3D DOSE RECONSTRUCTION TO INSURE CORRECT EXTERNAL BEAM TREATMENT OF PATIENTS

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Abstract—Radiation therapy treatments have become increasingly more complicated. There are multiple opportunities for humans, machines, software, and combinations thereof to result in a treatment error that could be of significance. Current methods for quality assurance are often abstract in nature and may have unclear underlying assumptions as to what is assumed to be working correctly, or may depend upon the diligence of persons to discover errors from a review of the treatment plan. Here, an example will be shown of a direct method to reconstruct and demonstrate the dose and the dose distribution delivered to a particular patient. By measuring the radiation fields that come out of the accelerator, and using the measurement as input to a 3-dimensional (3D) dose algorithm, the delivered patient dose is determined and presented in a manner similar to the treatment plan. The intended treatment plan dose may be directly compared. Using this feedback mechanism, there is less abstraction and dependence upon the diligence of individuals checking multiple steps in a treatment process, and assumptions can be clearly stated. With this system, the dose is determined and presented minimizing assumptions and dependence upon other systems. © 2007 American Association of Medical Dosimetrists.

KEY WORDS: Quality assurance, Intensity modulated radiation therapy, Electronic portal imaging device, Dose computation.

INTRODUCTION

Mistakes made by human errors, software errors, and equipment failures, can be unpredictable. In an increasingly complicated environment, the possibility of mistakes resulting in the incorrect treatment of patients with external radiation therapy becomes and has always been a serious issue. As persons can fail to discover mistakes in a review, methods that can demonstrate the dose will not be dependent upon people noticing an error from an inspection. Intensity modulated radiation therapy (IMRT), in particular, requires some sort of measurement of the output fields, as the delivery process using a multileaf collimator (MLC) to achieve field modulation is no longer a simple easily verifiable process.

Here, we show an example of a direct process that will reconstruct the dose to the patient from measurement of the delivered radiation fields to serve as a quality assurance method.1–5 The need for quality control and other methods were categorized in 2 earlier reports.1–3 The method reported here demonstrates the dose in a direct manner such that the dosimetry of the treatment plan and the delivery system has to be functioning properly for the correct dose to be reconstructed.

Equally important will be to understand what is not checked by the method reported here. It is strictly a method to test the dosimetry and delivery system. It is assumed that the accelerator is calibrated correctly for the referenced field size, typically an open 10 × 10-cm field, and that the correct dose is delivered for the 10 × 10-cm open field for a stated monitor unit (MU). It is assumed that the correct energy is selected for treatment. It is further assumed that the treatment fields will be placed on the patient correctly, as only the dosimetry and delivery systems are verified here. In this report, we will present an example verification case that illustrates the clinical application of this verification method.

METHODS AND MATERIALS

A treatment plan was done with an Eclipse planning system (Varian Medical Systems, Palo Alto, CA) for a head-and-neck IMRT case with 7 treatment fields. The aS500 Electronic Portal Imaging Device (EPID) of a Varian medical linear accelerator is used to integrate each treatment portal during a treatment run without the presence of the patient. Using the Varis software system from Varian, the resulting integrated field file names can contain the names of the treatment beams, which will allow for an automated association of each integrated file with the treatment portal using the following software employed for dose reconstruction. A software system called Dosimetry Check (Math Resolutions, LLC, Columbia, MD) is used to reconstruct the dose to the patient from the measured portal images.

Due to the large size of 7 IMRT treatment portals for the example case used for this report, a carriage shift was required for the Varian MLC to deliver each field in 2 halves. This resulted in 2 integrated files for each of the 7 treatment portals. With the beam names imbedded in these file names, Dosimetry Check can automatically add each of the 2 halves together to complete the treatment...
portal. Otherwise, the user will have to manually select
the files for a particular beam that are then added if more
than 1 is selected. Figure 1 shows the IMRT fields for
this example case, as integrated by the EPID and after
adding the 2 carriage shifts for each field together. The
fields are clearly heavily modulated, necessitating small
segments in some cases.

The treatment plan is downloaded by means of the
Dicom RT protocol to the Dosimetry Check program (an
RTOG download is also possible). The information in-
cludes the computed tomography (CT) scans, the out-
lined regions of interest, including the skin boundary
outline, the beam geometry in regard to the position of
isocenter in the CT scan set and machine angles, and the
planning system’s computed 3D dose matrix. The 3D
dose matrix is used to compare to the dose that is
reconstructed here by Dosimetry Check.

The EPID is placed about 105 cm from the source at
its closest position so that it can intercept as large a field
as possible. Otherwise, the distance does not matter, as
distances from 104 to 155 cm have been used, and
produce the same results. Using x-ray film at close dis-
cances, such as the blocking tray, has not been studied to
establish a useable range. In addition to the treatment
portals, a 10 × 10-cm field is integrated for a known MU
setting, 100 MU in this case, for calibration. The 10 ×
10-cm field should be integrated under the same condi-
tions as the treatment field, although the inverse square
law could be applied. The signal value averaged over a
small area at the central axis (CA) position (a 0.5-cm
square) of the 10 × 10 field images is used to normalize
all the other integrated field images by multiplying all the
integrated pixels by the ratio of the MU used for the 10 ×
10 field to the integrated pixel value for 10 × 10 at the
CA. The 10 × 10-cm field may also be used to find the
CA of the integrated field images upon the assumption
that the EPID is not moved between the IMRT field and
the 10 × 10 cm field. If the EPID is known to drift a
great amount with gantry rotation, then a 10 × 10-cm
field will be needed at each gantry angle. In the example
reported here, all fields were irradiated with the beam
pointed toward the floor.

The calibration condition for the accelerator in terms
of the dose delivered to depth for a stated source-to-surface-
distance (SSD) for the 10 × 10-cm field size for a stated
MU must be defined (typically units are calibrated at depth
with 100 cm either to the surface or to the point at depth).
The treatment portal images are then processed by decon-
volution with a point spread function (kernel) to reduce the
images to an in air fluence array. A unit called Relative
Monitor Units (RMU) was invented for this fluence.

The kernel is found by a fitting process that uses as
input the integrated field images for a series of increasing
square field sizes. The kernel parameters are optimized

Fig. 1. The images of the 7 IMRT fields for the example case, integrated by the EPID.
in a least-squares fit that will transform the square field images into a fluence array that will result in the correct dose computed in a water phantom.

This method does not model the accelerator head or field segments. Rather, this is what is measured. The deconvolution process first transforms the 2D images into 2D spatial frequency space. There, the spatial frequencies are multiplied by the kernel (which is the inverse of the point spread function for the EPID). Segment size is therefore not a consideration, as the problem is transformed to processing the spatial frequencies. Having fitted a mathematical form\(^2\)–\(^4\) for the deconvolution kernel, it is expected to hold for the normal range of spatial frequencies encountered. Successful results as shown here and elsewhere indicates that this assumption holds.\(^2\)–\(^4\)

In the kernel fitting processes, a 3D dose calculation engine using a pencil beam algorithm computes the dose in phantom using the respective fluence array to define each square field. In this manner, the fluence is consistent with the 3D dose algorithm employed.

An additional operation is still needed. Calibration of the EPID with a flood field will remove the in-air off-center ratio (OCR) profile from the fluence fields. This must be restored by multiplying the integrated pixels with a measured in-air off-center axis ratio as a function of radius, assuming circular symmetry.\(^3\) It is therefore important that the radiation field be properly symmetrical at the time of the flood field calibration. The Varian EPID integration system has the option of multiplying the integrated result by a measured profile at \(D_{\text{max}}\) in water for the largest field size. There has not been a significant difference noted between using images with the in-water factor vs. multiplying in the in-air OCR. Further, the user also has the option of supplying the Varian software with the in-air profile instead of an in-water profile.

Using the above deconvolution kernel and integrated 10\(\times\)10-cm field for normalization, the absolute dose in patient or phantom can be computed from the integrated field images.\(^3\) In the software program, each beam is associated with the integrated field and the dose is computed to the patient using the downloaded beam geometry and CT scans.

The assumption is that the proper dose is delivered to an open 10\(\times\)10-cm field for the given MU (because the beam images are normalized to the 10\(\times\)10 field) and that the correct energy beam is being delivered. It is of course further assumed that the treatment beams are on the proper place on the patient. Here, only the dosimetry and delivery is being tested.

The accuracy of the 3D dose algorithm was tested by comparing about 300 points that were measured in a Rando Phantom irradiated by a large field in an earlier report.\(^1\)\(^,\)\(^6\) The measured and computed points agreed to...
Fig. 3. Coronal plane through isocenter showing reconstructed dose (magenta) and plan dose (dotted green) for isodose values of 5800, 4640, 2900, and 1160 cGy (100%, 80%, 50%, 20%, respectively).

Fig. 4. Sagittal plane through isocenter showing reconstructed dose (magenta) and plan dose (dotted green) for isodose values of 5800, 4640, 2900, and 1160 cGy (100%, 80%, 50%, 20%, respectively).
slightly less than 3% at one standard deviation (SD) relative to the CA dose at $D_{\text{max}}$. This 3% will include any measurement error (the difference between measured and computed is considered); nonetheless, serves as a reasonable upper estimate of the accuracy of the algorithm. One would expect other classes of similar dose algorithms in treatment planning systems to perform at this level of system performance or better. Therefore, in comparing the reconstructed dose to the planning system dose, it would not be reasonable to use a criterium smaller than 3%.

RESULTS

After the dry run for measurement of the fields, it takes about 15 minutes for an experienced user of the software to process all files involved. About another 30 minutes is required to compute the statistics and displays shown below (a 1.6-GigaHertz computer running under Windows XP was used for this example case).

Results of the various methods of dose comparison between the dose reconstructed here and the planned dose are shown in Figs. 2 through 11.

Figures 2 through 4 are isodose plots in the transverse, coronal, and sagittal planes, respectively, here chosen to run through isocenter. The reconstructed dose is shown in magenta and the planned dose is in green and dotted. Because the Eclipse system downloaded the dose for 29 fractions, the reconstructed dose for a single fraction was multiplied here by the factor of 29. As seen, the isodose curves for a few selected dose values are close together. However, in a low-dose gradient area, 2 curves could be separated by a large distance; however, such separation would only represent a small consequential difference in dose. For this reason, gamma analysis provides a useful tool for evaluating 2 dose distributions.7

Shown in Figs. 5 through 7 are the same planes, with the regions of a gamma value of 1.0 or more tinted red. Regions with a gamma value of 1 or less represent areas where the dose agrees within 3% (3% of 5800 cGy in this case) or it is less than a 3-mm distance to a point where the dose is the same. The distance is searched in all 3 dimensions so that this is a 3D gamma analysis. The results show good agreement in the high-dose low-gradient areas. Larger disagreements in high-gradient areas and low-dose areas are less consequential.

Figure 8 is a gamma volume histogram shown for the whole patient volume, represented by the CT scan set and skin boundary outline. Similar histograms may be shown for any selected region of interest outlines. The percentage of points less than a gamma value of 1.0 is a useful statistic, here 88.3% is less than or equal to 1.0. Further, the same statistic can be computed for points with a dose greater than some dose threshold. For points with a dose $\geq 20\%$ of the target dose of 5800 cGy, 85.9% of the volume is $\leq 1.0$.

Figure 9 shows a 3D room central point perspective view of the gamma isosurface of value 1.0 under the transparent skin boundary surface.

Figure 10 is a dose difference volume histogram. Perhaps of less use than the gamma analysis because it does not consider distance to a point in high-gradient areas, the dose difference stated as a standard deviation may serve as a useful statistic.4 The value computed here is 3.6% of the target dose at 1 SD.

Figure 11 shows the DVHs computed for selected region of interest volumes. For the same points, the DVH computed from the planning system 3D dose matrix is shown as a corresponding dotted line. However, there may be some differences in the volumes reconstructed from the same set of 2D contours due to different algorithms used to generate volumes from contours. Dosimetry Check, for example, does shape interpolation at 1-mm intervals between contour planes. The treatment of voxels that contour lines run through may be different. For this reason, a direct comparison with the DVH computed by the planning system is not done.

DISCUSSION

The results show that the method of reconstructing the dose to the patient can serve as a tool for quality assurance of the delivery of dose and dosimetry. Only a few assumptions are made as to the proper performance of equipment. Further, the method directly demonstrates the dose to the patient rather than rely on an inference that the dose is correct if some other abstract measurements or process is correct. Differences in dose due to differences in the delivery system are directly demonstrated, which cannot so easily be inferred from comparing predicted individual beam profiles to measured profiles or composite dosimetry in a phantom.

However, any measurement is accompanied by uncertainty. Whether the small differences seen here are due to measurement error or represent real differences was beyond the scope of this report. The treatment delivery system is also a dynamic system and it cannot be assumed that it will produce the same results if the measurements are repeated. This would require the accumulation of statistics in a broader study—again beyond the scope of this report. This method only tests the dosimetry and delivery at the time the fields are measured. Other means would have to be used to guarantee consistent daily treatment. An integrating array that one could treat through would be a desirable improvement.

The basic dosimetry and calibration of the accelerator must be performed and tested separately. Other means must be used to verify the correct placement of the fields on the patient, such as comparing portal images.

Several methods and statistics have been demonstrated here for developing a pass-fail criteria for a positive quality control result.1,3,7
Fig. 5. Transverse plane through isocenter showing the areas of gamma value of 1.0 or greater for 3% or 3 mm.

Fig. 6. Coronal plane through isocenter showing the areas of gamma value of 1.0 or greater for 3% or 3 mm.
This method is more likely to produce a false negative than a false positive. A false positive would require a measurement or processing error to mask a treatment or dosimetry error, or an assumption made that turns out to be false.

A false negative would be produced by any error in the dose reconstruction process, from measurement of the fields, to the processing of the field data, to the association of the field data with the treatment beams. Experience has shown that some diligence in installing the method and training of people is required to get positive results. A negative result would trigger an investigation.

If a negative result is produced, then each field can be studied individually. The fields are all ready measured separately. However, the 3D dose matrix from the planning system is a simple sum of all fields, and the contribution from individual fields cannot be separated out of the sum. Therefore, the planning system would have to be used to compute a 3D dose matrix for an individual field and that plan downloaded separately, one plan for each treatment field (by field, we mean the sum of all segments for a treatment portal). The process for evaluating the plan for an individual field proceeds identically to that of the composite plan process showed above, as it makes no difference how many fields are in a plan.

**Wedges**

A plan with physical wedges will require special consideration when using an EPID to integrate the fields. A wedge changes the spectrum of the beam by differentially absorbing x-rays of different energies. The EPID responds differently to a different spectrum. This was shown to result in large errors for the wedge factor.\(^3\) Hence, the dose would be reported to be wrong, resulting in a false negative. However, it was also noted that the dose profile under a 60° physical wedge was correct.\(^3\) This indicates that the effect of an increasingly thick absorber diminishes the effect on the measuring EPID.

![Fig. 7. Sagittal plane through isocenter showing the areas of gamma value of 1.0 or greater for 3% or 3 mm.](image)

![Fig. 8. Gamma volume histogram for the skin boundary volume.](image)
If a plan uses wedges, there are several strategies available. One would be to use x-ray film instead of an EPID. A second possibility would be to first test the EPID with each wedge to determine if the profile is correctly reproduced. One would expect failure to be more likely with the 15° wedge than the 60°. If the profiles are not accurate, then only x-ray film is a resort. If the profiles are accurate, then only the wedge factor needs to be dealt with.

There are 2 possible ways to deal with the wedge factor. One would be to use a separate deconvolution.
kernel for each wedge, where the wedge factor is corrected by fitting the kernel using wedged fields. This has the disadvantage of having to select a separate kernel for wedged fields, with a separate kernel for each wedge, or to otherwise correct each wedged field with multiplication by a correction constant determined for that wedge. Another method would be to measure the field fluence with an ion chamber normalized to a 10 × 10-cm open field measurement, and renormalize the EPID integrated pixel values accordingly. A software function is provided to do this. This will have the disadvantage of having to either make a separate measurement, or to having to build an ion chamber into the EPID detector in some noninterfering fashion so that the measurement could be made during the integration run.

Hardening of the beam by wedges is also known to have a small effect on the dose at depth, which is not accounted for in the method. But this effect is known to be small—of the order of 1%—and need not be considered here.1

Lastly, physical wedges are generally not used for IMRT treatments where the reported method is of greatest use.

Compensators

Compensator-based IMRT was the first form of IMRT reported in the literature6 as currently done by solving for the intensity of field pixels individually as a solution to a mathematical formulation of a treatment objective function, and remains an alternative to MLC-based field modulation. Unlike the case of the wedge above, it is not going to be practical to test each compensator to see if the correct dose profile is produced by an integrating EPID. Because it is known that a large error in wedge factors occurs, even though the profile might be correct, a wedge being a special case of a compensator, there is going to be some range of thickness, from 0 to that thickness, where the dose profile is not going to be correct. For compensator-based IMRT, some other detector, such as x-ray film, will have to be used to measure the fields for input to the method. The EPID cannot be used unless some constant thickness can be added to all the compensators, and it is demonstrated that a single renormalization factor is all that is needed for correct dose prediction. Investigation of these considerations was beyond the range of this report.

Dose algorithm

Lastly, some consideration of the algorithms used with the method and the planning system may be of small consequence. However, experience with this method with the Pinnacle (Philips, Andover, MA) and Eclipse treatment planning systems has shown that the dose can be verified at the 3% level. This would indicate that current algorithms are consistent to this level, which is sufficient for screening for consequential dose errors. Larger differences might be encountered in areas known to be prone to algorithmic errors, such as interface regions.

CONCLUSIONS

The method reported here can serve as an unambiguous and direct quality control method for patients undergoing radiation therapy to help guarantee and document that the treatments are correct. The method constitutes a feedback loop that demonstrates the dose the patient will receive if the treatment fields are correctly placed on the patient. In so doing, there is not a dependency on persons discovering an error through a review of the planning and treatment process. The method is sensitive enough to avoid large consequential dose errors to patients.

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REFERENCES