

The gamma evaluation method as a routine QA procedure of IMRT

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ABSTRACT

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BACKGROUND: The conventional QA procedures dedicated to 3D CRT are unsatisfactory if the dMLC is in operation. In the case of IMRT not only should the dose on the beam axis, but also its distribution in the total plane perpendicular to the beam be taken under control. The comparison between the predicted and the observed fluence can be achieved using the gamma method. It takes into consideration the dose difference and the spatial displacement between analyzed points to provide a γ -index as a result of comparison.

AIM: The aim of the investigation was to develop the procedure of IMRT verification based on the gamma algorithm.

MATERIALS AND METHODS: 700 patients have been irradiated using IMRT since 2002. Over 1500 images recorded on the film and/or EPID have been analyzed with the help of self-made software. Histograms of γ -value and the γ -images have been created for each field. The fields have been classified depending on tumour location and the method of dose delivery, to obtain an average result for each class. We have performed a comparison of γ -histograms acquired with the help of different methods of recording.

RESULTS: We have observed a correlation between results of verification obtained with the help of the gamma algorithm and the method of intensity modulation.

CONCLUSION: Gamma evaluation allows one to find local hot-spots caused by irregularities in leaf motion or the tongue-and-groove effect.

KEY WORDS: IMRT, QA, Gamma evaluation, Gamma index, DTA

BACKGROUND

The concept of IMRT is an intentional diversification of dose distribution for the purpose of the best irradiation of the tumour body and simultaneous protection of the organs at risk [1]. The front of the IMRT beam in contrast with 3D CRT can be strongly undulated and the traditional QA procedures are not able to monitor the dose in the total plane perpendicular to the beam axis. There are several circumstances having an effect on the point dose really absorbed by tissue. The most important is the dynamic mode of MLC, which is required for dose diversification and pre-planned for intensity modulation. Of course, we cannot predict random failures and accidents which have great weight in the difference between

expected and acquired dose. However, from time to time there are occurrences which are undesired but repetitive and influence the local dose accumulated by the absorbent. They take place during proper execution of the treatment plan, as has been carefully discussed by Ping Xia and Lynn J. Verhey [2].

The irradiation leakage through a slit between adjacent leaves can be only statistically taken into account, which may be done by a treatment planning system (TPS). We can observe straight lines with high dose (hot-spots), whose escalation depends on the time of exposure. In order to reduce the dose leakage the border between leaves is in fact not a straight line [2].

Figure 1 presents the cross-section of the MLC and the intensity profile perpendicular to the direction of leaf motion. If the velocity of the two next leaves is very different, one can observe the “tongue and groove” effect. It occurs if significant dose gradients perpendicular to the leaf direction are expected. An example of this situation is presented in Fig. 2: the narrow region (white line) with dose much below the expected value. In the isocentre plane they are about 1 mm wide. Functional motions of irradiated tissues and ineffectiveness of patient immobilization usually reduce this undesired effect if the treatment consists of many fractions and the plan contains multiple gantry positions [3]. However, it is difficult to answer the question: Is the acquired dose distribution always really acceptable in the case of proper realization of the IMRT plan?

There have been several methods of IMRT verification proposed and strongly recommended in previous publications [4, 5, 6, 7]. The investigators suggest detailed control of the IMRT plans, comparison of optimal distributions obtained by TPS and different methods of calculation. Chui et al. [8] present some useful tests which help to keep the MLC in a good condition, which is the key to proper IMRT execution. On the basis of the above-mentioned papers of Depuydt et al. [5] and Low et. al [7] we have developed our own procedure for clinical dosimetry of IMRT treatment. It takes advantage of the gamma evaluation method to compare predicted dose distributions with images recorded on the treatment unit.

AIM

The aim of the investigation was to develop an independent procedure of IMRT verification which would be performed on each treatment unit regardless of whether portal dosimetry is available or not. The procedure should be based on gamma algorithm and be viable in a clinical environment.

MATERIALS AND METHODS

In many radiotherapy departments the predicted dose distribution and the acquired one are compared using the gamma evaluation method. As a result of analysis the matrix of $\gamma(\vec{r}_c)$ is obtained. For each reference point

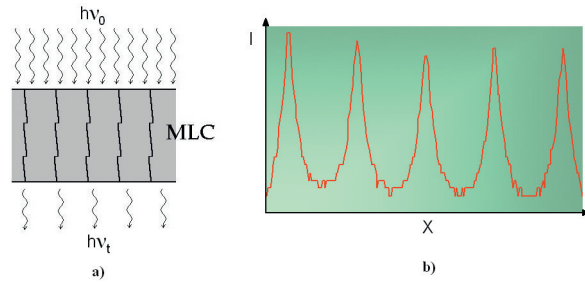


Fig. 1. The irradiation leakage through a slit between adjacent leaves of the collimator. a) the crosssection of the MLC ($h\nu_0$ – incident radiation, $h\nu_1$ – transmitted radiation), b) intensity profile perpendicular to the MLC direction

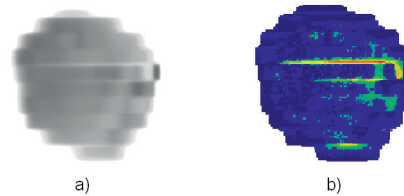


Fig. 2. The tongue-and-groove effect: a) dose distribution recorded on the film, b) corresponding γ -image

with respect to all measurement points \vec{r}_m the series of $\gamma(\vec{r}_c, \vec{r}_m)$, values is calculated using formula (1):

$$\gamma(\vec{r}_c, \vec{r}_m) = \sqrt{\frac{|\vec{r}_c, \vec{r}_m|^2}{DTA^2} + \frac{|D(\vec{r}_m) - D(\vec{r}_c)|^2}{\Delta D^2}} \quad (1)$$

where:

$|\vec{r}_c, \vec{r}_m|$ – distance between analyzed points,

$|D(\vec{r}_m) - D(\vec{r}_c)|$ – dose difference,

DTA, ΔD – scaling factors equal to 3 mm and 3.3% respectively.

As a final result $\gamma(\vec{r}_c)$ the minimum of $\gamma(\vec{r}_c, \vec{r}_m)$, is chosen for each \vec{r}_c .

In our cancer centre IMRT was started in 2002. In the beginning treatment plans were prepared using CadPlan/Helios, and recently Eclipse (Varian Medical Systems Inc, Palo Alto, CA). Varian Clinac 2300CD and 23Ex linear accelerators with Mark II 80MLC and

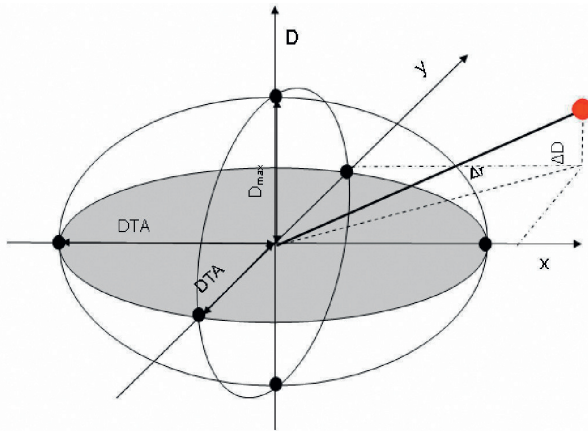


Fig. 3. The concept of gamma verification [5]: x, y, D – spatial and dose dimensions; DTA – distance-to-agreement; D_{max} – max dose deviation; $\Delta r, \Delta D$ – local spatial and dose divergence of the analyzed point

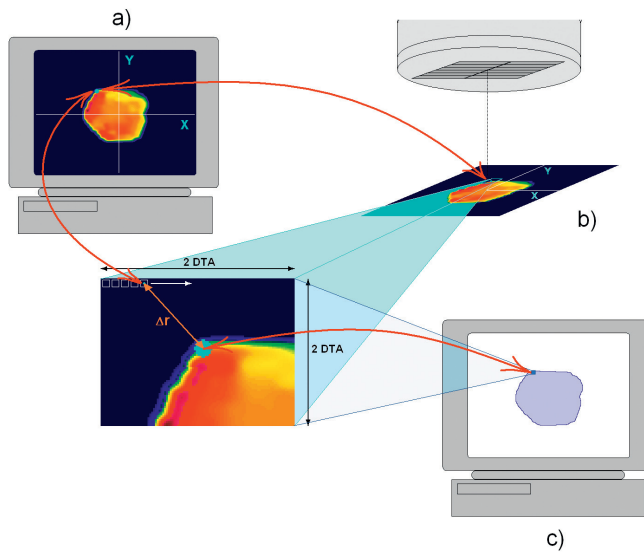


Fig. 4. The verification procedure: a) predicted dose image (film/portal) – treatment planning system, b) film/portal exposure – treatment unit, c) gamma calculations – home-made software

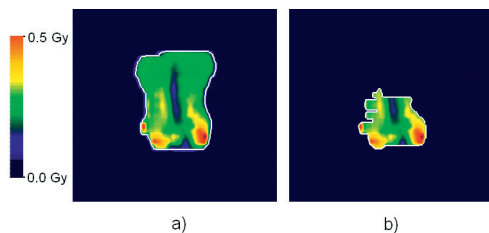


Fig. 5. The influence of border-dose parameter on the field area: a) 40% of max. dose, b) 20% of max. dose

Millennium 80MLC collimators were used for treatments.

The first step of the verification procedure (the verification mechanism is shown in Fig. 4) is to irradiate the solid water phantom (RW3, PTW Freiburg) with a dosimetric film inserted (X-Omat V, Kodak). For simplicity, the gantry and collimator rotation are set to be zero. However, the number of MUs and the leaf position sequence are imported from the original treatment plan. The source-to-film distance is equal to 100 cm and the beam axis is perpendicular to the surface of the phantom. The film is placed at a depth equivalent to 5 g/cm² in the plane parallel to the phantom surface. Depending on the film dosimetric characteristics (the most linear dependence between the optical density and the dose for XOmat V is between 20 cGy and 80 cGy, the dose variations outside this range are difficult to perceive) it is sometimes necessary to decrease the number of MUs. One film for each therapeutic field is irradiated and then developed in a semi-automatic developer (AX300 SE Alphatek Corporation Inc., Broadview, IL). The developed films (VXR-12 plus, VIDAR Corporation) are digitized (Mephysto MC², PTW Freiburg).

The specification of the therapeutic field border is difficult in the case of intensity modulation. The field aperture is typically defined as a 2D matrix of points accumulating no less than 50% of the maximal dose in the plane perpendicular to the beam axis. In the case of IMRT technique this approach could entail the loss of information about part of the field, especially if the modulation of intensity is great. We decided to reduce the criterion down to 20 percent (compare Fig. 5). Areas with dose below 20% situated inside the contour are understood as integral parts of the field, similarly to shielded space in the case of standard radiotherapy.

The comparison of the dose plane measured on the treatment unit to the predicted dose distribution is performed on self-made software GammaEval (BORLAND Delphi) based on a gamma algorithm. Distance-to-agreement (DTA) and acceptable dose deviation (ΔD) are set to 3 mm and 3.3% of the local reference dose respectively. In accordance with equation (1) calculations are performed for each reference point \vec{r}_c successively with re-

spect to all \vec{r}_m and \vec{r}_c passes an examination when $\gamma(\vec{r}_c) = \min \gamma(\vec{r}_c, \vec{r}_m) \leq 1$. In theory calculations should be performed for all \vec{r}_m but in practice we believe that every measurement point at a distance greater than 3 x DTA from \vec{r}_c can be passed over.

As a result of comparison we obtain a coloured γ -image which is a visualization of the 2D matrix of $\gamma(\vec{r}_c)$. The fusion of the image and the corresponding DRR helps to recognize and localize irregularities of dose accumulated by the target and organs at risk. However, the quantitative estimation of field irradiation is possible by courtesy of γ -histograms which combine information about the γ -index value (the quality of irradiation) with the area of the corresponding part of the field (see Fig. 6).

The alternative method of IMRT verification is portal dosimetry. Only the a-Si portal imagers are supported by portal dosimetry. The calibration procedure of our detectors has been performed according to the manufacturer's instructions: the dependence of field size on the signal intensity as well as beam diagonal profile have been measured. All the measurements have been done at 105 cm source-to-detector distance (SDD), as was suggested. Using the electronic portal image device (EPID) we are not able to exactly measure the dose distribution, which is the most important difference between the described method and film dosimetry. However, the quantitative parameter of the detected signal has been defined. Its value is set to be 0.907 CU (calibration units), while 100 MUs of photon, 10 cm x 10 cm field size beam is generated and the SDD is set as mentioned above.

The QA of IMRT treatment begins with verification plan preparation and predicted portal images calculation, which is performed with the help of the TPS. Before the exposure, the EPID was zeroed, and the dark field and flood field were measured.

We have a possibility to perform the evaluation of acquired images with the help of portal dosimetry software integrated in TPS as well as to compare the signal recorded by the imager with a predicted image using our home-made software. In contrast to commercial software we obtain not only the 2D matrix of γ -index values, but also differential and cumulative histograms of γ -value.

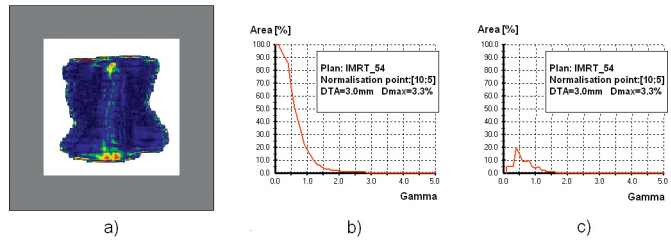


Fig. 6. Results of gamma verification: a) γ -image, b) cumulative histogram, c) differential histogram

Before our software was implemented into clinical practice, several tests of its propriety were performed. The concept of GammaEval software validation was to compare the calculated dose distribution to itself. To simulate the film-recorded dose distribution we had to rewrite the original fluence in PTW file format. It was done with the help of commercial software (Matlab). Validation of portal signals verification was easier to perform, because the predicted and acquired images were exported from TPS in the same file format.

When simulating portal-acquired images verification we always achieved 100% agreement. On the other hand, when verifying the film-simulated signals we observed negligible deviations between $\gamma = 0$ and $\gamma = 0.3$ probably caused by interpolation and dose value rounding performed during the file conversion procedure.

In clinical practice we sometimes observe that the final results of verification are incorrect. First we try to recognize the location of the error and precisely describe what kind of tissue it refers to. If the deviation between the planned and acquired dose is negligible, the physicians usually decide to continue the treatment. They also sometimes decide to reduce the total number of fractions if the dose for organs at risk (OAR) is much higher than expected. However, if the reduction of the total dose is not possible because of the therapeutic effect of the treatment, we have to change the treatment plan or perform the verification on the parallel treatment unit.

RESULTS

Depending on the size of optimized fluence it is sometimes necessary to split the delivery into two or three partially overlapping subfields (multiple carriage group fields). Performing

gamma analysis for multiple carriage group fields we have observed that the radiation leakage has an important contribution to the effectively accumulated dose, especially in the region of the overlapping part of the field, and this way can induce significant errors. We have performed the γ -examination for single carriage group fields and multiple carriage group fields separately. Our investigation shows that the deviations between the calculated dose and the dose absorbed by the tissue take place more frequently for multiple carriage group fields, but the level of divergences is lower (Fig. 7). The fields furthermore have been classified depending on the tumour location. Table I presents average results obtained for each class. One can conclude that the best agreement between the calculated and achieved dose distribution is observed for brain or prostate locations. Only about 14% and 13.1% of the total field area respectively do not pass the criteria of correctness.

The results obtained for H&N tumours are worse and the corresponding parameter is equal to 17.2% in this case.

Fig. 8 presents an example of cumulative -histograms obtained for the same therapeutic field with the help of two different methods of acquisition: film dosimetry and portal dosimetry. We have observed that usually the results of gamma evaluation when portal-acquired images are combined with predicted portal images are considerably better than if the dose recorded on film is compared to the calculated one. The field area corresponding to the γ -value from the range of 0.8–1.2 is as much as 10% smaller for portal dosimetry, although the typical deviation is about 4%.

Unfortunately, comparison of the results of verification performed by commercial software (Eclipse 7.3) and GammaEval is only available using γ -images because histograms are not created by the treatment planning sys-

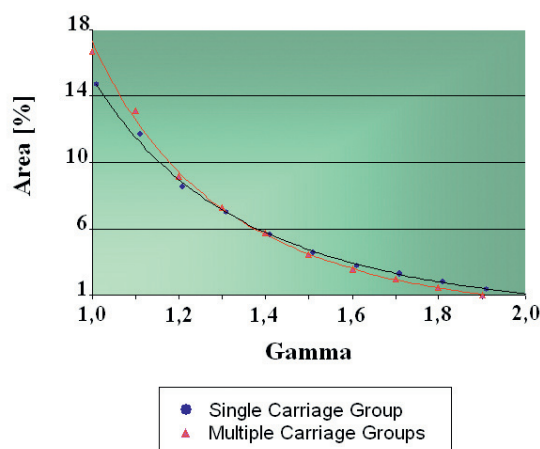


Fig. 7. The average results of gamma verification obtained for single carriage group fields and multiple carriage group fields

tem. Furthermore, it was also not possible to compare results for multiple carriage groups: Eclipse verifies each subfield separately, because the portal-acquired images are always saved when the beam-off signal occurs.

DISCUSSION

The interpretation of the results is strongly related to all circumstances of the verification procedure: method of image acquisition, signal resolution, calculation parameters, and others. When calculating γ -value according to Eq. 1 both DTA and ΔD can be chosen arbitrarily. Choosing different values than proposed, one can achieve a different value of $\gamma(\vec{r}_c, \vec{r}_m)$, and furthermore the new pass-fail criteria should be defined. However, it does not change the concept of the gamma algorithm. The tool we have created allows one to perform calculations for any value of DTA and ΔD , which can be chosen arbitrarily by the user.

Table 1. Average results of gamma verification obtained for different tumour location

Location	Number of fields	Field area with $\gamma \leq 1\%$	Std. Deviation
Head & Neck	308	17.2	4.8
Brain	150	14	3.9
Prostate	25	13.1	3.5
other	40	15.1	5.3

Interpretation of gamma value also depends on the understanding of Eq. 1. The ΔD parameter may be interpreted differently in the algorithm: in our algorithm its value is related to the calculated local dose $D(\vec{r}_c)$ (or local value of predicted portal image for portal dosimetry) and the equation should be preferably presented in a much more appropriate form:

$$\gamma(\vec{r}_c, \vec{r}_m) = \sqrt{\frac{|\vec{r}_c, \vec{r}_m|^2}{DTA^2} + \frac{|D(\vec{r}_m) - D(\vec{r}_c)|^2}{\Delta D^2 \times D(\vec{r}_c)^2}} \quad (2)$$

The consequence of verification with globally calculated ΔD (traditional interpretation of gamma algorithm) is an identical acceptable deviation of all \vec{r}_m , the same for low and high expected dose. In this case the dose deviation comparable with measured dose $D(\vec{r}_m)$, for example $\Delta D = 3.3$ cGy, would be accepted for both $D(\vec{r}_m) = 1.0$ Gy (3.3% deviation) and $D(\vec{r}_m) = 0.1$ Gy (33% deviation).

It is essential to emphasize that normalization of calculated and recorded dose matrixes must be performed inside regions of homogeneous dose. We have observed discrepancies if the normalization point was set at dose gradients and/or below the space between adjacent leaves of the collimator. Typically, in the case of standard 3D CRT, the dose distributions were normalized on the beam axis. In the case of IMRT verification this procedure may lead to overestimation of overall error, especially if the signal is at high resolution, because the beam axis is always located between central leaves of the MLC. To be independent of this “normalization effect” we perform calculations several times, the coordinates of the normalization point changing bit by bit with a fixed step. This way the random results are eliminated and as a final result of verification we take the best γ -histogram.

As reported, the results obtained when portal dosimetry is in operation usually look better in comparison with film dosimetry. There are at least two different reasons for this.

The first one is much better resolution of the acquired signal in the case of film dosimetry, which allows one to see even point or thin line errors (caused for example by the tongue-and-groove effect).

It is in fact limited only by the film structure and digitizer technical capacities. On the

other hand, film dosimetry requires a lot of work and demands high precision, especially when the film is being processed (developer temperature, reagents concentration, duration of developing procedure). If the calibration and film processing conditions are different, the results of gamma verification become worse.

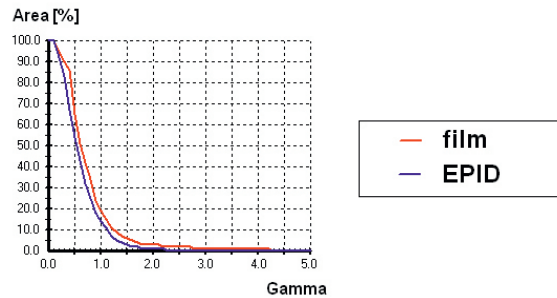


Fig. 8. Results of verification performed with the help of GammaEval for two different methods of acquisition: red line – film, blue line – portal dosimetry

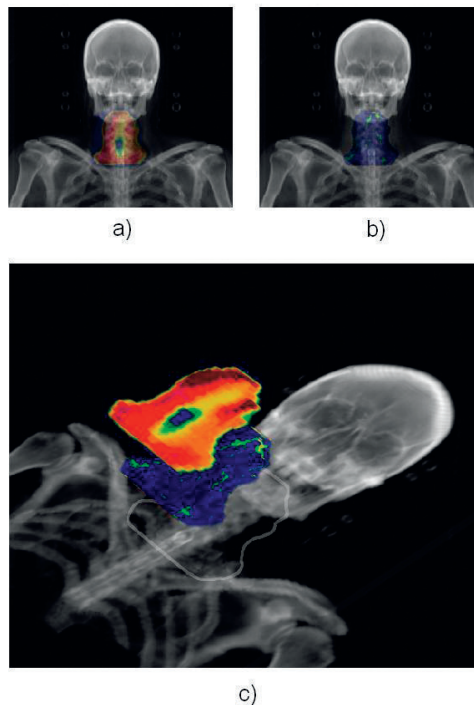


Fig. 9. The cross-examination of DRR and γ -image (concept) a) the fusion of DRR and dose image, b) the fusion of DRR and image, c) 3D-view

CONCLUSIONS

We believe that the gamma evaluation method is a reliable and effective instrument for IMRT treatment verification. It makes possible the qualitative and quantitative monitoring of treatment. The procedure discussed in this paper can be performed on each treatment unit, even if portal dosimetry is not available. The key to proper interpretation of results is understanding of some critical algorithm parameters. We have proposed measures which eliminate the disadvantage of the normalization effect. However, the only reliable approach seems to be absolute dose comparison.

We have observed that sometimes the quantitative outcomes of the analysis are not efficient and may suggest wrong conclusions. If the γ -histogram is worse than the average result, we propose performing cross-examination of the γ -image with DRR, which is shown in Fig. 9. The fusion of images is especially suggested for multiple carriage group fields and very small fields. We are currently investigating the procedure for precise location and considering the irregularities in dose accumulated regularly by the target and/or organs at risk while the proper IMRT treatment is performed. We are also planning to perform some additional tests which will give us more knowledge about differences between single carriage group and multiple carriage group fields execution and their influence on local dose delivery.

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