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Improvements in dose calculation accuracy for small off-axis targets in high dose per fraction tomotherapy

Abstract

Purpose: A recent field safety notice from TomoTherapy detailed the underdosing of small, off-axis targets when receiving high doses per fraction. This is due to angular undersampling in the dose calculation gantry angles. This study evaluates a correction method to reduce the underdosing, to be implemented in the current version (v4.1) of the TomoTherapy treatment planning software. **Methods:** The correction method, termed "Super Sampling" involved the tripling of the number of gantry angles from which the dose is calculated during optimization and dose calculation. Radiochromic film was used to measure the dose to small targets at various off-axis distances receiving a minimum of 21 Gy in one fraction. Measurements were also performed for single small targets at the center of the Lucy phantom, using radiochromic film and the dose magnifying glass (DMG). **Results:** Without super sampling, the peak dose deficit increased from 0% to 18% for a 10 mm target and 0% to 30% for a 5 mm target as off-axis target distances increased from 0 to 16.5 cm. When super sampling was turned on, the dose deficit trend was removed and all peak doses were within 5% of the planned dose. For measurements in the Lucy phantom at 9.7 cm off-axis, the positional and dose magnitude accuracy using super sampling was verified using radiochromic film and the DMG. **Conclusions:** A correction method implemented in the TomoTherapy treatment planning system which triples the angular sampling of the gantry angles used during optimization and dose calculation removes the underdosing for targets as small as 5 mm diameter, up to 16.5 cm off-axis receiving up to 21 Gy.

Keywords

per, high, targets, axis, off, small, accuracy, calculation, dose, fraction, tomotherapy, improvements

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Improvements in dose calculation accuracy for small off-axis targets in high dose per fraction tomotherapy

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Purpose: A recent field safety notice from TomoTherapy detailed the under-dosing of small, off-axis targets when receiving high doses per fraction. This is due to angular undersampling in the dose calculation gantry angles. This study evaluates a correction method to reduce the under-dosing, to be implemented in the next version of the TomoTherapy treatment planning software.

Methods: The correction method, termed 'Super Sampling' involved the tripling of the number of gantry angles from which the dose is calculated during optimization and dose calculation. Radiochromic film was used to measure the dose to small targets at various off-axis distances receiving a minimum of 21 Gy in one fraction. Measurements were also performed for single small targets at the center of the Lucy phantom, using radiochromic film and the Dose Magnifying Glass (DMG).

Results: Without super sampling, the peak dose deficit increased from 0% to 18% for a 10 mm target and 0% to 30% for a 5 mm target as off-axis target distances increased from 0 to 16.5 cm. When super sampling was turned on, the dose deficit trend was removed and all peak doses were within 5% of the planned dose. For measurements in the Lucy phantom at 9.7 cm off-axis, the positional and dose magnitude accuracy using super sampling was verified using radiochromic film and the DMG.

Conclusion: A correction method to be implemented in the TomoTherapy treatment planning system which triples the angular sampling of the gantry angles used during optimization and dose calculation removes the under-dosing for targets as small as 5 mm diameter, up to 16.5 cm off-axis receiving up to 21 Gy.

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I. INTRODUCTION:

Helical tomotherapy delivers intensity modulated radiotherapy using a continuously rotating fan beam of 6 MV photons modulated by a 64 leaf binary multileaf collimator (MLC).^{1,2}

50 Helical tomotherapy allows for delivery of highly conformal dose distributions and has recently been studied as a potential delivery mechanism for intracranial stereotactic radiosurgery (SRS).³⁻⁵ SRS is characterized by the delivery of high doses per fraction to small targets. Helical tomotherapy intracranial SRS dose distributions generally have similar homogeneity index to conformal arc dose distributions from linac cone-based systems, have
55 improved dose conformality and reduced dose gradient outside of the target.¹¹ The major advantage of helical tomotherapy for intracranial SRS delivery however is that it offers the potential to deliver SRS to multiple intracranial targets using one treatment setup position, thus vastly increasing the efficiency of SRS delivery.⁵

A recent Field Safety Notice was released by TomoTherapy (#5089, Jan 2011) identifying
60 a deficiency of the dose calculation algorithms in the TomoTherapy Hi-Art treatment planning system (TPS).¹² The notice stated that for small targets, away from machine isocenter receiving a high dose per fraction, the TPS dose calculation over-predicts the dose. The over-prediction increases with decreasing actual modulation factor and decreasing target size and increasing distance away from machine isocenter¹². The magnitude of over-prediction was
65 up to 25% for a 5 mm target, 12 cm off axis with an actual modulation factor of 1. This overestimation is a direct result of the TPS approximating the continuous gantry rotation during delivery as a series of 51 projections per gantry rotation. The gantry period is limited to a maximum of 60 s, therefore when a high dose per fraction delivery is required, the MLC leaves are open for as long as possible. This means that if a leaf is open, it is generally open
70 for close to the full projection time and arc (7.06°). For a target at the machine isocenter, this is not a problem, since the intersecting beamlet is from the central MLC leaf which rotates about that target. However for off-axis targets, the intersecting beamlet does not rotate about the target thus the beamlet scans across the target for the duration of the projection, blurring the dose distribution. In this context, the term 'off-axis' refers to any point
75 that is away from the machine isocenter in the left-right and anterior-posterior directions. The geometry of the situation is shown in Figure 1. The TPS calculation assumes that the beamlet dose is delivered from a stationary location at the center of the projection, for the

duration of the projection. This under-sampling of the angular gantry locations results in the over-prediction of the dose for long leaf open times.

80 Accurate measurement of the dosimetric effect of angular under-sampling in dose calculation is difficult as dosimetric effects only present for small targets. Measurement of doses to small targets is challenging due to the lack of electronic equilibrium, detector size and perturbation effects and large output variation with field size.⁶⁻⁸ This report investigates the magnitude of the dosimetric effect of the angular under-sampling and describes and
85 tests a method for correcting the angular under-sampling in the TomoTherapy TPS dose calculation, to be implemented in a forthcoming version of the TPS. The dose calculation accuracy with and without the correction method was tested using radiochromic film and a high-resolution 1D silicon diode strip detector, the Dose Magnifying Glass (DMG).

II. METHODS & MATERIALS:

90 A. Angular under-sampling correction

The dose over-prediction is a result of the TPS not taking into account the blurring of the dose deposited by beamlets as the gantry rotates and is observed in off axis targets and long
leaf open times. The correction method is to simply increase the number of source positions used for dose calculation during the optimization (every 10th iteration) and during the final
95 dose calculation. The number of projections was increased from 51 to 153, meaning that the beamlet dose is calculated every 2.35° as opposed to every 7.06° .

B. Dosimetric verification

To test the TPS correction method, a series of measurements of the dose to small, off-axis targets were performed. The under-sampling inaccuracy only presents itself in small targets,
100 therefore two small target sizes were investigated - 10 mm and 5 mm diameter spheres. A planning kVCT of the TomoTherapy 'Cheese' phantom was obtained. This was imported into the Pinnacle RTPS (Philips Radiation Oncology Systems, Fitchburg, USA). Four 10 mm diameter targets and four 5 mm targets were delineated. The targets were centered on the same central coronal plane, spaced approximately 5-6 cm apart in the LR direction. The
105 image set and contours were transferred to a research TomoTherapy RTPS system which

utilizes a GPU instead of a cluster for optimization and dose calculation.^{13,14} The phantom was aligned such that the phantom center was 3 cm off-axis in the LR direction and aligned with machine isocenter in the SI and AP directions. This meant that the target spheres were placed 0, 5.5, 10.5, and 16.5 cm off-axis in the LR direction.

Two sets of treatment plans were created. A separate treatment plan was created for each target. The first set of plans, termed 'no SS', used the current, clinical method of 51 projections for the optimization and final dose calculation. The second set, termed 'SS', used 153 projections for the optimization and final dose calculation. Each plan was designed to deliver a peak dose of 23.25 Gy to a target, with a minimum peripheral target dose of 21 Gy. This dose level was selected as it is the desirable dose level used for intracranial SRS in our department. The optimization method derived by Soisson et. al.⁵ was used to obtain peaked dose distributions with a sharp dose fall-off outside of the target. The field width was 1 cm and the pitch was 0.100 and 0.090 for the 10 mm and 5 mm diameter target plans respectively. The modulation factor was varied so that the minimum target dose was at least 21 Gy but no more than 21.25 Gy. Modulation factors were typically close to 1.00, to reduce treatment time and to result in treatment plans in which the MLC leaves, if open, were generally open for the full projection. This is the worst case scenario for the under-sampling problem since the leaf is open for the full projection rotation, thus provides a robust test of the correction method. A total of 30 iterations were run for each plan.

Radiochromic film has been shown to be a useful dosimeter for small radiation fields due to its high spatial resolution and minimal perturbation of the dose.^{9,10} Therefore Gafchromic EBT2 film was used to measure the dose delivered to each target in the Cheese phantom. Separate measurements and films were used for each target delivery. Previous work has shown the utility of EBT2 film for high-dose-per-fraction measurements such as those for SRS.¹⁵ The film was calibrated from 0 - 32 Gy using a 6 MV photon beam from a Varian 600C/D (Varian Medical Systems, Palo Alto, USA). For measurement, each 20.32 x 25.40 cm² sheet of film was cut into five strips of approximately 4.06 x 25.40 cm². The films were aligned with the target to be measured and the full treatment was delivered, with no scaling of the treatment dose. All films were scanned one at a time on a Epson Expression 10000XL in transmission mode, using an opaque template to ensure consistent film location on the scanner bed and to minimize lateral scattering of light from the scanner source. Films were scanned at 72 dpi in 48 bit color mode.

A second set of plans was created for delivery to the Lucy SRS phantom (Standard Imaging, Middleton, USA) mounted to the couch using a Radionics Interfix couch mount (Integra, Plainsboro, New Jersey, USA). Two planning kVCT scans were taken of the Lucy phantom. The first consisted of the Lucy phantom with the film cassette in the coronal plane and a 'dummy' sheet of EBT2 film in place so the film could be visualized on the CT scan. The second image consisted of the Lucy phantom with a custom-made Lucite insert designed to hold the Dose Magnifying Glass (DMG) in the coronal plane such that the detector spanned along the left-right axis. The DMG in the Lucy phantom is shown in Figure 2. Due to the physical extent of the Radionics couch mount and the Lucy frame/phantom apparatus, in all cases, the setup was such that the center of the Lucy phantom was 9.7 cm off-axis in the anterior-posterior direction. The geometry of the angular under-sampling problem is such that an offset in the anterior-posterior direction should have the same effect as an offset in the left-right direction. The position of the phantom centered 9.7 cm off-axis is a realistic target position for patients treated with this frame apparatus.

The DMG is a high spatial resolution, 128-channel, 1D detector strip.¹⁶ Each channel is a 20 μm wide silicon diode. The separation between each channel is 200 μm , such that the 128 channels span a 25.6 mm long strip with a measurement point every 0.2 mm. The DMG is connected to a PC via a TERA chip which allows read out of all 128 channels at up to 500 Hz. For all measurements the read out frequency was 10 Hz. That is, the signal from each channel was obtained every 100 ms.

The response (counts per unit dose) of the DMG was obtained using a 6 MV photon beam from a Varian 600C/D. A uniformity correction is required to give the response of the outside channels relative to the central channels. The detector response in a calibration field is a combination of the individual detector channel sensitivity and the variation in the linac beam profile at the location of a channel of interest. A geometric shift method as described by Wong et. al. was implemented to separate these two components¹⁷. The whole DMG was irradiated with a 5 x 40 cm^2 field at $D_{1.5\text{cm}}$ in solid water with 10 cm solid water for back scatter with the DMG centered in the middle of the radiation field. The DMG was then shifted 1 mm both to the left and right and irradiated with the same beam. This allows each channel to then 'see' the same beam as it's neighboring channels. Using interpolation out from the central channel, the relative responses can then be obtained. The final outcome of this process is the counts per unit dose for each channel that is applied

to each measurement. The measurement data consists of 128 count readings corresponding to each channel with a sample every 100 ms. The DMG has inherent angular dependence of approximately 15% when irradiated edge-on and 7% when irradiated from the back side. The angular dependence has been well characterized such that the relative response is known for radiation incident from every 10° for the full 360° .¹⁷ A simple first order correction was used to account for this. The start gantry angle and gantry rotation period are known from the treatment sinogram. The gantry angle as a function of time was then used to obtain the relative response as a function of time during the delivery which was then corrected. The angular-corrected samples were then summed to give the total accumulated counts per channel. Each channel was then converted to dose using the counts per unit dose conversion factor.

All measurements were delivered three times on a TomoTherapy Hi-Art treatment machine and included a fine MVCT scan for registration purposes. The MVCT slice width was reduced to 1 mm to improve the resolution available for positional alignment.

III. RESULTS:

A. Effect of SS on treatment planning parameters

The expected treatment time was recorded for all cheese phantom plans. Figure 3 shows the treatment duration for the 10 mm and 5 mm diameter targets, both for No SS and SS plans. When SS was not used, the treatment time was relatively constant for the 10 mm targets, and increased slightly with distance away from machine isocenter for the 5 mm targets. When SS was used however, the treatment time is seen to increase significantly with target distance away from machine isocenter, for both sets of targets. This is a direct result of the optimizer, when given the extra dose calculation information from the increased angular sampling, determining that the target volume may be under-dosed thus the treatment time needs to increase to ensure sufficient target dose is achieved. A second point of difference between the plans optimized with and without SS was the level of modulation performed. The plans optimized with SS exhibited a lower actual modulation factor than the no SS plans, with the leaf open time histogram showing a higher proportion of leaves open for the full projection time.

B. Off-axis dose accuracy

200 The peak measured dose was compared with the peak planned dose for each of the off-axis targets planned and delivered on the cheese phantom. The difference, normalized to the planned dose, is shown in Figure 4. Without supersampling turned on, for both targets diameters, as the target center moves away from the machine isocenter, the difference between the planned and measured peak dose increases. That is, as the target moves away from isocenter, the amount by which the calculated dose is over-predicted increases. The over-prediction is greater for the 5 mm target which is expected since the smaller dimensions are more susceptible to dose blurring effects. When supersampling was turned on for both optimization and dose calculation, the measured dose and the calculated doses are in much better agreement, with no visible trend of over-prediction as the target locations move away from the machine isocenter. Both target diameters exhibit reduced agreement between the planned and measured peak doses for the 10.5 cm off-axis target with SS on, where the measured dose is approximately 5% lower than the planned dose. All other SS measured doses agreed with the planned dose within 3%.

C. Lucy phantom

215 Figure 5 shows the planned and film-measured coronal dose planes in the center of the Lucy phantom. The film measured dose in the centre of the target can be seen to be less than the planned dose for the no SS plans. Figure 6 shows the measured and planned left-right dose profiles through the center of the 5 mm and 10 mm targets delivered to the Lucy phantom, with the target center approximately 9.7 cm off-axis. It can be seen that the supersampling used for both optimization and dose calculation again improves the agreement between the measured and the calculated doses. Figure 6(a) and (c) show that the measured SS peak dose is still less than that planned, by approximately 5%, which is similar to the results seen in Figure 4 for the 10.5 cm off-axis target.

225 Figure 6 also shows the positional alignment of the delivered dose. The film-measured dose planes and calculated dose plans at the coronal film plane were registered using the marking pins in the Lucy phantom film cassette. For the DMG measurement, the profiles were aligned using the visible extent of the DMG on the planning KVCT. All measured

dose profiles align well with the planned dose profiles, within the positional error bars which represent the extent of the dose calculation voxels in the treatment planning system.

IV. DISCUSSION

The presented results show that angular under-sampling in helical tomotherapy treatments of small targets away from machine isocenter leads to an over-prediction of the target dose. The difference between the calculated dose and the actual delivered dose increases with distance away from machine isocenter. An improvement to the optimization and dose calculation algorithm, which triples the number of gantry positions at which the dose is calculated, has been shown to remove the trend of over-prediction with off-axis distance for targets as small as 5 mm diameter, up to 16.5 cm off-axis. Although the measurements show that the under-sampling trend was removed, some discrepancies were still observed between the planned and measured dose with SS turned on. We are unable to explain why the measured peak dose at approximately 10 cm off-axis is consistently less than the planned dose. Any remaining differences are most likely due to positional misalignment, since the small target dimension and high dose gradient characteristics of these measurements increase the chance of misaligning the dosimeters. The Lucy phantom measurements (at 9.7 cm off-axis anterior-posterior) can be approximately compared with the Cheese phantom measurements (at 10.5 cm off-axis left-right). Measured with film, the dose discrepancy in the Lucy phantom at 9.7 cm off-axis was reduced from 9.3% to 4.7% (10 mm target) and 17.8% to 0.7% (5 mm target). The dose discrepancy in the Cheese phantom at 10.5 cm off-axis was reduced from 10.9% to 4.3% (10 mm target) and from 16.7% to 3.2% (5 mm phantom).

Figure 7 shows the distance the centre of an individual beamlet travels across a target as a function of the off-axis distance. It can be seen that for a 51 projection dose calculation, when the target is greater than 5 cm away from machine isocenter, the distance travelled is greater than the width of an individual leaf, thus is greater than the spatial resolution of dose delivery. When the number of projections increases to 153, the distance travelled over the target decreases significantly such that it's not until the target is 15 cm away from machine isocenter that the distance travelled exceeds the dose delivery resolution. However, Figure 7 is an approximation based on an infinitesimally small pencil beamlet. In reality the beamlet has a FWHM that increases with distance from the source due to divergence. Thus

as the source rotates, the geometric FWHM of the beamlet intersecting with the target, at the position of the target, at any one time is changing. The inset figure in Figure 7 shows the FWHM of a beamlet at the location of the four off-axis targets as a function of gantry angle, which adds to the lateral dose coverage. The beamlet FWHM was calculated based on a nominal beamlet geometric FWHM of 0.625 cm at iso-center with divergence. The geometric FWHM of the beamlet thus changes with distance from the source. For targets in the range of -16.5 - +16.5 cm off-axis (the full range that the beamlets see from the full 360°), the beamlet FWHM ranges from 0.50 cm to 0.75 cm when 51 projections are employed. Thus, the exact dose coverage deficiency due to the motion of the centre of the beamlet is a combination of gantry angle and off-axis target distance.

The measurements performed in the Lucy phantom show improved agreement between the calculated dose and the measured dose for the plans optimized and calculated when SS is used, using two different dosimeters in a realistic patient geometry. The two different dosimeters were able to be registered to the planning CT such that the positional accuracy of the dose distribution could be verified to within one dose voxel ($0.195 \times 0.195 \times 0.25 \text{ mm}^3$). This verifies both the positional and dosimetric accuracy of the tomotherapy delivery system for the intracranial SRS treatments investigated in this study.

V. CONCLUSION

Tomotherapy's approximation of continuous gantry rotation during delivery using static gantry angles leads to dose calculation discrepancies for small targets positioned away from machine isocenter receiving high doses per fraction. Angular under-sampling during dose calculation leads to an over-prediction of the calculated dose. A correction method which triples the number of gantry angles used in both optimization and final dose calculation has been tested for 10 mm and 5 mm diameter spherical targets up to 16.5 cm away from machine isocenter. The measured doses showed the method removed the trend of over-prediction of the peak dose thus improved the accuracy of dose calculation for small, off-axis single targets receiving high doses such as those seen in stereotactic radiosurgery procedures. There still however remained an unexplained over-prediction of the dose at 10.5 cm off-axis of approximately 5%, which is significantly reduced compared with the plans calculated and delivered without the correction method.

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REFERENCES

- ¹T. R. Mackie, T. Holmes, S. Swerdloff, P. Reckwerdt, J. O. Deasy, J. Y., B. Paliwal, and T. Kinsella, "Tomotherapy: A new concept for the delivery of dynamic conformal radiotherapy," *Med. Phys.* **20**(6), 1709-1719 (1993)
- ²R. Jeraj, T. R. Mackie, J. Balog, G. Olivera, D. Pearson, J. Kapatoes, K. Ruchala, and P. Reckwerdt, Radiation characteristics of helical tomotherapy, *Med. Phys.* **31**(2), 396404 (2004).
- ³J. Penagaricano, Y. Yan, C. Shi, M. Linskey and V. Ratanatharathorn, "Dosimetric comparison of Helical Tomotherapy and Gamma Knife Stereotactic Radiosurgery for single brain metastasis," *Radiat. Oncol.* **1**(1), 1-26 (2006)
- ⁴C. Han, A. Liu, T. E. Schultheiss, R. D. Pezner, Y. Chen, J. Y. C. Wong, "Dosimetric comparisons of helical tomotherapy treatment plans and step-and-shoot intensity-modulated radiosurgery treatment plans in intracranial stereotactic radiosurgery," *Int. J. Radiat. Oncol. Biol. Phys.* **65**(2), 608-616 (2006)
- ⁵E. T. Soisson, P. W. Hoban, T. Kammeyer, J. M. Kapatoes, D. C. Westerly, A. Basavatia and W. A. Tomé, "A technique for stereotactic radiosurgery treatment planning with helical tomotherapy," *Med. Dosim.* **36**(1), 46-56 (2011)
- ⁶A. J. D. Scott, A. E. Nahum, and J. D. Fenwick, "Using a monte carlo model to predict dosimetric properties of small radiotherapy photon fields," *Medical Physics* **35**, 4671-4684 (2008).
- ⁷M. L. Taylor, T. Kron, and R. D. Franich, "A contemporary review of stereotactic radiotherapy: Inherent dosimetric complexities and the potential for detriment," *Acta Onco-*

logica **50**, 483-508 (2011).

⁸I. J. Das, G. X. Ding, and A. Ahnesjö, "Small fields: Nonequilibrium radiation dosimetry," *Medical Physics* **35**, 206–215 (2008).

320 ⁹O. A. García-Garduño, J. M. Lárraga-Gutiérrez, M. Rodríguez-Villafuerte, A. Martínez-Dávalos, and M. A. Celis, "Small photon beam measurements using radiochromic film and monte carlo simulations in a water phantom," *Radiotherapy and Oncology* **96**, 250–253 (2010).

¹⁰E. E. Wilcox and G. M. Daskalov, "Evaluation of Gafchromic EBT film for cyberknife dosimetry," *Medical Physics* **34**, 1967–1974 (2007).

¹¹E. T. Soisson, M. P. Mehta, W. A. Tomé, "A Comparison of Helical Tomotherapy to Circular Collimator-Based Linear-Accelerator Radiosurgery for the Treatment of Brain Metastases", *Am. J. Clin. Oncol.* **34**(4), 388-394 (2011)

¹²Tomotherapy Incorporated, "Field Safety Notice Medical Device Correction #5089", January 26th 2011

¹³W. Lu, "A non-voxel-based broad-beam (NVBB) framework for IMRT treatment planning", *Phys. Med. Biol.* **55**(23), 7175-7210 (2010)

¹⁴Q. Chen, W. Lu, Y. Chen, M. Chen, D. Henderson, E. Sterpin, "Validation of GPU based TomoTherapy dose calculation engine", *Med. Phys.* **39**(4), 1877-1887 (2012)

330 ¹⁵N. Hardcastle, A. Basavatia, A. Bayliss and W. A. Tomé, "High dose per fraction dosimetry of small fields with Gafchromic EBT2 film", *Med. Phys.* **38**(7), 4081-4085 (2011)

¹⁶J. H. D. Wong, M. Carolan, M. L. F. Lerch, M. Petasecca, S. Khanna, V. L. Perevertaylo, P. Metcalfe and A. B. Rosenfeld, "A silicon strip detector dose magnifying glass for IMRT dosimetry," *Med. Phys.* **37**(2), 427-439 (2010)

340 ¹⁷J. H. D. Wong, T. Knittel, S. Downes, M. Carolan, M. L. F. Lerch, M. Petasecca, V. L. Perevertaylo, P. Metcalfe, M. Jackson and A. B. Rosenfeld, "The use of a silicon strip detector dose magnifying glass in stereotactic radiotherapy QA and dosimetry", *Med. Phys.* **38**(3), 1226-1238 (2011)

FIG. 1. Schematic diagram showing the effect of under-sampling of the angular gantry locations. A quadrant of the gantry rotation is shown. For targets at machine isocenter (green), the intersecting beamlet rotates about the center of the target. For targets away from machine isocenter (red), the intersecting beamlet scans across the target since the center of rotation is not at the location of the target. Expected dose profile shapes are shown at the bottom left.

FIG. 2. (a) The DMG in the custom made acrylic insert for the Lucy phantom (b) the DMG mounted in the Lucy phantom.

FIG. 3. Treatment duration for (a) 10 mm diameter targets and (b) 5 mm diameter targets for the cheese phantom plans

FIG. 4. Difference between the planned and the measured peak target doses in the Cheese phantom, normalized to the planned peak dose for (a) 10 mm diameter targets and (b) 5 mm diameter targets. The error bars represent the standard deviation of three measurements.

FIG. 5. Planned and measured dose maps from the Lucy phantom at 9.7 cm off-axis. Dose maps represent the dose in Gy to the central, coronal plane of the Lucy phantom. The white tick marks represent 1 cm increments.

FIG. 6. Planned and measured dose profile through the center of the targets. (a) and (b) show the EBT2 film measured profiles through the 10 mm and 5 mm targets respectively. (c) and (d) show the DMG measured profiles through the 10 mm and 5 mm targets respectively. The error bars on the EBT2 film and DMG data are the standard deviations of three measurements.

FIG. 7. Distance the centre of a beamlet traverses over a target during one projection for 51 and 153 projections. The horizontal dotted line represents the width of one leaf when projected to isocenter. The inset figure shows the beamlet FWHM for each target at each gantry angle. This FWHM is the lateral dose coverage of each beamlet.