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Research and application of beam auto-modeling method for three-dimensional conformal radiation therapy treatment planning system

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Abstract: To insure the accuracy of irradiation, the three-dimensional conformal radiation therapy (3DCRT) treatment planning system (TPS) should, based on the types and orientations of the beam, automatically calculate the source position and the field size of the irradiation beam to cover the target (tumor) volume while avoiding excessive irradiation of the surrounding tissues. A beam modeling method was developed: first, based on the derived matrix of coordinate system transformation, the 3D patient model is tranfered from the patient coordinate system to the beam coordinate system; then, the iso-center coordinate of the target is caculated, from which the coordinate of the beam source is derived; finally, the target volume is projected onto the iso-center plane to identify the field size of the irradiation beam. By implementing of this method, the beam modeling module was developed with features of 3D visualization and a friendly user interface. The accuracy and efficiency of this module was verified based on the benchmark cases of the AAPM report 55 and the clinical patient image data by testing and comparing it with the beam modeling module of a commercial 3DCRT TPS.

Key words: radiation therapy; beam modeling; coordinate system transformation; visualization

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三维适形放疗计划系统中照射射束自动建模方法的研究和实现

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摘要:三维适形放疗计划系统需要根据射束的不同类型和方向,在照射位置和区域上覆盖目标(肿瘤)区域的同时避免周围正常组织的过量照射,确保照射精度.发展了一种射束建模方法:①通过推导坐标系转换矩阵,

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将病人影像坐标系统下的三维病人模型转换到射束坐标系下；②计算照射目标中心点坐标并以此确定照射距离和射束源位置；③根据照射目标在等中心平面上的投影确定照射区域大小. 应用该方法，实现了具有三维可视效果和友好交互性的射束模块. 采用美国医学物理协会 55 号报告（AAPM 55）中基准测例和临床病人影像数据，与某商业适形放疗计划系统的射束模块进行测试比较，验证了本方法的有效性和正确性.

关键词：放射治疗；射束建模；坐标系转换；可视化

0 Introduction

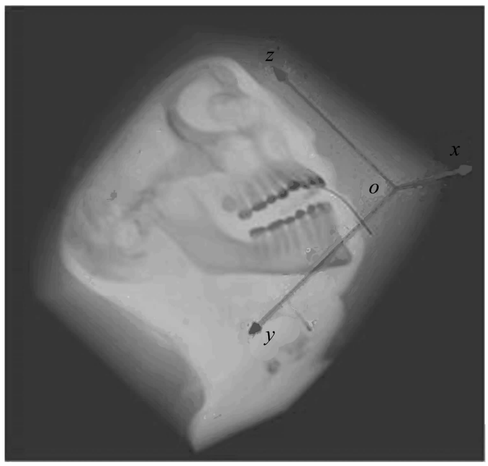
Three-dimensional conformal radiation therapy (3DCRT) has been considered the most exciting development in radiation oncology since the introduction of computed tomography imaging into treatment planning[1]. The goal of 3DCRT is to achieve maximal therapeutic benefit expressed in terms of a high probability of local control of disease with minimal side effects. Physically this often equates to the delivery of a high dose of radiation to the tumor or target region whilst maintaining an acceptably low dose to other normal tissues, particularly those adjacent to the target[2-3]. Two key issues are necessary to achieve the above aim. ① The field size of the irradiation beam needs to be in good agreement with the projection of planning target volume (PTV) along the irradiation direction. ② The dose rate inside and on the surface of PTV need to be uniform[2]. Despite there being some commercial treatment planning systems (TPS) for 3DCRT developed to achieve that goal, these software are bounded to specific medical linear accelerators and cannot be adaptive to domestic radiation therapy facilities. In this paper, based on the developed 3D-reconstruction algorithm developed by the authors[4], we focus on solving the first key issue of 3DCRT. The beam modeling method has been designed and an independent beam modeling module has been developed by the implementation of this method. The benchmark testing has been carried out by comparing it with other commercial TPS to ensure the validity and efficiency of this module.

1 Methodology

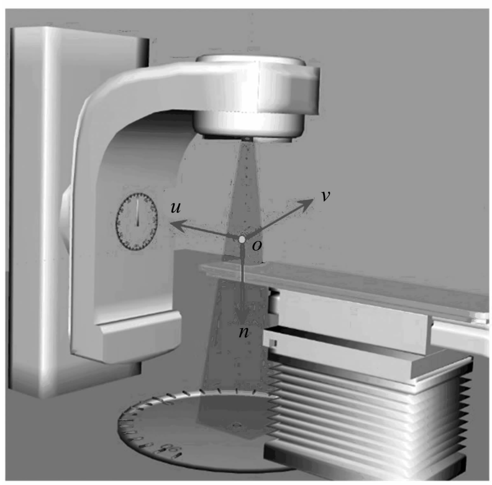
1.1 Definition of coordinate system

Coordinate system is the foundation of beam modeling[5]. To determine the position of the beam source and the field size along the irradiation direction, it is necessary to transfer the 3D patient model from patient coordinate system (PC) into the beam coordinate system (BC). The definitions of PC and BC are:

Patient coordinate system (PC), as a global reference coordinate system in TPS, is constructed based on a CT dataset scanned from a patient. Fig. 1(a) shows the schematic diagram of PC. Suppose that



(a) patient coordinate system



(b) beam coordinate system

Fig. 1 Irradiation coordinate system

a patient is supine on the couch, head near to gantry of accelerator. The origin is defined at the upper left corner of the first slice that is near his feet. Axis Y of patient coordinate is vertical to the floor and its positive direction is towards to the patient's back. Axis Z is parallel to the couch and its positive direction is from the patient's feet to head. Axis X is defined by the right-hand screw rule and its positive direction is from the patient's right side to left side. Under the patient coordinate system, the orientation of the radiation beam is determined by three angles of the accelerator machine: gantry angle, turntable angle and collimator angle.

Beam coordinate system (BC), as a local coordinate system, is constructed based on the orientation of the irradiation beam. Each beam has its own beam coordinate system. The origin of BC is at the iso-center (iso) of a beam. Axis n (corresponding to Axis Y) is the beam axis. The direction of Axis n is along the direction of irradiation. Axis v (corresponding to Axis X) and Axis u (corresponding to Axis Z) is determined by the right-hand screw rule. The iso-plane is on the u-v plane. Fig. 1(b) shows the schematic diagram of a BC which belongs to a beam in the initial state (gantry angle, turntable angle and collimator angle are all 0°).

1.2　Procedure of beam modeling

There are three types of irradiation in external radiotherapy. They are fixed source-to-axis distance (SAD) irradiation, fixed source-to-surface distance (SSD) irradiation and angular speed control (ARC) irradiation. ARC in fact is a kind of rotation SAD beam with controlled angular speed. Its beam modelling method is the same as that of SAD.

The flowchart of beam modeling is shown in Fig. 2.

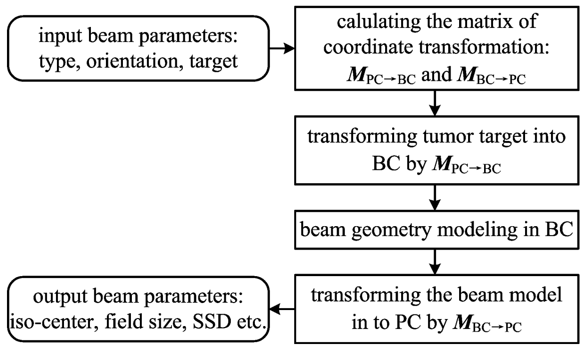


Fig. 2　Flowchart of beam modelling

According to the coordinate system definition of the PC and BC, the matrix of coordinate transformation from PC to BC is

$$\boldsymbol{M}_{PC\to BC} = \boldsymbol{R} * \boldsymbol{T}(-\mathrm{iso}_x, -\mathrm{iso}_y, -\mathrm{iso}_z) \quad (1)$$

where R is the rotation transformation matrix from PC to BC, iso_x, iso_y iso_z are coordinates of the iso-center, and $\boldsymbol{T}$ is iso transformation matrix from the origin of BC to that of PC:

$$\boldsymbol{T} = \begin{bmatrix} 1 & 0 & 0 & -\mathrm{iso}_x \\ 0 & 1 & 0 & -\mathrm{iso}_y \\ 0 & 0 & 1 & -\mathrm{iso}_z \\ 0 & 0 & 0 & 1 \end{bmatrix} \quad (2)$$

$$\boldsymbol{R} = \begin{bmatrix} v_x & v_y & v_z & 0 \\ n_x & n_y & n_z & 0 \\ u_x & u_y & u_z & 0 \\ 0 & 0 & 0 & 1 \end{bmatrix} \quad (3)$$

Based on Eqs. (2) and (3), Eq. (1) is expanded to Eq. (4):

$$\boldsymbol{M}_{PC\to BC} = \begin{bmatrix} v_x & v_y & v_z & -(v_x \cdot \mathrm{iso}_x + v_y \cdot \mathrm{iso}_y + v_z \cdot \mathrm{iso}_z) \\ n_x & n_y & n_z & -(n_x \cdot \mathrm{iso}_x + n_y \cdot \mathrm{iso}_y + n_z \cdot \mathrm{iso}_z) \\ u_x & u_y & u_z & -(u_x \cdot \mathrm{iso}_x + u_y \cdot \mathrm{iso}_y + u_z \cdot \mathrm{iso}_z) \\ 0 & 0 & 0 & 1 \end{bmatrix} \quad (4)$$

In (4), $\boldsymbol{v}$, $\boldsymbol{u}$, $\boldsymbol{n}$ as three direction vectors of the coordinate axis of BC, are defined by orientation of the irradiation beam. They can be calculated in the following steps:

① Initializing three vectors: $\boldsymbol{v}_0 = [1, 0, 0]^T$, $\boldsymbol{n}_0 = [0, 1, 0]^T$, $\boldsymbol{u}_0 = [0, 0, 1]^T$.

② Rotating $\boldsymbol{v}_0\boldsymbol{u}_0\boldsymbol{n}_0$ around Axis Z with gantry angle (α) to get $\boldsymbol{v}_1\boldsymbol{u}_1\boldsymbol{n}_1$:

$\boldsymbol{v}_1 = \boldsymbol{R}_Z(\alpha) * \boldsymbol{v}_0$; $\boldsymbol{n}_1 = \boldsymbol{R}_Z(\alpha) * \boldsymbol{n}_0$; $\boldsymbol{u}_1 = \boldsymbol{R}_Z(\alpha) * \boldsymbol{u}_0$. The rotation matrix is

$$\boldsymbol{R}_Z(\alpha) = \begin{bmatrix} \cos\alpha & -\sin\alpha & 0 \\ \sin\alpha & \cos\alpha & 0 \\ 0 & 0 & 1 \end{bmatrix} \quad (5)$$

③ Rotating $\boldsymbol{v}_1\boldsymbol{u}_1\boldsymbol{n}_1$ around Axis Y with negative turntable angle (β) to get $\boldsymbol{v}_2\boldsymbol{n}_2\boldsymbol{u}_2$:

$$\boldsymbol{v}_2 = \boldsymbol{R}_Y(-\beta) * \boldsymbol{v}_1;\ \boldsymbol{n}_2 = \boldsymbol{R}_Y(-\beta) * \boldsymbol{n}_1;$$
$$\boldsymbol{u}_2 = \boldsymbol{R}_Y(-\beta) * \boldsymbol{u}_1.$$

The rotation matrix is

$$\boldsymbol{R}_Y(-\beta) = \begin{bmatrix} \cos\beta & 0 & -\sin\beta \\ 0 & 1 & 0 \\ \sin\beta & 0 & \cos\beta \end{bmatrix} \tag{6}$$

④ Rotating $\boldsymbol{v}_2\boldsymbol{n}_2\boldsymbol{u}_2$ around $\boldsymbol{n}_2$ with collimator angle (γ) to get $\boldsymbol{vun}$:

$$\boldsymbol{v} = \boldsymbol{v}_2\cos\gamma + (\boldsymbol{n}_2 \times \boldsymbol{v}_2)\sin\gamma + \boldsymbol{n}_2(\boldsymbol{n}_2 \cdot \boldsymbol{v}_2)(1-\cos\gamma);$$
$$\boldsymbol{u} = \boldsymbol{u}_2\cos\gamma + (\boldsymbol{n}_2 \times \boldsymbol{u}_2)\sin\gamma + \boldsymbol{n}_2(\boldsymbol{n}_2 \cdot \boldsymbol{u}_2)(1-\cos\gamma);$$
$$\boldsymbol{n} = \boldsymbol{n}_2.$$

The matrix of BC to PC is the inversed matrix of $\boldsymbol{M}_{\mathrm{PC}\to\mathrm{BC}}$: $\boldsymbol{M}_{\mathrm{BC}\to\mathrm{PC}} = \boldsymbol{M}_{\mathrm{PC}\to\mathrm{BC}}{}^{-1}$.

Based on the matrix of coordinate transformation, the beam geometry in BC is illustrated in the following steps:

① Fixing the beam source at the negative axis of n and its distance to origin is default value (100 cm).

② Transforming the surface model of target tumor into BC by $\boldsymbol{M}_{\mathrm{PC}\to\mathrm{BC}}$.

③ Perspective projecting the target 3D model onto the iso-plane to get field size (X_1, X_2, Y_1, Y_2).

④ Beam geometry modeling by connecting the beam source to four vertexes of field size.

⑤ Using two clipping planes to clip the beam geometry model as a quadrangular frustum pyramid model (shown in Fig. 3) for a better visualization effect of a beam model. The distance from the pre-clipping plane and post-clipping plane toiso-plane is labelled as AOD and POD, respectively.

Based onthe above steps, the beam source coordinates and the field size can be calculated for SAD irradiation. The other important parameters needed in the dose calculation can also be calculated, including the iso-center coordinates and the source-to-surface distance (SSD). For the SAD irradiation, the iso-center is determined by the geometry center of the target model. SSD is the distance from the beam source to the intersection point which is determined by the beam axis passing through the surface of the patient skin model. The beam axis is a line connecting the beam source point with the iso-center, which is labelled as Axis n in BC. To calculate the intersection point, an algorithm of uniform-level octree subdivision was employed to locate the triangle cell that intersected the beam axis[6]. Then the coordinate of intersection point is obtained by linear interpolated calculation from the corrdinates of three vertices of the located triangle cell. As for the SSD irradiation, the iso-center is just the intersection point on surface model. SSD is set at 100 cm as default.

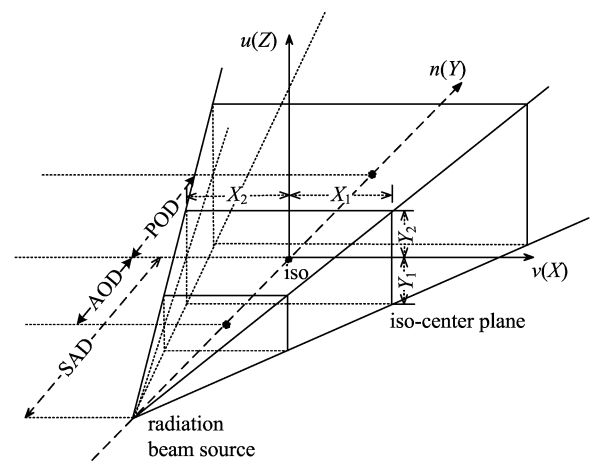


Fig. 3 Geometry model of irradiation beam

1.3 Module development

An independent module of irradiation beam has been designed based on object-oriented programming (OOP). The framework of module design is shown in Fig. 4, which includes three layers. The first layer is the graphical user interface (GUI) providing interactive interfaces to set beam parameters such as irradiation orientation, types, target and visibility of models. The second layer is the business layer (BL) providing the function class. The super-class vtkBeamActor provides the abstract interfaces to the user. Three sub-classes inherit the super-class and implement abstract interfaces. The third layer is a class foundation layer which supports the above two layers and consist of Microsoft Foundation Classes (MFC)[7] and the open-source visualization toolkit (VTK)[8].

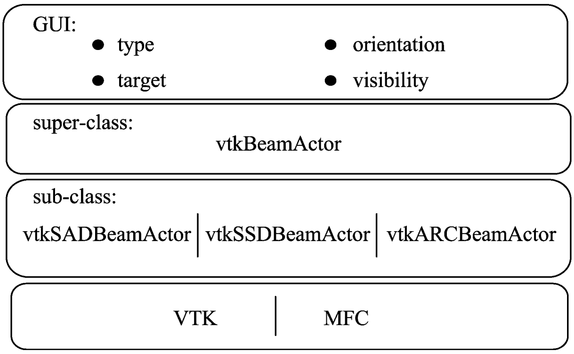


Fig. 4　Framework of beam module

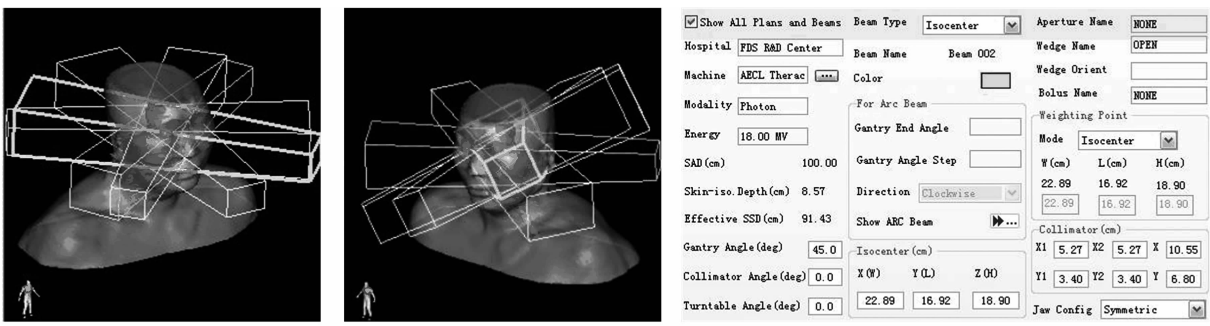
2　Results and discussion

2.1　Results

The module was tested and compared with the corresponding module on a commercial TPS Venus. The test cases consist of the benchmark cases from AAPM 55 report[9] and several clinical patient cases. Due to space constraint, the results of only one clinical test case from a patient with nasopharyngeal carcinoma (NPC) were reported in this study. The dataset contained 96 CT slices scanned from head and chest. The thickness of a CT slice is 5 mm. A file of RT-structure which contained contoured points of target tumor and OARs was also included. The test dataset was imported into our module as well as Venus TPS. Two forward plans were designed for testing. The first plan tests the function of SAD irradiation and the second plan tests the function of SSD irradiation. A comparison of the above two plans was carried out on Venus. The beam modeling results are shown in Fig. 5(a) and (b). The GUI of irradiation parameters are shown in Fig. 5(c). The detailed parameters of results are listed in Tab. 1 and Tab. 2.

2.2　Discussion

The parameters in Tab. 1 and Tab. 2 are calculated results of auto beam modeling. The coordinates of iso-center is calculated from the geometry center of the reconstructed target model; the value of SSD in SAD irradiation is the distance from the calculated beam source and intersection point on the surface of the model. By comparing it with commercial TPS Venus, the relative error of iso-center coordinates was no more than 0.8% both for SAD and SSD irradiation. The relative error of SSD in table was no more than 0.17%.



(a) coplanar SAD beam models　(b) non-coplanar SSD beam models　(c) GUI of beam parameters setting panel

Fig. 5　Modeling results of SAD and SSD irradiation

Tab. 1　Beam parameters of SAD irradiation

gantry angle	SSD in this study	SSD in Venus/cm	% Rel. error of SSD	iso-center in this study/cm	iso-center in Venus/cm	% Rel. error of iso-center
0°	91.14	91.18	0.04	(22.89,16.92,18.90)	(22.84,16.87,18.75)	(−0.22,−0.29,−0.8)
45°	91.43	91.44	0.01	(22.89,16.92,18.90)	(22.84,16.87,18.75)	(−0.22,−0.29,−0.8)
120°	92.86	92.70	−0.17	(22.89,16.92,18.90)	(22.84,16.87,18.75)	(−0.22,−0.29,−0.8)
240°	91.10	91.11	0.01	(22.89,16.92,18.90)	(22.84,16.87,18.75)	(−0.22,−0.29,−0.8)
315°	90.79	90.95	0.17	(22.89,16.92,18.90)	(22.84,16.87,18.75)	(−0.22,−0.29,−0.8)

Tab. 2 Beam parameters of SSD irradiation

gantry angle	turntable angle	iso-center in this study/cm	iso-center in Venus/cm	% Rel. error of iso-center
0°	30°	(22.86,8.06,18.90)	(22.84,8.05,18.75)	(−0.09,−0.12,−0.8)
45°	30°	(28.94,10.86,18.90)	(28.89,10.82,18.75)	(−0.17,−0.37,−0.8)
120°	0°	(29.07,20.49,18.90)	(29.16,20.52,18.75)	(0.30,0.15,−0.8)
240°	0°	(15.18,21.36,18.90)	(15.14,21.33,18.75)	(−0.26,0.14,−0.8)
315°	30°	(16.38,10.41,18.90)	(16.44,10.47,18.75)	(0.36,0.57,−0.8)

The beam parameters of our module showed a good agreement with those of commercial TPS. It is also noticeable that the offset value of iso-center coordinate Z is always −0.8%. The reason may be the inconsistent definition of PC origin between commercial TPS and our beam module. In Venus, the Z coordinate of the origin was defined at the middle of a CT slice. However, it was defined at the bottom of a CT slice in our developing radiotherapy plan system. This would cause a difference of a half of the thickness of a CT slice, that is, 0.15 cm in our test dataset.

3 Conclusion

A beam modeling method has been developed to simulate and visualize the different types and orientations of irradiation in external radiation therapy. It would be helpful for TPS user to optimize the irradiation orientation as well as provide the important parameters for accurate dose calculation. The main contributions of this study could be concluded as follows:

(Ⅰ) By introducing the local coordinate system: beam coordinate system (BC), the complexity of beam geometry modeling could be decreased effectively.

(Ⅱ) The matrix of coordinate transformation between BC and PC was deduced based on the definition of coordinate system.

(Ⅲ) Beam parameters were calculated and compared with a commercial TPS. The effect and accuracy of beam modeling method had been verified.

(Ⅳ) Using OOP technology and VTK, a beam modeling module has been developed which shows the good 3D visualization effect and friendly user interface.

In addition to its application in radiotherapy TPS, the principle of coordinate transformation of this beam modeling method is also appropriate for other application fields. In our further work, the modeling method of beam accessories such as block and MLC (multi-leaf collimator) will be investigated.

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