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The design, physical properties and clinical utility of an iris collimator for robotic radiosurgery

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Abstract

Robotic radiosurgery using more than one circular collimator can improve treatment plan quality and reduce total monitor units (MU). The rationale for an iris collimator that allows the field size to be varied during treatment delivery is to enable the benefits of multiple-field-size treatments to be realized with no increase in treatment time due to collimator exchange or multiple traversals of the robotic manipulator by allowing each beam to be delivered with any desired field size during a single traversal. This paper describes the Iris variable aperture collimator (Accuray Incorporated, Sunnyvale, CA, USA), which incorporates 12 tungsten–copper alloy segments in two banks of six. The banks are rotated by 30° with respect to each other, which limits the radiation leakage between the collimator segments and produces a 12-sided polygonal treatment beam. The beam is approximately circular, with a root-mean-square (rms) deviation in the 50% dose radius of <0.8% (corresponding to <0.25 mm at the 60 mm field size) and an rms variation in the 20–80% penumbra width of about 0.1 mm at the 5 mm field size increasing to about 0.5 mm at 60 mm. The maximum measured collimator leakage dose rate was 0.07%. A commissioning method is described by which the average dose profile can be obtained from four profile measurements at each depth based on the periodicity of the isodose line variations with azimuthal angle. The penumbra of averaged profiles increased with field size and was typically 0.2–0.6 mm larger than that of an equivalent fixed circular collimator. The aperture reproducibility is ≤0.1 mm at the lower bank, diverging to ≤0.2 mm at a nominal treatment distance of 800 mm from the beam focus. Output factors (OFs) and tissue-phantom-ratio data are identical to those used for fixed
collimators, except the OFs for the two smallest field sizes (5 and 7.5 mm) are considerably lower for the Iris Collimator. If average collimator profiles are used, the assumption of circular symmetry results in dose calculation errors that are $< 1$ mm or $< 1\%$ for single beams across the full range of field sizes; errors for multiple non-coplanar beam treatment plans are expected to be smaller. Treatment plans were generated for 19 cases using the Iris Collimator (12 field sizes) and also using one and three fixed collimators. The results of the treatment planning study demonstrate that the use of multiple field sizes achieves multiple plan quality improvements, including reduction of total MU, increase of target volume coverage and improvements in conformity and homogeneity compared with using a single field size for a large proportion of the cases studied. The Iris Collimator offers the potential to greatly increase the clinical application of multiple field sizes for robotic radiosurgery.

1. Introduction

Traditionally, fixed divergent circular collimators have been employed to shape x-ray beams used for radiosurgery. Such collimators have very low collimator transmission, sharp penumbrae and perfect field size reproducibility. These characteristics are important because large-dose single-fraction or hypofractionated treatments require submillimetric geometric beam delivery accuracy and steep dose gradients in order to spare adjacent organs at risk. The CyberKnife® Robotic Radiosurgery System (Accuray Incorporated, Sunnyvale, CA, USA) is designed to deliver radiation to targets located anywhere in the body. This system consists of a 6 MV X-band LINAC mounted on a robotic manipulator that is aligned continually throughout each treatment fraction by a stereoscopic x-ray image guidance system. The specifics of the system have been described elsewhere (Antypas and Pantelis 2008, Kilby et al 2009). Each treatment beam is shaped using a fixed circular collimator selected from a set of 12 that produce beam diameters ranging from 5 to 60 mm at a nominal treatment distance of 800 mm from the therapeutic x-ray target (table 1). Field sizes are selected manually during treatment planning, and the dose distribution is calculated for each beam using either a correction-based algorithm (ray tracing using effective path length heterogeneity correction) or a Monte Carlo dose calculation algorithm (Hol et al 2008, Ma et al 2008, Muniruzzaman et al 2008, Wilcox and Daskalov 2008). Both of these algorithms assume circular beam symmetry in their current implementation (MultiPlan® Treatment Planning System v3.5, Accuray).

The CyberKnife System allows up to three fixed collimators to be combined in any single treatment plan. Often, a single fixed collimator is used for all beams, with the field size usually representing a compromise between the requirement of achieving high-dose conformity and steep dose gradients (which is better achieved using smaller collimators), and that of minimizing the total number of monitor units (MU) and treatment beams (which is better achieved using larger collimators). Treatments combining two collimators are reasonably common and have been previously reported (King and Cotrutz 2005). A recent treatment planning study has demonstrated that combining two collimators in the treatment of early non-small-cell lung cancer reduces the required total MU by an average of 31% (range, 4–56%) compared with equivalent treatment plans generated using a single field size (Pöll et al 2008). This improvement is achieved by targeting smaller collimators at the periphery of the target volume to maintain high-dose conformity and steep dose gradients, and larger collimators at the central part to reduce the total number of beams and MU needed. Treatment delivery
Table 1. Characteristics of the fixed circular collimators provided with the CyberKnife System.

<table>
<thead>
<tr>
<th>Characteristic</th>
<th>Description</th>
</tr>
</thead>
<tbody>
<tr>
<td>Field diameter at 800 mm SSD</td>
<td>5, 7.5, 10, 12.5, 15, 20, 25, 30, 35, 40, 50, 60 mm</td>
</tr>
<tr>
<td>Source to distal surface distance</td>
<td>400 mm</td>
</tr>
<tr>
<td>Thickness</td>
<td>70–87 mm</td>
</tr>
<tr>
<td>Material</td>
<td>Tungsten (high purity alloy)</td>
</tr>
<tr>
<td>Transmission</td>
<td>&lt;1% at all points cf central axis of unblocked 60 mm field</td>
</tr>
<tr>
<td>Internal bore geometry</td>
<td>Straight for 5 and 7.5 mm, divergent for all others</td>
</tr>
</tbody>
</table>

using multiple fixed collimators with the CyberKnife System is automatically divided into multiple traversals of the robotic manipulator around the patient, one traversal for each of the required collimators. In the first traversal, all of the beams with the first field size are delivered. The collimator is then exchanged for the second size, either automatically, using a robotic collimator exchange system, or manually, and the process is repeated for the second field size and so on for a possible third. This method increases the total robot traversal time compared with a single-collimator treatment, and introduces a treatment pause associated with each collimator exchange. These two factors offset some of the treatment time reduction that may be obtained with multiple field sizes, and, in practice, may deter the use of multiple field sizes. This is also the reason for the practical upper limit of three field sizes per treatment plan set by the vendor.

The rationale for an iris collimator that allows the field size to be varied during treatment delivery is to enable the benefits of multiple-field-size treatments to be realized with no increase in treatment time due to collimator exchange or multiple traversals of the robotic manipulator by allowing each beam to be delivered with any desired field size during a single traversal. In addition, the practical upper limit of three field sizes per treatment could be removed, allowing the full range of field sizes to be added to the degrees of freedom considered by the inverse plan optimization algorithm. A variable aperture collimator producing circular therapeutic fields has been previously described (Graves et al 2007), but this device is designed for small animal radiotherapy using a micro-CT scanner with a beam energy of 120 kVp and is not suitable for radiosurgery using a 6 MV beam. This paper describes an iris collimator for the CyberKnife System based on a novel mechanical design (Echner 2006, 2008), including its design, physical properties and a beam commissioning procedure. The clinical utility of the iris collimator is assessed by comparing treatment plans generated using this collimator with those generated using fixed collimators.

2. Technical design description

2.1. Design goals

The high-level design goals for an iris collimation system were

(a) the radiation fields should be effectively circular, and the range of field sizes should match those currently available using fixed collimators;
(b) radiation transmission through the device should be no greater than with the current fixed collimators;
(c) field sizes should be positioned reproducibly;
(d) any increase in penumbral width relative to that of fixed collimators of equivalent field sizes should be small;
Table 2. Circularity and penumbra results for a 50 mm field size obtained with single-bank cast collimator prototypes. Penumbra is expressed as the average 20–80% width minus that measured with a fixed circular collimator (i.e. the additional penumbra). The transmission penumbra is greater with the MCP96 alloy polygonal collimators than with the tungsten circular collimator so these values can only be used to compare the polygonal alternatives. The ‘50% max/min width ratio’ is the ratio of maximum to minimum diameters of the 50% dose decrement line measured on each film.

<table>
<thead>
<tr>
<th>Depth (mm)</th>
<th>Parameter</th>
<th>6-sided static</th>
<th>12-sided static</th>
<th>24-sided static</th>
<th>6-sided dynamic</th>
<th>Circular</th>
</tr>
</thead>
<tbody>
<tr>
<td>100</td>
<td>50% max/min width ratio</td>
<td>1.14</td>
<td>1.03</td>
<td>1.01</td>
<td>1.01</td>
<td>1.01</td>
</tr>
<tr>
<td></td>
<td>Average penumbra (mm)</td>
<td>1.51</td>
<td>1.52</td>
<td>1.46</td>
<td>2.04</td>
<td>0.00</td>
</tr>
<tr>
<td>200</td>
<td>50% max/min width ratio</td>
<td>1.13</td>
<td>1.01</td>
<td>1.01</td>
<td>1.01</td>
<td>1.01</td>
</tr>
<tr>
<td></td>
<td>Average penumbra (mm)</td>
<td>1.80</td>
<td>1.92</td>
<td>1.80</td>
<td>2.43</td>
<td>0.00</td>
</tr>
</tbody>
</table>

(e) the device should not reduce the available treatment workspace around the patient compared with treatments delivered using fixed collimators;

(f) the additional weight of the device should not degrade the alignment accuracy of the robotic manipulator;

(g) the device should be interchangeable with the current fixed collimators, so that either collimation option can be selected for any patient.

2.2. Device description

2.2.1. Initial concept and evaluation. Initial experiments were conducted to assess the circularity and penumbra requirements by comparing the dose distributions delivered using a 50 mm diameter fixed circular collimator with those delivered using cast collimators formed as 6-, 12- and 24-sided regular polygons using the MCP96 alloy. The collimators were mounted on a conventional 6 MV LINAC (Primus, Siemens Medical Solutions, Erlangen, Germany) and measurements were made using radiochromic film positioned normal to the beam central axis at a distance of 800 mm and at depths of 100 and 200 mm in a solid water-substitute phantom. In addition to static experiments, a dynamic experiment was conducted to investigate the dose distribution achievable with a continuously rotating polygon. This was performed using the six-sided cast collimator rotated in 2° increments through a total angle of 60° about the central axis with an equal MU setting delivered at each angle. The beam circularity was quantified as the ratio of maximum to minimum diameters of the 50% dose decrement line measured on each film. In addition, the penumbra was measured as the average 20–80% dose decrement width. The results are presented in table 2 and show that a six-sided static polygon delivers a substantially non-circular dose distribution, while a six-sided rotating polygon delivers a circular dose distribution but at the cost of increased penumbral width. As expected, circularity improved as the number of sides increased, while the penumbra was very similar for all static designs. The 12-sided static design was determined to achieve the optimum combination of circularity, penumbral width and complexity, and selected for further investigation. The circular collimator also did not deliver a perfectly circular beam, which was probably due to a combination of underlying beam asymmetry and measurement uncertainty.

Two 12-sided iris collimator designs were examined, using either one or two collimator banks (figure 1). These were based on designs originally described by Echner (2006, 2008). The two-bank design was considered likely to offer considerable advantages with respect to
several of the design goals. The most significant of these was that, in this case, radiation leakage between the collimator segments could be limited without the need for a tongue-and-groove or similar design, because the two banks were rotated by 30° with respect to each other. Therefore, the gap between any two collimator segments in the lower bank was always covered by the body of a segment in the upper bank, and vice versa. A mechanical advantage was gained in the two-bank design because the height-to-width ratio of each segment required for the same total collimator thickness was reduced compared to a single-bank design. This ratio affected the susceptibility of segments to deform under torsional stress induced by motion or gravity, and therefore, the two-bank design was likely to improve field size reproducibility and overall device reliability. In addition, eliminating the need for a tongue and groove design reduced the risk of adjacent collimator segments coming into contact with each other during operation, thereby reducing the likelihood of binding. A two-bank, 12-sided static collimator concept was tested using a set of cast alloy collimator experiments similar to those described above. These experiments were performed using both a conventional 6 MV LINAC and a CyberKnife System. The 50% dose diameter ratio was 1.00–1.02 at depths 15–200 mm and source-to-film distances (SFDs) 800 and 1000 mm. Similar measurements performed using a fixed circular collimator gave 50% dose ratio values of 1.00–1.01. Based on these results, a fully mechanical design was produced based on the two-bank 12-sided design concept.

2.2.2. Final design description. The Iris™ variable aperture collimator (Accuray) incorporates twelve 60 mm tall prism-shaped tungsten–copper alloy (18 g cc\(^{-1}\)) segments in two banks of six for shaping the radiation beam (figure 2). The bank on the beam entry side of the collimator is called the upper bank and the bank on the beam exit side the lower bank. The banks are rotated by 30° with respect to each other. Each segment attaches to a cross roller linear bearing table that constrains the segment motion. Because of the linear bearings, the segments do not rotate as the aperture is opened or closed; instead, the segments translate into and out of the beam. Springs are used on the upper bank to tension the assembly in the closed direction to minimize backlash in the drive system. Also attached to the segment assembly is a roller bearing that engages a cam plate in the central part of the Iris Collimator assembly (figure 3). This cam plate has 12 slots, with one slot for each roller bearing. The cam plate is rotated by a single motor which, through the roller bearings, pushes the segment assemblies along the linear bearings. The radial location of the roller bearing in the slot is used to adjust
Figure 2. Illustration of the major mechanical components of the Iris Collimator. (a) External view of the assembled device. (b) A view from beneath the lower segment bank toward the x-ray target, which illustrates how the gap between any two collimator segments in one bank is covered by the body of a segment in the other bank, thereby limiting the intersegment transmission.

the ratio of the opening of the upper bank to the lower bank. This ratio is so set that the upper bank opens to 80% of the aperture of the lower bank to give a stepwise approximation of the beam divergence.

The Iris Collimator aperture is measured by linear variable differential transformer (LVDT) transducers. The LVDT transducers were selected due to their high accuracy and reproducibility, low mechanical wear characteristics, radiation tolerance and low susceptibility to electrical noise. Two transducers are used for redundancy and are driven by two non-adjacent, non-opposing segments in the lower bank. Additionally, two precision limit switches are used to detect contact at the two ends of segment travel (i.e. at the fully closed and fully
Figure 3. A view of the drive mechanism, showing only two segments in each bank. Rotational movement of the cam plate causes the roller bearing attached to the base of each segment assembly to translate along its slot, causing each segment to move along its linear bearing. A single rotation movement of the cam, therefore, forces all 12 segments to move linearly.

Figure 4. Isodose lines for the 40 mm Iris Collimator aperture measured using radiochromic film in a plane normal to the beam axis at SFD 800 mm and depth 15 mm. The 15° radial lines illustrate the dose profiles extracted from each film. The isodoses range from 10% (outermost isodose curve) to 90% in 10% increments.
open positions). These are activated by two other non-adjacent and non-opposing segments in the lower bank. These switches provide a means to calibrate the aperture size and also to ensure that this calibration remains valid over time. During the aperture calibration process, the segments are positioned at each end of travel using the limit switches, and the values of LVDTs are recorded. Before every treatment fraction, the Iris Collimator is moved to the two limits of aperture and the LVDT values are read and compared with the values recorded during aperture calibration. Treatment can start only if the current and recorded readings agree within 0.1 mm (aperture size defined at the lower bank). This provides a quality assurance check of the aperture calibration before every treatment fraction. In addition, the two LVDT values are constantly monitored during the treatment. Their values must correspond to the desired aperture and agree with each other within a tolerance of 0.1 mm. Treatment is interrupted immediately if this check fails. The Iris Collimator controller consists of redundant microcontrollers that process the LVDT and limit switch data to position the segments using a pulse-width modulated motor driver. As a safety mitigation, power to the motor driver is removed when the radiation beam is on. The Iris Collimator controller communicates to the system control PC over a controller area network (CAN) bus connection.

The Iris Collimator assembly attaches to the treatment head through the use of a pneumatically actuated exchange mechanism with three precision clamping assemblies and redundant back-up latches. A 22.2 mm thick tungsten alloy shielding disk is located between the exchange mechanism and the segments, adjacent to the exchange mechanism. This disk is used to minimize leakage in the patient plane. A stainless steel shell encapsulates the mechanical assembly for rigidity, alignment and protection of the internal components. In combination with the robotic manipulator, this mechanism allows the entire Iris Collimator assembly to be exchanged with fixed collimators without manual handling.

The distal face of the Iris Collimator (i.e. the bottom surface of the lower bank) is at the same distance from the x-ray target as the distal face of a fixed collimator, which is 400 mm. However, the outer dimensions of the Iris Collimator assembly are slightly larger than the fixed collimator housing. The increased outer dimensions do not reduce the number of positions (nodes) from which the treatment beams can be delivered, and the CyberKnife System’s proximity detection software routine adjusts its geometric model that checks for collision avoidance depending on the type of collimator in use. These checks are performed in both the treatment planning and treatment delivery software packages to ensure that no part of the treatment delivery system conflicts with a three-dimensional avoidance zone within which the patient and treatment couch are positioned. Both collimator types are also equipped with contact sensors as additional safety measures. The range of non-coplanar beam directions available with the Iris Collimator is identical to that available with fixed collimators.

The complete Iris Collimator assembly—including the exchange mechanism, shielding disk and the external shell—has a mass of 21.5 kg, compared with 14.2–14.6 kg for the equivalent assembly containing a fixed collimator. This is within the payload capacity of the robotic manipulator.

3. Physical properties

3.1. Radiation characteristics

3.1.1. Method. The radiation beam properties of the Iris Collimator were measured using radiochromic film placed between solid water-substitute phantom slabs. The high resolution, fast data acquisition time (<8 Gy exposures) and energy independence of GAFChromic® EBT film (ISP Corporation, NJ, USA) were utilized to investigate and characterize the circularity,
penumbra and collimator transmission of the Iris-collimated beam (Butson et al 2003, 2006). Analysis codes were written in Matlab® (The MathWorks, Natick, MA, USA) to convert the 16-bit film images scanned at 300 dots per inch (dpi) in the red channel of an Epson V700 RGB optical transparency scanner (Seiko Epson, Nagano, Japan) into dose maps. Raw scan images were pre-processed using an 11 × 11 pixel median filter to remove noise. The size of the median filter was selected to correspond roughly to the size of the stereotactic diode detectors that are routinely used to acquire the beam commissioning data for CyberKnife LINACs because some of our investigations involved comparisons of film-acquired dose profiles against those acquired with a diode detector in a water phantom. Visual inspection of the filtered images was performed using a computer animation to overlay the raw and filtered image profiles extracted along successive horizontal lines in order to verify that the median filter did remove the undesirable noise without altering the field profile shapes. After the filtered image is converted to a dose map, the 50% isocontour is extracted on a subpixel scale using two-dimensional (2D) interpolation. The geometric center of the 50% isocontour is selected as the beam’s central axis location.

3.1.2. Circularity. The full range of field sizes were examined using film exposures at 800 mm SFD and at 15, 100 and 200 mm depths. Radial dose profiles were extracted from each film at 15° intervals (figure 4). Since the position of the central axis was computed on a subpixel scale, it was therefore not coincident with one of the grid points of the film image file. Consequently, the radial profiles must also be extracted via 2D interpolation. The rms deviation of the 50% dose radius for each field size is shown in figure 5, along with similar measurements for fixed circular collimators. The fixed collimators do not produce perfect circularity, as was previously observed in experiments conducted on a conventional LINAC (section 2.2.1). The observed increase in rms deviation at very small field apertures for both fixed and Iris collimators is influenced by the inherent limitation in scan resolution (e.g. for a 5 mm field size a 1% deviation is equivalent to 0.05 mm but a 300 dpi scan has

![Figure 5. Comparison of the 50% isodose circularity for fixed and Iris collimators measured at SFD 800 mm and depths 15, 100 and 200 mm.](image-url)
0.085 mm pixel size). Although the non-circularity is greater for the Iris Collimator than for the fixed collimator for all field sizes, the typical rms deviation of the 50% dose radius about the mean was <0.8% of the mean radius, and this corresponds to a variation in the 50% dose radius ranging from <0.02 mm for a 5 mm aperture to <0.25 mm for a 60 mm aperture. The implications of this residual non-circularity in beam shape for beam commissioning and accuracy of dose calculation are considered in section 3.3.

3.1.3. Penumbra. Penumbra was measured using a set of film experiments similar to those described in 3.1.2 as the 20–80% dose decrement width along each radial dose profile, and the average assessed over all dose profiles at each depth. The average penumbra data demonstrated that the Iris Collimator penumbra increased with field size and was typically 0.2–0.6 mm larger than that of an equivalent fixed circular collimator (figure 6). This was an expected consequence of the stepwise approximation of a divergent collimator shape because of the increase in transmission penumbra. In addition, the penumbra along any individual radial direction was found to differ by a few tenths of a millimeter compared with the profile along different radial directions (rms deviation approximately 0.1 mm at 5 mm field size, increasing to about 0.5 mm at 60 mm field size). This is a result of the non-circularity in the beam shape, and its implications for beam commissioning and dose calculation accuracy are considered in section 3.3.

3.1.4. Collimator transmission. The collimator transmission was measured using EBT film at SFD 800 mm and depth 15 mm, using an exposure of 50000 MU. The measurements were performed with the Iris Collimator both fully closed and open to the maximum aperture setting while the central axis was blocked, and also for a blank (i.e. without a beam aperture) fixed collimator of 87 mm thickness (figure 7). The maximum leakage rate beneath a segment was found to be 0.05% of the dose rate measured on the central axis of an open maximum
Figure 7. Measurement of Iris Collimator transmission using radiochromic film at SFD 800 mm and 15 mm depth. (a) Film image showing the leakage pattern for a closed Iris aperture. The central dark spot arises because the aperture cannot be fully closed. The small residual aperture at the center of the closed Iris Collimator was blocked using lead wire. (b) Dose profiles extracted at the two positions shown in (a) (profile 1 through central axis, profile 2 below profile 1), and a similar film obtained with a blank fixed collimator. The central peak in the Iris collimator profile is explained by the inability of the aperture to fully close, and resulting transmission through the lead wire used to block this region.

field size, compared with 0.12% for a fixed collimator. A pattern of increased inter-segment leakage was observed, but the maximum leakage in this region rose only to 0.07% as a result of the rotated two-bank design. Therefore, the secondary collimator transmission through the Iris Collimator was reduced by a factor of approximately 2 at all points relative to the fixed collimators, owing to the greater tungsten thickness and the two-bank design. It should also
be noted that the blank fixed collimator represents the maximum thickness of any of the fixed collimators, which range from 70 to 87 mm.

3.2. Mechanical characteristics

3.2.1. Field size reproducibility. The Iris Collimator control system is designed to achieve an aperture reproducibility of \( \leq 0.1 \text{ mm} \) at the lower bank, diverging to \( \leq 0.2 \text{ mm} \) at a nominal treatment distance of 800 mm from the beam focus. The reproducibility achievable with each system is assessed using a machine vision system (ATS Automation Tooling Systems, Cambridge, ON, Canada) prior to shipment by the vendor. This system measures the distance between all three opposing pairs of segments in the lower bank over repeated runs, between each of which the aperture is opened and closed by the control system. For each run, the average distance is calculated. The maximum deviation allowed between this measurement in any single run and the nominal aperture setting is 0.1 mm.

3.2.2. Pointing accuracy. The net difference in mass between the Iris collimation subassembly and the fixed collimator subassembly is about 7 kg. To compensate for this increase in weight, the robotic manipulator uses different positional offset corrections during treatment with the Iris or fixed collimators. The differences in the two sets of corrections are typically less than 0.5 mm throughout the manipulator workspace. After these corrections are applied, the overall system targeting accuracy, as assessed by an end-to-end test (Antypas and Pantelis 2008, Yu et al 2004), has been found to satisfy the same acceptance test criteria as the fixed collimators (total targeting error \( \leq 0.95 \text{ mm} \)).

3.3. Treatment planning system commissioning

3.3.1. Commissioning procedure. For a regular dodecagon, the average radius can be analytically computed as a function of the length of the radial line segment that is the perpendicular bisector of one of the sides of the polygon. Therefore, a linear superposition of the minimum and maximum radial profiles with suitable weighting coefficients could yield the correct average profile. Such an approach, if possible, is attractive because the number of water phantom scans required in a plane transverse to the beam axis would be only two. The current vendor-recommended approach to generating the representative off-axis profile for the fixed circular collimators involves averaging the data from two orthogonal scans (0\(^\circ\) and 90\(^\circ\)) in order to incorporate the underlying beam asymmetry. For a 12-sided beam, one could use the same approach to sample two very different regions of the beam while simultaneously acquiring the maximum and minimum profiles by scanning at 0\(^\circ\) and 105\(^\circ\). However, this approach does not work because the Iris Collimator comprises two, stacked, six-sided apertures. There is a symmetry-breaking aspect to this arrangement because half of the 12 sides are defined by the upper hexagonal bank and the other half by the lower half. The 50% isodose contour may appear as a regular dodecagon having a cyclical pattern of maximum and minimum radii with a period of 30\(^\circ\) as one moves azimuthally about the beam axis; however, an examination of the lower isodose contours shows the existence of a hexagonal pattern with a period of 60\(^\circ\) superposed in phase with the primary period of 30\(^\circ\) (figure 8(a)). Figure 8(b) is a stylized representation of the isodose lines which shows that over a single 60\(^\circ\) period, four distinct profiles are encountered:

(a) large radius and large penumbra,
(b) small radius and large penumbra,
Figure 8. The radial distance of isodose lines as a function of azimuthal angle: (a) measured data for a 40 mm Iris Collimator field obtained using 120 water phantom scans showing an approximately 12-sided (30°) periodicity is observed at higher doses (40–80%) decreasing to a 6-sided (60°) periodicity at lower doses, and (b) a stylized representation showing the existence of four distinct combinations of penumbra and beam radius.

(c) large radius and small penumbra,
(d) small radius and small penumbra.

These four profiles occur in the sequence described above and are separated by 15° intervals around the beam axis, i.e. 0°, 15°, 30° and 45°. Measurement noise and the underlying beam asymmetry, as demonstrated by the 90% isodose in figure 8(a), obscure this idealized pattern in the measured data. The four profiles are not special, in that any set of four profiles $0° + \Delta \theta + r \times 60°$, $15° + \Delta \theta + s \times 60°$, $90° + \Delta \theta + t \times 60°$, $105° + \Delta \theta + u \times 60°$, where $r, s, t$ and $u$ are integers, will in principle also sufficiently sample the profile space. The vendor’s recommended approach to acquiring the representative equivalent circular profile is the simple (equally weighted) averaging of profiles measured at 0°, 15°, 90° and 105° (figure 9), which is based on combining this observed pattern with the need to sample the underlying beam asymmetry. Data collected using radiochromic film images of the beams show that any arbitrary angle added to this set of four angular positions also produces a suitable set that will yield the same average profile (e.g. 5°, 20°, 95° and 110°).
Figure 9. Beam profiles for the 60 mm Iris Collimator aperture measured at SSD 800 mm and depth 15 mm. (a) The full beam profiles measured at 0°, 15°, 90° and 105°. (b) A magnified view of the individual profiles and the average of these four profiles. The maximum distance to agreement between the average and any individual profile is approximately 0.5 mm.

In all other respects (i.e. output factor (OF) and tissue-phantom-ratio (TPR) data acquisition) the commissioning procedures for the Iris Collimator are identical to those used for fixed collimators as described elsewhere (Antypas and Pantelis 2008, Sharma et al 2007). The only substantial difference observed between these measured datasets for Iris and fixed collimators is that the OFs for the smallest field sizes (5 and 7.5 mm) are considerably lower for the Iris Collimator. Relative to the 60 mm collimator, the 5 mm field size OF is typically around 0.57 for the Iris Collimator compared to around 0.71 for the fixed collimator, and at 7.5 mm field size, these values become approximately 0.78 and 0.84, for Iris and fixed collimators, respectively, based on measurements made with a PTW 60012 diode detector (PTW, Freiburg, Germany). This is because the total collimator thickness is much greater for the Iris Collimator (120 mm) than for small fixed collimators (about 70 mm), while the source-to-distal collimator surface is identical (400 mm). For the smallest field sizes, the x-ray focal spot is partially obscured at the OF measurement point, and because of the difference
in collimator thickness, this effect is greater for the Iris Collimator than the equivalent fixed collimator.

3.3.2. Dose calculation accuracy. Beam models of the Iris Collimator were generated and commissioned within the treatment planning system using the method described above. Dose calculations for single beams in a uniform solid water-substitute phantom were compared against EBT film measurements made in a plane normal to the beam axis at 100 mm depth and SFD 800 mm. For the largest, and therefore, most non-circular field size of 60 mm, 98.4% of pixels passed a 1 mm/1% gamma criterion and the mean distance to agreement was 0.1 mm. For a 10 mm field, these values were 99.0% and 0.1 mm, respectively.

The impact of the circular symmetry assumption for dose calculation of multiple overlapping treatment beams in clinical treatment plans was assessed by recalculating a series of treatment plans using a modified dose calculation algorithm to simulate the effect of a 12-sided polygonal beam. This was performed using a standard ray-tracing algorithm in which TPR, OF and off-center ratios (OCRs) measured with fixed circular collimators were used to calculate the dose at each point P. To simulate a polygonal beam shape, the radial distance, \( r' \), used to look-up the OCR value for dose calculation at point P was generated as \( r' = r \left( \frac{r_c}{r_{edge}} \right) \), where \( r \) is the geometric distance from the central axis to P, \( r_c \) is the radius of a circular field with the same area as the polygonal field and \( r_{edge} \) is the distance from the central axis to the polygonal field edge along a radius passing through P. This has the effect of achieving a 12-sided isodose distribution, and also stretching the penumbral width near the polygonal vertices while compressing it near the flat sides, so that the non-circularity of each beam is exaggerated. Five treatment plans were calculated in this manner, including isocentric and non-isocentric cases with field sizes ranging from 20 to 60 mm. The maximum difference in isodose line position observed at any point compared with a circular beam dose calculation was 0.5–1.0 mm in each case. None of these plan comparisons demonstrated any clinically significant differences between the circular and polygonal dose calculations. It should be noted that the polygonal approximation dose calculation algorithm was used for this experiment only, and is not included in the vendor-supplied treatment planning system.

4. Treatment planning study

4.1. Method

Nineteen treatment plans were generated using the Iris Collimator, and also using one and three fixed collimators. Each plan was generated based on a clinical case that had previously been treated with the CyberKnife System using a single fixed collimator, and the same field size was used for the single-collimator plans generated in this study. With the Iris Collimator, all 12 possible field sizes were included in the optimization process. For the three-collimator treatment plans, the field sizes were selected manually based on the frequency distribution of field size usage in the Iris Collimator plan. The one and three collimator selections were made manually based on clinical experience, and the optimality of these collimator selections for the test cases studied is not guaranteed. Clearly, this would require generating plans with each of the 12 collimators, and each combination of 3 collimators, for each case. For our study, the collimator selection by human experts leaves a slight uncertainty with respect to the one and three collimator results. However, the difficulty in establishing the optimum collimator subset reflects a real problem in treatment planning, where combinatorial optimization over all subsets is clearly impractical. Hence, this demonstrates an advantage to optimization using all 12 collimator sizes with the Iris Collimator.
Table 3. Overview of the treatment planning cases. In each case, dose constraints were applied to the target volume, relevant organs at risk and shells surrounding the target volume to ensure that all solutions were at least comparable to the original, baseline single-collimator treatment plan in all respects. In addition, a constraint was placed on the total MU setting on the same basis.

<table>
<thead>
<tr>
<th>Primary optimization objective</th>
<th>Case name</th>
<th>Description</th>
<th>Target volume (cm³)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Minimize total MU</td>
<td>Lung 1</td>
<td>Peripheral lesion</td>
<td>35.9</td>
</tr>
<tr>
<td></td>
<td>Lung 2</td>
<td>Peripheral lesion</td>
<td>6.0</td>
</tr>
<tr>
<td></td>
<td>Lung 3</td>
<td>Central lesion</td>
<td>104</td>
</tr>
<tr>
<td></td>
<td>Lung 4</td>
<td>Central lesion</td>
<td>77.3</td>
</tr>
<tr>
<td></td>
<td>Pancreas</td>
<td>Pancreas</td>
<td>159</td>
</tr>
<tr>
<td></td>
<td>Liver</td>
<td>Liver lesion</td>
<td>877</td>
</tr>
<tr>
<td>Maximize target coverage</td>
<td>Spine 1</td>
<td>T-spine vertebral body</td>
<td>10.0</td>
</tr>
<tr>
<td></td>
<td>Spine 2</td>
<td>C-spine extradural lesion</td>
<td>0.7</td>
</tr>
<tr>
<td></td>
<td>Spine 3</td>
<td>T-spine vertebral body</td>
<td>10.0</td>
</tr>
<tr>
<td></td>
<td>Spine 4</td>
<td>T-spine vertebral body</td>
<td>12.0</td>
</tr>
<tr>
<td>Maximize conformity (minimize CI)</td>
<td>Acoustic neuroma 1</td>
<td>Acoustic neuroma</td>
<td>0.4</td>
</tr>
<tr>
<td></td>
<td>Acoustic neuroma 2</td>
<td>Acoustic neuroma</td>
<td>0.8</td>
</tr>
<tr>
<td></td>
<td>Head &amp; neck</td>
<td>Base of tongue</td>
<td>43.4</td>
</tr>
<tr>
<td></td>
<td>Brain metastasis</td>
<td>Single brain metastasis</td>
<td>2.2</td>
</tr>
<tr>
<td>Maximize target uniformity (minimize HI)</td>
<td>Prostate 1</td>
<td>Prostate gland</td>
<td>74.2</td>
</tr>
<tr>
<td></td>
<td>Prostate 2</td>
<td>Prostate gland</td>
<td>93.7</td>
</tr>
<tr>
<td></td>
<td>Prostate 3</td>
<td>Prostate gland</td>
<td>74.7</td>
</tr>
<tr>
<td></td>
<td>Prostate 4</td>
<td>Prostate gland</td>
<td>72.5</td>
</tr>
<tr>
<td></td>
<td>Prostate 5</td>
<td>Prostate gland</td>
<td>53.9</td>
</tr>
</tbody>
</table>

For each field size, a number of candidate beams were considered in the optimization problem. This number varied with the total number of field sizes considered, i.e. for a single field size there were 1200 candidate beams, for 3 sizes there were 1000 candidate beams for each size (3000 total candidate beams) and for 12 field sizes there were 500 candidate beams per aperture size (6000 total candidate beams). The candidate beams were distributed evenly across the treatment nodes, and the beams from each node targeted randomly selected points on the target volume surface.

In each case, beam weights were optimized using a sequential multiple objective optimization algorithm (Schlaefer and Schweikard 2008). This optimization method enabled the definition of maximum dose and MU constraints that could not be violated by any plan solution. The solution was then constructed through a sequence of optimization steps. Each step involved finding the global minimum of a single-term cost function corresponding to a dose objective for any one volume, or setting the objective to the total MU setting. The resulting minimum value was applied as an additional constraint, with optional relaxation, to all further steps. This entire series of constraints and optimization steps, collectively termed an optimization script, was defined as appropriate for each clinical case and then was used to generate a treatment plan using both the fixed and Iris collimators for each case.

The clinical cases, target volumes and optimization objectives associated with each case are described in table 3. The cases are grouped into four categories by their primary planning objective.

(a) Minimize total MU setting—applied to a selection of lung, pancreas and liver cases.
(b) Maximize target volume coverage—applied to a selection of spinal treatment cases.
(c) Maximize the conformality of the prescription isodose to the target volume—applied to intracranial and head and neck cases. These plans were analyzed using a conformity index (CI) defined as the ratio of total volume of the prescription isodose to target volume enclosed within the prescription isodose (a perfect value is unity).

(d) Minimize dose heterogeneity across the target volume—applied to prostate cases. These plans are analyzed in terms of a homogeneity index (HI), defined as the ratio of the maximum point dose within the target volume to the prescription dose. In each case, dose is prescribed to an isodose line achieving at least 98% PTV coverage.

Because the beams are targeted randomly, each plan was generated 10 times to obtain an average result for the planning objective with each collimation option.

4.2. Results

The results for these four categories are summarized in figure 10. Figure 10(a) shows that a reduction in total MU of 30–75% is generally achievable when multiple field sizes are combined compared to a single field size. The majority of the MU reduction is achieved when three field sizes are used, but further improvements of up to 6% are observed in 4/6 cases
when all 12 field sizes are employed. These cases correspond to a range of target volume from 6.0 to 104 cm³. The distribution of weighted (non-zero MU) beams between field sizes after optimization for a typical case (Lung 4) is illustrated in figure 11(a), which shows that ten of the available field sizes were utilized by the optimization algorithm. The obvious exception to the trend of decreasing MU is the liver lesion, in which the MU setting increased slightly as the number of field sizes increased. This is a very large target volume (877 cm³) with a maximum dimension (approximately 140 mm) much greater than the largest field size (60 mm), and thus, primarily only the largest field size was utilized by the optimization algorithm, as shown in figure 11(b). Therefore, when one field size is primarily selected during optimization, a single-collimator treatment plan is expected to provide a better solution than a 12 field size solution, simply because the number of candidate beams with that field size is much larger (1200 versus 500).

Figure 10(b) shows that target volume coverage improvements of 5–10% were achieved for vertebral body targets (spine cases 1, 3 and 4) when all 12 field sizes were used. In each case, a continued improvement was observed as the number of field sizes increased. Figure 11(c) shows a typical distribution of weighted beam sizes in these cases. Spine case 2, which is a very small extra-dural target volume (0.7 cm³) located within 1 mm of the spinal cord, shows no improvement with increasing number of field sizes. In this case, the plan required a very steep dose gradient that effectively limited the optimization algorithm to selecting just the smallest field sizes (as shown in figure 11(d)), and therefore, using more than just the 5 mm field size conferred little benefit.
Iris collimator for robotic radiosurgery

Figure 11. Distribution of weighted (non-zero MU) beams among field sizes when all 12 sizes were available during plan optimization. (a) Lung 4, (b) liver, (c) spine 1, (d) spine 2, (e) head and neck and (f) prostate 3. Each plan was generated ten times. The plots show the mean value from all 10 optimizations as a thick horizontal bar, the 25–75% range as a box and the full range as an error bar.

Figure 10(c) demonstrates substantial conformity improvements in 2/4 cases when multiple field sizes were used. In the head and neck case, which has the largest and most complex target volume, CI was reduced from 1.78 with 1 field size to 1.35 with 12 field sizes. A wide range of weighted beam sizes was utilized in the Iris Collimator plan, which is shown in the weighted beam distribution for this case in figure 11(e). For the two acoustic neuroma cases, there was no benefit associated with multiple field sizes for the same reason as with the spine case 2 discussed above.

Finally, figure 10(d) illustrates that substantial improvement in target volume HI was achieved in every case when multiple field sizes were used. The prescription isodose as a percentage of the maximum point dose increased from 79–84% for the single field size plans to 90–92% when 12 field sizes were used, resulting in ≤10% dose variation between the maximum point dose and the dose encompassing 98% of the PTV in each case. A wide range of weighted beam sizes was utilized in the Iris Collimator plans, which is shown in the weighted beam distribution for one prostate case in figure 11(f).

5. Discussion

This paper describes a novel iris collimator design that employs two banks of six collimator segments rotated by 30° with respect to each other to produce a 12-sided polygonal treatment beam for the CyberKnife Robotic Radiosurgery System. This design achieves the high level
design goals outlined in section 2.1, as demonstrated by the data presented in sections 2.2.2 and 3, yet is fundamentally simple. In particular, the rotated two-bank design has very low collimator transmission and effectively eliminates leakage between the collimator segments without the need for a tongue-and-groove or similar design, and the drive mechanism allows all 12 segments to be positioned using a single motor.

The 12-sided Iris Collimator beam projection results in a small non-circularity in the dose distribution delivered to a plane orthogonal to the beam axis. This is observed as an rms deviation in the 50% dose radius of \(<0.8\%\) (corresponding to \(<0.25\) mm at the 60 mm field size) and a variation in the penumbral width of approximately 0.1 mm (rms) at the 5 mm field size, increasing to approximately 0.5 mm (rms) at 60 mm. A method has been described by which the average dose profile can be obtained from four profile measurements at each depth based on the periodicity of the isodose line variations with azimuthal angle, and therefore, a slightly different commissioning procedure is required for the Iris Collimator compared with fixed circular collimators. If average collimator profiles are used, the assumption of circular symmetry results in dose calculation errors that are generally \(<1\) mm or \(<1\%\) for single beams across the full range of field sizes. Single beam comparisons are expected to show the worst-case differences between the circular approximation and the actual Iris Collimator beam. For treatment plans involving many non-coplanar beams, which are the norm for the CyberKnife System (e.g. a typical CyberKnife radiosurgery treatment plan has 100–200 beams), the polygonal nature of an individual Iris Collimator beam is expected to be blurred in the overlapping of multiple beams incident upon the lesion, resulting in a combined error due to the circular beam approximation that is generally smaller than the single-beam error. This is consistent with treatment plans generated using a standard circular beam model and the 12-sided beam shape approximation, which show worst-case isodose line differences of 0.5–1.0 mm.

Other than profiles, the Iris Collimator beam characteristics are not substantially different from those of the corresponding fixed collimator field sizes, with two exceptions:

(a) The small field (5 and 7.5 mm) OFs are considerably lower for the Iris Collimator because the greater height of the Iris Collimator obscures more of the focal spot at small apertures.  
(b) The collimator transmission is significantly lower at all points under the Iris Collimator, resulting in a reduction of greater than 50% in the secondary collimator leakage dose rate. This is due to a combination of the greater thickness of tungsten (120 mm with Iris Collimator versus 70–87 mm with fixed collimators) and the rotated dual-bank Iris Collimator design, which prevents any substantial leakage between the shielding segments.

The results of the treatment planning study demonstrate that the use of multiple field sizes leads to a range of plan quality improvements, including reduction of total MU, increase of target volume coverage and improvements in conformality and homogeneity compared with using a single field size for a large proportion of treatment indications. Importantly, total MU reductions of 30–75% are achievable for the majority of cases in which this is the primary planning objective. The subset of cases in which no significant plan quality improvement was conferred by multiple field sizes involved either very small (0.4–0.8 cm$^3$) or very large (877 cm$^3$) target volumes, which resulted in primarily the smallest or largest field sizes being selected by the beam weight optimization algorithm. In every other case (target volumes 2.2–159 cm$^3$), the use of multiple field sizes provided a better plan quality than that achievable with a single fixed collimator. Much of the plan quality improvement in each case could be obtained using just three field sizes, with a smaller incremental increase offered by using all 12 field sizes. Although a three-field size treatment can be delivered using fixed collimators,
there is a considerable treatment time advantage of using the Iris Collimator because in this case the robotic manipulator performs only one traversal as opposed to three, and no pause is required to exchange collimators during treatment.

Practical experience with the Iris Collimator obtained since it entered routine clinical use in July 2008 has confirmed these benefits. Fuller et al (2008) have previously described the delivery of heterogeneous dose distributions using two fixed collimators for localized prostate cancer, which involves preferentially boosting the periphery of the prostate while sparing the urethra. They have recently compared treatment plans generated by this two-collimator approach with those generated using the Iris Collimator and demonstrated that total MU reduction (mean 25% in 10 cases), treatment beam reduction (mean 27%) and increase in achievable boost volume (mean increase of 10% in the prostate volume receiving 150% or more of the prescription dose) are obtained with Iris Collimator (Fuller et al 2009). Even greater reductions in MU and treatment beams have been shown for head and neck, lung, pancreas and axilla treatments compared with plans generated using fixed collimators (Lee et al 2009). Unlike the results presented in this work, the planning comparisons reported by both Fuller et al (2009) and Lee et al (2009) include a change in the plan optimization algorithm as well as the collimation system.

The Iris Collimator design allows, in principle, the delivery of a continuous range of field sizes rather than the currently available discrete set of 12. The pattern of plan quality improvements observed in this study demonstrates an incremental decrease in benefit as the number of available field sizes increases. This suggests that a relatively smaller clinical advantage would be gained by increasing the number of field sizes beyond 12 within the current range of 5–60 mm. The results do suggest, however, that increasing the maximum field size to a value >60 mm is likely to be beneficial for the treatment of larger target volumes (with a threshold volume likely to be somewhere in the range 159–877 cm³). Such an aperture increase would require a re-evaluation of the number of collimator segments required to form an effectively circular beam, which may impact the overall design.

The Iris Collimator is designed to complement fixed collimators, and the automated exchange mechanism, Xchange™ Robotic Collimator Changer (Accuray), enables the two collimation systems to be switched at any time between treatments without manual handling. The larger OFs of the two smallest fixed collimators (5 and 7.5 mm) compared with the corresponding Iris Collimator field sizes suggest that treatments delivered using just a small field size (such as for very small target volumes within the central nervous system) may be better delivered with a fixed collimator owing to the shorter treatment time and also the perfect field size reproducibility. However, for all other indications, it is possible that the improvements in plan quality associated with the routine use of multiple field sizes made possible by the Iris Collimator may result in a declining use of fixed collimators for CyberKnife Systems equipped with both options. Of course, the practical significance of the Iris Collimator will only be fully established after extensive user experience with this new collimation system.

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