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A Technique for Stereotactic Radiosurgery Treatment Planning with Helical Tomotherapy

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Abstract

Purpose—The purpose of this study was to develop an efficient and effective planning technique for stereotactic radiosurgery using helical tomotherapy.

Methods and Materials—Planning CTs and contours of twenty patients, previously treated in our clinic for brain metastases with Linac-based radiosurgery using circular collimators, were used to develop a robust Tomotherapy planning technique. Plan calculation times as well as delivery times were recorded for all patients to allow for an efficiency evaluation. In addition, conformation and homogeneity indices were calculated as metrics to compare plan quality to that which is achieved with conventional radiosurgery delivery systems.

Results—A robust and efficient planning technique was identified to produce plans of radiosurgical quality using the TomoTherapy treatment planning system. Dose calculation did not exceed a few hours and resulting delivery times were less than an hour, which allows the process to fit into a single day radiosurgery workflow. Plan conformity compared favorably with published results for Gamma Knife radiosurgery. In addition, plan homogeneity was similar to Linac-based approaches.

Conclusions—The TomoTherapy planning software can be used to create plans of acceptable quality for stereotactic radiosurgery in a time that is appropriate for a radiosurgery workflow that requires that planning and delivery occur within one treatment day.

Keywords

Stereotactic Radiosurgery; Tomotherapy; Brain Metastases; Treatment Planning

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I. Introduction

Tomotherapy has shown potential as a precision stereotactic radiosurgery delivery system^{1–3}. Recent work carried out by our group has shown that the on-board MVCT can be used for stereotactic localization and the system is capable of sub-millimeter delivery accuracy if used in conjunction with a precise intracranial stereotactic patient positioning system¹. In addition, planning studies have shown that Tomotherapy produces conformal plans for small targets^{4,5}. Since the delivery is non-isocentric, it could potentially provide an efficiency advantage in the case of multiple intracranial targets or large irregularly shaped targets that would require multiple isocenters when using conventional SRS. Although Tomotherapy lends itself well to fractionated stereotactic procedures, such as stereotactic radiotherapy (SRT) and stereotactic body radiotherapy (SBRT), the unique challenges involved with single fraction delivery and the use of an invasive fixation system are the focus of this particular work.

Stereotactic localization can be achieved with Tomotherapy using image guidance in combination with rigid head fixation. We therefore, propose to use of an invasive head frame for fixation of the patient, while on-board MVCT is used for stereotactic localization and an optical tracking system is used for shift verification. Moreover, to avoid difference in table sag between CT and Tomotherapy, we propose that a thin slice pretreatment MVCT is employed in the SRS treatment planning process. We therefore have designed the following workflow: After head ring placement, the patient is set up on the Tomotherapy couch using a Tomotherapy specific tabletop frame docking device (InterFix by Integra Radionics, Burlington, MA) for acquisition of the thin slice reference/planning MVCT image data set. An array of passive infrared markers is placed on the frame for optical tracking (Dynatrac by 3D Line Medical Systems, Milano, Italy). Camera and couch coordinates are recorded at the time of initial imaging to aid in repeat set-up of the patient. After scanning, the thin slice MVCT image is sent to 3rd party treatment planning software where it can be fused with an MRI for target delineation. A treatment plan is generated on the Tomotherapy Treatment Planning Station (TPS) using the MVCT for the dose calculation. The patient is then repositioned on the couch, using the couch and camera coordinates recorded previously, and a pre-treatment verification MVCT is performed. An MV-MV fusion of the planning and pre-treatment verification images is completed to localize the target for delivery. If a shift is required, the shift is applied to the couch while secondary verification of the magnitude and direction of the shift is performed via optical tracking.

From a treatment planning perspective, generating a deliverable radiosurgery plan with the time requirements introduced by this workflow is challenging. Conventional SRS planning can be performed quickly employing straightforward planning approaches, in which known beam arrangements are used to create spherical and ellipsoidal shaped dose distributions. Tomotherapy involves a sophisticated inverse planning technique with a relatively lengthy dose calculation. Emphasis must therefore be placed on making good initial choices in planning parameters to avoid a time-consuming trial-and-error process. This work aims to identify an SRS planning technique for Tomotherapy that can be performed quickly yielding high quality plans for the treatment of brain metastases, which compare well to other forms of radiosurgery.

II. Materials and Methods

Test Patients

Twenty patients with brain metastases, previously treated with cone-based linear accelerator SRS, were used to test for the efficacy and robustness of the proposed Tomotherapy planning technique. Patient details are shown in Table I. Patients selected had between one

to six intracranial metastases with target volumes ranging from 0.02 ml to 10.78 ml with a median volume of 0.38 ml and an average volume of 1.06 ml. Patients' original planning CT and MR-defined target volumes were used for Tomotherapy comparison planning.

Planning Goals

The goal of Tomotherapy planning was to generate clinically acceptable plans within a reasonable planning time. A clinically acceptable plan is one in which dose conforms tightly to the target and dose falloff into extra-target normal tissues is as steep as possible given the delivery technique. In other forms of radiosurgery, doses are typically prescribed to 50–80% of target maximum dose depending on the SRS modality to ensure the tumor boundary is on the steepest part of the dose gradient. The work by Lax⁶ shows that allowing for heterogeneous target coverage reduces the extra-target dose, which is one of the guiding principles SRS and has been shown to hold true for Tomotherapy as well.^{7,8} Hence, any Tomotherapy SRS planning technique should generate plans that are sufficiently inhomogeneous to allow for a steep dose gradient beyond the target despite the strong tendency of the Tomotherapy optimization algorithm to produce homogeneous dose distributions that are typically desirable for fractionated radiotherapy. The resulting planning techniques should be simple and result in treatment delivery times that are comparable to or shorter than those of conventional SRS techniques.

Treatment Planning Software

All plans were run using version 3.1.4 of the TomoTherapy Hi-Art treatment planning system (TPS). This software release has two primary advantages over previous versions. First, the gradient of the dose falloff in the longitudinal direction is increased since the first and last delivered beamlets start closer to the tumor boundaries. Second, version 3.1.4 corrects an anomaly in volume element summation present in earlier versions, which impacts DVH generation for small (< 2 ml) target volumes and regions at risk in steep dose gradient regions⁹.

Planning Technique

Contouring—Based on our own experience and previously published studies^{8,10}, it was determined that additional non-anatomic planning structures would be needed to help create a dose distribution with the desired characteristics. Here the goal was to define useful planning structures that would improve the speed of plan optimization while keeping contouring time to a minimum. For the final technique, two different sets of added planning structures were identified; one for small targets (volume ≤ 2 ml) and another for large targets (volume > 2 ml). In addition to the physician-defined target, the following contours were added:

1. A 3mm ring peripheral to the target to facilitate increased dose fall off in adjacent tissue.
2. A 2.5 mm diameter central sub-volume (CSV) placed at the center of mass of the target.
3. For targets in the plane of the eyes, both eyes must be contoured and be directionally blocked to prevent any beamlets from entering through them.

For larger targets having a volume larger than 2 ml, the following structures may also be needed:

1. An additional inner sub-volume (SV) having a volume greater than one voxel but less than 2 ml.

2. An inner shell of 2mm thickness whose outer edge is formed by the periphery of the target.

The planning volumes for the two cases have been illustrated in Figure 1.

Initial Plan Parameters—All plans were calculated using the “Fine” dose grid, resulting in a dose voxel size of $1.41\text{mm} \times 1.41\text{mm} \times 1.25\text{mm}$ (where image pixels are directly used as dose voxels). A field width of 10 mm was used for treatment planning to maximize the rate of dose falloff at the superior and inferior boundaries of the target.

The modulation factor determines the range of allowable intensity values for the leaves of the multileaf collimator (MLC) and is defined as the ratio of maximum to mean MLC leaf opening times for the leaves that are open. As shown in Figure 2, increasing modulation can be used to increase dose inhomogeneity within the target. Figure 3 shows the corresponding increase in the steepness of the peripheral dose gradient. However, increasing modulation also proportionally increases the gantry period (when it is above its minimum value), and therefore leads to increased delivery times. Plan quality and efficiency must be considered together to determine an ideal modulation factor. A modulation factor of 1.7 was chosen for all plans to obtain a balance between dose inhomogeneity and treatment time although a range between about 1.5 and 2.0 would be acceptable for most cases.

For helical delivery, pitch refers to the fraction of the field width the couch moves into the bore per rotation of the gantry. This parameter thus defines the degree of beam overlap between each rotation and determines the number of rotations over which a target voxel will be within the beam projections. For SRS planning, there are two main considerations in terms of pitch. First, when pitch is reduced, the gantry period is decreased because an increased number of rotations are available to deliver dose to a given voxel and each rotation delivers a smaller dose. Second, as the dose per fraction increases, the gantry period increases. Therefore in SRS, where the dose per fraction is high, it might not be possible to deliver the entire dose while keeping the gantry rotation period within the 60 second limit and multiple “passes” might be necessary to deliver the prescribed dose. If one is intent on keeping the delivery to a single pass, then the pitch should be as small as possible. However, from a workflow perspective a small pitch increases the beamlet computation time. Hence, it will be inefficient to choose a pitch that is too small. For a patient with four metastases, Figure 4 shows the beamlet computation time as a function of pitch for a clinically used 16-node cluster. The goal here was to determine the largest pitch that will allow for delivery of the entire dose in a single pass. Ideal pitches consistently produce a gantry period between 50 and 60s.

Table II shows the range of pitch values that allow for single pass delivery as a function of the minimal peripheral dose and a modulation factor of around 1.7. However, if the pitch results in a gantry period of slightly longer than 60 seconds, a small decrease in Modulation Factor (MF) can be used to increase the gantry speed as described by Woch et al.¹¹.

Plan Objective and Optimization

Through iteration, it was determined that the planning objectives shown in **Tables IIIa** and **IIIb** reliably yield inhomogeneous dose distributions for small ($< 2\text{ ml}$) and large ($> 2\text{ ml}$) targets, respectively. The aim is to “weight” the minimum dose to the target and the maximum dose to the surrounding 3mm ring structure higher than all other target goals so that the prescription isodose coincides with the target boundary. The maximum dose to the target is set to the maximum allowed (120Gy) to ensure that it is removed from the optimization. The maximum dose in the target is then set by the CSV, which is set to 125% of the prescribed dose to mimic prescribing the 80% line in conventional SRS. For larger

targets, instead of assigning optimization goals to the target structure itself, one uses a larger sub-volume at the center of the target (SV) and the 2mm outer shell as “Tumor” structures to better shape the dose, while the target structure is given a “Sensitive Structure” type for target DVH evaluation. Two additional elevated weights are given to the maximum dose to the SV and the uniform (max and min) dose to the 2mm inner shell to ensure the shell is at the prescribed dose while the SV is at 125% of the prescribed dose. The CSV is still included and prescribed to 125% of the prescription dose to aid in plan scaling. The cutoff where the small volume technique breaks down is somewhat dependent on target shape and location but any target greater than 2 ml may require the large volume technique to achieve the desired dose distribution.

Controlling the dose near the outer edge of the target with the 2mm shell helps prevent this peripheral dose from becoming elevated, as shown in Figure 5 for a large target planned with both the small volume and large target techniques. This elevated dose is caused by simultaneously trying to keep the target dose above the prescription and trying to create a fast dose fall-off outside the target – a phenomenon known as “Gibbs ringing”¹². The frequency of this ringing is such that small targets do not suffer from it and the 2mm shell is unnecessary.

This planning strategy works for reasons directly related to the TomoTherapy planning system’s objective function. The TomoTherapy TPS uses an objective function that favors homogeneous dose in structures designated as targets. The optimizer objective function is given by¹³

$$O(\Psi) = \sum_{t \in T} \left(\frac{\alpha_t}{N_t} \sum_{i=1}^{N_t} \beta_{t,i}^{\max} \beta_{t,i}^{\min} (D_t^p - D_i^d)^2 \right) + \sum_{r \in R} \left(\frac{\alpha_r}{N_r} \sum_{i=1}^{N_r} (\varphi_{r,i}^{\max} + \varphi_{r,i}^{dvh}) (D_r^p - D_i^d)^2 \right) \quad (1)$$

where

O denotes the objective function

Ψ is the energy fluence vector (with one entry per beamlet)

T is the set of structures designated as targets

R is the set of structures designated as sensitive structures

D_x^p is the prescribed dose to structure x , always 0 for sensitive structures

D_i^d is the dose deposited in voxel i

α_x is the importance value for structure x

N_x is the number of voxels in structure x

$\beta_{x,i}^{\max}$ is target structure x ’s maximum dose penalty if voxel i is greater than x ’s maximum dose, otherwise 1

$\beta_{x,i}^{\min}$ is target structure x ’s minimum dose penalty if voxel i is less than x ’s minimum dose, otherwise 1

$\varphi_{x,i}^{\max}$ is sensitive structure x 's maximum dose penalty if voxel i is greater than x 's maximum dose, otherwise 0

$\varphi_{x,i}^{dvh}$ is sensitive structure x 's DVH point penalty if voxel i violates x 's DVH constraint, otherwise 0

This objective function favors uniformity in targets due to the left summation of the two summations. For a target, even if no constraint is violated, all of its voxels still contribute to the sum since in that case $\beta_{x,i}^{\max}=1=\beta_{x,i}^{\min}$, and their individual contribution to the overall sum is equal to their difference from the intended prescription dose, which is the same for all voxels in the target. For sensitive structures, which are governed by the right summation, only voxels that violate a constraint make a nonzero contribution to the objective function.

In order to achieve non-uniform dose distributions, it is therefore necessary to counter the tendency of the optimizer to generate a homogeneous dose distribution within the target as well as to control how the nonuniform dose is distributed within it. The extra planning structures considered here thus accomplish three goals:

1. Drive the optimizer to create inhomogeneous target dose distributions.
2. Allow control over how dose is distributed within the target.
3. Ensure sufficient beamlets are enabled to cover the target. (A beamlet is a point in the leaf sinogram)

For small targets (see **Table IIIa**), the original target structure is not used as the prescription structure, so the plan is not normalized to its DVH dose during optimization. Instead, the CSV is used as the prescription structure and is prescribed to more than the intended dose. Since the CSV is small, uniformity in it is trivially achieved and its constraints are easily satisfied. The CSV thus serves only to normalize the plan to a known scale. The question then becomes one of adequately covering the treatment target, and controlling the shape of the inhomogeneous dose distribution.

If only the CSV were used as a target, not enough beamlets would be enabled to cover the treatment target well. At least some beamlets that do not directly strike the CSV are needed. These “grazing beamlets” strike the target without striking the CSV and provide coverage of voxels toward the periphery of the treatment target. For this reason, the treatment target itself is also included as a target structure so that sufficient beamlets are enabled at the start of optimization. However, if the “grazing” beamlets are not constrained the optimizer will adjust them such that uniformity within the target is achieved, therefore a 3mm ring is added around the treatment target as a sensitive structure to ensure a steep dose fall-off beyond the target.

The technique for larger targets (see **Table IIIb** for constraints) is to build a pair of structures inside the treatment target to control the dose distribution within the target. The SV is the size of a small target (2 ml or less) to achieve a high dose in the central portion of the target, while the smaller CSV inside it allows one to scale the plan. The target becomes a “Sensitive Structure” and in its place a 2mm thick ring structure whose outer edge is formed by the periphery of the treatment target is used as a “tumor” structure in the optimization.

As indicated in the **Table IIIa and IIIb**, the CSV structure (or one of the CSV structures in the case of multiple targets) is used for the prescription. In general, for multiple targets the CSV for the largest treatment target should be used for prescription but one can use any other CSV if desired to aid in plan scaling. The ability to achieve target coverage at the 80%

prescription isodose level depends on the target volume. If the target is too small to achieve the desired gradient, the maximum achievable target inhomogeneity becomes apparent in the first few iterations of plan optimization and the dose distribution for that target can be scaled to ensure the target is not overdosed.

For plan scaling, the easiest is to use the CSV as a hard constraint in the optimization to scale the entire planned dose to the treatment target up or down as needed. However, in the case of multiple treatment targets, the CSV technique alone works best for the treatment target to which the prescription CSV belongs. For the other treatment targets, it is usually necessary to use a combination of three parameters: increasing the importance of the treatment targets, increasing the maximum dose penalty to the ring structure (to scale the dose down), or increasing the minimum dose penalty to the target (to scale the dose up). To maintain control of the plan, it is best to move in small increments when increasing importance and penalty values. If the plan starts moving in an undesirable direction, it was found best to cancel and start the optimization again as opposed to continuing to employ the same techniques.

All plans were run until the plan showed that 100% of each target was getting close to the prescribed dose with a reasonable gradient; typically a few hundred iterations (with each iteration requiring 5–7 seconds) but no restrictions were placed on this except that the planning time was kept to a time that would reasonably fit into our proposed radiosurgery workflow. In addition, in many cases, increased iterations led to a more homogenous tumor dose so plans were not left to run unattended.

Multiple Plans

In the case of multiple treatment targets, it is sometimes beneficial to plan targets separated by a large longitudinal distance as separate plans. Not only does this improve delivery efficiency in the case that there are several unused rotations between tumors during delivery, but it also simplifies the planning. For example, a hard constraint on one target will then not compromise the ability to deliver a desired dose to other targets. If targets in different areas of the brain are prescribed to different doses, allowing for increased pitch for the lower dose treatment targets can further reduce delivery and computation time. The effects of splitting the plan are shown in Table IV for a subset of patients.

Plan Quality Comparison

Although the detailed plan comparison between the Tomotherapy SRS plans and linear accelerator based SRS plans is outside the scope of this work, for the purpose of comparing this technique to more established radiosurgery techniques, the prescription isodose volume to tumor volume ratio (PITV)¹³, conformation number (CN), and a homogeneity index (HI) were calculated for all 59 targets. In addition, the volume of normal tissue receiving 12Gy was reported as an indication of dose falloff. The CN, originally described by van't Reit et al.¹⁴, takes into account the relative locations of the prescription isodose volume and the target volume by incorporating target coverage. Currently both metrics are regularly reported in the literature as objective measures of conformity so both will be considered here to allow for comparison to other work. The PITV and CN are defined below:

$$PITV = \frac{V_{PITV}}{V_T} \quad (2)$$

and

$$CN = \frac{(V_{T,P})^2}{V_T \times V_{PIV}} \quad (3)$$

Where V_{PIV} is the prescription isodose volume, V_T is the target volume, and $V_{T,P}$ is the volume of the target covered by the prescription isodose. The homogeneity index (HI) here is calculated using:

$$HI = D_{max} / D_{Rx} \quad (4)$$

Where, D_{max} is the maximum dose to that target and D_{Rx} is the prescribed dose.

Since Gamma Knife radiosurgery is often perceived of as the gold standard of treatment delivery, a comparison is made to the data collected by Nakamura *et al.*¹⁵ for 1338 lesions treated with the Gamma Knife unit. Nakamura *et al.*¹⁵ used what is referred to in their paper as the “New Conformity Index” which is the inverse of the conformity index originally proposed by van’t Reit *et al.*¹⁴ and Paddick¹⁶. For the purpose of comparing to this study, the inverse CN is considered. To compare the plan homogeneity achieved with this technique to pre-existing radiosurgery planning techniques, the average, range and median HI were also compared to the original linear accelerator based SRS plans. In addition, 12 Gy-volumes were compared to the original linear accelerator based SRS plans to determine any changes in normal tissues dose between the techniques. The 12 Gy-volume has been associated with the symptomatic radiation risk of necrosis (S-NEC) by several authors^{17–20} and can be used an indication of dose falloff into normal brain.

III. Results

Planning Technique Robustness and Delivery Parameters

A reliable and efficient planning technique was identified for stereotactic radiosurgery treatments on Tomotherapy. Final plan details for the 20 patients studied are shown in Table V. For many patients, delivery time is improved with Tomotherapy over conventional linear accelerator based SRS treatment plans in which multiple isocenters are employed. Duration of linac-SRS delivery can be estimated by multiplying the number of isocenters by the time it takes to deliver each isocenter, which is typically between 15 to 20 minutes in our clinic. Thus, for example, a 4-lesion delivery requiring one isocenter per lesion requires a total delivery time of 60 to 80 minutes. However, many lesions themselves require more than one isocenter and therefore these estimates represents a lower bound on the delivery time of linac-based SRS. Times are typically more than 120 minutes for multiple lesions requiring multiple isocenters per lesion. As can be seen from Table V, the maximum delivery time for a Tomotherapy-based SRS was 66 minutes, with a median time of 31.2 minutes for the group of patients studied, which compares favorably with the estimated lower bounds for the delivery times for linac-SRS. Furthermore, it was always possible to deliver the entire prescribed dose in a single treatment fraction with the plan parameters described here.

Tomotherapy delivery time is quite predictable, given that this is simply the number of gantry rotations multiplied by the gantry period, which, as already mentioned can be manipulated via adjusting the pitch to be in the range of 50–60 seconds. Table II indicates that pitches between 0.13 and 0.18 give a gantry period in this range for delivered doses of 15–24 Gy (with a pitch of 0.13 corresponding to the high end of the dose range). It would appear that even for metastases scattered throughout the longitudinal extent of the brain with

no gaps to allow for multiple plans, the treatment time would be unlikely to exceed 90 minutes, rivaling alternate delivery methods.

Finally, beamlet computation time with the 16-node cluster was not more than a few hours for any patient and should fit into a radiosurgery workflow. It is anticipated that this time will only decrease as computational capabilities improve over time.

Comparison to Conventional Radiosurgery

In order to determine the credibility of any planning technique, it is important to ensure that it reliably produces plans of radiosurgical quality. We aimed to achieve with the Tomotherapy system a conformity that is comparable to pre-existing radiosurgery systems with dose falloff that is as steep as possible. To measure conformity, data for the 59 lesions studied here is compared to data reported by Nakamura *et al.*¹⁵ for 1338 lesions treated with Gamma Knife radiosurgery. They report median target coverage, median PITV, and median inverse conformation numbers of 0.97, 1.67, and 1.78, respectively. For the Tomotherapy data set, these median values were 0.99, 1.59, and 1.66, respectively. In addition Nakamura *et al.*¹⁵, split their data in to four volume quartiles and reported median PITV values for each. The first quartile included volumes ranging from 0.01 to 0.25 ml, the second includes 0.26 ml–1.5 ml, the third includes 1.6 ml–5.5 ml, and finally, the fourth includes 5.6 ml–55.9 ml. They report median PITV ratios of 3.43, 1.85, 1.53, and 1.35 for the four quartiles, starting with the first quartile. Using the same volume cutoffs, we report median ratios of 2.26, 1.51, 1.44, and 1.18 for the same four quartiles, which compare favorably to the Gamma Knife values. Moreover, these results also compare favorably with a study by Gutierrez *et al.*²¹, who made the same comparison for brain metastases treated in combination with whole brain radiation therapy. They report PITV values of 2.22, 1.34 and 1.17 for the second, third and fourth quartiles respectively. Using the conformation number, we also looked at average values for three additional subgroups, < 0.1 ml, 0.1–1 ml, and > 1 ml. Average conformation numbers for the three groups were 3.84, 1.88, and 1.46, respectively. Both results show an improvement of conformity with volume. Although comparatively worse conformity is observed for very small metastases, Nakamura *et al.*¹⁵ note that since no complications were observed for patients with smaller volume lesions (< 1 ml), conformity is not as critical as it is for larger lesions where a much larger volume of normal brain is irradiated with decreasing conformity.

Homogeneity was compared to the original linear accelerator based plans using circular collimators to show that the planning technique described here does allow for adequate inhomogeneity to increase peripheral dose falloff. Results are shown in Table VI. Achieved homogeneity indices compared favorably with linear accelerator based SRS. Median homogeneity for the linac-based SRS and Tomotherapy plans, respectively, were 1.25 and 1.24 with average values of 1.30 and 1.24.

The 12 Gy-volume has been postulated to correlate with the risk of S-NEC^{17–20}. Due to the small size, and predominantly spherical shape of the targets in this patient cohort, the changes in low dose volumes were not dramatic. The average increase for the 60 targets was 1.1 ml (range –1.9 ml to +5.2 ml), with an average total increase of 3.4 ml (range –1.9 ml to +12.1ml) over the 20 patients. Clearly, the increase in the 12 Gy-volume is dependent on number and size of the targets. The effect of these changes on relative risk of S-NEC will be explored in future work.

IV: Discussion

The work presented above shows that our proposed Tomotherapy SRS planning technique allows one to develop plans for multiple brain metastases that can be delivered efficiently in

a time frame that is comparable to or shorter than what is currently achievable using linac-based SRS or Gamma Knife SRS. Multiple sites can be treated within a single plan and that there is no need to have each target positioned at the isocenter as with conventional SRS techniques.

This work only considers the case of brain metastases, which are often well circumscribed spherical targets. It is possible that the planning technique described here might not be applicable for other diseases treated with SRS. For instance for lesions where dose homogeneity is important (for example those infiltrating normal brain) the special techniques described here to achieve a greater degree of inhomogeneity may not be useful. In addition, targets having a complex shape might require higher modulation, and hence pitch will have to be reduced to allow for delivery in a single fraction.

It is important to note that this technique applies to the Tomotherapy optimization algorithm as currently implemented. If there are changes to the algorithm described in equation (1), the described technique will have to be scrutinized and a new planning technique may have to be developed. All beamlet calculation and optimization time estimates only pertain to calculations made with the 16-node cluster currently in operation at our clinic. With a newer cluster design, it would be expected that dose calculation times would be shorter. It may even be possible to calculate on 512×512 resolution images, reducing the voxel size and to improve plan accuracy. We found that we could not increase resolution and calculate beamlets in a reasonable time frame with our current cluster configuration. The addition of dynamic jaws as described by Chen *et. al.*²² could improve delivery efficiency by allowing for increased couch speed between targets in the case of multiple targets. Future work will aim to look at the impact of dynamic jaws on delivery efficiency.

Since all of the plans here were preformed using CT datasets for patients treated with linear accelerator based SRS, detailed comparison of these plans to the clinically used linac-based SRS plans will be considered in future work and is beyond the scope of this technical paper. Here, we report on the detailed development of the planning technique for this nonstandard application of the Tomotherapy system.

V: Conclusions

This work describes the development and evaluation of a new planning technique for stereotactic radiosurgery for brain metastases using Tomotherapy. An efficient planning technique has been described that will in turn allow for efficient treatment delivery. We have presented here one possible solution, of a multitude of possible solutions, which yields reliable results that are consistent with SRS planning goals. This work also establishes the feasibility of a one-day SRS workflow on Tomotherapy since planning time can be kept to 2 hours or less for a number of targets, which compares well with planning times necessary for multiple targets in conventional linear accelerator based SRS.

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Abbreviations

SRS	Stereotactic radiosurgery
TPS	Treatment Planning System
DVH	Dose Volume Histogram

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Figure 1.

Examples of added contours for treatment planning. Small target volumes require only a ring and a very small central subvolume (CSV) while larger tumors require two subvolumes (the CSV and a larger SV), a ring inside the periphery of the volume, and another ring external to the target.

Figure 2.
Dose inhomogeneity, as measured with a homogeneity index, as a function of modulation factor for a single metastasis.

Figure 3.

Change in maximum dose to the target and the minimum dose at 3mm from the target with increasing modulation factor for one target. Results show that as the intra-target hot spot increases, peripheral fall off is steeper.

Figure 4. Beamlet computation time as a function of pitch for one patient with 4 metastases using a clinically configured planning station and cluster.

Figure 5.

Two dose distributions for a 10.5 ml target. Although both dose distributions are inhomogeneous, in panel (a) the highest dose is located peripherally, while in panel (b) it is located centrally. The result on the right is achieved by setting harder constraints on the 2mm outer ring structure (2mm RS), shown in purple, whose outer edge is formed by the periphery of the treatment target, and using the additional larger sub-volume (SV), shown in red, at the center of the target to force a higher dose at the center.

Table 1

Information regarding targets and prescription for test patients.

Patient #	Target #	Target Location	Dose (Gy)	Volume (ml)	Total Volume (ml)
1	1	<i>Rt Post</i>	<i>21</i>	<i>0.52</i>	<i>0.52</i>
2	4	<i>Lt Parietal</i>	<i>18</i>	<i>0.32</i>	<i>6.64</i>
		<i>Lt Parietal</i>	<i>15</i>	<i>5.81</i>	
		<i>Rt Frontal</i>	<i>18</i>	<i>0.16</i>	
		<i>Rt Temp</i>	<i>18</i>	<i>0.35</i>	
3	3	<i>Rt Post Front</i>	<i>21</i>	<i>0.61</i>	<i>3.77</i>
		<i>Lt Medial</i>	<i>21</i>	<i>1.24</i>	
		<i>Rt Ant Front</i>	<i>21</i>	<i>1.92</i>	
4	3	<i>Cerebellar</i>	<i>21</i>	<i>0.45</i>	<i>1.86</i>
		<i>Lt Post Front</i>	<i>21</i>	<i>0.34</i>	
		<i>Lt Ant Front</i>	<i>21</i>	<i>1.07</i>	
5	3	<i>Lt Frontal</i>	<i>18</i>	<i>0.82</i>	<i>2.77</i>
		<i>Rt Parietal</i>	<i>14</i>	<i>0.65</i>	
		<i>Rt Cerebellar</i>	<i>18</i>	<i>1.30</i>	
6	6	<i>Lt Cerebellar</i>	<i>20</i>	<i>0.93</i>	<i>3.23</i>
		<i>Rt Occipital</i>	<i>20</i>	<i>0.60</i>	
		<i>Rt Frontal</i>	<i>20</i>	<i>1.45</i>	
		<i>Lt Frontal</i>	<i>20</i>	<i>0.08</i>	
		<i>Lt FrontPar</i>	<i>20</i>	<i>0.12</i>	
7	3	<i>Rt Temporal</i>	<i>20</i>	<i>0.04</i>	<i>5.78</i>
		<i>Lt Parietal</i>	<i>15</i>	<i>2.48</i>	
		<i>Rt Cerebellar</i>	<i>18</i>	<i>0.04</i>	
8	4	<i>Rt Occipital</i>	<i>15</i>	<i>3.26</i>	<i>0.59</i>
		<i>Lt Cer Post</i>	<i>15</i>	<i>0.07</i>	
		<i>Lt Cer Inf</i>	<i>15</i>	<i>0.40</i>	

Patient #	Target #	Target Location	Dose (Gy)	Volume (ml)	Total Volume (ml)
		<i>Lt Cer Ant</i>	17	0.04	
		<i>Rt Frontal</i>	21	0.07	
9	2	<i>Rt Occipital</i>	20	0.84	7.17
		<i>Lt Frontal</i>	20	6.33	
10	6	<i>Rt Front Up</i>	21	1.79	2.93
		<i>Lt Frontal</i>	21	0.43	
		<i>Rt Front Low</i>	21	0.16	
		<i>Rt Front Med</i>	21	0.03	
		<i>Rt Thalamus</i>	21	0.07	
		<i>Lt Temp</i>	17	0.46	
11	3	<i>Rt Frontal</i>	21	0.15	0.32
		<i>Lt Midbrain</i>	15	0.10	
		<i>Rt Temp</i>	21	0.08	
12	4	<i>Paracentral</i>	15	10.79	11.26
		<i>Rt Occ Lat</i>	21	0.12	
		<i>Rt Occ Med</i>	21	0.21	
		<i>Lt. Parietal</i>	21	0.14	
13	3	<i>Lt. Frontal</i>	18	1.92	2.17
		<i>Rt Cerebellar</i>	23	0.10	
		<i>Rt Paracellar</i>	23	0.15	
14	4	<i>Lt Parietal</i>	21	0.06	0.22
		<i>Lt Post Front</i>	21	0.04	
		<i>Rt Low Front</i>	18	0.10	
		<i>Lt Ant. Front</i>	21	0.02	
15	3	<i>Rt Occipital</i>	21	0.14	0.24
		<i>Lt Occipital</i>	24	0.05	
		<i>Lt Parietal</i>	24	0.05	

Patient #	Target #	Target Location	Dose (Gy)	Volume (ml)	Total Volume (ml)
16	2	<i>Rt Frontal</i>	18	2.61	3.02
		<i>Lt Temp</i>	21	0.41	
17	2	<i>Rt Temp</i>	21	1.62	3.54
		<i>Lt Frontal</i>	21	1.92	
18	1	<i>Lt Cerebellar</i>	15	5.13	5.13
19	2	<i>Lt Frontal</i>	24	1.03	1.23
		<i>Rt Temp</i>	24	0.20	
20	1	<i>Medial</i>	21	1.19	1.19

Table II

Pitch values as a function of the prescribed dose to allow for delivery in a single pass (keeping the gantry period under one minute via lowering pitch) with a minimum of beamlet computation time. In other words, these pitches are as high as possible to stay within the one minute gantry period limit.

Maximum Prescribed Dose (Gy)	Pitch
15	0.18
18	0.17
20	0.14
21	0.13–0.14
24	0.13

Table III

Table IIIa: Initial optimization constraints for tomotherapy radiosurgery planning for small (<2 ml) targets. Note that the target maximum dose is set to a very high dose to avoid any penalty being imposed on target voxels receiving a high dose during optimization.						
Tumor Settings						
Type	I	Max Dose	Max Dose Penalty	DVH %	DVH Dose	Min Dose Penalty
Target	1	120 Gy*	1	100	Prescription	100
CSV**	1	Prescription/0.8	1	100	Prescription/0.8	1
Sensitive Structure settings						
Type	I	Max Dose	Max Dose Penalty	DVH %	DVH Dose	DVH Penalty
3mmRing	1	Prescription	100	50	Prescription/2	1

Table IIIb: Initial optimization constraints for Tomotherapy radiosurgery planning for large (>2 ml) targets. Note that the target structure itself is designated as a sensitive structure.						
Tumor Settings						
Type	I	Max Dose	Max Dose Penalty	DVH %	DVH Dose	Min Dose Penalty
CSV*	1	Prescription/0.8	1	100	Prescription/0.8	1
2mmRS	1	Prescription	1	100	Prescription	100
SV	1	Prescription/0.8	1	100	Prescription/0.8	Same as CSV
Sensitive Structure settings						
Type	I	Max Dose	Max Dose Penalty	DVH %	DVH Dose	DVH Penalty
Target	1	120Gy**	1	100	Prescription	100
3mmRing	1	Prescription	100	50	Prescription/2	1

I=Importance, CSV=Central Subvolume

* Maximal allowed dose to be prescribed

** Indicates one of these structure types is used as the prescription

I=Importance, CSV=Central Subvolume 2mmRS = 2mm Ring Structure

* Indicates that one of these structure types is used as the prescription

** Maximal allowed dose to be prescribed

Table IV

Table shows a comparison in delivery time for a single plan and for multiple plans for patients with a large (>5cm) longitudinal gap between targets. Significant time savings are shown with the use of multiple plans in these cases.

Patient	No. Plans	Active Rotations	Time (min)	Time Saved(min)
4	1	71.9	72.1	40.3
	2	33.2	31.8	
8	1	70.1	58.6	28.9
	2	29.4	29.7	
11	1	58.8	54.1	16.2
	2	38.9	37.9	
16	1	85.3	69.8	39.2
	2	32.3	30.6	
19	1	59.7	59.9	31.8
	2	28.8	28.1	

Table V

Table shows pitch, modulation factor, number of rotations, total length, delivery time and beamlet computation time for 10 patients. It is important to note that the plans were done on version 3.1.4 and time estimates are from a clinically configured DC. For patients where multiple plans were used, multiple values are given if they differed between plans.

Patient	No. of Plans	No. Targets	Max Dose	Pitch	Active Rots.	Mod. Factor	Gantry Period	Delivery Time (min)	DC2 Beamlet Comp Time
1	1	1	21	0.14	12.8	1.7	57	12.3	0:33:31
2	1	4	18	0.17	44.1	1.6	56	41.4	2:46:30
3	1	3	21	0.14	53.4	1.7	55	49.2	2:06:33
4	2	3	21	0.14	33.2	1.7	54,60	31.8	1:28:04
5	1	3	18	0.16	61.1	1.7	56	57.2	1:42:43
6	1	6	20	0.14	74.5	1.7	53	66.0	2:44:00
7	1	3	18	0.17	46.5	1.7	54	42.1	1:52:54
8	2	4	21,17	0.14,0.17	29.3	1.5,1.7	60	29.7	1:08:20
9	1	2	20	0.15	30.7	1.7	53	27.3	2:06:18
10	1	6	21	0.14	60.0	1.7	60	60.0	2:37:09
11	2	3	21	0.14	38.9	1.6	58	37.9	0:51:28
12	1	4	21	0.14	50.0	1.7	50	41.9	3:02:09
13	1	3	23	0.13	29.0	1.7	60	29.2	1:46:47
14	1	4	21	0.14	47.1	1.7	57	44.9	1:03:00
15	1	3	24	0.13	46.3	1.7	60	46.5	0:59:03
16	2	2	21,18	0.13	32.3	1.7	56	30.6	1:57:19
17	1	2	21	0.13	19.2	1.7	53	17.1	2:08:36
18	1	1	15	0.18	16.7	1.7	55	15.5	1:22:36
19	2	2	24	0.13	28.7	1.7	56,60	28.1	1:04:57
20	1	1	21	0.14	13.9	1.7	55	13	0:47:15

Table VI

Comparison of the homogeneity index (maximum dose/prescription dose) achieved for a linear accelerator based planning approaching using circular collimators and helical tomotherapy.

	Linac	Tomotherapy
Maximum HI	2.02	1.52
Minimum HI	1.06	1.07
Median HI	1.25	1.24
Average HI (+/- St. Dev.)	1.30 +/- 0.17	1.24 +/- 0.09