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Comparison of conventional and Monte Carlo dose calculations for prostate treatments

D. Fraser¹, C. Mark¹, F. Cury², A. Chang¹, F. Verhaegen¹
¹Medical Physics Unit, McGill University, Montréal, Canada
²Radiation Oncology Department, Montreal General Hospital, Montréal, Canada
fverhaegen@medphys.mcgill.ca

Abstract. Monte Carlo (MC) calculations are rapidly finding their place in clinical dose assessments. We investigated conformal prostate dose distributions as calculated by MC, and compared them to several analytical dose calculations. The treatment distributions for twenty prostate cancer patients, treated with 18 MV 3D conformal radiation therapy, were retrospectively assessed. The BEAM code based on EGSnrc was used to model the beam from which phase space files were used as input into the XVMC algorithm. This was compared to conventional treatment planning system calculations (CADPLAN) with and without inhomogeneity corrections. Results indicate that the CADPLAN generalized Batho Power Law, modified Batho Power Law, and equivalent tissue-air ratio methods contain inaccuracies in calculated dose to 95% of the prostate planning target volume of 3.5%, 3.3%, and 2.9%, respectively. The greatest discrepancies in the organs at risk were seen in the bladder where the inhomogeneity correction methods all predicted that 50% of the prescribed dose covered an average of 8.2% more of the bladder volume than that predicted from the MC calculation. Water equivalent MC and water equivalent CADPLAN calculations revealed important discrepancies on the same order as those between heterogeneous MC and heterogeneous CADPLAN calculations. The data indicate that the effect of inhomogeneities is greater in the target volume than the organs at risk, and that accurately modeling the dose deposition process is important for each patient geometry, and may have a greater impact on the dose distribution in the prostate region than correcting an analytical algorithm for the presence of inhomogeneities.

1. Introduction
The trend towards dose escalation in smaller target volumes has increased the drive for greater accuracy in dose calculations. Dose coverage is affected by tissue inhomogeneity not only in the planning target volume (PTV) but also in nearby organs at risk (OAR). The conventional method to predict the dose to a patient is to assume a homogeneous water medium. Inhomogeneous objects are then accounted for analytically by treating inhomogeneities as a perturbation of the dose under the same beam conditions in a homogeneous water phantom. More recently, particle transport methods have been incorporated into treatment planning systems (TPS) either explicitly, or through convolution kernels. Accounting for tissue differences with improved algorithms reduces the uncertainty in absolute dose.
In this study dose calculations with four analytical algorithms from a commercial TPS, and two Monte Carlo (MC) methods, are compared for twenty external beam 3D conformal radiation therapy (3D-CRT) treatment plans for prostate cancer.

2. Methods

2.1. Patient Planning

3D-CRT treatment plans for twenty prostate cancer patients were generated using an 18 MV coplanar five field beam geometry: one anterior field (gantry at 0°); two parallel opposed lateral fields (gantry at 90°, 270°); and two posterior oblique fields (gantry at 110°, 250°). Each field was shaped with a 52 leaf collimator, and depending on the patient, the use of 15°-30° physical wedges in the oblique fields were used to protect the femoral heads. The patient model was determined from CT images, and the plans were manually optimized with an analytical algorithm based on a water equivalent patient without heterogeneities. The PTV was defined from the CT image as the prostate volume plus a 7 mm margin.

2.2. Retrospective Dose Calculations

Analytical dose calculations were performed with CADPLAN v6.0 (Varian Medical Systems Inc., Palo Alto, CA). CADPLAN’s double pencil beam model (PB-wat), derived from broad beam data, is used to calculate dose distributions in water equivalent material. Inhomogeneities are considered via three common correction-based methods: the Batho Power Law (Batho); the modified Batho Power Law (MBatho); and the equivalent tissue-air ratio (ETAR). Batho is an empirical correction factor method that uses tissue-maximum ratios (TMR) for high energy beams, raised to a power that depends on the medium’s electron density relative to water. It was originally developed for dose calculations in water below a single slab of lung tissue [1]. MBatho differs in its definition of depth. In high energy photon beams the build-up region can be several centimeters thick in which the TMR values are not valid. The modified method uses only the descending part of the TMR curves by adding the depth of maximum dose to the depth used in the previous generalized Batho method. ETAR scales the depth and radius of the TAR(z, A) derived in a unit density water medium according to the relative effective electron density (along the primary ray path) of the inhomogeneous medium [2]. The treatment plan (beam geometry and monitor units) was developed for the homogeneous water equivalent material scenario, PB-wat. The second, third, and fourth calculations multiply the dose value from the PB-wat method with an inhomogeneity correction factor determined using the Batho, MBatho, and ETAR methods, respectively.

Monte Carlo radiation transport techniques use numerical methods to model the physical processes which govern interactions between radiation and their environment. These processes are described by probability distributions so that a first-principles approach is used. XVMC [3,4] is a fast 3D photon MC code based on the Voxel Monte Carlo [5] algorithm originally developed for electron beams. XVMC uses several simplifications and approximations to increase computational efficiency in the range of energies and materials encountered in radiation therapy. These include: (1) a fast electron transport algorithm and a fast ray tracing technique; and (2) the dose from low energy scattered photons and bremsstrahlung photons produced in the phantom matrix is estimated with a kerma approximation. The EGSnrc/BEAM [6] code was used to model a Clinac 2300 linear accelerator, and score phase space files for each beam in the treatment plan. The beam model used a 17.8 MeV electron pencil beam of radius 1.0 mm, and has previously been validated such that off axis ratios for a 10x10 cm² matched measurements at the depth of maximum dose and at 10 cm within 1 % [7]. We also verified the beam model against the PB-wat algorithm for simple open field geometries, and only showed differences that were greater than the MC statistical error in the penumbra and build up regions. It is expected that the TPS’ multileaf collimator model will further increase these differences with MC. The dose distribution for each beam was calculated with XVMC using a maximum
statistical error of 2 %, for an average simulation time of 60 minutes (five fields) on a single 2.7 GHz 64 bit AMD processor. XVMC distributions were calculated based on inhomogeneous CT voxel material and density assignments (XVMC-het), as well as assuming a homogeneous water equivalent patient (XVMC-wat). In-house software was used to calculate dose volume histograms for all dose calculations [8]. The absolute dose resulting from the XVMC simulations was calculated by relating the dose per incident particle to the dose per monitor unit (MU) from the linear accelerator calibration, and then multiplying by the number of MUs as specified in the plan. A MC simulation under calibration conditions (10x10 cm², 100 cm SSD, water phantom) was performed and the dose per incident particle at the depth of maximum dose was related to the calibration value of 1.00 cGy/MU in water. This approach ignores the effect of backscatter towards the monitor ion chamber. However, it has been shown that for field sizes larger than 5x5cm² such as the ones used in this research, this effect is negligible [9].

Analysis of twenty patients is performed with a comparison of dose-volume based indices. The dose (cGy) covering 95 % of the PTV volume, D 95%PTV, is used for the target, while in order to represent the shallow dose gradient regions in the OARs – rectum and bladder, the calculated doses are compared via the volume (cm³) receiving 50% of the prescribed dose, V 50%PD.

3. Results

Figure 1 is a graph of the difference in D 95%PTV between algorithms. The values are normalized to the prescribed dose (either 6600 cGy or 7200 cGy). The average and one sigma standard deviations of the differences are also indicated. The values of ΔD 95%PTV calculated by CADPLAN are within 8.6 % with an average value of 3.5 %, and those calculated by XVMC-wat are within 2.4 % with an average value of 1.7 %. All calculations predicted a higher dose to the PTV than the full MC calculation that considered inhomogeneities. It has been shown that in cases where the electron density of an overlying inhomogeneous layer is greater than that of water the power-law method over estimates the dose [10]. Additionally, this method assumes semi-infinite slab geometry. In the treatment plans assessed, the two lateral and/or posterior oblique beams reach the PTV usually after passing through a portion of the femurs. These have finite dimension, are generally only partially in the field, and have a greater electron density than water. The Batho methods are therefore not ideally suited for this treatment plan geometry.

The average differences in D 95%PTV between PB-wat and Batho, MBatho, and ETAR are 0.9 %, 1.1 %, and 1.4 %, respectively. The average difference in D 95%PTV between XVMC-wat and XVMC-het is 1.7 %. The difference between analytical algorithms is of the same order as that between MC algorithms. CADPLAN’s Batho algorithms average the electron density over each 1 cm interval between the dose point of interest and the source, thereby lowering the effective electron density of any bone intersecting the beam path. Similarly, the ETAR algorithm scales the depth by averaging the electron density along the volume elements in a direct path from the source to the point of interest, and scales the radius by multiplying the beam’s equivalent circular radius by a weighted average of the electron density along that path. The weighting is determined from the difference between the scatter air ratio values at the point of interest and that at each point along the beam path. These methods do not incorporate inhomogeneities from adjacent horizontal volume elements, and reduce the effects of relatively small inhomogeneities in otherwise water-similar media. Figures 2.a and 2.b provide an example of one of the larger roles inhomogeneities had on the PTV in our data set (patient 13). The difference in D 95%PTV between XVMC-het and XVMC-wat is 1.7 %, and between MBatho and PB-wat is 1.0 %. However, the MC data differs from the analytical algorithms by approximately 6 % in the PTV. This substantial difference is seen in all organs for patient 13 in Figure 2.a, as well as in the PTV for all patients in Figure 1. The maximum value of ΔD 95%PTV for the analytical inhomogeneity corrections in Figure 1 is 7.8 % (Batho) with an average of 3.5 % (Batho).
\[ \Delta D_{95\% \text{PTV}} = (D_{95\% \text{PTV}})_{\text{algorithm}} - (D_{95\% \text{PTV}})_{\text{XVMC-het}} \]

Patient 13: Dose Volume Histograms

Patient 13: PTV Dose Volume Histograms

Patient 2: Rectum Dose Volume Histograms

50\% of the prescribed dose

Patient 18: Bladder Dose Volume Histograms

50\% of the prescribed dose

Figure 1: Percentage difference in dose covering 95\% of the PTV volume, \( \Delta D_{95\% \text{PTV}} \), normalized to the prescribed dose.

Figure 2: Cumulative DVHs for a) all organs and b) the PTV for patient 13, c) the rectum for patient 2 where inhomogeneities had the largest impact on the rectum distribution, and d) the bladder for patient 18, which illustrates a typical case where the calculation method impacted the distribution.
Figures 3 and 4 are graphs of the difference between algorithms in $V_{50\%PD}$ for the rectum and bladder. The values are normalized to the total organ volume. The majority of $\Delta V_{50\%PD}$ for the rectum are negative, indicating that XVMC-het predicted dose coverage over a larger volume. For the bladder all $\Delta V_{50\%PD}$ are positive, indicating that XVMC-het predicted dose coverage over a smaller volume.

Among the four analytical calculations, the average and range of $\Delta V_{50\%PD}$ for each of the rectum and bladder separately are essentially the same. In MC simulations, the $\Delta V_{50\%PD}$ for XVMC-wat is generally very small and has an average value of -0.1 % for the rectum and 0.4 % for the bladder. These indicate that the effect of inhomogeneities on the dose to the rectum and bladder may be smaller than to the PTV. This may be explained from the inadequate lateral scattering model in the analytical calculations such that the dose to the rectum and bladder, which are tangential to four out of five beams, remain relatively the same. Similar results were found by Yang et al. [11] for 10 MV photons in a coplanar geometry when comparing pencil beam homogeneous and inhomogeneous algorithms, and by du Plessis et al. [12] where Batho and ETAR methods were accurate within 3 % of MC simulations for a prostate model. Yang et al. also found that MC and pencil beam algorithms predicted doses to the CTV within 3 % of each other, which corresponds with the results in Figure 1. Figure 2.c graphs the cumulative DVH of the rectum for patient 2 (and shows $V_{50\%PD}$) who was an exception with a relatively large (-4.85 %) discrepancy between XVMC-wat and XVMC-het. In the XVMC-wat simulation for this patient, the 50 % isodose line passes through the rectum, but the CT image showed that the rectum was filled with gas. Because the rectum lies in the descending portion of the depth dose curve, the presence of a gas cavity increases the penetration depth of the beam allowing for a greater rectum volume to be encompassed by the 50 % isodose line in the XVMC-het simulation. Figure 2.d graphs the cumulative DVH of the bladder for patient 18. This is a typical case in the data set demonstrating the small differences between homogeneous and inhomogeneous CADPLAN calculations, and the large differences between MC and analytical calculations.

Figure 3: Difference between rectum volumes contained within the 50 % prescribed isodose line. The volumes are normalized to the total organ volume.
Prostate sites are similar to water due to the occurrence of mainly soft tissue, despite the presence of bone in the femoral heads. Although there is a small 1.7% difference in $D_{95\%PTV}$ for the PTV, and essentially no difference in $V_{50\%PD}$ for the OARs when inhomogeneities are considered, there is consistently a large difference between analytical and MC methods, indicating that the dose model has a greater impact than inhomogeneity corrections for the beam geometry used. In addition, because of the large range in $V_{50\%PD}$ for both OARs and $D_{95\%PTV}$ for the PTV, the impact of the dose algorithm can only be approximated from averages, but the inaccuracies of analytical methods should be considered on a patient by patient basis from more accurate techniques.

\[
\Delta V_{50\%PD} = (V_{50\%PD})_{\text{algorithm}} - (V_{50\%PD})_{XVMC-het}
\]

Figure 4: Difference between bladder volumes contained within the 50% prescribed isodose line. The volumes are normalized to the total organ volume.

4. Conclusion:
The dose distributions of six dose calculation algorithms were compared in twenty prostate treatment plans. Of the three inhomogeneity correction techniques the ETAR correction factor most closely matched the MC dose calculations. Both analytical and MC inhomogeneity correction algorithms had a greater impact on the PTV than the OARs, when compared with their respective water equivalent calculation. Simulating the patient as a homogeneous water equivalent medium in MC calculations only marginally better matched the water equivalent analytical calculation, indicating that for this cohort of patients and beam geometry, the beam model and dose calculation method have a greater impact than do inhomogeneities. The standard deviations and ranges displayed in Figures 1, 3, and 4 are testament that the differences in dose calculations between MC and analytical algorithms are unpredictable and highly influenced by patient geometry to the point that MC dose calculations for this type of treatment are recommended.

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