Monitor unit calculation for an intensity modulated photon field by a simple scatter-summation algorithm

This article has been downloaded from IOPscience. Please scroll down to see the full text article.


(http://iopscience.iop.org/0031-9155/45/3/401)

View the table of contents for this issue, or go to the journal homepage for more

Download details:
IP Address: 152.92.171.92
The article was downloaded on 15/10/2010 at 16:33

Please note that terms and conditions apply.
NOTE

Monitor unit calculation for an intensity modulated photon field by a simple scatter-summation algorithm

L Xing, Y Chen, G Luxton, J G Li and A L Boyer
Department of Radiation Oncology, Stanford University, Stanford, CA 94305-5304, USA
E-mail: lei@reyes.stanford.edu (L Xing)

Received 30 September 1999, in final form 3 December 1999

Abstract. An important issue in intensity modulated radiation therapy (IMRT) is the verification of the monitor unit (MU) calculation of the planning system using an independent procedure. Because of the intensity modulation and the dynamic nature of the delivery process, the problem becomes much more involved than that in conventional radiation therapy. In this work, a closed formula for MU calculation is derived. The approach is independent of the specific form of leaf sequence algorithms. It is straightforward to implement the procedure using a simple computer program. The approach is illustrated by a simplified example and is demonstrated by a few CORVUS (NOMOS Corporation, Sewickley, PA) treatment plans. The results indicate that it is robust and suitable for IMRT MU verification.

1. Introduction

Intensity modulated radiation therapy (IMRT) has recently emerged as a new radiation therapy modality arising from the development of inverse planning and computer-controlled delivery using multivane collimators (MIMiC) or multileaf collimators (MLC). While IMRT improves the dose distribution for many clinically complicated cases, it also entails new techniques for treatment verification because of the intensity modulation of the incident photon beams. Among various issues related to verifying the treatment, a pressing problem is how to verify efficiently the calculation of monitor units (MU) from a commercial inverse planning system. An independent MU check is required for a patient treatment based on primitive machine data, for example output factor, TMR, etc. For a conventional treatment plan this is typically done using an independent manual calculation. Because of the lack of an effective tool for the MU calculation, ion chamber measurement has been employed to verify the patient dose in a surrogate phantom in many institutions, which is time consuming and costly.

The most straightforward way to verify the MU or dose calculation of an IMRT treatment planning system is perhaps to sum the fractional MUs corresponding to the segmented fields. The MU calculation for a multileaf collimated static field has been described in the literature (Georg and Dutreix 1997). A few attempts have been made to develop more efficient techniques for the verification of dynamic delivery. Boyer et al (1999) have investigated some theoretical aspects of the MU calculation for an intensity modulated field. Recently, Kung and Chen (1999) have developed an approach for the ‘forward’ dose verification. In their calculation, the MUs and leaf sequence files for a patient treatment were used as input to compute the dose at a given point in the patient. A modified Clarkson method was used to determine the dose for a given non-uniform fluence map. To the best of our knowledge, the work by Kung and Chen...
represented the first successful demonstration of an independent dose calculation for IMRT. The purpose of this paper is to bridge the gap between the intuitive approach of Kung and Chen and the theoretical framework by Boyer et al. (1999), and to provide a close formula for MU calculation in IMRT. The formula is general and suitable for any intensity modulated beam delivery method. The method will be illustrated for MLC modulated IMRT using a Varian accelerator (Varian Oncology Systems, Palo Alto, CA). The data for a few clinical IMRT cases will be presented.

2. Method

In this study we concentrate on IMRT based on fixed-gantry delivery. As will be seen below, the methodology is quite general and can be generalized for dynamic arc IMRT. For convenience, let us first consider the case of a single incident beam. Multiple beams can be dealt with similarly as long as their relative weights are known. Assume that a treatment field defined by the jaws can be partitioned into beamlets. Also, assume that there are $K$ segments in the dynamic treatment field. The field shapes of these segments are known from the given leaf sequence file. The dose at a given point in the patient is a summation of the contributions of all segments of treatment, i.e.

\[ D = \sum_{k} \text{MU}_k d_k = \text{MU} \sum_{k} f_k d_k \]  

(1)

where $\text{MU}_k$ is the number of monitor units delivered by the $k$th segment, $\text{MU}$ is the total monitor unit delivered by the beam, $f_k = \text{MU}_k / \text{MU}$ is the fractional MU of the $k$th segment and $d_k$ is the dose per unit MU from the $k$th segment. Let $d_{m}^0$ be the dose contribution from the $m$th beamlet when it is open, and $d_{m}^l$ be the leakage dose per unit MU from the $m$th beamlet when it is blocked by the MLC. If we denote the set of beamlets inside the open field area of the $k$th beam segment by $A_k$, the dose from the $k$th segment per unit MU is given by

\[ d_k = \sum_{m \in A_k} d_{m}^0 + \sum_{m \notin A_k} d_{m}^l. \]  

(2)

We introduce a notation

\[ \delta_{m,A_k} = \begin{cases} 1 & \text{if } m \in A_k \\ 0 & \text{if } m \notin A_k \end{cases} \]  

(3)

and further assume that the leakage dose can be related to the open beamlet dose by

\[ d_{m}^l = \alpha d_{m}^0 \]  

where $\alpha$ is the transmission factor. Then, by exchanging the order of summations, we can recast equation (1) as

\[ D = \text{MU} \sum_{m} C_m d_{m}^0 \]  

(4)

where

\[ C_m = \sum_{k} \left[ \delta_{m,A_k} + \alpha (1 - \delta_{m,A_k}) \right] f_k. \]  

(5)

Equation (4) relates the dose at a given point to the setting of the monitor units, MU, and is the main result of this paper. Equation (4) shows that, in order to obtain the dose of a multisegment
MU calculation by scatter-summation algorithm

Dynamic field, one only needs to perform the dose calculation for the contributing beamlets once. The increase in MUs required to deliver a modulated field is reflected by the \( C_m \) factor defined by the summation in equation (5). The summation represents the modulation of the dose at the point of interest, which is similar to a wedge factor or a compensator transmission factor, as described by Geis and Boyer (1996). However, the current definition takes into account the leaf leakage during the segments when the beamlets are blocked by the MLC. The calculation is facilitated by separating the open and blocked sets of beamlets, assuming that any beamlet is either completely open or blocked. Except for the beamlet size, which can be dependent on step size in a specific leaf sequence algorithm, our approach is independent of leaf sequence algorithms and delivery machines. All one needs for the calculation is the dose/MU and the transmission factor for each beamlet. The beamlet dose, \( d_0 \), can be calculated using a variety of methods, as simple as a Clarkson type of approach or as complex as Monte Carlo simulation.

The remaining problems are computational. To obtain the absolute value of MU using equation (4), it is required to calculate the dose contributions from the beamlets. Given a collection of beamlets, there are many ways to calculate their dose contributions to a point. Instead of dividing the field into a series of annular rings as demonstrated by Kung and Chen (1999), we proceed by using the beamlets as the elementary building blocks. This decomposition seems to be more natural since it utilizes the field boundaries defined by the MLC. In the following we describe a simple way to carry out the MU calculation based on a modified Clarkson method (Khan 1994, BIR 1996).

The calculation point can be anywhere inside the field. As in the conventional treatment, it is convenient to choose the isocentre as the verification point because the source-to-skin distance (SSD) of the beam can be easily obtained and no off-axis ratio is involved. In this case, we consider the contribution from the central four beamlets as the ‘primary dose’, denoted by \( D_p \), and the contributions from the rest of the beamlets as ‘scatter’. To be specific, let us assume that the beamlet size is 1 cm \( \times \) 1 cm. In reality each of the four beamlets in the central 2 cm \( \times \) 2 cm square can be either open or blocked. When all the four beamlets are open in the beam, the primary dose is given by

\[
D_p(\text{cGy}) = \text{MU} \times C_f(\text{cGy/MU}) \times S_i(l_{eq}) \times S_p(l = 2 \text{ cm}) \times \text{TMR}(d, l = 2 \text{ cm})
\]  

(6)

where \( C_f \) is the calibration factor of the linac, \( S_i \) is the collimator scatter factor obtained by in-air measurement, \( S_p \) is the phantom scatter factor defined as the ratio of the dose rate for a given field at reference depth to the dose rate at the same depth for the reference field size with the same collimator opening, \( l_{eq} \) is the equivalent square of the field defined by the opening jaws, and \( \text{TMR}(d, l) \) is the tissue-to-maximum ratio for a square field of side \( l \) at the depth \( d \) (Khan 1994). The total scatter dose is calculated by summing up the contributions of all non-central beamlets. Each non-central beamlet acts as a scatter source and its contribution to the isocentre can be computed using the modified Clarkson method (Khan 1994, BIR 1996). To the first-order approximation, the scatter contribution from a beamlet centred at a distance \( r_m \) from the isocentre is given by \( D_m = D_m^a a^2 \). Here \( a \) is the beamlet size and \( D_m^a \) is the scatter dose from a unit area at the centre of the \( m \)th beamlet. To obtain \( D_m^a \), one may first compute the scatter dose from a circular ring (Khan 1994, BIR 1996) with radius \( r_m \) and then divide it by the area of the ring, \( 2\pi r_m \Delta r_m \). To improve the accuracy, a beamlet can be further divided into a number of sub-beamlets and the dose from these sub-beamlets can be dealt with in a similar manner as described above. A sum of the contributions of the corresponding sub-beamlets gives the scatter dose of the beamlet. We found that a sub-beamlet size of 0.5 cm \( \times \) 0.5 cm can yield satisfactory results consistent with experimental measurements for all clinical cases we have tested and further reduction in the sub-beamlet size does not give notable improvement.
Of course, equation (4) is quite general and allows one to use any other method for MU calculation based on the beamlet information.

The case of multiple incident beams can be dealt with similarly. One can proceed along two directions to verify the system calculation. One is to use the prescription information and fractional weighting of each beam to derive the monitor units for each beam. These MUs can then be compared with the system calculation. When the information about fractional weighting is not available (e.g. currently the Corvus system does not provide the relative weights of the incident beams), one can proceed to use MU$_j$ for each field provided by the treatment planning system to calculate the dose at a point using

$$D = \sum_{j=1}^{J} D_j = \sum_{j=1}^{J} \text{MU}_j \left( \sum_{m=1}^{M} C_{m,j} d_{m,j}^0 \right)$$

(7)

where the index $j$ has been added to label each individual incident beam, and $J$ is the total number of beams. The dose given by equation (7) is then compared with that given by the treatment planning system. One should note that this method only verifies the composite dose, not that of individual beam.

3. Results

3.1. Simplified examples

Let us consider the delivery of a simple fluence profile shown in figure 1(a). There are a few ways to generate the fluence map by segmented delivery; two are shown in figures 1(b) and 1(c). In the first delivery scheme, the MLC leaf pair delivers the three fluence blocks in sequences. In this case, $f_1 = 1/4$, $f_2 = 1/2$ and $f_3 = 1/4$. Substituting the $f_k$s into

Figure 1. An example to illustrate the MU calculation in a dynamic delivery. For convenience, each block in the figure corresponds to 1 MU. (a) Three-beamlet intensity profile to be delivered. The beamlets are numbered 1, 2, and 3 from left to right. (b) The first delivery scheme. (c) The second delivery scheme.
MU calculation by scatter-summation algorithm

Figure 2. An example to illustrate the MU calculation in a dynamic delivery. For convenience, each block in the figure corresponds to one MU. (a) Five-beamlet intensity profile to be delivered. The beamlets are numbered 1 to 5 from left to right. (b) A four-segment delivery scheme.

Using equation (4) and equation (5) we have

\[
MU = \frac{4D}{(1 + 3\alpha)d_1^0 + 2(1 + \alpha)d_2^0 + (1 + 3\alpha)d_3^0}.
\]

(8)

If we ignore the transmission dose, the above result indicates that in order to deliver a dose \(D\) to a point using the segmented delivery shown in figure 1(b), the delivery time is four times longer than that required by a delivery using a physical compensator. This agrees with an intuitive calculation based on the simple shape of the fluence map. For the delivery scheme shown in figure 1(c), it can be similarly shown that the delivery time is two times longer than
that required by a physical compensator delivery if the leaf transmission is ignored. Note that the transmission doses in the two delivery schemes are different. The transmission is less if the delivery is more efficient.

To give another example, consider the delivery of a slightly more complicated fluence profile shown in figure 2(a) and the delivery scheme shown in the figure 2(b). The delivery has four segments with \( f_1 = 2/7 \), \( f_2 = 1/7 \), \( f_3 = 1/7 \) and \( f_4 = 3/7 \). According to equations (4) and (5) we have

\[
MU = \frac{7D}{(3 + 4\alpha)d_1^0 + (4 + 3\alpha)d_2^0 + (2 + 5\alpha)d_3^0 + (5 + 2\alpha)d_4^0}
\]

(9)

When the point of prescription is on the central axis, it takes 7/5 times longer than that required by a physical compensator delivery if the leaf transmission is ignored. This agrees again with an intuitive calculation based on the shape of the fluence map.

3.2. Clinical examples

The above method has been applied to over 20 clinical IMRT cases, including head and neck, brain and prostate patients, as well as in the surrogate phantoms for chamber measurement. The clinical IMRT plans were generated using the CORVUS inverse planning system (Xing et al 1999). The leaf sequence files were read into the software module. The fractional MU factors, \( f_k \), and the primary and scatter doses are then computed. In table 1 we show the detailed calculation for a nine-field prostate phantom plan together with measurement data obtained using a 0.147 cm\(^3\) IC-10 ionization chamber (Wollhöfer Dosimetrie, Schwarzenbruck, Germany) following the recommendations of the AAPM protocol (AAPM 1983). The measurement for each field in a cylindrical water phantom (Xing et al 1999) was used as a surrogate for measurements in the patient. For comparison, the primary and the scatter doses for each field are separately listed along with the total contribution of the field.

Table 2 summarizes the results of five clinical IMRT cases. Both the original patient plan and the corresponding surrogate phantom plan data are listed. An ion chamber measurement for each patient in a cylindrical water phantom is also presented in the table 2. In all cases, the calculated dose agreed with the CORVUS calculation within 4%, except for the 6 MV liver case where the deviation is greater than 7%. According to the report of AAPM Task Group 40

<table>
<thead>
<tr>
<th>Gantry angle (degrees)</th>
<th>Corvus calculation (cGy)</th>
<th>Primary dose (cGy)</th>
<th>Scatter dose (cGy)</th>
<th>Beam dose (primary+scatter) (cGy)</th>
<th>Ion chamber measurement (cGy)</th>
</tr>
</thead>
<tbody>
<tr>
<td>30</td>
<td>45.4</td>
<td>41.9</td>
<td>3.0</td>
<td>45.0</td>
<td>43.8</td>
</tr>
<tr>
<td>65</td>
<td>22.8</td>
<td>20.5</td>
<td>1.9</td>
<td>22.5</td>
<td>24.1</td>
</tr>
<tr>
<td>95</td>
<td>30.0</td>
<td>27.8</td>
<td>1.8</td>
<td>29.6</td>
<td>28.8</td>
</tr>
<tr>
<td>135</td>
<td>29.4</td>
<td>26.2</td>
<td>2.0</td>
<td>28.2</td>
<td>29.0</td>
</tr>
<tr>
<td>225</td>
<td>33.7</td>
<td>30.4</td>
<td>2.1</td>
<td>32.5</td>
<td>33.3</td>
</tr>
<tr>
<td>265</td>
<td>25.3</td>
<td>23.4</td>
<td>1.9</td>
<td>25.3</td>
<td>24.5</td>
</tr>
<tr>
<td>295</td>
<td>20.7</td>
<td>18.3</td>
<td>1.8</td>
<td>20.1</td>
<td>20.8</td>
</tr>
<tr>
<td>330</td>
<td>46.3</td>
<td>43.0</td>
<td>2.9</td>
<td>46.0</td>
<td>45.1</td>
</tr>
<tr>
<td>Total dose</td>
<td>253.6</td>
<td>231.7</td>
<td>17.5</td>
<td>249.2</td>
<td>249.6</td>
</tr>
</tbody>
</table>
MU calculation by scatter-summation algorithm

Table 2. Summary of dose verification for five IMRT patients.

<table>
<thead>
<tr>
<th>Disease site</th>
<th>Patient plan, Corvus dose (cGy)</th>
<th>Patient plan, current calculation (cGy)</th>
<th>Phantom plan, Corvus dose (cGy)</th>
<th>Phantom plan, current calculation (cGy)</th>
<th>Ion chamber measurement (cGy)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Prostate #1 (15 MV)</td>
<td>222.6</td>
<td>232.5</td>
<td>224.0</td>
<td>226.7</td>
<td>231.3</td>
</tr>
<tr>
<td>Prostate #2 (15 MV)</td>
<td>208.4</td>
<td>209.1</td>
<td>220.0</td>
<td>216.4</td>
<td>227.0</td>
</tr>
<tr>
<td>Brain (4 MV)</td>
<td>173.3</td>
<td>173.5</td>
<td>196.0</td>
<td>192.9</td>
<td>197.7</td>
</tr>
<tr>
<td>Liver (6 MV)</td>
<td>234.3</td>
<td>251.4</td>
<td>237.0</td>
<td>238.3</td>
<td>231.1</td>
</tr>
<tr>
<td>Abdomen (15 MV)</td>
<td>179.0</td>
<td>177.7</td>
<td>178.0</td>
<td>176.2</td>
<td>176.3</td>
</tr>
</tbody>
</table>

(Kutcher et al 1994), a disparity should be resolved before commencing or continuing a treatment if the independent calculation differs by more than 5% from the treatment plan. Our calculation did not take into account the contour variation. In the liver case, we noticed that there was a significant variation of the skin contour within the treatment fields. The irregular skin surface led to a large discrepancy in the final dose. An ion chamber measurement was used to resolve the discrepancy in this case.

4. Summary

We have provided a close formula for MU calculation from a prescribed dose and a given leaf sequence file for IMRT treatment. The MU is expressed as a function of the contributing beamlet doses and the dynamic modulation factors. The formula is quite general and can be implemented as an independent MU verification tool for any leaf sequence algorithm and delivery machine. It may also find an application in dose calculation from a known leaf sequence file and finite pencil beam kernels. The approach was illustrated in verification of the isocentre dose for CORVUS treatment plans and our results indicated that it is robust and efficient for clinical use.

Acknowledgments

We wish to thank J Kung of the University of Chicago for communicating his results and for useful discussions. We would also like to thank B Curran of the NOMOS Corporation for technical support. This work was supported in part by grants from the Whitaker Foundation and the National Cancer Institute (no CA43840).

References


Geis P and Boyer A L 1996 Use of a multileaf collimator as a dynamic missing-tissue compensator Med. Phys. 23 1199–205


BIR (British Institute of Radiology) 1996 Central axis depth dose data for use in radiotherapy Br. J. Radiol. (suppl 25) 1–183

Khan F 1994 The Physics of Radiation Therapy (Baltimore: Williams & Wilkins)

