x', y', z')\) corresponds to the shifted and rotated beamlet distribution.

6 Final dose distribution: The final dose distribution \((D_f)\) is obtained by superposing the distributions calculated for each projection of the treatment as

\[
D_f(x, y, z) = \sum_{p=1}^{P_f} D_p(x, y, z),
\]

where \(P_f\) is the total number of projections of the sinogram.

2.2 Test of the tool

The dose verification tool was evaluated in the static and dynamic modes by comparing measured and calculated values.

2.2.1 Static tests

Using the static mode of the system, measurements were performed in a water tank (Model MP3, PTW Freiburg, Germany) using a photon diode detector (diode type p, Model 6008, PTW Freiburg, Germany) at 15 mm depth with a SSD of 85 cm, located with its axis perpendicular to the central axis of the beam. This detector was selected due to its favourable characteristics in terms of size \((V=0.03 \text{ mm}^3)\) and less than 1% energy dependence in sensitivity [10]. It should be noticed that as the TomoTherapy system uses a flattening filter free linac, the off-axis spectral variation is less than 5% [11].

The tests consisted in the comparison of measured and calculated OFs, as well as transverse and longitudinal dose profiles for different MLC configurations, using a fixed jaw setting of 5 cm.

The results were obtained at symmetric static field settings in the X direction from 1.25 cm (2 leaves opened) to 40 cm (64 leaves opened). In addition to the rectangular fields, irregular MLC patterns (e.g. every other pair of leaves was set to be open) were used to compare measured and calculated dose profiles.

Figure 2 shows the sinogram used for the measurement of OFs and planar dose profiles for different field openings in the X direction. Initially, the sinogram has 30 projections where all the leaves remain closed to provide sufficient time for the stabilization of the beam output before starting the measurements. Then, for every 10 projections, an additional amount of leaves were opened symmetrically until all leaves were open. A different projection time was used for each set of measurements, as the profile scan required a longer time than the measurement of OFs.

2.2.2 Dynamic tests

The dose verification tool was tested in the three dynamic modes (topographic, rotational and helical) for a simple MLC/jaws configuration, i.e. a 2.5x2.5 cm\(^2\) square field size projected at the isocenter. The different treatment parameters used for the evaluation of each mode can be seen in Table 1. The integral dose was calculated in the virtual cylindrical phantom (Figure 1(a)) and delivered to the “cheese”-phantom (Figure 1(b)) using the same parameters and the same MLC configuration. The phantom, provided by TomoTherapy (Accuray Inc., Sunnyvale, USA), is a cylinder of solid water that can be divided in two halves and has holes for locating ionization chambers. The dose was measured with a film \((10" \times 12" \text{ EDR2 Ready Pack, Kodak, Rochester, NY USA})\) positioned at the coronal plane that divides the “cheese”-phantom into two equal halves. The irradiated films were then developed and scanned for later analysis.

### Table 1

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Unit</th>
<th>Topographic(*)</th>
<th>Rotational</th>
<th>Helical</th>
</tr>
</thead>
<tbody>
<tr>
<td>(\theta_l)</td>
<td>°</td>
<td>0</td>
<td>0</td>
<td>0</td>
</tr>
<tr>
<td>(Y_l)</td>
<td>cm</td>
<td>-5</td>
<td>0</td>
<td>-5</td>
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<tr>
<td>(T)</td>
<td>s</td>
<td>10</td>
<td>10</td>
<td>10</td>
</tr>
<tr>
<td>(P_r)</td>
<td>-</td>
<td>10</td>
<td>10</td>
<td>10</td>
</tr>
<tr>
<td>(Y_r)</td>
<td>cm</td>
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<td>1</td>
<td>1</td>
</tr>
<tr>
<td>(P_f)</td>
<td>-</td>
<td>100</td>
<td>20</td>
<td>100</td>
</tr>
</tbody>
</table>
Table 2
Systematic and random temporal variations introduced in the parameters of the treatment.

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Systematic</th>
<th>Random</th>
</tr>
</thead>
<tbody>
<tr>
<td>Output (O)</td>
<td>(O(t) = O)</td>
<td>(\pm 0.02 \cdot O)</td>
</tr>
<tr>
<td>Couch velocity (V)</td>
<td>(V(t) = 0.98 \cdot V + 2 \cdot 10^{-4} \cdot t) if (t \leq 100) s</td>
<td>(\pm 0.04 \cdot V)</td>
</tr>
<tr>
<td></td>
<td>(V(t) = 1.02 \cdot V - 2 \cdot 10^{-4} \cdot (t - 100)) if (t &gt; 100) s</td>
<td></td>
</tr>
<tr>
<td>Gantry speed ((\omega))</td>
<td>(\omega(t) = 0.02 \cdot \omega \cdot \sin (t \cdot \frac{2\pi}{100}) + \omega)</td>
<td>(\pm 0.04 \cdot \omega)</td>
</tr>
</tbody>
</table>

2.3 Parameter variations

It has been suggested, as part of a QA program of the TomoTherapy system, to check whether the output variation remains within \(\pm 2\%\) of the long time average \([12,13]\). However, it was found that this was not always accomplished, as seen in Figure 8. Motivated by this fact, the effects of single parameter variations in the total dose for a simple helical case were studied.

In the case of angular variations in gantry rotational speed the effects on the delivered dose blur out in helical delivery and a looser tolerance is allowed \([14]\). Therefore, a random variation of \(4\%\) on gantry rotational speed was chosen. It has been shown \([15]\) that a \(5\%\) couch speed variation produces a deviation of the longitudinal radiation field symmetry of more than \(2\%\). For this reason a random variation of \(4\%\) on couch velocity was used.

The dose delivered to the virtual phantom was calculated in a simple reference case using the helical mode. The treatment parameters were: \(2.5\) cm jaw opening, 4 leaves open, \(20\) s gantry period, \(50\) projections per rotation, \(1\) cm displacement of the couch per gantry rotation, \(0^\circ\) initial gantry angle, \(5\) cm displacement of the phantom’s position prior to the treatment, and a total number of projections of 500. This was used as the reference case, in which the output(O), gantry speed (\(\omega\)) and couch velocity(V), were assumed to be constant during the treatment. Then, the calculation was repeated, but this time random and systematic temporal variations of the parameters were introduced. Random variations were added to all the studied parameters, while systematic sinusoidal and linear variations in time (t) of the gantry speed and couch velocity, respectively, were also introduced. For the velocity, the couch started at \(2\%\) lower than planned and accelerated during the first half of the treatment time while decelerating during the second half. This was done in order to meet the planned position at the end of the treatment while maintaining the couch absolute position below the tolerance of \(1\) mm throughout the delivery process. In the case of the gantry speed a systematic sinusoidal variation according to the position of the linac was introduced, assuming that the velocity would increase while the linac is descending and decrease in the opposite direction. The individual variations are displayed in Table 2. Finally, profiles in the longitudinal and transverse directions of the reference and perturbed distributions were compared in the coronal plane of the virtual cylindrical phantom (shown in Figure 1(a)) in the X and Y axis.

3 Results

3.1 Static mode

Figure 3 shows the comparison of measured and calculated profiles in the X direction for different MLC openings. The jaws projected a field size in the Y direction of \(5\) cm to the isocenter plane for every evaluated field. The extended source, determined with the method described in 2.1, which was used for these calculations, was found to have radial symmetry and a FWHM of \(0.72 \pm 0.05\) mm.

Profiles are found to be in good agreement with the values measured with a diode. All studied cases pass a globally-defined gamma analysis with acceptance criteria of \(2\%/2\) mm. However, small differences were observed. For example, the maximum dose was underestimated by the calculation method in the smallest field size evaluated (\(1.25\) cm in the X direction). Mild disagreement is also found in the low-dose tail of the largest field size (\(20\) cm in the X direction), for which the measured dose was slightly underestimated.

The differences between measured and modelled OFs (Figure 4) remain within \(\pm 2\%\) of the normalization value taken at the reference field size of \(10\times5\) cm\(^2\). The maximum deviation was \(-2.0\%\), corresponding to the smallest field size evaluated equal to \(1.25\) cm in the X direction. The differences between modelled and measured OFs for the rest of
the evaluated fields remain below 1%. almost no difference in OFs was observed for field sizes larger than 5x5 cm².

Planar dose profiles using the static mode of the system were evaluated in both directions (X and Y) for simple symmetric rectangular fields. Figure 5 shows the performance of the tool modelling of dose profiles in the main axes of the plane perpendicular to the central axis (CAX). The relative intensity of the profiles is displayed on a logarithmic scale. From this figure, it can be seen that the shape of the profile penumbras and low-dose tails are not the same. The tool is able to correctly predict the profile shapes to within ∼0.5% of the maximum dose, well within the gamma acceptance criteria.

Good agreement (gamma passes at 2%/2mm) between the measured and calculated profile can also be seen in Figure 6. The maximum dose under each pair of open leaves was predicted very accurately, however, a slight overestimation of 1.5% was observed for positions where the beam was blocked by the closed leaves. There, the dose was slightly overestimated by the calculation tool. Also, it can be seen that certain penumbra regions within the pattern (specially on the right side with respect to the CAX) appear to be “shifted” from the measured positions with a maximum shift corresponding to ∼2 mm. It is not clear if these discrepancies are a result of detector effects, interpolation and voxel averaging issues or inaccuracies related to the MLC position and shape.

3.2 Dynamic mode

Figure 7 shows the results of the dynamic tests. All the evaluated cases show good agreement (gamma passes at 2%/2mm) with the measured profiles in the longitudinal direction. The distributions were normalized to the dose at the center of the phantom. It can be seen that the measured profiles are not smooth in the “plateau” region, but show some discrepancies from the calculated results. Relative 2-D dose distributions in the transverse plane of the virtual phantom for the three dynamic modes show some artefacts.

3.3 Parameter variations

In Section 2.3 the specific variations introduced to the beam output, the speed of the gantry and the couch velocity are given. These can also be visualized on the left side of Figure 9. On the right side of the figure, the result of introducing these variations can be seen. Figure 9(b) shows that a random variation of ±2% in the output of the beam throughout the treatment does not have considerable influence on the calculated dose. Similar results are obtained by considering an uncertainty in the gantry speed of 4% of the nominal value and a sinusoidal systematic variation with a 2% amplitude from the nominal value. In this case the discrepancies on the transversal axis are greater than for the output variation, nevertheless, they still remain within tolerance
levels. The couch speed variation, corresponding to a systematic variation (accelerated movement in the first half of the treatment and deceleration of the couch for the second half) plus a random component (±4% of the nominal value), shows a larger impact on the final outcome of the treatment when compared to the other studied parameters, particularly in the transverse direction. The discrepancies in this case still pass the gamma analysis with acceptance criteria of 3%/3mm for doses higher than 20% of the maximum. Larger deviations were observed in the low dose region for the transverse profile but passed the gamma analysis with a tolerance of 4%/4mm.
study that the penumbras in the $\pm 2\sigma$ or more of the beam parameters throughout the treatment. The parameters, exhibiting the effect of temporal variations in one of the “plateau” regions were not flat as expected for constant profiles for these simple configurations, one can notice that configurations.

4 Discussion and conclusions

After comparing measured and calculated OFs and dose profiles for different field configurations in the static and dynamic delivery modes, it was found that they were within the acceptance criteria of 2%/2mm globally defined.

Most of the small discrepancies between measured and calculated profiles and OFs in the static mode (see Figures 3 and 4) can be associated with interpolation, convolution or voxel averaging effects (the calculation grid was of 1x1x1 mm$^3$). Nevertheless, the influence of other effects related to the MLC hardware (i.e., the exact shape and position of the leaves) could not be discarded as the approximation of perfect collimators was used.

Figure 5 shows that the penumbras in the $X$ and $Y$ directions are well modelled by the combination of extended source and PBK effects. Furthermore, the inclusion of the leakage contribution in the $X$ direction was shown to be sufficient to model the “umbra” of the distributions down to $\sim 1\%$ of the maximum dose. The fact that the agreement in the $Y$ direction is very good, is based on the fact that the jaws that define the field size in the longitudinal direction were made of tungsten of up to 23 cm in thickness and, therefore, the contributions due to the transmission of photons through the collimators in this direction was negligible. As seen in Figure 6, the tool was also found to be well suited to model dose distributions in asymmetric and complex field configurations.

From the dynamic tests, it was found that the verification tool was able to reproduce measured profiles within 2%/2mm for simple beam configurations. By looking at the measured profiles for these simple configurations, one can notice that the “plateau” regions were not flat as expected for constant parameters, exhibiting the effect of temporal variations in one or more of the beam parameters throughout the treatment. The fact that the couch speed was set to 1 mm/s (fast in comparison with clinically used values: 0.3-0.5 mm/s) contributes to enhance the dosimetric effect (plateau variations) of the parameter variations. The artifacts in the relative 2-D dose distributions can be attributed to interpolation and grid size effects.

The dose verification tool is not only useful as a dose verification system, but can also be used to study more deeply this dynamic delivery technique and identify the individual role of the different beam parameters in the final outcome of the treatment, as the overall performance of the system depends on the performance and synchrony of the MLC, gantry, couch and linac.

Only the couch velocity variation introduced (see Figure 9) yielded deviations greater than 3%/3mm from a reference helical distribution (for which the parameters were assumed to be constant during the complete treatment), in the low dose region of the transversal profile. Random variations of the beam output (of $\pm 2\%$) had almost no influence in the total dose (with a maximum observed deviation of 0.7% from the reference value). The effect of output variations (random and systematic) in the total dose for different clinical cases has been described in the literature [16,17]. The cancellation of systematic uncertainties and averaging of random variations throughout the treatment, made the recommended $[12] \pm 2\%$ threshold for beam output variation too conservative. These conclusions are in agreement with the presented results. The same was found for the rotational speed variation.

The developed tool offers a very useful approach to study the TomoTherapy delivery system and to test its performance using the different available modes (static, topographic, rotational and helical). Therefore, it can be used as a supplemental dosimetric QA procedure for the system. For other TomoTherapy users the source model and PBK should be re-commissioned according to the procedures described by Caprile and Hartmann in 2009 [8]. It was possible to visualize the calculated distributions not only in one and two-dimensional representations (as shown in Figure 7), but also in three-dimensional intensity maps and isosurfaces. The availability of this feature is indeed desirable in certain cases (e.g., the use of isosurfaces to study and identify variations in the “thread” effect by using different pitch values).

The performance of this tool was successfully evaluated in the different modes of the TomoTherapy system. This dose calculation tool was able to predict accurately OFs and planar dose distributions in water from very small to large field sizes and, therefore, this procedure can be implemented as a supplemental check for the already established dosimetric QA program of the system, providing an independent dose calculation tool. Nevertheless, the approach developed in this investigation still lacks more flexibility regarding phantom positioning and composition, which can be addressed in future studies.
Figure 9. Effect of random and systematic parameter variations on a test helical case. Left: variations introduced for individual parameters, according to Table 2, for the output (a), couch velocity (c) and gantry speed (e). Right: Comparison between calculated coronal profiles $X$ (red) and $Y$ (blue), with (dashed line) and without (continuous line) considering the variation in output (b), couch velocity (d) and gantry speed (f).
Acknowledgements

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References