Dosimetry Check- Portal Dosimetry System

For true 3D IMRT and IMAT dose verification and dose reconstruction with absolute dose using available EPID, ion chamber array, or diode array for conventional linacs, and the TomoTherapy detector array for TomoTherapy:

- Measure the radiation fields pre-treatment.
- OR measure the fields during treatment for exit (transit) dose reconstruction.
- Using measurement is working backwards to compute the dose to the patient.
- Compare to the plan dose.

Dose calculation algorithm used in Dosimetry Check

The program uses a pencil beam algorithm. The pencil beam is developed from measured beam data and Monte Carlo calculated kernels. A single poly-energetic kernel is used to represent the pencil. There were two considerations for making this choice. First we would like the algorithm to be as fast as possible. Calculating each energy separately with a spectrum and mono-energetic kernels would reduce the speed significantly. It is our philosophy that if greater accuracy is desired, one may as well use a Monte Carlo algorithm, so that there is a fast algorithm for quick results and a slow one for greater accuracy. Second, for the Dosimetry Check application, we have to consider the input information. We can only measure dose in a plane perpendicular to a beam, not the spectrum, at individual pixels covering the radiation field. Therefore we need an algorithm that can start with the intensity distribution in reference to dose.

In words, dose_c is the computed constant that converts everything that follows to give the calibrated dose for the calibration field size, SSD, and depth. The units of the pencil kernel therefore to not really matter. Dose_c is simply the ratio of the specified dose rate, usually 1.0 cG/mu, for some field size, SSD, and depth, to the result computed for the same field size, SSD, and depth.

The Off Axis Correction is table developed from measured diagonal fan line data taken at depth for the largest field size. This two dimensional table is simply the ratio of measured over computed values stored as a function of the tangent of the angle with the central ray, t above, and depth. The depth de is the effective depth of the point of calculation P. de is computed by tracing along a ray from the point of entry into the patient body, and summing the incremental path length times the density at the location. This table provides a means of accounting for the change in beam penetration off axis since the kernel was developed from data on the central ray.

The Sad/Spd squared term is the inverse square law, where Sad is the source axis distance of the treatment machine, typically 100cm. Spd is the distance from the source to the plane point P is in, along the central ray.

Next follows the integration of the pencil beam kernel over the area of the field. This formula is of course only symbolic. We can only integrate numerically on a computer, but the formula does show the mathematical idea. The kernel K(r,de(r,q))/2pr is the dose at the radius r at depth to the point P from the incremental area dr rdq. de(r,q) in cylindrical coordinates centered at P equals

$de(x_r,y_r)$ in Cartesian coordinates and is the equivalent depth along the diverging ray through the differential area element at the surface to the plane perpendicular to the central ray that contains the point P. We actually don't have this kernel K, what we do have is the integration of K from 0 to r which is the dose at P from the circular disk of radius r at depth. So we actually take a difference here over dr in the numerical computation. Note that the r term cancels out and we have taken the 2p term outside of the integral.

The differential area element is weighted by the field intensity Field at (r,q), here shown in the Cartesian coordinates (x_r,y_r) at the distance sad. For the Dosimetry Check program, this is looked up directly from the pixel value at that location from the measured field. In comparison program GenerateFieldDoseImage computes this value at each pixel location as the product of the monitor units, the scatter collimator factor, the in air off center ratio table, and the attenuation of any shielding blocks. For general treatment planning we would here also model the attenuation of any wedge, compensators, and any other devices that effect the field fluence such as a multi-leaf collimator.

The one over the maximum of one or the in air off axis ratio (OCR) term is to correct the kernel for having been derived from data that contains the effect of the OCR. Here we divide it out as long as the OCR has a value greater than one. The effect of most flattening filters on the OCR is that the in air fluence increases initially with radius. We want to correct for that effect. Otherwise for points on the central axis we could be effectively applying the effect of the OCR twice during integration over the area of the field, once built into the kernel and then formally through the above Field term.

But in the corner of the largest field the OCR value often decreases due to a cut off of the field due to the primary collimator. We do not want to cancel out that effect in those areas. The correction here is a compromise to achieve some deconvolution of the OCR term out of the kernel term. As the OCR is stored in terms of the tangent angle a ray makes with the central ray, we divide the radius at depth by the SSD used to measure the fields from which the kernel was derived from plus the effective depth along the ray from the differential area element to the plane perpendicular to the central ray that contains the point of calculation P.

Note that there are two effective depths used in the above equation. One is along the ray to the point of calculation P, de(x,y). The other is inside the integration (sum) and is traced along the ray from the surface to the differential area element dr rdq = $de(x_r,y_r)$. The ray tracing along each ray is only done once. A fan line grid covering the area of the beam is created. Each fan line is ray traced through the patient with the storing of the accumulated equivalent depth at intervals along each ray. The spacing between the fan lines is set at a level half way through the patient to the dose matrix spacing value that the user can reset from the default value. Below this plane the fan lines are diverging further apart and above the plane the fan lines are converging closer together. Node points are distributed along the fan lines at equally spaced intersecting planes perpendicular to the central ray, and it is at these node points that the equivalent depths are stored. During the above numerical integration over each plane, the nearest ray is used to look up the equivalent depth. The same node points that hold the equivalent depths also define the points to be calculated. All other points are interpolated within this diverging matrix.

Let us note here the approximations that are made. The integration is in the plane perpendicular to the central ray that contains point P. Rigorously it can be argued that the integration should be in the plane perpendicular to the ray from the source to the point P. For a worst case, the largest field size of 40x40, the ray at the edge of the field on a major axis makes an angle of 78.7 degrees instead of 90 degrees, an 11.3 degree difference, which is not going to make a large difference. The Off Axis Correction above will tend to cancel out such approximations. To be consistent, the Off Axis Correction is computed using the same algorithm. Using the inverse law correction for the slant depth along the ray from the source to point P made no difference from using the vertical distance parallel to the central ray. Another approximation is made below in the generation of the pencil kernel. It is assumed that at a depth d and radius r, the dose will be the same for a cylinder of radius r and for a diverging cone of radius r at the same depth. This is the same approximation made in the concept of Tissue Air Ratio.

Rectangular arrays of equally spaced points are generated for planar images, and a three dimensional lattice of points is generated for 3d perspective room views. The dose at these points are computed for each beam by interpolating within the above diverging fan line array of points generated for each beam. This means that in general the dose is interpolated between the eight corners of the diverging box that a point is inside. We will interpolate with less than eight points provided that the sum of the weights from the points is greater than 0.5, as some node points might fall outside of the patient. Once computed, each beam saves its dose matrix to a disk file, unless the beam is changed in which case a new fan line dose matrix array is created.

Dose Volume Histograms

You can compute the dose volume histogram for outlined regions of interest. The dose is that computed by DosimetryCheck, not by the other treatment planning system. Shown below is the volume histogram popup. We will refer you to the RtDosePlan manual for details on the use of this option, under the Plan section. Dose volume histograms are provided here as there may be some use in comparing to the planning system. In summary, you can select the same volume for different plans, as well as different volumes. Hitting the compute button will start the computation which you can stop when the plotted curves have settle down.

While the computation is running more points are being computed. When any volume reaches a point to volume density of 1000 (1000 points per cc) that volume will not be computed anymore. The computation will stop if all volumes reach that density.

Dose Comparison Tools

Several tools are provided for comparing and showing the dose distribution that may have been downloaded from the treatment planning system. These tools can be used for enhancing your understanding of any dose difference. The tools are found under the Evaluate pull down.

Display Point Doses

Show Point Dose Values

You can click the mouse on images and get the computed Dosimetry Check (DC) and Treatment Planning System (TPS) plan (also called foreign) dose

Compare 1D Dose Profile

You can compare the dose profile between the planning system and reconstructed dose for any line through the patient space. Do you this by selecting any reformatted image through the patient, and then positioning a line on that plane

Compare 2D Isodose Curves

You can select to view isodose curves together from the downloaded TPS dose along with the dose computed here. Or show one or the other, or show the difference between the two dose sets. The dose difference is the absolute value between the DC computed and TPS dose.

Compare Dose in 3d Room View

To engage this control with a current frame, you must first make a 3d room perspective view of the primary image set of the current plan or an image set fused to the primary image set

Gamma Method 2D and 3D

The gamma method is a method to compare two dose distributions Reference to the DC reconstructed Dose

Gamma Volume Histogram

A volume histogram of the gamma function can be computed

An accumulative distribution or differential distribution may be shown depending upon the choice of a toggle button. The distribution may be shown for any selected outline region of interest

Show 2D and 3D TPS Dose

There is an option to simply display the dose from the treatment planning system (the reference dose). Select either the Show 2D TPS Dose or the 3D control (see above compare 3D control) under the Evaluate pull down. You then type in the dose value for the isodose curve to be displayed. Then hit the "Display in Current Frame" button. All the isodose curves will be shown in the same color. You can select the color, and also tint the dose for the current isodose value which you can select with the mouse from the list. This also will allow you to show the dose on a fused image set. The isodose of reference and the corresponding Isodose line from DC check is displayed together for comparison.

***M*onitor Unit Check**

A monitor unit check can be reported. The monitor unit check is accomplished with the following formula:

DC mu = plan mu X plan dose / DC computed dose

Where plan dose is the planning system reported dose to a point from a single beam,

DC computed dose is the dose that DosimetryCheck computes at the same point,

Plan mu is the monitor units from the planning system for the beam, and

DC mu is the monitor units found from considering the ratio of the dose to the point.

The DC mu is not to be used in treating a patient. This report is only for verification purposes. Any large difference should be investigated and corrected.

Beam Dose Specification Point

A beam dose specification point is defined by Dicom RT items (300A, 82) and (300A, 84), the location of the point and the dose. If both are present for each beam, then you can base the mu check report on that information. As the beam contribution and monitor unit should be in the Dicom RT plan file which you imported, you should have not to enter any further information. The dose specification point is for one fraction. Each beam will have a separately specified point. The points might or might be at the same physical location.

Specific point

You can alternately select a point from the list of specific points. The beam contribution of the dose to that point from the planning system will not be known from the Dicom RT download, and so you will have to enter those values in the example popup shown below from referring to the planning system. The monitor unit should be known from the Dicom RT download, Dicom tag (300A, 86). If not, then you will have to enter those values as well

Technical Specs

1 3D In-Vivo Patient Exit Dosimetry

2 Transit Dose Export to any 3rd Party Software.

3 Uses an independent 3D dose calculation algorithm to calculate dose

4 Can perform secondary check like any other Mu Check software.

5 Directly demonstrates the dose to the patient using the Volumetric CT Data Set

6 Can use customers measured beam data to match TPS output

7 User defined dose matrix

8 Calculates dose using the entire CT Data set

9 Uses EPID as the measurement device for both absolute calibration and intensity maps

10.No additional hardware requirements

11 Provides 0.25mm sample resolution

12 Displays composite dose for all beams

13 Displays dose of each single beam

14 Displays dose in a 3D Dose cloud

15 Displays dose in the Coronal, Sagital or Axial planes

16 Can reformat CT Image in any plane and display dose

17 Calculate and display 1D profiles in any plane through the CT data set

17 Displays reconstructed calculated dose and planning system dose in any generated CT image

18 Displays 3D volumetric comparative Dose Volume Histograms

19 Displays 3D Gamma Volume Histograms Structure by Structure

20 Imports TPS Calculation Points (Dicom RT)

21 User can add additional 3D points in DC

22 Calculated points anywhere in the CT data set and displays calculated vs. measured

23 User defined inhomogeneous correction

24 Provides MLC QA tools using the EPID: distance function and profile function on EPID image.

25 Measure distance function to determine leaf position

26 Deconvolution Algorithm to remove scatter from the EPID

27 Contouring tools

MlcGapTest Program
For LeafLine QA

Introduction

This program is provided to measure the 50% edge of a field using an EPID or equivalent imaging system. This can be used to verify the leaf position of a multi-leaf collimator (MLC). The program can produce a report for a defined test, such as one or a series of parallel openings. Following several journal papers, we will call this a GardenFence test to distinguish from what we are calling Picket Fence below.

The program also provides analysis for Picket Fence, which we here define to be that in AAPM report 72, Task Group 50, on page 31. With Picket Fence we analyze the width of the gap detected and the center of the gap (which can be a dip or a bump in the dose) for each leaf.

Picket Fence

The Picket Fence test is defined in AAPM Report No. 72, Task Group No. 50, page 31.

The name of this test is not assigned as "Picket Fence" in that report. We are using that name to distinguish from Garden Fence above, and choosing these names because of references we found in two papers published in Medical Physics that used this terminology.

Garden Fence

We define Garden Fence as a series of one or more rectangular fields that are separated. This can be used to verify the leaf position of a multi-leaf collimator (MLC) and also to measure the leaf gap.