RADIOThERAPY DOSE CALCULATIONS IN THE PRESENCE OF HIP PROSTHeses

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Abstract—The high density and atomic number of hip prostheses for patients undergoing pelvic radiotherapy challenge our ability to accurately calculate dose. A new clinical dose calculation algorithm, Monte Carlo, will allow accurate calculation of the radiation transport both within and beyond hip prostheses. The aim of this research was to investigate, for both phantom and patient geometries, the capability of various dose calculation algorithms to yield accurate treatment plans. Dose distributions in phantom and patient geometries with high atomic number prostheses were calculated using Monte Carlo, superposition, pencil beam, and no-heterogeneity correction algorithms. The phantom dose distributions were analyzed by depth dose and dose profile curves. The patient dose distributions were analyzed by isodose curves, dose-volume histograms (DVHs) and tumor control probability/normal tissue complication probability (TCP/NTCP) calculations. Monte Carlo calculations predicted the dose enhancement and reduction at the proximal and distal prosthesis interfaces respectively, whereas superposition and pencil beam calculations did not. However, further from the prosthesis, the differences between the dose calculation algorithms diminished. Treatment plans calculated with superposition showed similar isodose curves, DVHs, and TCP/NTCP as the Monte Carlo plans, except in the bladder, where Monte Carlo predicted a slightly lower dose. Treatment plans calculated with either the pencil beam method or with no heterogeneity correction differed significantly from the Monte Carlo plans. © 2003 American Association of Medical Dosimetrists.

Key Words: Dose calculation, Hip prosthesis, Monte Carlo, Superposition, Pencil beam.

INTRODUCTION

The high density and atomic number of hip prostheses relative to water yield challenges for radiotherapy dose calculation when beams pass through these structures. Several authors have quantitatively calculated or measured the effects of such prostheses or high atomic number interfaces on dose distributions.1–12 This topic has created enough attention to warrant the formation of an American Association of Physicists in Medicine (AAPM) Task Group to examine the subject of management of patients with inserted high-Z materials.13 Sauer12 performed Monte Carlo and superposition calculations for phantom geometries; however, a comparison of both simple and advanced dose calculation algorithms in a patient computed tomography (CT) geometry has not been performed.

Most hip prostheses are made from cobalt-chrome alloys,14 because they are considered to have the best combination of corrosion resistance, fatigue resistance, and mechanical strength.15 However, both titanium and stainless steel hip prostheses are also available.1 The prostheses vary in size and may have solid or hollow femoral heads.1 Hollow prostheses tend to be larger. Only solid prostheses were considered here.

The aim of this work was to compare dose distributions in the vicinity of hip prosthesis materials in both phantom and patient geometries as calculated by the Monte Carlo, superposition, and pencil beam algorithms. The clinical significance of the dose distribution differences in the patient geometry was investigated.

CT imaging of hip prostheses causes streak artifacts due to aliasing.16–19 These artifacts reduce the diagnostic quality of the images and provide erroneous density information needed for dose calculation. However, the effects of image reconstruction on target delineation and dose calculation are beyond the scope of this article.

METHODS AND MATERIALS

Calculations using various dose calculation algorithms were performed in phantom and patient geometries containing high atomic number and density materials. The phantom calculations were performed to study the dosimetric effects of the prostheses under well-defined conditions. The patient calculations were performed to study the clinical significance of the resultant dose distributions obtained under realistic conditions. Three dose calculation algorithms were used: Monte Carlo, superposition, and pencil beam. For reference,
Dose calculation methods

The Monte Carlo method was chosen because of its recognition as the most accurate dose calculation method available. The superposition method was selected, because it is the most accurate broadly available algorithm, and the pencil beam method was selected because of its status as one of the most widely used correction methods. The Monte Carlo code used was EGS420 with usercodes BEAM21 and DOSXYZ.22 The energy cutoffs were AE = ECUT = 700 keV and AP = PCUT = 10 keV. Validation of the Monte Carlo codes with both other Monte Carlo results and experimental results has been performed.23,24 To allow comparison between Monte Carlo and other dose calculation algorithms, the Monte Carlo dose-to-material values (except in the prostheses itself) were converted to dose-to-water using the method of Siebers et al.25 The superposition calculations used Pinnacle’s (ADAC Laboratories, Milpitas, CA) collapsed-cone convolution24 algorithm. The physical density of the prosthesis (rather than the relative electron density) was used as specified in the Pinnacle3 Physics Guide. The pencil beam code was developed in-house, based on that of Mohan and Chui.26 The pencil beam algorithm physics employed is similar to that of the ratio of tissue-air-ratios (RTAR) method described in Metcalfe et al.27

Phantom calculations

A 2 × 2 × 2-cm³ block of iron (ρ = 8.0, Z = 26) or titanium (ρ = 4.5, Z = 22) centered on the beam axis at 4-cm depth was inserted into a 40 × 40 × 40-cm³ water phantom by adding CT numbers corresponding to the correct density in the CT image set. (The CT number was determined using the material density and the CT-to-density conversion table.) The phantom is shown in Fig. 1. Because of comparable densities and effective atomic numbers, the calculations with the iron represent those of cobalt-chrome prostheses and yield similar dosimetric results. The voxel size for the calculations was 0.4 × 0.4 × 0.2 cm³.

A 6-MV 100 cm SSD 10 × 10-cm² field was incident on the phantom. Sibata et al.1 showed that the perturbations of the high-Z prostheses were slightly more pronounced for 6 MV than 18 MV; thus, 6-MV calculations are presented here.

From the dose distributions calculated, central-axis depth dose curves and profiles through the center of metal block (5-cm depth) were extracted.

Patient calculations

Due to the CT image reconstruction limitations for the high-density and atomic number prosthesis materials described above, it was not possible to use an artifact-free CT image set of a patient with a hip prosthesis. Thus, a 28-mm diameter sphere with a density of 8.0 g cm⁻³, representing the head of a typical hip prosthesis,14 was inserted into the right femoral head of an existing CT image set of a patient, replacing the normal hip. The subsequent dose calculations used the existing treatment plan for the patient, including the same number of monitor units for each field. This treatment plan was composed of 18-MV isocentric fields in the standard 4-field box configuration. The voxel size for the calculations was 0.4 × 0.4 × 0.4 cm³.

The analysis of the dose distributions obtained by the various algorithms used isodose distribution comparison, DVH comparison, and radiobiological indice calculation. The TCP calculation method is described elsewhere28 using a γ₅₀ value (the percent increase in TCP per percent increase in dose at TCP = 50%) of 1.0 for a prostate tumor.29 The NTCPs for the rectum and prostate were calculated using the Lyman model.30 values tabulated in Burman et al.,31 and data taken from Emami et al.32

RESULTS

Phantom calculations

Depth dose curves for the phantom geometry using the different calculation methodologies are shown in Fig. 2. Neither the superposition nor pencil beam methods predict the increase in electron backscattering (and to a lesser extent, photon backscattering) from the high atomic number material. Note that the magnitude of this backscattering is dependent on the beam energy and the prosthesis composition and density. The finite voxel size underestimates the true interface dose, because converting a CT grid to a dose calculation grid is an averaging process. Because of this, smoother density interfaces exist in the dose calculation grid than in the CT grid.
Fig. 2 shows that immediately beyond the prosthesis, the superposition and pencil beam methods predict a higher dose than that calculated by Monte Carlo. Further from the prosthesis, the differences between the algorithms decrease.

The increased mass scattering power of the prosthesis relative to that of water causes the dose increase at the proximal edge of the prosthesis. This increased lateral electron scattering within the prosthesis means that fewer electrons cross the distal prosthesis/water interface, causing the rebuild-up near the distal edge of the prosthesis. Neither the superposition nor the pencil beam method accounts for rebuild-up in electron fluence past the prosthesis. It is interesting to note that with the addition of an opposed field, the total magnitude of the dose enhancement will be reduced.

Monte Carlo is the only algorithm that explicitly accounts for the effect of different atomic composition of the prosthesis relative to water. The higher atomic number causes the photon spectrum to be attenuated more by photoelectric and pair production interactions and less by Compton interactions than an equivalent mass of water. The different interaction probabilities within the prosthesis mean that relatively fewer scattered photons per unit mass exit the prosthesis. From this phenomenon, we would expect superposition and pencil beam methods to predict a higher dose than Monte Carlo outside the prosthesis. The differing atomic number of the prosthesis with respect to water also changes the spectrum of the photons transmitted through the prosthesis, further affecting dose downstream of the prosthesis.

Lateral dose profile curves for the different calculation methodologies are shown in Fig. 3. Neither the superposition nor the pencil beam method predicts the increase in dose, due to an increased scattering of electrons in the metal and the resulting higher dose adjacent to the prosthesis.

Patient calculations

Isodose plans for the prosthesis geometry as calculated by the Monte Carlo, superposition, and pencil beam algorithms are shown in Fig. 4. Dosimetric differences in the vicinity of the prosthesis are seen. Also of note is the 60-Gy line through the rectum, which for the pencil beam calculation remains straight, whereas the Monte Carlo...
Carlo and superposition 50-Gy lines curve slightly inward through the rectum. This difference is due to the pencil beam algorithm’s inability to account for the electronic disequilibrium in the lower density media.

The difference in the 3D dose distributions can be represented by DVHs, as shown in Fig. 5. This figure shows that for the prostate, rectum, and bladder, the pencil beam calculations were consistently higher than the superposition calculations, which in turn were generally slightly higher than the Monte Carlo calculations. The higher pencil beam results for the prostate and bladder are consistent with the phantom calculation results. Pencil beam-predicted dose differences in the rectum are due to the lack of accounting for electronic disequilibrium in the lower density media rather than the effect of the prosthesis.

The differences in the 3D dose distributions can be quantitatively represented by a single value if predicted clinical outcome is used. The TCPs and NTCPs of the rectum and bladder are given in Table 1. This table shows that the TCPs calculated by the 3 algorithms are similar, although the NTCPs calculated by superposition are higher than those of Monte Carlo, and the pencil beam NTCPs are higher still. The predicted outcome values in Table 1 are not surprising, based on the DVHs of Fig. 5.

**DISCUSSION**

The accuracy of the dose calculation algorithms in the vicinity of hip prosthesis materials is proportional to their complexity of physical process modeling. The limitation of superposition at interfaces with different atomic numbers is observed near hip prostheses. An improvement could be the addition of kernels generated in different media as suggested by Papanikolaou and Mackie, or the corrections of Sauer. However, further than a few millimeters from the boundary, superposition results do not significantly differ from those of Monte Carlo.

The pencil beam method does not predict dose well near the interface. However, beyond the interface, the dose difference between pencil beam and Monte Carlo calculations decreases. These results are consistent with those of Sibata et al., who showed that the equivalent path length correction method yields acceptable dosimetric results at a distance of 5 cm or greater from the prosthesis.

The Monte Carlo, pencil beam, and superposition dose calculation methods were significantly more accurate than the no-heterogeneity correction method.

Dose distribution results for a patient plan with a hip prosthesis showed the pencil beam method calculating a higher dose than superposition. In general, the superposition dose was slightly higher than that of Monte Carlo.

It should be noted that the calculations performed here used high CT numbers (up to 8000). Often, the CT...
scanners assigned a maximum CT number (4096) that would artificially reduce the density of the prosthesis. Also, no streak artifacts (commonly seen on CT scans of hip prostheses) were present.

The results presented in this paper reinforce the concept that increasingly rigorous physics modeling in dose calculation algorithms will more accurately predict dose. Monte Carlo is the only radiotherapy dose calculation algorithm that can account for dose variations at interfaces with different atomic numbers. The Monte Carlo and superposition methods both accounted for the electronic disequilibrium present in the lower density rectum. The pencil beam method does not account for electronic disequilibrium. For the prostate case, the prosthesis was a sufficient distance from the target so that interface effects did not affect the dose to the target. However, this may be a problem for other sites, such as paraspinal treatments, where surgical rods have been inserted close to the vertebrae and tumor.

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REFERENCES