Independent monitor unit calculation for intensity modulated radiotherapy using the MIMiC multileaf collimator

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(Received 12 February 2002; accepted for publication 6 June 2002; published 19 August 2002)

A self-consistent monitor unit (MU) and isocenter point-dose calculation method has been developed that provides an independent verification of the MU for intensity modulated radiotherapy (IMRT) using the MIMiC (Nomos Corporation) multileaf collimator. The method takes into account two unique features of IMRT using the MIMiC: namely the gantry-dynamic arc delivery of intensity modulated photon beams and the slice-by-slice dose delivery for large tumor volumes. The method converts the nonuniform beam intensity planned at discrete gantry angles of 5° or 10° into conventional nonmodulated beam intensity apertures of elemental arc segments of 1°. This approach more closely simulates the actual gantry-dynamic arc delivery by MIMiC. Because each elemental arc segment is of uniform intensity, the MU calculation for an IMRT arc is made equivalent to a conventional arc with gantry-angle dependent beam apertures. The dose to the isocenter from each 1° elemental arc segment is calculated by using the Clarkson scatter summation technique based on measured tissue-maximum-ratio and output factors, independent of the dose calculation model used in the IMRT planning system. For treatments requiring multiple treatment slices, the MU for the arc at each treatment slice takes into account the MU, leakage and scatter doses from other slices. This is achieved by solving a set of coupled linear equations for the MUs of all involved treatment slices. All input dosimetry data for the independent MU/isocenter point-dose calculation are measured directly. Comparison of the MU and isocenter point-dose calculated by the independent program to those calculated by the Corvus planning system and to direct measurements has shown good agreement with relative difference less than ±3%. The program can be used as an independent initial MU verification for IMRT plans using the MIMiC multileaf collimators.

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Key words: IMRT, monitor unit calculation, quality assurance, MIMiC multileaf collimator

I. INTRODUCTION

In delivering external beam radiotherapy (EBRT), the accelerator beam-on time or monitor units (MU) for a given treatment plan provides a link between the planned dose and the actual dose delivered to a patient. Uncertainties incurred in computing the MU will affect the dose delivered to the entire volume of irradiation for patients receiving EBRT. It is standard practice in EBRT to verify independently the calculation of MU prior to the start of a patient treatment. For intensity modulated radiotherapy (IMRT), an independent calculation of MU becomes difficult due to the complex relationship between the MU and the beam-intensity modulation shape as well as the technique used to generate the intensity modulation. Since the beam intensity modulation and its delivery technology are inherently complex as compared to conventional EBRT, independent verification of MU for IMRT plans have been carried out mostly by measurement of point doses in a separate verification phantom. Several research groups have recently examined the issues involved in the MU calculation for intensity modulated (IM) photon beams and have shown that it is feasible to perform an independent calculation of MU. The ability to perform such an independent calculation of MU has the potential to help streamline the patient treatment quality assurance (QA) and to reduce the human resources needed in maintaining a quality IMRT program.

Investigations of independent MU calculations have been focused mainly on IM beams at fixed gantry angles (gantry-static). The intensity modulation for a gantry-static beam is generated typically by a multileaf collimator (MLC) using the step-and-shoot or sliding window techniques. Some theoretical issues involved in the MU calculation for gantry-static IMRT beams have been discussed by Boyer et al. Typically, a gantry-static intensity-modulated beam is considered as composed of a series of partially overlapping open beams. Each open beam, or a segment of the IM beam, has uniform beam intensity but differing aperture dictated by the shape of intensity modulation. Kung et al. has applied the Clarkson scatter summation technique to calculate the dose to central axis (CA) from the open beam segments and has shown that an independent cal-

calculation of MU for IM beams can be obtained within this framework. By further considering each open-beam segment as a sum of elemental beamlets, Xing et al.\textsuperscript{11} has derived a general formalism for calculating MU with gantry-static IM beams and demonstrated its utility with several clinical examples. For IMRT delivery with the sliding window technique, which requires greater spatial resolution in beamlet size, the pencil beam algorithm has been used in the independent MU verification calculation.\textsuperscript{9}

Intensity modulation has also been generated with gantry-dynamic photon beams. Multiple overlapping gantry arcs with gantry-angle dependent beam apertures collimated dynamic MLC have been used in the proposed intensity modulated arc therapy (IMAT).\textsuperscript{21} As opposed to IMAT, the Peacock system (Nomos Corporation) delivers IMRT using a slit beam with its beam intensity modulated dynamically by a binary multileaf intensity modulator (MIMiC) over the path of a single gantry arc.\textsuperscript{5,22–24} For the Peacock IMRT system, the gantry arc is modeled by multiple gantry-static IM beams at intervals of 5° or 10° in the planning process. In principle, the independent MU calculation methodology developed for the gantry-static IM beams can be applied to these planned IM beams. This, however, does not correspond to the actual gantry-dynamic delivery. In the actual delivery, the intensity modulation of the planned IM beams is converted to leaf patterns that are suitable for arc delivery as opposed to 5° or 10° steps. The impact arising from the difference between the planned and the delivered intensity modulation depends on the number of gantry-static beams used in the arc modeling and on the patient geometry. It would increase when fewer gantry-static beams are used in planning and/or when the patient anatomy exhibits large spatial variations in its external contour and tissue heterogeneity. The aim of this work is to develop an independent MU calculation program for gantry-dynamic IMRT with the MIMiC multileaf collimator, using intensity modulations that correspond to the actual gantry-dynamic arc delivery.

**II. MATERIALS AND METHODS**

**A. The Peacock gantry-dynamic IMRT system**

The Peacock IMRT system consists of an “inverse” treatment planning system (Corvus Version 3) and a computer-controlled intensity modulation device, MIMiC, that can be attached to (and removed from) an existing linear accelerator (Varian Clinac-2100C in our setup). The system has two unique features in its delivery of an IMRT treatment. First, it delivers the IMRT in the accelerator’s gantry-dynamic mode with the photon beam intensity modulated dynamically by the tertiary MIMiC multileaf collimator. The MIMiC consists of 20 pairs of tungsten leaves, which can be moved in or out of the beam in the direction parallel to the axis of gantry rotation (Fig. 1). The transit time for leaf motion is on the order of 100 to 150 ms (Ref. 21) and the intensity modulation for a given beamlet is considered as binary. These leaves partition a rectangular beam into 40 smaller beamlets. The beamlet size, which determines the spatial resolution of intensity modulation, can be set to approximately 1×1 cm (1 cm mode) or 1×2 cm (2 cm mode) at the level of isocenter by using a mechanical stopper. The binary leaf motion designed for MIMiC limits the maximum treatment field size to approximately 20×2 cm or 20×4 cm, at the isocenter, for each arc delivery. In the remainder of this paper, the volume irradiated by such an arc is termed as a treatment slice. Second, when treating larger lesions, multiple treatment slices may be required to provide an adequate coverage of tumor volume in the field length direction (see Fig. 2 for an illustration of a treatment that requires two treatment slices). The field borders of the arcs in the adjacent slices are matched dosimetrically at the axis of gantry rotation. In actual treatment, the field matching is achieved by moving the treatment table, with patient being immobilized to the table, in the direction parallel to the axis of the gantry rotation between successive arc deliveries. The Peacock system, therefore, delivers IMRT in an arc-by-arc or slice-by-slice fashion analogous to the acquisition of conventional CT images. This analogy has made IMRT using the Peacock system known also as serial tomotherapy in the literature.\textsuperscript{5,25}

The Corvus planning system models an intensity-modulated arc as multiple gantry-static beams evenly spaced over the length of the arc. The angular separation between the adjacent beam central axes is typically chosen to be 10°
or $5^\circ$ To allow the gantry to achieve a stable rotational speed, the MIMiC leaves are kept closed for the beams in the first and the last $10^\circ$ segment of the arc. The intensity modulation of the planned beams is optimized by using the simulated annealing algorithm based on a user-defined dose-volume prescription. For dose delivery, the planning system converts the nonuniform beam intensity at the planned gantry angles to MIMiC leaf patterns (opening or closing) as a function of gantry angle, suitable for the arc delivery.\textsuperscript{22–24} The actual dose delivered to a patient by each arc depends on both the MIMiC leaf patterns and the MU assigned to the arc. While the leaf patterns determine the intrinsic spatial dose distribution, the MU assigned to each arc affects the absolute dose delivered to the entire volume of irradiation. In the following, we focus on the verification calculation of the MU for a given set of leaf patterns as planned by the Peacock IMRT system.

B. MIMiC arc delivery

To obtain a true independent MU verification calculation, leaf patterns that correspond to the actual arc delivery of a Peacock IMRT plan were used in this work. For a given arc delivery, the intensity modulation of the gantry-static beams planned by the Corvus planning system is decomposed into multiple elemental beams evenly spaced over the planned arc.\textsuperscript{22} Each elemental beam is chosen to represent a $1^\circ$ arc segment [the MIMiC has an angular resolution of $0.25^\circ$ (Ref. 22)]. Each elemental beam has uniform beam intensity but irregular beam aperture defined by the opening of its 40 beamlets. For example, suppose the angular spacing between the planned beam axes is $10^\circ$, then a planned gantry-static beam is decomposed into 10 elemental beams spaced over a $10^\circ$ arc segment centered at the central axis of the beam. If a beamlet in the gantry-static beam has a beam intensity of 100%, the corresponding beamlet would be open in all 10 elemental beams. If a beamlet has a planned intensity of only 50%, the beamlet would be open for only the five elemental beams near the center of arc segment and be closed for other elemental beams near the edges of the arc segment. These elemental beams, which closely approximate the actual delivery of a Peacock IMRT plan, are used for the MU calculation. The MU calculation presented here for such an IMRT arc is, in fact, equivalent to a conventional arc in EBRT, except that the beam aperture is now gantry-angle dependent and that the open-beam portion can be discontinuous.

C. Dose calculation algorithm

A dose calculation algorithm based on tissue-maximum-ratio (TMR) and scatter-maximum-ratio (SMR) was used in this work. It differs from the finite pencil beam algorithm used by the Corvus planning system. Dose per monitor unit to the center of an arc (point $P$ in Fig. 1) is calculated as the sum of doses given by all elemental beams. For a given elemental beam at gantry angle $\theta$, the dose per MU to $P$ is given by

\[
D_p(\theta) = \sum_{i=1}^{40} d_i(t_i, \theta),
\]

where $d_i(t_i, \theta)$ denotes the dose per monitor unit to point $P$ given by the beamlet $i$ and $t_i$ is the depth of tissue above the isocenter plane (a plane perpendicular to the beam central axis across the isocenter) for beamlet $i$. For the beamlets that are immediately adjacent to the beam central axis, $d_i(t_i, \theta)$ includes the primary as well as the scattered dose. For beamlets that are not adjacent to the CA, $d_i(t_i, \theta)$ includes only the scattered dose. To avoid the uncertainties involved in the extrapolation of TMR and output factors to zero field-size, the beam formed by the four beamlets immediately adjacent to the CA was considered as the “primary” beam. The rest of the beamlets would contribute only the scattered dose. Dose per MU to the central axis from the primary beam is given by

\[
d(t, \theta) = \frac{D_{SSD}(d_{\text{max}}, 10 \times 10) \times (SAD + d_{\text{max}})}{SAD}^2 
\times \text{OF}(m \times l) \times T \times \text{TMR}(t, m \times l),
\]

where $D_{SSD}(d_{\text{max}}, 10 \times 10)$ denotes the dose per MU to the $d_{\text{max}}$ of the calibration field under the SSD calibration geometry. The primary beam dimensions $(m \times l)$ equals to $2 \times 1.64 \text{ cm}$ and $2 \times 3.36 \text{ cm}$ for the 1 cm and 2 cm MIMiC mode, respectively. The output factor and TMR of the primary beam are, respectively, designated $\text{OF}(m \times l)$ and $\text{TMR}(t, m \times l)$. SAD (source-to-axis distance) is 100 cm in our case. $T$ is a modifying factor for output factor when some beamlets within the primary beam are closed (often the case in the Peacock IMRT plan). It is defined as a quotient of the central axis dose of a primary beam with partial leaf closure to that of the full primary beam. For ideal beam collimation exhibiting zero leaf transmission or tongue and groove effects, $T$ would be equal to $n/4$ where $n$ is the number of beamlets open. In practice, $T$ is influenced by design parameters, and so becomes dependent upon the actual beamlet pattern. In this work, $T$ was measured for the two MIMiC beams.
modes for all seven symmetrically independent beamlet configurations. The results are given in Table II.

Dose to CA from beamlets that are not adjacent to the CA (i.e., primary beam is not included) was calculated by using the Clarkson scatter integration algorithm\(^26\) (symbols used are illustrated in the beam's eye view for one beamlet in Fig. 3). The SMR was deduced self-consistently from broad beam TMR and the primary beam TMR (see Appendix for details). Dose per MU to CA from beamlet \(i\) is given by

\[
d_i(t, \theta) = D_{SSD}(d_{max} \times 10) \times \left( \frac{SAD + d_{max}}{SAD} \right)^2 \times OF(m \times 1) \times \frac{\Delta \phi}{2\pi} \sum_{j=1}^{n} \left[ \frac{1}{(SAD + d_{max})} \right]_{j} \times \text{SMR}(t, r_{in}^{(j)})
\]

where \(\Delta \phi\) is angular resolution used in the Clarkson sector integration, \(n\) denotes the total number of scatter sectors. Note that the summation on the right-hand side of Eq. (3) can be precalculated for all beamlets over a range of possible tissue depths. The precalculated effective scatter-maximum ratio for each beamlet can be stored in tabular form as a function of tissue depth. It makes the MU calculation computationally efficient since only a table look-up of beamlet scatter is needed in the real-time calculation for a given IMRT plan. As will be discussed further in the next section, the accuracy of beamlet scatter calculation can also be improved independently within the same formalism.

For IMRT plans that require multiple treatment slices, the dose to the center of each arc consists of the dose given by the arc itself and the off-slice scattered and leakage doses from arcs delivered at other treatment slices. The calculation of the dose by the arc itself is given by Eqs. (2) and (3). The off-slice scattered dose was calculated by using Eq. (3) with appropriate beamlet scatter calculated for the open beamlets of the off-slice treatment arcs. The off-slice leakage dose to a treatment slice depends on the distance between the two treatment slices and on the MIMiC operating mode due to the setting of accelerator jaws (Fig. 4). In the 1 cm mode, the distance between the CA of two adjacent arcs is 1.64 cm, within the open aperture of the accelerator jaws. Therefore the leakage to an immediately adjacent slice is determined solely by the MIMiC leaf transmission. The leakage to the next treatment slice, however, is determined by the combined transmission of accelerator jaw and the MIMiC leaf. In 2 cm mode, off-slice leakage to the immediately adjacent treatment slice is determined by the combined transmission of accelerator jaw and the MIMiC leaf. Leakage to the next treatment slice is determined solely by accelerator jaw transmission.

D. Monitor unit calculation formalism

For an IMRT plan that requires only one treatment slice, the MU for the arc can be calculated easily once the dose to the center of gantry rotation is calculated. Let \(\mu_{\theta}\) denote the MU given per degree of gantry rotation (assume the gantry speed and dose-rate are constant) and \(D_{p}\) denote the prescription dose to the center of gantry rotation, then

\[
MU_{\theta} = \frac{D_{p}}{\sum_{\theta} \sum_{i} D_i(t, \theta)}.
\]

The total MU for the arc is the product of the arc length (in unit of degree) and \(MU_{\theta}\). Note that Eq. (4) can also be used to calculate the dose delivered to \(P\) if the MU for the arc is known. For example, using the MU reported by the Corvus planning system, one can calculate the dose to \(P\) and compare it to that calculated by Corvus or to that measured by ionization chamber at the same position.

For IMRT plans that require multiple treatment slices, the off-slice scattered dose can be large, especially when the transverse dimension of the treatment volume is large (4–9% of the total dose delivered to each treatment slice). The off-slice scatter is directly proportional to the MU delivered at those off-slice positions. Mathematically, one can write

\[
D_{p_i} = \sum_{j} MU_j \cdot d_{ij} \quad \text{for} \quad i = 1, 2, ..., N,
\]

where \(i\) denotes treatment slice, \(D_{p_i}\) denotes the dose to the center of gantry rotation for treatment slice \(i\), \(d_{ij}\) denotes the dose per MU delivered to the calculation point of treatment slice \(i\) due to the arc delivered at treatment slice \(j\), and \(MU_j\) is the MU given at the treatment slice \(j\). To determine the

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**Fig. 3.** A beam’s eye view of the geometry used in scatter dose computation using the Clarkson scatter sector-integration technique for an open beamlet.

**Fig. 4.** Illustration, in Beam’s eye view, of the positions of adjacent MIMiC arcs with respect to the position of accelerator jaws for the 1 cm (A) and 2 cm (B) mode of arc delivery. The CA of an immediately adjacent arc is within the beam collimated by the accelerator jaws (5 cm at the isocenter) for the 1 cm mode but is under the shield of the jaws for the 2 cm mode.
MU at each treatment slice self-consistently, one needs to solve the coupled linear equations of (5). For a plan that requires \(N\) treatment slices, we have

\[
\begin{pmatrix}
\text{MU}_1 \\
\text{MU}_2 \\
\vdots \\
\text{MU}_N
\end{pmatrix}
= \begin{pmatrix}
d_{11} & d_{12} & \cdots & d_{1N} \\
d_{21} & d_{22} & \cdots & d_{2N} \\
\vdots & \vdots & \ddots & \vdots \\
d_{N1} & d_{N2} & \cdots & d_{NN}
\end{pmatrix}^{-1}
\begin{pmatrix}
\text{D}_p^1 \\
\text{D}_p^2 \\
\vdots \\
\text{D}_p^N
\end{pmatrix}.
\]

As mentioned previously, Eq. (6) can also be used to verify point doses when the MU at various treatment slices are known. Equations (4) and (6) form the basis for the independent MU calculation verification to be presented later in this paper.

**E. Beam data acquisition and IMRT plan verification**

The input beam data needed for the MU calculation were measured directly for the 6 MV photon beam (Varian Clinac-2100C, SN228) being used for the Peacock IMRT. The output factor of the MIMiC collimated primary beams, defined as the quotient of the radiation output at the depth of maximum buildup of a MIMiC collimated beam to that of the reference calibration beam defined by the accelerator’s secondary jaws, was measured using a Capintec-RP-05P small-volume ionization chamber (nominal active volume 0.07 cm\(^3\), SN: CIL079287) in polystyrene phantoms. The TMR for the beam defined by the opening of the four central leaves of the MIMiC was measured directly using the same Capintec chamber in polystyrene phantoms. The TMR for other broad beams, collimated by the accelerator jaws, were calculated from the percent depth dose (PDD) and the output factors measured in water using the RFA300 beam scanning system. The calculated TMR were verified against the measured TMR at selected field sizes.

The delivery of a Peacock IMRT plan has been verified extensively during the commissioning of the IMRT system. Dose delivered to different types of verification phantoms was measured for a variety of IMRT plans using radiation detectors that include small volume (1 \(\times\) 1 \(\times\) 1 mm\(^3\)) LiF TLDs, radiographic verification films, and the small volume ionization chamber. The Corvus planning system requires an MU calibration factor for each arc delivery mode (1 cm or 2 cm). This was based on the ratio of the average measured dose to the average planned dose during commissioning. For the MU verification study reported in this work, the point doses to the isocenter of various IMRT plans were all measured by using the small volume ionization chamber in a rectangular polystyrene verification phantom.

**III. RESULTS**

**A. Input data for dose calculation**

Dosimetry data needed for the MU calculation include (1) the output factor and TMR for the MIMiC collimated primary beams, (2) the \(T\) factor that quantifies the output change due to the closure of leaves within the primary beam, and (3) the broad beam TMRs and output factors needed for deriving the SMRs. Table I lists the measured output factors for the primary beams. The TMR for the corresponding primary beams are plotted in Fig. 5. The circles in Fig. 5 represent the measured TMR while the solid line represent the TMR calculated from the percent depth dose with equivalent square field sizes that best fit the measured data. The \(T\) factor, derived from the measured outputs for various closure patterns of the four center leaves in arc delivery, are given in Table II. The \(T\) factor is practically the same for both MIMiC operating mode. The \(T\) factor expected for perfect partial beam collimation is listed in the third column in Table II. Note that the measured \(T\) factors are smaller than the corresponding ideal values. We believe that the reduction is caused primarily by the tongue-and-groove structure on the leaf end and leaf sides that attenuates the primary beam.
reaching the measuring chamber. It could also be caused by less scatter dose due to the tongue-and-groove and electron inequilibrium. The transmission factor at off-slice positions was measured to be 0.45% in 1 cm mode for immediately adjacent treatment slices and is practically zero for the next treatment slices. For 2 cm mode, the transmission for the adjacent treatment slices is negligible. However, the transmission for the next treatment slice is about 0.42% due to the finite length of the MIMiC leaves.

The TMR calculated from PDD and output factors were compared to TMR measured at field sizes of 10×10 and 5×5 cm². Excellent agreement was found. The SMR for broad circular beams were obtained according to Eq. (A4) given in the Appendix. Self-consistency was checked by comparing the TMR calculated from the derived SMR. Using the SMR, the scattered dose to CA [Eq. (3)] from each of the 40 beamlets was precalculated at 61 tissue depths for both 1 cm and 2 cm arc modes and for several off-slice treatment positions.

B. MU/isocenter point-dose calculation and comparison

Other than the measured dosimetry input data shown above, there are no other adjustable parameters in the MU calculation formalism. An in-house MU verification program has been written to perform independent MU verification calculation for Peacock IMRT plans. For each IMRT plan, the program reads the basic plan information exported by the Corvus system regarding the MIMiC operating mode, total arc length, number of treatment slices, patient external contour, isocenter location, and the intensity modulation patterns for the planned gantry-static beams for each arc. The intensity modulation patterns of the planned beams are then decomposed into beam apertures of elemental arc-segments corresponding to the arc delivery at each table position. The program computes the dose per MU to the center of gantry rotation at each treatment slice based on the planned leaf pattern and the corresponding patient external contour. It has the option to compute MU if the desired dose to the center of gantry rotation of each treatment slice is known or to compute the dose if the MU for each treatment slice is known. Since verification measurement by ionization chamber of an IMRT plan gives the dose directly, the calculated dose using the planned MU can be compared directly to that measured by ionization chamber and that given by the Corvus planning system.

1. Idealized Peacock IMRT plans

The simplest test plans for checking the accuracy and the validity of the dose calculation model used in the MU calculation program are plans that have constant beam intensity. For example, a single Peacock arc with a constant beam aperture defined by the opening of the four center MIMiC

<table>
<thead>
<tr>
<th>Beamlet patterna</th>
<th>Measured T factor</th>
<th>Ideal T factor</th>
</tr>
</thead>
<tbody>
<tr>
<td><img src="image" alt="Beamlet pattern" /></td>
<td>0.0062</td>
<td>NA</td>
</tr>
<tr>
<td><img src="image" alt="Beamlet pattern" /></td>
<td>0.218</td>
<td>0.25</td>
</tr>
<tr>
<td><img src="image" alt="Beamlet pattern" /></td>
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</tr>
<tr>
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<td>0.466</td>
<td>0.50</td>
</tr>
<tr>
<td><img src="image" alt="Beamlet pattern" /></td>
<td>0.425</td>
<td>0.50</td>
</tr>
<tr>
<td><img src="image" alt="Beamlet pattern" /></td>
<td>0.728</td>
<td>0.75</td>
</tr>
<tr>
<td><img src="image" alt="Beamlet pattern" /></td>
<td>1.000</td>
<td>1.00</td>
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</tbody>
</table>

*aShaded area indicates the radiation is blocked by the corresponding beamlet.

<table>
<thead>
<tr>
<th>Table III. Comparison of dose to isocenter: Single treatment slice with constant beam intensities.</th>
</tr>
</thead>
<tbody>
<tr>
<td>MIMiC mode</td>
</tr>
<tr>
<td>-----------------</td>
</tr>
<tr>
<td>1 cm</td>
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<tr>
<td></td>
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<tr>
<td>2 cm</td>
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leaves (closed, of course, in the first and the last 10° segment of the arc) would provide a test of the dose calculation accuracy using the measured primary beam TMR, the output factors, and the handling of irregular patient geometry. It also provides a known limiting condition for checking the internal consistency of the MU calculation program. A similar plan with the open beam aperture defined by more than the four center leaves would provide a test of both the accuracy and self-consistency of the scatter calculation algorithm, without the need of discerning the effects of complex intensity modulation patterns. Four test plans with constant beam apertures were created using the beam utility module of the Corvus planning system for a rectangular verification phantom. A treatment delivery diskette for each plan was created by the Corvus system in the same way as for normal patient IMRT plans. Table III compares the doses calculated by the verification program to that measured by ionization chamber and to that given by the Corvus planning system. Doses calculated by the verification program are found to be in good agreement with the measured dose (<±1%). The dose given by the Corvus planning system is consistently lower (approximately 2.5%) than the measured dose. This systematic difference is due primarily to the MU calibration factor introduced in the Corvus planning system, which is designed to make the average planned dose to agree with measured dose for different types of IMRT plans.

2. IMRT validation plans

Validation of the MU verification program for Peacock IMRT plans was performed on the verification phantoms so that corresponding measurement can be made. IMRT plans with single and multiple treatment slices were created for the 1 cm and 2 cm MIMiC mode. The plan is designed to deliver a concave high radiation dose surrounding a critical structure. Table IV compares the isocenter point doses for IMRT plans with one, two, and three treatment slices. The total dose calculated for each treatment slice is found to be in good agreement with both the measured and the Corvus computed doses (<±1.5%).

The influence of off-slice scatter and leakage in a multiple-treatment-slice IMRT plan is illustrated in Tables V and VI for the two three-treatment-slice plans shown in Table IV. The comparison is made between the calculation and measurement only since the corresponding information is not given by the Corvus plans. The ionization chamber measurements show that the scatter and leakage contribution from
immediately adjacent arcs is on the order of 7% and 5% of the total dose delivered to a given treatment slice for the 1 cm and 2 cm mode, respectively. The scatter and leakage contribution from the next adjacent treatment slice is about 2.5% in both 1 cm and 2 cm mode. If the off-slice scatter and leakage were neglected in the MU verification calculation, an error of up to 10% could result in the calculated MU for a three-treatment-slice IMRT plan. Therefore, the off-slice scatter and leakage has to be taken into account appropriately in an independent MU verification program for Peacock IMRT plans. Tables V and VI also offers some insight on the accuracy of off-slice scatter dose calculation by the verification program. The total dose to a calculation point on a given treatment slice consists of the dose given by the arc delivered on the treatment slice (in-slice arc) and the scattered and leakage dose from the arcs delivered off-slice. Tables V and VI show that the dose calculated for the in-slice arc agrees well with the corresponding measured dose ($\pm 1.6\%$). The scattered and leakage dose calculated from the off-slice arc show relatively large differences (from about 1% to 17%) as compared to the corresponding measured dose. When the calculation point is inside the high dose region, the off-slice scattered and leakage dose accounts for only a small portion of the total dose to that point. Thus the relatively large difference seen in the scatter dose calculation has only insignificant effect on the comparison between the calculated and the measured dose for each given slice. However, when a calculation point is outside the high dose region, scatter and leakage dose become the dominate contribution to the total dose and the accuracy of scatter calculation would need to be improved in order to maintain an accurate comparison with measurement. We will discuss this point further in the next section.

### 3. MU calculation for patient plans

The validation tests have shown that the MU/isocenter point-dose calculated on the verification phantom agrees well with both the measured and the Corvus planned values. The MU verification program has also been applied to many different patient plans. Table VII shows the MU calculated for five IMRT patient plans. For each patient plan, a phantom plan has also been created on the verification phantom for experimental verification. The MU calculated for the phantom plans of the five corresponding patient plans is also shown in Table VII. For patient plans, the calculated MUs can only be compared to that given by the Corvus planning system. For the phantom plans, comparison can be made with both the measured and the planned values. For multiple-treatment-slice phantom plans, the measured MUs (the MUs that would produce the planned doses at the respective isocenter of each treatment slice) are deduced self-consistently from the measured doses using a modification of Eq. (6) as follows:

$$\begin{pmatrix}
\frac{\text{MU}_{\text{meas1}}}{\text{MU}_{\text{Corvus1}}} \\
\frac{\text{MU}_{\text{meas2}}}{\text{MU}_{\text{Corvus2}}} \\
\vdots \\
\frac{\text{MU}_{\text{measN}}}{\text{MU}_{\text{CorvusN}}}
\end{pmatrix} = \begin{pmatrix}
m_{11} & m_{12} & \cdots & m_{1N} \\
m_{21} & m_{22} & \cdots & m_{2N} \\
\vdots & \vdots & \ddots & \vdots \\
m_{N1} & m_{N2} & \cdots & m_{NN}
\end{pmatrix}^{-1} \begin{pmatrix}
D_{p1} \\
D_{p2} \\
\vdots \\
D_{pN}
\end{pmatrix},$$

(7)

where $m_{ij}$ denotes the measured dose contribution to treatment slice $i$ by the arc delivered at treatment slice $j$ with $\text{MU}_{\text{Corvus} j}$. $D_{pi}$ denotes the Corvus planned dose to the center of gantry rotation of treatment slice $i$. As shown in Table VII, good agreements exist between the calculated, the Corvus planned, and the measured MU, with relative differences less than $\pm 3\%$. It indicates that the MU verification program can provide a robust and independent verification of MU computed by the Corvus planning system. When the calculated MU differs by more than 3% of the planned MU, further investigation such as ionization chamber measurement should be initiated.

### IV. DISCUSSION

Intensity modulation of the photon beams used in EBRT has greatly improved our ability in designing dose distributions for complex target volumes. The resulting IMRT has the potential to reduce the unwanted doses to healthy tissues
and therefore affords a possibility to escalate the dose to the intended target volume in EBRT. The implementation of IMRT also poses new challenges to the established quality assurance program in EBRT arising from both the design and the delivery of intensity modulated photon beams. Issues related to the commissioning, ongoing system QA, and patient treatment QA of an IMRT system have been the subjects of active research in IMRT. In this work, an independent MU/isocenter point-dose verification calculation program is developed for IMRT plans delivered in gantry-dynamic mode using the MIMiC multileaf collimator, as a part of the patient treatment QA tool. The program utilizes input dosimetry data that can be measured directly and the MIMiC leaf patterns that correspond to the actual gantry-dynamic dose delivery. The approach can be readily applied to Tomotherapy using binary intensity modulators. It can also be adapted to other gantry-dynamic IMRT deliveries such as IMAT. It should be emphasized, however, that this verification calculation program checks only one aspect of an IMRT plan, i.e., the MU or the isocenter point dose. Other aspects of the plan such as the creation of beam intensity modulation, its translation to instructions for the delivery device, and patient setup should be verified independently in a comprehensive patient treatment QA program. In the following, we discuss some of the limitations and possible improvements of MU calculation program developed in this work.

A main restriction of the current MU verification program is that the dose calculation point needs to be on the central axis at the center of gantry rotation. Off-axis dose calculation is possible, for example, by using the off-axis-ratio, but the simplicity of the approach would be lost. Off-axis dose verification calculation is an important part of a comprehensive IMRT QA program. Current practice relies primarily on phantom measurements using radiographic films or other detectors at selected points. Off-axis verification calculation can be performed by using an existing and independent treatment planning system. The approach utilized in this work for modeling the arc delivery of Peacock IMRT plans is being implemented in the conventional 3D treatment planning system at our institution for off-axis dose calculation verification. This approach would ease the restriction of making dose calculation at the isocenter.

Peacock IMRT plans often place the isocenter outside the high dose region in order to gain a better spatial resolution of the intensity modulation by MIMiC. In these cases, a significant portion of dose to the isocenter comes from the scattered dose. If accurate isocenter dose is to be calculated, an accurate algorithm for scatter-dose calculation is needed for the MU verification program. In fact, any valid comparison with the Corvus planning system also requires the scatter calculation by the Corvus planning system to have the same order of accuracy. For these plans, using an independently commissioned planning system would enable the comparison to be made in the high dose region. Otherwise a measurement in a verification phantom with the measurement point placed in the high dose region should be made.

The broad-beam TMR used in this work was deduced from the accelerator-jaw-collimated beams. Strictly speaking, it should have been deduced from broad beams collimated by the MIMiC collimator since the MIMiC is used for defining intensity modulated beams in the Peacock IMRT.

### Table VII. Comparison of MU for patient IMRT plans.

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<th>Plan</th>
<th>MIMiC mode</th>
<th>Treatment slice</th>
<th>MU_{Corvus}</th>
<th>MU_{Calc}</th>
<th>MU_{meas}</th>
<th>MU_{Calc} - MU_{Corvus} \times 100</th>
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delivery (while the accelerator jaws are maintained at a fixed position). The use of MIMiC collimated broad-beam TMR would give a more accurate and consistent account of the combined collimator scatter in the calculation of the scattered dose from the MIMiC-defined beamlets. The scatter from all MIMiC beamlets can also be measured directly. Since the Peacock IMRT typically uses only one photon energy, the measurement of the scatter to CA for all 40 beamlets (actually 10 when taking into account of the symmetry about the central axis) at various tissue depths can be accomplished without excessive labor. The measured beamlet-scatter would greatly improve the accuracy of the scatter-dose calculation. An effort in this regard is currently being pursued.

The Peacock IMRT delivery system requires the gantry rotation speed and the accelerator dose-rate be constant during the arc dose delivery. To minimize the variation in the gantry rotation speed, the first and the last 10° segments of an IMRT arc are planned with MIMiC leaves being closed, i.e., not to be used in the IMRT dose delivery, so that gantry can reach a stable rotational speed. The MIMiC controller is also equipped with independent gantry angle sensors which check, in real-time, the variation of gantry rotation speed throughout the arc delivery. When the variation of gantry speed is greater than a preset tolerance value, the MIMiC controller generates a gantry-speed interlock that immediately shuts off the dose delivery by the accelerator. The Peacock system does not provide additional interlock for monitoring the dose rate fluctuations. It is the responsibility of physicists to make sure the accelerator used for MIMiC delivery can maintain a relatively constant dose rate in the arc mode. The MU verification program developed in this work also assumes a constant gantry rotation speed and dose rate. Therefore, the comparison between the calculated and the measured doses does not take into account possible small, but within the preset tolerance, variations in gantry speed and dose rate that may be present during the arc delivery. For these reasons, experimentally measured dose has been taken as the reference of the true dose to the target volume in all our comparisons.

V. CONCLUSION

An independent MU/isocenter point-dose calculation program has been developed for verifying the MU of gantry-dynamic IMRT plan using the MIMiC multileaf collimator. Evaluation of the program indicates that the MU calculated by the program agrees well with that calculated by the Corvus planning system and with that deduced from point dose measurement (<±3%). The program provides a quick and independent initial verification of the MU for gantry-dynamic IMRT plans. Measurement is called for if the difference between the calculated MU and the planned MU is greater than 3%.

APPENDIX

To avoid the uncertainties associated with the extrapolation of the zero-field TMR and output factor, the beam defined by the opening of the four MIMiC leaves immediately adjacent to the central axis (CA) (see Fig. 1) is treated as the “primary” beam for computing dose to CA. Using the measured TMR of the primary and other broad beams, corresponding SMR of a broad beam can be defined self-consistently as follows.

For an isocentric beam of circular beam aperture \( r \) incident perpendicularly on a semi-infinite homogeneous phantom, the dose to the isocenter at a depth of \( d \) below the phantom surface can be written as

\[
D(d,r) = D(d,m \times l) + S(d,r - m \times l),
\]

where \( D(d,m \times l) \) denotes the dose to isocenter from a beam of field size \( m \times l \) centered at the central axis and \( S(d,r - m \times l) \) denotes the dose contributed by the remainder of the field, i.e.,

\[
S(d,r - m \times l) = D(d,r) - D(d,m \times l).
\]

Equation (A2) can be rewritten as

\[
\frac{S(d,r - m \times l)}{D(d,m \times l)} = \frac{D(d,r)}{D(d,m \times l)} \times \frac{D(d_{\text{max}},r)}{D(d_{\text{max}},m \times l)} - \frac{D(d,m \times l)}{D(d_{\text{max}},m \times l)},
\]

where \( D(d_{\text{max}},r) \) and \( D(d_{\text{max}},m \times l) \) denote the doses to the isocenter from isocentric beams of field size \( r \) and \( m \times l \), respectively. Invoking the standard definition of TMR and output factor, we have

\[
\text{SMR}(d,r) = \text{TMR}(d,r) \frac{\text{OF}(r)}{\text{OF}(m \times l)} - \text{TMR}(d,m \times l),
\]

where

\[
\text{SMR}(d,r) = \frac{S(d,r - m \times l)}{D(d_{\text{max}},m \times l)}
\]

is the scatter-maximum ratio defined self-consistently for the broad beam and \( \text{OF}(r) \) is the output factor for field size \( r \). By rearranging Eq. (4), the TMR for the broad beam is given by

\[
\text{TMR}(d,r) = \frac{\text{OF}(m \times l)}{\text{OF}(r)} \left\{ \text{TMR}(d,m \times l) + \text{SMR}(d,r) \right\}.
\]

For irregular fields, the Clarkson sector integration algorithm for scatter can be employed to calculate an effective SMR and \( \text{OF}(r) \). The dose per MU to the isocenter of such a field is then given by

\[
\hat{D}(d,r) = \hat{D}_{SSD}(d_{\text{max}},10 \times 10) \text{OF}(m \times l) \left( \frac{SAD + d_{\text{max}}}{SAD} \right)^2 \times \left\{ \text{TMR}(d,m \times l) + \text{SMR} \right\},
\]

where \( \hat{D}_{SSD}(d_{\text{max}},10 \times 10) \) denotes the dose per MU to the \( d_{\text{max}} \) of the calibration field under the SSD calibration geometry and \( \text{SMR} \) is the effective SMR for the irregular field. Dose calculations in this work are based on Eq. (A6).