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Robotic stereotactic radioablation of breast tumors: Influence of beam size on the absorbed dose distributions

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HIGHLIGHTS

- Stereotactic body radiation therapy of breast tumors is analyzed using Monte Carlo simulation.
- The influence of beam collimation on the absorbed dose distributions is determined.
- Large field sizes increase target dose uniformity and midlevel doses to healthy structures.
- Skin dose is greatly affected by changes in beam collimation.

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ABSTRACT

Robotic stereotactic radioablation (RSR) therapy for breast tumors has been shown to be an effective treatment strategy when applied concomitantly with chemotherapy, with the purpose of reducing the tumor volume thus making it more amenable for breast conserving surgery. In this paper we used Monte Carlo simulation within a realistic patient model to determine the influence that the variation in beam collimation radius has on the resultant absorbed dose distributions for this type of treatment. Separate optimized plans were obtained for treatments using 300 circular beams with radii of 0.5 cm, 0.75 cm, 1.0 cm and 1.5 cm. Cumulative dose volume histograms were obtained for the gross, clinical and planning target volumes as well as for eight organs and structures at risk. Target coverage improves as the collimator size is increased, at the expense of increasing the volume of healthy tissue receiving mid-level absorbed doses. Interestingly, it is found that the maximum dose imparted to the skin is highly dependent on collimator size, while the dosimetry of other structures, such as both the ipsilateral and contralateral lung tissue are basically unaffected by a change in beam size.

1. Introduction

Robotic stereotactic radioablation (RSR) is a clinically-established technique that makes use of a plurality of radiation beams of small cross-section, usually less than 2 cm in radius, aimed at a target volume from tens to hundreds of different positions, while using a pair of orthogonal digital radiological images to monitor the position of the target (Adler et al., 1997). While initially used for the treatment of brain lesions, the technique has been gradually extended to other treatment sites including brain, prostate and lung tumors (Slotman et al., 2008). Recently, this modality has gained attention for the treatment of breast tumors in patients who, because of the size of their tumor, do not qualify for breast conserving surgery (Bondiau et al., 2009). A recent Phase I clinical trial was completed that determined the suitability of this treatment approach, concomitant with neo-adjuvant chemotherapy, as well as the recommended values for the prescription dose and tolerance limits for the healthy structures surrounding the tumor (Bondiau et al., 2013). The trial was designed as a 5-level dose escalation study with a total of 25 patients evenly distributed in each dose level, using a Cyberknife system (AccuRay Inc, Sunnyvale CA) as the treatment machine. The median tumor volume was 22 cm^3. The number of beams for each treatment ranged from 48 to 231 with an average of 136. In a previous publication the authors of the clinical trial reported a mean collimator diameter of 21 mm, ranging from 15 mm to 35 mm. The highest pathologic complete response (pCR) rate was reached at the 25.5 Gy dose level and a total of 23 patients were able to undergo breast conserving surgery after the treatment. As with other radiotherapy techniques, a thorough understanding of the dosimetric characteristics that result upon the irradiation of the breast and chest wall using beams with a relatively small cross section is needed in

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order to both fully exploit the advantages that may be offered, and equally important, to assess the potential disadvantages of said technique. Monte Carlo simulation as applied to the calculation of absorbed dose distributions has proven to be the gold standard in radiotherapy, particularly for those challenging cases in which significant tissue inhomogeneities exist, as is the case in the irradiation of thorax, where bone and lung tissue are in close proximity to the target volume (Vanderstraeten et al., 2006). Monte Carlo simulation is therefore an invaluable tool when trying to determine the characteristics of the absorbed dose distributions that result when particular treatment parameters, such as the beam collimation, are changed.

In this work, Monte Carlo simulation, a realistic patient model and an optimization algorithm are used to analyze the effect that different beam collimation radii have on the resultant absorbed dose distributions.

2. Materials and methods

2.1. Patient model

We used the Rensselaer Polytechnic Institute (RPI) female phantom (Zhang et al., 2009) with a digitally added tumor in the left breast as our patient model. A portion of this phantom, which has a resolution of 0.25 cm in each direction, was extracted at the thorax level. As we are working with fairly small radiation beams the phantom was up-sampled in order to produce a resolution of 0.125 cm in each direction. An ellipsoidal tumoral mass representing the gross target volume (GTV) was digitally added with its three major axes having dimensions of 1.5 cm, 1.5 cm and 1.0 cm. In order to generate the clinical and planning target volumes, CTV and PTV respectively, the guidelines established in a Phase I clinical trial (Bondiau et al., 2013) were followed, specifically: the CTV was formed by adding a 0.5 cm margin around the GTV whereas for the PTV a 0.25 cm margin was added to the CTV. Therefore, the volume of the GTV is 9.375 cm^3, that of the CTV is 25.125 cm^3, while the PTV has a volume of 37.125 cm^3. Ten different materials were used to represent the different tissues present in the phantom, including compact bone, soft tissue, striated muscle, adipose and glandular tissue, skeletal muscle, lung tissue, blood, and water to model the heart contents. The tumor material was modeled as consisting of soft tissue. The composition of each material was taken from ICRU (1989). It is assumed that there is air surrounding the phantom, so the electron contamination arising from the interaction of the x-ray beams on their way to the patient is fully taken into account.

2.2. Treatment setup

A total of 300 circular beams were used in each of the treatments presented in this work. Four beam radii were modeled, namely 0.5 cm, 0.75 cm, 1.0 cm and 1.5 cm. The x-ray beam is modeled as originating from a point source, with a source-to-axis distance of 80 cm, as in the Cyberknife system. The x-ray spectrum from a Varian 6 MV medical accelerator was taken from the literature (Garnica-Garza, 2008). The Monte Carlo-calculated $TPR_{20/10}$ for a 60 mm beam diameter at an SAD of 80 cm using this x-ray spectrum is 0.63. A separate software developed at our institution was used to determine the positions from which each of the beams were aimed at the tumor. The software tries to minimize beam overlap at the entrance while keeping the number of times each voxel in the target is "visited" by all the beams as uniform as possible (Garnica-Garza, 2013).

2.3. Monte Carlo simulation parameters

The Monte Carlo code PENELOPE (Salvat et al., 2006) and the auxiliary set of subroutines from the PENEASY suite (Sempau, 2006) were used to determine the absorbed dose distributions in the voxelized RPI phantom from each beam. In all the simulations, both the photon and electron cutoff energies, below which no transport takes place, were set at 10 keV. Transport parameters c1 and c2, mean free path between hard elastic collisions and the maximum fraction of energy spent by an electron in any given step respectively, were set at 0.1, as recommended by the developers of the code (Salvat et al., 2006). Enough histories were run in each simulation to keep the average uncertainty in those voxels receiving at least 50% of the maximum dose at or below the 1% level. As four different beam radii were modeled, a total of 1200 dose matrices were calculated.

2.4. Treatment plan optimization

The Cimmino iterative relaxation algorithm (Cimmino, 1938; Censor et al., 1988) was used to determine each beam weight, according to a user-defined set of treatment goals. The software was compiled using the PGI *pgf90* FORTRAN compiler (The Portland Group, Lake Oswego OR) and is capable of running in parallel using the open Multi-Processing (open-MP) protocol. Our implementation of this algorithm is discussed in detail elsewhere (Pérez-López and Garnica-Garza, 2011; Facundo-Flores and Garnica-Garza, 2013; Garnica-Garza, 2013). The prescribed dose of 25 Gy and the set of treatment goals shown in Table 1 were taken from the clinical trial (Bondiau et al., 2013), except for the sternum, rib cage and heart wall, for which no dosimetry was reported. For these structures, the prescription goals were chosen after several preliminary optimization runs. A treatment plan was generated for each beam collimation radius, and for each optimized treatment cumulative dose volume histograms (cDVH) were determined for each of the structures shown in Table 1. The non-tumor integral dose (NTID) was also calculated for each plan. The software was run on an Intel i5 processor simultaneously using 4 available threads, with each optimization taking about 50 hours of computer time.

3. Results and discussion

While 300 beams were made available to the optimization algorithm for each treatment plan, the algorithm was free to turn off any number of beams in order to meet the treatment objectives. The final number of beams used in each treatment plan were 270,

Table 1
Treatment goals for the optimization of the breast RSR treatment. The upper and lower absorbed dose limits are denoted by D_L and D_U in Gy. W is the weight assigned to each structure or volume, and the total sum of them is 1.0. (I) stands for ipsilateral and (C) for contralateral.

Structure	D_L	D_U	W
GTV	25	28	0.20
CTV	25	28	0.20
PTV	25	28	0.20
Rib cage	0	5	0.02
Sternum	0	5	0.01
Breast (I)	0	25	0.10
Breast (C)	0	10	0.10
Lung (I)	0	10	0.05
Lung (C)	0	10	0.02
Heart wall	0	10	0.05
Skin	0	15	0.05