Optimization of Dose Distribution for LINAC Based Radiosurgery Using Elliptical Collimators

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Summary

The use of non-circular collimators is considered on all or some of the arcs used for radiosurgery on a linear accelerator in order to obtain conformal dose distribution for non-spherical lesions using a single isocenter. An extension of software to allow for use of non-circular collimators and a mathematical optimization based on prescribed doses on the lesion surface and on points in normal tissue (critical structures) are presented. Tests of the optimization method on simulated cases indicate that several boosts from selected positions along the arcs, superimposed on an optimized arc configuration allows one to obtain a highly conformal dose distribution with simple elliptical inserts. The optimization method can be applied to any type of collimation and is particularly effective with variable dose-rate machines.

Keywords: LINAC radiosurgery; dose distribution.

Introduction

Conventional Linear Accelerator (LINAC)-based neuroradiosurgery employs convergent noncoplanar arcs and circular collimators to generate highly focussed, quasi-spherical dose distributions centered around a stereospatially-defined anatomical target point that is aligned coincident with the isocenter of the LINAC beam. In many clinical cases, anatomical lesion volumes have irregularly-shaped geometry, and irregular fields need to be shaped to conform to these irregular target lesion volumes. Although geoconformal dose distributions and protection of adjacent critical structures can be achieved to some extent by modifying arc and couch configuration, length, position and non-segmented arc dose weighting, these maneuvers may only significantly impact on the lower isodose level regions of a dose distribution. Alternatively, the use of multiple overlapping targets is another approach that can result in a composite field with enhanced geoconformal dosimetry in the higher isodose regions. The inherent field overlap of the multiple target method can introduce dose inhomogeneities that may increase radiotoxicity [6].

The need to achieve geoconformal dosimetry without increasing dose volume inhomogeneity requires new approaches. Leavitt et al. [3] and McGinley et al. [4] presented a computer-controlled system of dynamically-adjustable, 4-jawed collimators for stereotactic radiosurgery, while other adjustable multi-leaf micro-collimator systems for multi-angle fixed point and shoot fractionated, stereotactic precision radiotherapy are also in use. Such dynamic or static field molding techniques can improve conformal dosimetry or may mitigate against dose inhomogeneity, but require a significant technical effort. Also Schlegel et al. [10] have presented a manual multileaf collimator which is adjusted for each arc based on a lesion projection solid model.

Suggested by Serago et al. [11] and, independently, by the present authors [8], an alternative approach is the use of non-circular collimation. However, the ability to accurately simulate elliptical fields and to evaluate the degree of geoconformal dosimetry and dose homogeneity achieved in tandem, requires appropriately compliant treatment planning software [13]. Accordingly, we present the extension of our radiosurgery treatment planning and virtual simulation system [12] to non-circular collimation, along with our model for concurrent mathematical optimization of geocon-
formal dose distribution for irregular lesions using a single-target, non-circular collimation approach. This optimization system is intended to identify and simulate preferred combinations of: 1) dose-weighted treatment arc segment configuration and collimation parameters, that best correspond to a desired homogenous, geoconformal dose distribution.

Methods and Materials

1. Dose Computation for an Elliptical Collimator

Computation of dose to any point in the brain both represents a superposition of contributions from all arcs and is affected by the introduction of non-circular collimators. For circular collimators, the point dose contribution, \(D\), (arising from a gantry rotation arc segment, i.e. arc increment) may be calculated according to Rice et al. [9]:

\[
D(d,s) = MS(c)(SAD/SSD + d)^2 TMR(d,p) R(\rho;c) \tag{1}
\]

where:
- \(d\) = geometric depth of the calculation point below the surface of a patient’s scalp, along the central axis (CAX) of the beam,
- \(s\) = radial distance perpendicular from CAX to a calculation point,
- \(\rho\) = radial distance from CAX to the edge of the circular field projected at depth \(d\),
- \(R\) = off-axis ratio or beam profile for a collimator \(c\),
- \(M\) = monitor units delivered for the arc increment,
- \(S(c)\) = total scattering or output factor for the collimator \(c\),
- \(SAD\) = source axis distance,
- \(SSD\) = source to skin distance along the CAX,
- \(TMR\) = tissue-maximum-ratio (for depth \(d\) and field size \(\rho\)).

It is assumed that the LINAC is calibrated to 1 cGy/MU at the depth of maximum dose, and that, in accordance with conventional algorithms for brain irradiations, tissue inhomogeneity corrections are not needed.

To extend the computation to non-circular collimators, the radial beam profile is replaced by a 2D-profile, \(\delta(r,\phi;c)\). The angle \(\phi\), defined in Fig. 1, subtends the projected collimator major axis with the direction from the CAX to the calculation point. Radial dose profiles were measured for a series of angles around the center of the ellipse, usually 30° apart, using film positioned at the isocenter. In order to obtain a profile, the dose point is projected along the beam line to the depth of the LINAC isocenter, and a linear interpolation between two adjacent table entries (positions) is used to estimate a profile falling between those two table positions. We have found it convenient to convert the film-derived digital optical density data to a 2D dose distribution by using an H&D curve and by then extracting twelve radial profiles from it. Precision-fashioned on a milling machine, our actual elliptical lead collimator insert has a rounded rectangular cross-section, rather than a true ellipse, and has a three stage taper for divergence.

In order to compute the dose contributions from arc increments (segments) to points inside the skull, the beam projection is translated to stereotactic frame coordinates. (The stereotactic coordinate system is defined by standard stereotactic frame axes: \(OZ\) – along the couch pointing to the cranial vertex; \(OX\) – pointing to the right with the patient supine; and \(OY\) anterior/posterior transecting the frame origin). Next, we define \(V\), as the intersection of the gantry rotation plane with the beam’s eye view-plane, \(II\), shown in Fig 1. Thus, if the gantry axis is \(G\), and the beam central axis is \(P\), then the reference axis, \(V\), is given by \(V = G \times P\). This choice leaves the axis \(V\) pointed to the left side of the head (see Fig. 2), provided the table is perpendicular to the gantry rotation plane. For the table at \(\theta\), \(G\) is calculated in stereotactic frame coordinates \(OXYZ\) by applying a rotation of the ‘table’ vector \(Z\) about a vertical axis \(Y\) which transects the isocenter. In the \(OXYZ\) stereotactic frame system we get \(G = (- \sin \theta, 0, \cos \theta)\). With \(P_1, P_2, P_3\), being the components of the “beam” vector, the reference axis is given by:

\[
V = \begin{bmatrix}
-\sin \theta & 0 & \cos \theta \\
\end{bmatrix}
\begin{bmatrix}
P_1 \\
P_2 \\
P_3 \\
\end{bmatrix}
= (- P_3 \cos \theta, P_1 \cos \theta + P_3 \sin \theta, - P_2 \sin \theta) \tag{2}
\]

The program projects a dose point \(M\) along the X-ray source beam line and onto the beam’s eye view plane that transects the target, computes the distance of the projection (\(M’\) in Fig. 1) to the target (coincident with the isocenter), and solves Eq. (2) for the angle \(\Phi\) between \(IM’\) and \(V\). The collimator orientation is given by the angle \(\Psi\) between \(V\) and the major axis of the ellipse (IA in Fig. 1).