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Patient-specific quality assurance for helical tomotherapy: An evaluation of two different detector systems.

Carl Bladh-Johansson

Master's Thesis
Hallaryd, December 24, 2010

Supervisors: Marika Enmark, Kristoffer Petersson, Joakim Medin, Petra Ambolt, Tommy Knöös

Whether I shall turn out to be the hero of my own life,

or whether that station will be held by anybody else,

these pages must show.

David Copperfield (1850) by Charles Dickens

Carl Bladh-Johansson

Patient-specific quality assurance for helical tomotherapy: An evaluation of two different detector systems.

Intensity-modulated radiation therapy (IMRT) has during the recent years become a frequently used technique in radiotherapy. Lately IMRT has also been combined with rotational beam delivery, with a simultaneously moving couch, a technique called tomotherapy. The great benefit with this treatment method is the ability to give highly conformal treatments, with high dose to the tumour, while sparing organs at risk. Patient-specific quality assurance (QA) of the generated tomotherapy treatment plan is essential to ensure that the patient will receive the planned dose.

During the last years the use of electronic array dosimeters in the QA process has increased substantially, since they provide almost instantaneous readout in terms of absolute dose distribution. In the present work one such array dosimeter has been implemented, namely the Delta^{4}® (ScandiDos AB, Uppsala, Sweden). The Delta^{4}® has subsequently been compared with a two-dimensional array dosimeter called MapCHECK 2 (Sun Nuclear Corporation, Melbourne, USA), that is already in clinical use at Skåne University Hospital in Lund. The gamma evaluation method has been used for comparison of dose distributions.

25 patient plans were recalculated on CT-scans of the phantoms that constituted the two dosimetry systems. Delivery QA plans for pretreatment evaluation were then created and measured with the two dosimetry systems respectively. The pass rate *i.e.* the percentage of diodes passing the gamma evaluation was calculated for each plan for different gamma criteria and thresholds. The pass rate was then used to compare the dosimetry systems with each other.

The results from the measurements showed a much poorer agreement between planned- and measured dose distribution than expected for the MapCHECK 2 system whereas the Delta^{4} system showed a more decent agreement. Even though the Delta^{4} system proved to be superior the MapCHEK 2 it showed on a noteworthy weakness in dose difference agreement, often below 85 % for a 3 % dose difference criterion. Though on the other hand the Delta^{4} showed an almost perfect agreement in distance-to-agreement, which keep the pass rate of the gamma evaluation on a still acceptable level.
For the MapCHECK 2 an alternative way of calculating the gamma evaluation (using Van Dyk percent difference) was also evaluated which gave a gamma pass rate comparable with that for the Delta^{4}. The Van Dyk percent difference is however highly controversial, since it does not reflect local dose differences which might be more relevant for organs at risk.

The principle conclusion of this thesis is that the Delta^{4} system is superior the MapCHECK 2 system in validating treatment plans for tomotherapy. It should however be stressed that the Delta^{4} system, in its current state, is not the optimal system for patient-specific QA since the agreement in dose difference is somewhat low.

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Hallaryd, December 24, 2010

Carl Bladh-Johansson

Abbreviations and Acronyms

1D	One-dimensional
2D	Two-dimensional
3D	Three-dimensional
ADCL	Accredited dosimetry calibration laboratory
CT	Computed tomography
DICOM	Digital imaging and communications in medicine
DTA	Distance to agreement
DQA	Delivery QA
HU	Hounsfield Unit
IGRT	Image-guided radiation therapy
IMRT	Intensity-modulated radiation therapy
MLC	Multileaf collimator
MU	Monitor unit
MV	Mega voltage
MVCT	Mega voltage computed tomography
PMMA	Poly(methyl methacrylate)
PRF	Pulse repetition frequency
QA	Quality assurance
SAD	Source-to-axis distance
TH	Threshold
TPS	Treatment planning system
VMAT	Volumetric modulated arc therapy

Contents

1 Introduction

Cancer is, according to the Swedish National Board of Health and Welfare (Socialstyrelsen), the second most common cause of death in Sweden with more than 22 000 deaths per year, corresponding to approximately 25% of all deaths.[1] Cancer is also a leading cause of death worldwide and according to the World Health Organisation (WHO) it accounted for 7.9 million deaths, approximately 13% of all deaths, in 2007.[2]

Treatment of cancer aims to cure, prolong life and improve quality of life for patients. Principal treatment methods are surgery, radiotherapy and chemotherapy. These methods are rather often used complementary to each other *e.g.* radiotherapy can be given as adjuvant therapy after surgery to kill the remaining cancer cells or as neoadjuvant therapy given before surgery to shrink the tumour to make its resection easier. Radiotherapy has for many years been an important modality in Sweden for treatment of cancer and in the present situation it is represented in almost half of all treatments.[3] The main purpose of radiotherapy is to accurately deliver an absorbed dose to a specified target (tumour) to either efficiently cure the patient (referred to as curative treatment) or to shrink the tumour to prevent and relieve suffering and to improve quality of life (referred to as palliative treatment). While doing this one must also minimise the absorbed dose to the normal healthy tissue to avoid unnecessary damage. This means that high geometrical and dosimetrical precision is necessary for a successful treatment. To better achieve these goals new treatment methods for cancer are continuously being developed and introduced into the world of radiotherapy.

In the spring of 2009 a TomoTherapy Hi-Art® treatment system (figure 2) was installed at the radiotherapy department at Skåne University Hospital in Lund. Thus making Lund the first, and currently only, site in the Nordic countries that offers this treatment modality. The ring gantry-based Hi-Art System is intended to be used mainly for advanced treatments, such as head and neck tumours and for patients with extensive and/or complicated targets where target conformity is in very high demand. The great benefit with the tomotherapy unit is namely the ability to give highly conformal treatments, with high dose to the tumour, while sparing organs at risk.

A very important step, which must not be neglected, in the radiotherapy process is the execution of a patient-specific QA procedure. The purpose with the patient-specific QA procedure is to make sure of that the machine delivers the planned dose distribution to the patient. Ideally this QA procedure should be performed during the actual patient treatment, *i.e.* by using *in vivo* dosimetry methods. These methods are however rarely applied and most QA procedures are in fact performed prior to the treatment.[4][5] For pre-treatment verification of patient plans in helical tomotherapy ion chambers and film has traditionally been used for dosimetric QA. As with other intensity-modulated radiation therapy (IMRT) techniques, there is also a substantial interest in using electronic array dosimeters, since they provide almost instantaneous readout in terms of absolute dose distribution.[6] Currently there are several commercial available 2D array dosimeters, diode-based as well as ionization chamber based. These detectors have shown to be adequate when used for verification of IMRT patient plans delivered by linear accelerators but their applicability of verifying composite rotational dose distributions is still in need of further evaluation since they were not designed for rotational delivery modalities such as TomoTherapy.[6][7][8] An

apparent problem with using a single-plane array to verify composite rotational dose distributions is the fact that the 2D dose-distribution information available when the beam direction is perpendicular to the detector plane is gradually reduced to 1D as the incident angle approaches 90°.[6]

At Skåne University Hospital in Lund all patient-specific QA measurements for tomotherapy are currently being made by the use of a 2D array dosimeter, called MapCHECK 2™, manufactured by Sun Nuclear Corporation. Since this single-plane array dosimeter shows the same weakness to verify composite rotational dose distributions as described in the previous section, another dosimetric system, called Delta^4® was purchased in the spring of 2009 with the intention to replace the MapCHECK 2 for patient-specific QA on tomotherapy. The Delta^4 is a diode phantom with a novel geometry for QA of rotational based treatments manufactured by ScandiDos AB (Uppsala, Sweden). The phantom consist of a bi-planar 3D diode array dosimeter that is intended to preserve the dose distribution information regardless of the beam incidence angle.[6]

Although the Delta^4 dosimetric system was purchased in the spring of 2009 and despite several attempts to get the system working clinically no success had been achieved when this MSc Thesis was commenced, in the beginning of January 2010.

1.1 Aim

The primary aim of this MSc Thesis was to implement the Delta^4® detector system for helical tomotherapy delivery QA (*i.e.* patient-specific QA), at Skåne University Hospital in Lund. A secondary aim was to evaluate and compare the Delta^4 system (ScandiDos AB, Uppsala, Sweden) with the MapCHECK 2™ system (Sun Nuclear Corporation, Melbourne, USA) and to see if there are any possible advantages and/or disadvantages with the different systems.

2 Theory

2.1 Tomotherapy

Tomotherapy is a form of intensity modulated radiation therapy that combines rotational beam delivery with intensity modulation. Tomotherapy differ from other techniques that provides rotational beam delivery, such as RapidArc™ (Varian medical Systems, USA) and VMAT (volumetric- modulated arc therapy, Elekta AB, Sweden), by delivering the radiation slice-by-slice, continuously from all angles around the patient, trough a ring gantry. Tomotherapy (with the Greek prefix tomo-, which means "slice") literally means "slice therapy" and in practise, tomotherapy uses multiple rotations of "narrow" fan beam with a continually changing one-dimensional intensity profile to achieve full coverage of the entire tumour.[9]

Tomotherapy has been implemented both as a sequence of rotational modulated beams, with each rotation at a fixed couch position (known as serial tomotherapy), and as a beam following a helical trajectory relative to the patient (known as helical tomotherapy). The latter is the technique used by the TomoTherapy Hi-Art System® manufactured by TomoTherapy Incorporated, Madison, Wisconsin. The helical tomotherapy technique has an advantage over serial tomotherapy, since with a helical beam delivery there can be no "matchlines" containing hot or cold dose regions as can be a problem with a true slice-by-slice delivery like serial tomotherapy, especially if couch translation between slices is inaccurate.[9]

A close analogy to how the beams are optimised and delivered in the tomotherapy process is the image acquisition mode used in CT scanning. With CT imaging, a large number of physically created narrow beam attenuation profiles are mathematically projected back along each beam path to reconstruct the object's density distribution within a narrow slice. In the tomotherapy the required dose distribution is mathematically projected out of the patient and then used to create narrow beamlet intensity profiles. These beamlets (to the amount of tens of thousands) are then physically projected back into the patient during the slice-based beam delivery.[9]

2.1.1 The TomoTherapy Hi-Art System®

The TomoTherapy Hi-Art System®, depicted in figure 2, has a linear accelerator that produces an 6 MV X-ray beam. The energy is chosen to represent a good balance between rate of attenuation and lateral penumbra width, since these show opposite behaviour with increased or decreased energy. Generally a lower attenuation is sought, as with high energies, but at the same time a small penumbra is desirable as with lower energies.[9]The system has a source-to-axis distance (SAD) of 85 cm with a 85 cm, in diameter, bore through which the patient travels. The Hi-Art system comes with a built-in image detector, designed for single-slice CT imaging thus making inter- and intrafraction image guidance possible. The detector consists of xenon gas-filled ionization chambers encased in tungsten. For imaging purposes the linac is de-tuned to produce x-rays of energy at 3.5 MV and at the same time the pulse repetition frequency (PRF) is decreased to spare the patient some of the additional dose.

The X-ray beam is collimated to a fan beam geometry with a width of 40 cm and a fan beam thickness that can be varied between 1.0, 2.5 and 5.0 cm by

the use of symmetrical tungsten collimators. The 5.0 cm width is generally used only for long target volumes e.g. craniospinal axis or abdominal cavity treatments whereas the 1 cm width is used for small lesions such as brain metastases. 2.5 cm is thus the most frequently used collimator width. The total treatment time is approximately inversely proportional to the collimator width. Below the collimators there is a binary multileaf collimator (MLC) with 64 interdigitating leaves, with the options for each leaf to be either open or closed. The time for the leaves to open and close is approximately 20 ms.[9]

The Hi-Art system does not include a physical flattening filter, consequently the dose-rate is increased thus reducing treatment times, and the primary fluence originating from the treatment head is forward peaked until it is modulated by the MLC.[10] Intensity modulation is achieved by varying the fraction of time for which different leaves are open. For delivery and planning a full gantry rotation is divided into 51 projections, thus every projection corresponds to a gantry rotation of just 7°. Each projection is characterised by its own leaf opening pattern thus defining an individual intensity modulation pattern. The gantry rotates at a constant velocity during treatment with a period between 10 and 60 seconds per rotation and so the time per projection is 196 ms or greater.[10] The pattern of MLC leaf opening times, for all leaves at all angles, throughout successive rotations is contained in a delivery sinogram. In the sinogram each row represents leaf opening times for a single projection and one pixel in the map corresponds to one leaf (figure 1).

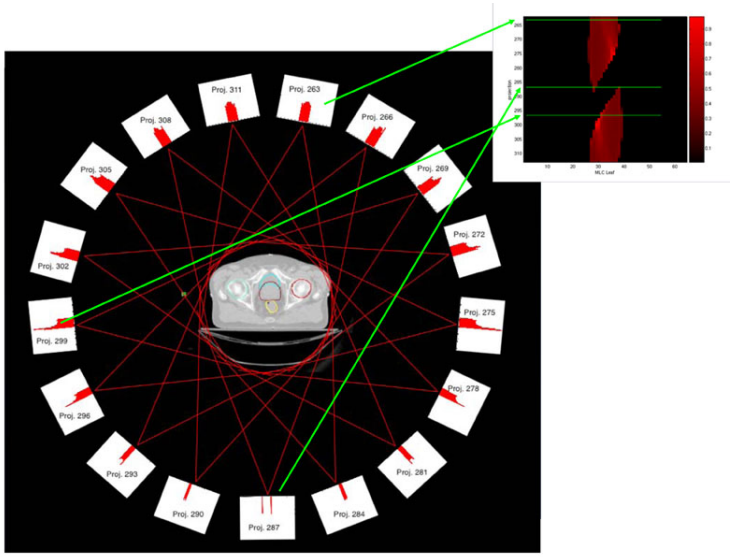

Figure 1: Illustration of dose delivery by the use of a delivery sinogram. In the sinogram (depicted in the small box in the upper right corner, where the y-scale = projection and x-scale = MLC leaf) each row represents leaf opening times for a single projection and one pixel in the map corresponds to one leaf. A full gantry rotation is divided into 51 projections.

The degree of modulation for a projection can be characterised through the modulation factor, which represents the ratio of the maximum to the average

leaf open times for the projection. The modulation factor is proportional to the overall treatment time and with appropriate physical constraints for the tomotherapy delivery, modulation factors between 1.0 and 4 are considered to be suitable. The maximum permissible modulation factor for any part of the delivery is specified, by the operator, during the treatment planning. The system will then, during the optimisation, ensure that the final modulation factor is less or equal to the specified maximum value. For complex geometries highly modulated treatments can often achieve more conform treatments but at the cost of prolonged delivery time. A higher modulation factor is considered unnecessary for less complex cases like relatively symmetrical targets close to the central axis of the patient.[9]

Another way to improve the modulation capabilities, especially in the superior/inferior direction, is to use a small pitch. The pitch is defined as couch movement per rotation in units of the collimator width. Consequently a collimator width of 5 cm and a pitch of 0.2 equals a couch translation of 1 cm for each gantry rotation. Typically the pitch value is small in helical tomotherapy, commonly between 0.1 and 0.5, such that the helix that the source traces out is a tight helix, resulting in overlap between adjacent rotations. A small pitch can thus compensate for superior/inferior resolution losses caused by a broad collimator width. A smaller pitch value does not automatically imply an increased treatment time, since a smaller pitch yields a higher amount of rotations thus a lower dose needs to be deposited per rotation and consequently the gantry rotations can be faster.[9]

Hi-Art machines has a very fundamental difference towards conventional linacs in that they do not work on a monitor unit-based system, but operates more like a cobalt unit. The output is therefore calibrated in terms of a reference dose rate, measured in units of cGy per minute rather than the conventional cGy per monitor unit.[10] Since the Hi-Art system is developed specifically for IMRT, flat fields can be obtained by beam modulation and consequently there is no need for a flattening filter. Because of this, the field has a cone-like profile in both lateral and transversal directions. Consequently dose rate as well as the shape of the cone-profile are very important for correct patient treatment and should be closely monitored.[11]

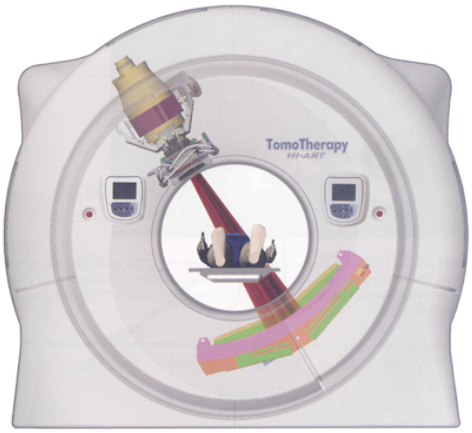

Figure 2: The TomoTherapy Hi-Art System® (Courtesy of TomoTherapy Inc.)

2.2 Comparison of Dose Distributions

Comparison of dose distributions can be done using a number of different methods. In this section dose difference, the distance-to-agreement and the gamma evaluation methods will be presented.

2.2.1 Dose Difference

The technique used most frequently by radiation oncology physicists to compare measured and calculated dose distributions is the dose difference method. The method superposes the measured and calculated isodose curves with a subsequent qualitative assessment of the acceptability of the calculation algorithm. Determination and presentation of the numerical difference between the measured and calculated dose distributions will then highlight regions of disagreement.[12]A difficulty with comparing dose difference distributions is that they are very sensitive in high-dose gradient regions. In such areas a small spatial error may result in a large, but clinically insignificant, dose difference. Hence dose differences alone might not be a suitable instrument to decide whether a point is acceptable or not.

2.2.2 Distance-To-Agreement, DTA

The distance-to-agreement (DTA) is specified as the spatial distance between a data point in the reference distribution and the closest point in the evaluated distribution that shares the same dose. DTA is a useful complement to dose difference measurements, especially when it comes to high dose gradient regions, and a useful tool to determine the acceptability of the dose calculation.[13]

The mathematical definition of the DTA, at the position $\vec{r_r}$, can be seen in equation 1, where $|\vec{r_e}-\vec{r_r}|$ is the spatial distance between evaluated and reference dose points. A value for the maximum accepted spatial difference, referred to as the DTA criterion, Δd, is often set between 2 to 5 mm. This limit corresponds to the tolerance for misalignment or rotation of the phantom at irradiation or the film while scanning. If the DTA value, at a specific point, is less than Δd the comparison passes at that point. If instead the DTA value is greater than Δd, the comparison fails for that point.[12] As pointed out by equation 1, the comparison is done for each data point.

The DTA concept has a weakness since it is overly sensitive when it comes to regions with low dose-gradients. In these areas a slight difference in dose will result in a large DTA value. This implies that neither measurement of dose difference or DTA should be used alone but rather in combination with each other.[14]

$$DTA(\vec{r_r}) = min|\vec{r_e}-\vec{r_r}|\forall\vec{r_e} \tag{1}$$

2.2.3 Gamma Evaluation

As stated in the distance-to-agreement section above, the use of only dose difference or DTA might in some cases be insufficient to decide whether a data point should be accepted or not. A solution to this issue was presented by D.A. Low et al. [13] in which they simultaneously incorporate the dose difference and

DTA criteria. The method provides a numerical quality index for each measured point, referred to as the gamma value. This value represent a measure of the disagreement in regions that fail the acceptance criteria and indicates the calculation quality in regions that pass.[15]

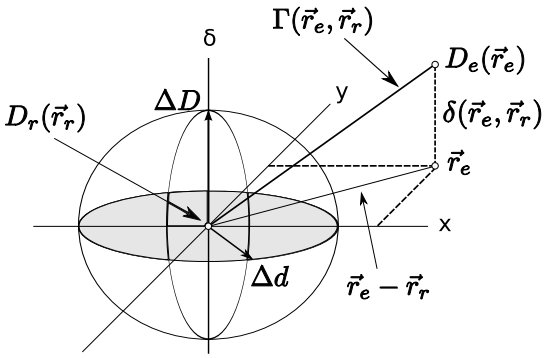

Figure 3: Two-dimensional geometric representation of dose distribution evaluation criteria using the combined ellipsoidal dose-difference and distance-to-agreement tests.[13]

The concept of the gamma evaluation asserts that the agreement between the two distributions, *i.e.* the dose difference and the distance-to-agreement, is acceptable if there exists a point of the calculated dose matrix that belongs to the ellipsoid (figure 3) having one point of the measured dose matrix for the centre and the assigned tolerance values for the radius, thus representing the acceptance criteria. In mathematical terms the ellipsoid is defined by the equation 2.

$$1=\sqrt{\frac{r^2(\vec{r},\vec{r_r})}{\Delta d^2}+\frac{\delta^2(\vec{r},\vec{r_r})}{\Delta D^2}} \tag{2}$$

where $r(\vec{r},\vec{r_r})$ is the spatial distance between the reference position and any other position and $\delta(\vec{r},\vec{r_r})$ is the dose difference between the reference position and any other position. Further Δd and ΔD are the acceptance criteria for distance-to-agreement and dose difference respectively. The right hand side in equation 2 can then be used to define the γ function, *i.e.* the minimum generalised Γ function in the set of evaluated points,

$$\gamma\left(\vec{r_r}\right) = min\left\{\Gamma\left(\vec{r_e},\vec{r_r}\right)\right\} \forall \left\{\vec{r_e}\right\}, \tag{3}$$

where

$$\Gamma\left(\vec{r_e},\vec{r_r}\right) = \sqrt{\frac{r^2\left(\vec{r_e},\vec{r_r}\right)}{\Delta d^2}+\frac{\delta^2\left(\vec{r_e},\vec{r_r}\right)}{\Delta D^2}} \tag{4}$$

is the generalised Γ function, computed for all evaluated positions $\vec{r_e}$ and reference positions $\vec{r_r}$. The expression $r\left(\vec{r_e},\vec{r_r}\right) = \mid \vec{r_e} - \vec{r_r} \mid$ is the spatial distance between the evaluated and reference dose points and $\delta\left(\vec{r_e},\vec{r_r}\right) = D_e\left(\vec{r_e}\right) - D_r\left(\vec{r_r}\right)$ is the difference between evaluated dose $D_e\left(\vec{r_e}\right)$ at position $\vec{r_e}$ and reference dose $D_r\left(\vec{r_r}\right)$ at position $\vec{r_r}$. [13] The pass-fail criteria will then be

$$\gamma(\vec{r_r}) \leq 1, \, calculation \, passes \tag{5}$$

$$\gamma(\vec{r_r}) > 1, \, calculation \, fails \tag{6}$$

The γ function is defined independently for each reference point thus no influence from any neighbouring reference points is present in the computation of γ in that point. Commonly used passing criteria for IMRT are $\Delta D = 3\%$ och $\Delta d = 3mm$ but other combinations exists as well. To provide accurate calculation of γ in steep dose gradient regions, the pixel spacing of the evaluated distribution needs to be sufficiently small. As a general rule, the spacing should be less than or equal to 1/3 of Δd.[14]

The angle between the δ-axis and $D_e\,(\vec{r_e})$, in figure 3, is called the γ-angle. The γ-angle can be useful for the interpretation of deviations since it indicates the parameter mostly influencing the γ-value, *i.e.* the dose difference or the DTA. The angles of 0° are defined on the dose-difference axis. For example, if the γ-angle is between $\frac{\pi}{4}$ and $\frac{\pi}{2}$ the index is dominated by the DTA criterion. The angle is always between 0 and $\frac{\pi}{2}$ since it is calculated with absolute values.[5]

2.3 Image Value-to-Density Calibration Table

A calibration table for image value-to-density, that matches the Hounsfield Unit values (HU-values) generated by the CT, against the measured physical density (in gram per cubic centimetre) for scanned object, has to be created in the TPS before any DQA plans can be made. Every table for image value-to-density is created so that it corresponds to a specific tomography device. For instance if DICOM CT data obtained from the build-in image detector of the Hi-Art system has been used when importing the phantoms (The Delta^4 and the MapCHECK 2) to the TPS an image value-to-density table has to be created for this device in order to make correct DQA-plans.[16]

The image value-to-density table for the Hi-Art system is obtained by first making a MVCT scan of the TomoTherapy Commissioning Phantom with the image quality set to fine, corresponding to a slice thickness of 2 mm. The phantom (figure 11) is cylindrical in shape with a diameter of 30 cm and a length of 18 cm. It has 20 holes of 2.8 cm diameter (wherefore it is also called the "Cheese" phantom) in which 12 different plugs of certified electron densities can be included for CT calibration. All other holes can be filled with solid water plugs.[17] The MVCT image is obtained with the 12 plugs of different electron densities inserted in the phantom. The image is then evaluated in the TPS software, where ROI's are made in the area of the plugs, so that HU-values for the corresponding electron densities can be determined. The image value-to-density table that was used in this thesis is represented in figure 4.

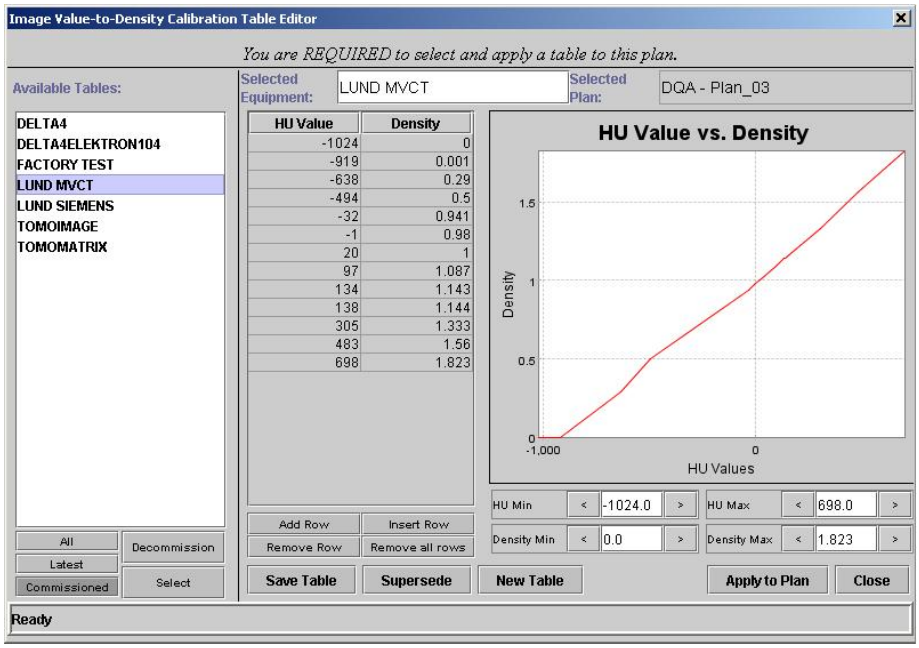

Figure 4: Image value-to-density calibration table

3 Material and Methods

3.1 Detectors

3.1.1 Delta^{4}®

The Delta^{4}® dosimetric system (figure 5), manufactured by ScandiDos AB, uses two orthogonal diode arrays embedded in a cylindrical poly(methyl methacrylate) (PMMA) phantom and associated computer software to measure and comparing the composite dose distribution for a complete helical tomotherapy treatment plan with the dose distribution predicted by the treatment planning system (TPS).[18] The crossed planes, that can be seen in figure 5, are obtained by a main detector board through the entire diameter of the phantom and two separated wing detector boards. The diameter of the PMMA phantom is 220 mm and the density is 1.19 g/cm^3 with a relative electron density of 1.147 g/cm^3. The Delta^4 3D-phantom consists out of four phantom quarters in which the detector arrays, (of the wing units and the main unit) are mounted in between.

The phantom consists of 1069 cylindrical p-type silicon diodes, each with an active area of 0.78 mm^2. The diodes are spaced at 5 mm intervals over the central 60 x 60 mm^2 area of each plane whereas outside the central area the diodes are placed 10 mm apart.[18]

For conventional linacs the recording of measured dose is triggered by an electronic synchronisation pulse, which is obtained from the accelerator console test point. This means that the electrometers will only be in recording mode for a short period of time, before, during and after the actual radiation pulse, thereby facilitating the use of time dependent four dimensional applications. However for helical tomotherapy the triggering mechanism is quite different since no synchronisation signal is available from the tomotherapy machine. Instead the measurement is triggered by the radiation pulse itself. The Delta^4 device will initially operate in a pulse searching mode and once a radiation pulse is detected by a diode the detectors will change to measurement mode and remain in that state as long as the next pulse arrives in a given time frame, corresponding to the pulse repetition frequency (PRF) of 300 ± 10 Hz. If no such pulse is detected the system will revert back to pulse searching mode.[6]

Another difference between the use of the Delta^4 device with a conventional linac and a tomotherapy unit is that for conventional linacs the gantry angle is independently sensed by using an inclinometer attached to the gantry or accelerator head. Thereby making it possible for the Delta^4 device to identify which control point of a dynamic arc delivery that is being delivered. The measured dose can then be associated with that control point and appropriate correction for gantry angle applied.[18] Also with the conventional treatment plan recalculated on the Delta^4 phantom beam parameters, such as segment size and depth of measurement for each diode, are known or can be calculated so that depth and field size corrections also can be applied on a segment level.[6] For tomotherapy however no beam geometry information is made available to the Delta^4 and instead an average rotational response correction of 1.010, as reported by ScandiDos AB, is applied to the dose measured by each detector. No volumetric dose interpolation is possible, hence the dose distribution is only evaluated in the two orthogonal measurement planes.[6]

The verification process requires the recalculation of treatment plans on a CT scan of the phantom. Scandidos AB recommends the use of a DICOM CT image

set provided by themselves, rather than an in-house CT scan of the phantom, to avoid errors and distortions due to the diodes in the CT scan and deviations in density caused by tracks on the printed circuit board. Such distortions may result in erroneous calculations in the treatment planning system.

The Delta^4 software comes with at set of common dose comparison tools such as dose volume histograms and γ analysis. The γ analysis uses the planned 3D dose distribution to calculate the gamma index, *i.e.* the search is not limited to planned data points in the detector plane. This is made possible by interpolating the planned dose for calculation of the gamma index. [19]

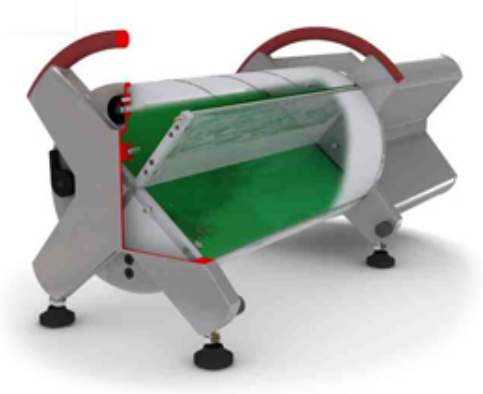

Figure 5: Depiction of the Delta^4 phantom. The front panel has been cut away to show the crossed detector planes. (Image courtesy of ScandiDos AB, Uppsala, Sweden).

3.1.2 MapCHECK 2™

Model 1177 MapCHECK 2™ (figure 6), manufactured by Sun Nuclear Corporation, is a two-dimensional radiotherapy dosimetry system for quality assurance. The MapCHECK 2 detector provides a quality assurance test of the linear accelerator's ability to successfully deliver a planned quality assurance dose map (called delivery QA or DQA) in a phantom. The DQA plan is not a measurement of the planned dose map that would have been delivered during the treatment of the patient but a recalculation, on a phantom, of the dose defined by the treatment planning system, TPS, which must be delivered by the linear accelerator.[20]

The MapCHECK 2 detector consists of 1527 diode detectors with a uniform detector spacing throughout the array of 7.07 mm, equalling a total octagonal detector array size of 32 x 26 cm. The sampling frequency of the MapCHECK 2 detector is 50 ms and each diode in the array has an active detector area of 0.64 mm^2 and an active detector volume of 0.000019 cm^3.[20]

Figure 6: Depiction of MapCHECK 2™ (Image courtesy of Sun Nuclear Corporation, Melbourne, USA).

To adapt the MapCHECK 2 for rotational and helical plan QA, *i.e.* RapidArc™, VMAT, and TomoTherapy® a 18 cm thick homogeneous water equivalent phantom, called MapPHAN-MC2™ (figure 7), using a virtual water construction is used.

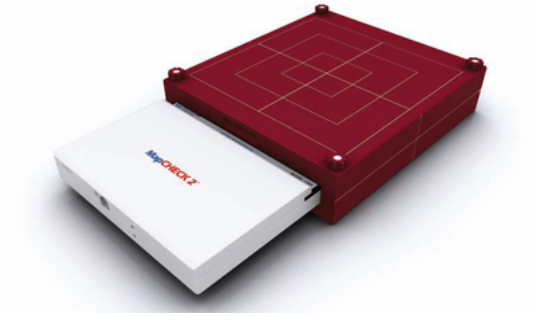

Figure 7: Depiction of MapPHAN-MC2™ with the MapCHECK 2 inserted. (Image courtesy of Sun Nuclear Corporation, Melbourne, USA).

Percent Dose Difference Calculation The MapCHECK 2 associate software offer two ways to calculate percent dose difference; Van Dyk percent difference and MapCHECK 2 percent difference. The MapCHECK 2 percent difference is chosen as default and is simply the allowed difference between co-located measurement and plan points. If the values of the measured and plan points having the same coordinates agree within the percent difference, the point passes. The Van Dyk percent difference is however the percent difference between any measured point and the corresponding plan point normalised to a common point (typically maximum dose point). The Van Dyk percent difference can also be used in the gamma analysis and will then be defined, in the MapCHECK software, as follows:

$$PDE_{k,l} = 100 \cdot \frac{P_{k,l} - M_{g,h}}{M_{norm}} \tag{7}$$

where $PDE_{k,l}$ (corresponding to $\delta\left(\vec{r_e}, \vec{r_r}\right)$ in equation 4) is the percent dose difference between the planned value $P_{k,l}$ at point (k, l) and the measured value

$M_{g,h}$ at point (g,h). M_{norm} is the measured value at the normalisation point, which is typically chosen as the maximum dose value in the plan.[20]
The measurement uncertainty, $e_{V\gamma}$, can then be derived as:

$$e_{V\gamma} = 100 \cdot \frac{M_{g,h}}{M_{norm}} \cdot \sqrt{e_{g,h}^2 + \left[\left(\frac{P_{k,l}}{M_{g,h}} - 1\right) \cdot e_{norm}\right]^2} \tag{8}$$

where $e_{g,h}$ and e_{norm} are the measurement uncertainties that are associated with the MapCHECK 2 calibration and absolute dose precision. The percentage acceptance criterion, ΔD_M (corresponding to ΔD in equation 4) is then revised to:

$$\Delta D_M^{'} = \Delta D_M + e_{V\gamma} \tag{9}$$

3.2 Steps Performed to Get the Delta^4 System Working

The Delta^4 detector system did not work properly together with the Hi-Art system when this MSc Thesis was commenced. The problem was mainly that the measured dose seemed to consistently be approximately 7-10 percent too high. Therefore a comprehensive troubleshooting was initiated, where every detail that could possibly affect the measurement procedure was investigated thoroughly.

The first step was to make a MVCT scan of the phantom by using the on board imaging device of the Hi-Art System. The phantom was scanned, with the detector boards replaced by slabs of PMMA to avoid distortions due to the diodes. The quality was set to fine resulting in a slice thickness of 2 mm. This CT-set was then exported as a patient structure to Oncentra Masterplan, where it was renamed to "_phantom" and exported back to the tomotherapy TPS where it now was interpreted as a phantom and thereby added to the list of phantoms.

DQA plans was then planned on this new phantom, instead of the recommended artificial CT-set manufactured by ScandiDos AB, and the image value-to-density calibration table, called "LUND MVCT" (see paragraph 2.3) used for plan adaptive and DQA measurements with the MapCHECK 2 was used. This procedure resulted in a better agreement between measured and planned dose, with an with a approximately 3 percent too high measured dose, compared with 7-10 percent with the artificial CT-set.

The second step was to make a rotational 1 Gy uniform plan consisting of a cylinder with a diameter and Length of 10 cm respectively. This plan was measured as a first step in every measurement session and was used to derive a daily correction factor, taking into account both the daily variation in the output of the machine and daily variations in the Delta^4 system. The daily correction factor was then applied to the nextcoming measurements.

The Delta^4 detector was also relative- and absolute calibrated and a directional calibration was also performed.

Since a MVCT scan was used, instead of the artificial CT-set, as planning material there were no fiducial markers available in order to fit the red lasers (used for patient/phantom setup) to the centre of the detector in the TPS system. As a solution to this the distance to the centre of the detector was measured from the right edge of the phantom, resulting in a distance of 23.80 cm. Consequently every DQA plan was planned so that the red laser, in the longitudinal direction, was positioned 23.80 cm from the right edge of the phantom (figure 8). The

lasers corresponding to the height and the lateral position were positioned where the PMMA slabs, that substituted the detector boards during the MVCT scan, intersected each other. In the TPS the mouse pointer was then placed where the lasers intersected, by first making sure to be in the correct slice, *i.e.* when the lasers became extended. The position (X,Y,Z), *i.e.* the difference between the red and green lasers (figure 9) , was noted and later specified as an offset in the Delta[4] software. For every DQA plan the phantom was moved so that as much as possible of the target volume was enclosed by the phantom and also as far as possible positioned at the centre of the phantom. In order to move the phantom the couch first needed to be removed and later replaced. The couch was positioned at a distance 7.35 cm below the lower corner of the phantom, corresponding to the distance measured at the time the MVCT scan of the phantom was made.

At the time of measurement, the phantom was positioned according to the red lasers and a MVCT scan was made of the arrangement in order to determine how much the couch was sagging (typically between 3-4 mm) due to the weight of the phantom. This was corrected for, but no differences regarding the lateral or longitudinal position were accounted for, since the matching process was considered not to be relied upon (since our MVCT scan of the phantom, used as reference material, lacked fiducial markers).

The steps mentioned above will be explained more thoroughly later on in this thesis.

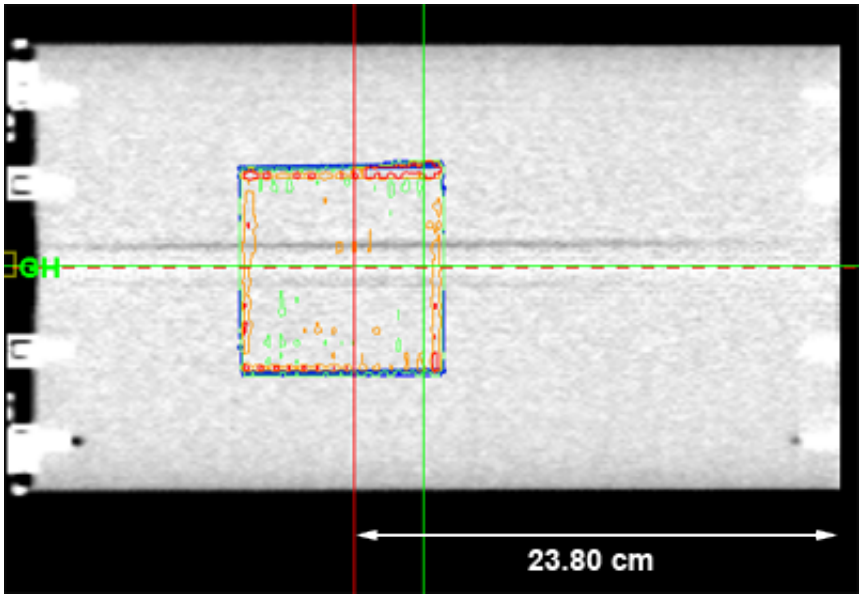

Figure 8: Every DQA plan was planned so that the red laser, in the longitudinal direction, was positioned 23.80 cm from the right edge of the phantom, corresponding to the centre of the detector plates.

3.3 Measurements

The QA measurement procedure consisted of a treatment plan that was transferred to a detector phantom. A dose matrice was recalculated on the phantom,

which was registered in the Hi-Art treatment system through a MVCT scan. The DQA plan was then executed and the calculated matrice was compared with the corresponding measured matrice.

3.3.1 Delta^{4}®

Before the Delta^{4}® system could be used for measurements together with the tomotherapy unit it had to be relative calibrated at a conventional linac (see separate part) and absolute calibrated at the tomotherapy unit. During the calibration the main and wing units were removed temporarily from the cylindrical phantom since the calibration was performed in a dedicated calibration phantom.[19]

The Delta^{4} measurements were initiated by first exporting the approved DQA plan and the calculated dose distribution, rescaled to a single fraction, from the tomotherapy Planning Station to the Delta^{4} software. The DQA plan was chosen and opened whereupon the dose yet again was rescaled to a single fraction, since when rescaling the dose in the TomoTherapy Planning software v3.1 only the treatment scheme sent to the tomotherapy treatment console is rescaled. The DICOM Phantom Dose object still contains the accumulated dose for the complete treatment. The next step in the measurement preparations was to correct for the phantom offset, since non-isocentric set up was used during the delivery QA set-up planning in the TomoTherapy Planning Station, i.e. the phantom was moved out of the machine's isocenter. The offset correction was done by entering the the offset, defined in the delivery QA set up, between the red phantom set up lasers and the machine's isocenter (green laser): transverse, left/right, up/down (figure 9). A MVCT scan of the phantom was then acquired so that the position of the Delta^{4} could be thoroughly adjusted after comparison between the MVCT scan and the planning-CT. The Delta^{4} device was then set to search mode by pushing the start button in the Delta^{4} software and the delivery QA procedure was executed from the TomoTherapy Operator Station.

Absolute Calibration The absolute calibration was commenced by a reference measurement in a static calibration field, $SSD = 80$ cm, with a calibration field size of 5 x 40 cm. The reference dose was determined with a calibrated ionisation chamber (Exradin A1SL, Standard Imaging Inc., Middleton, WI) in the A1SL calibration slab, made of PMMA, provided by Scandidos AB. The boards were positioned at a physical measurement depth of 4.3 cm (water-equivalent depth of 4.9 cm, with both backscatter and build-up surrounding the detector boards. Thereafter the ionization chamber calibration slab was substituted with the calibration slab that consisted of a calibration- frame and lock. The Delta^{4} main unit was inserted and irradiated with the same dose as during the reference measurement, without altering any couch settings. The calibration was repeated for the two wing units which were calibrated in pairs. The absolute calibration factor was determined for one detector (the central diode) per detector board.

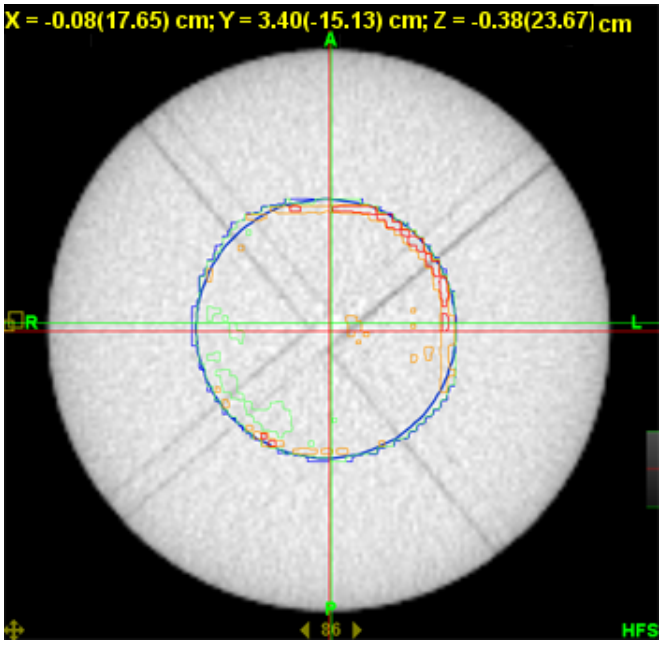

Figure 9: The offset between the red phantom set up lasers and the machine's isocenter (green laser): left/right, transverse, up/down.

Relative Calibration The Delta^4 calibration was made as an cross-calibration. At first the dose was determined with a farmer ionisation chamber, in an ionisation chamber calibration slab. Subsequently the ionisation chamber calibration slab was substituted by the calibration phantom, with the Delta^4 units inserted, whereupon a relative and an absolute calibration was performed. The relative calibration determined an individual relative sensitivity factor for every detector, that was used to compensate for detector-to-detector differences. The relative calibration was done by irradiating the detector board in different positions inside a large field with a constant number of monitor units, thus all detectors were consecutively moved into positions of known dose.

The calibration process was initiated by choosing "Relative Calibration" from the "Tools"-menu in the The Delta^4 ® software. The connected detector boards were then selected as well as the operative accelerator and beam quality, in this case 6 MV since that was the beam quality that corresponded to the tomotherapy unit. The field size was set to 26x26 cm and the SSD was set to 95 cm. The next step was to select the first detector board position in the software and align the calibration phantom to the corresponding position (displayed by the software). An offset measurement was done after which the measurement began by an irradiation of 100 MU. The procedure was then repeated for all detector positions, with the side scatter block used at the outermost positions.

Directional Calibration An optional choice in the Delta^4 calibration process is the performance of a directional calibration. This calibration aims to increase the accuracy of the measurements by determining the detector-to-detector variations regarding the rotational direction-dependency.[21] The procedure during

this calibration was to measure the dose in two detector board orientations: Normal and upside down. The hardware set up was identical to the set up during the relative calibration.

Daily Correction Factor Prior to every measurement session, a daily correction factor was derived that corrected for daily variations in the output of the machine, arising from variations in for example pressure, temperature and dose rate. This correction factor was derived by irradiating the Delta^4 detector with a rotational 1 Gy uniform plan (planned on the Delta^4 phantom), consisting of a cylinder with a diameter and Length of 10 cm respectively. The DQA plan had been planned so that the irradiation was centred over the central 60 x 60 mm^2 region of the detector, where the diode spacing is denser (5 mm compared to 10 mm). The mean deviation of the measured dose to the planned for the diodes could then be used as a daily output variation correction factor.[6]

To ensure that the prescribed dose of 1 Gy in the dose calculation truly was delivered by the machine a separate DQA plan was created on the TomoTherapy phantom (figure 11). The dose was measured in a point located 2 cm from the centre of the phantom and compared with the corresponding dose value from the dose calculation in the TPS system.

3.3.2 MapCHECK 2™

The patient-specific QA was performed with the MapCHECK 2 detector attached to the MapPHAN phantom and evaluated with the MapCHECK 5.00.00 software.

When starting up the MapCHECK program, with the MapCHECK powered on, a background measurement was taken automatically for 30 seconds. During this time the MapCHECK 2 measured the current in each detector and calculated a background rate. This rate was later, after a measurement, used to calculate a correction value for each diode by multiplying the background rate with the measurement time. The calculated background correction values were then subtracted from the measured dose values of each diode.[20]

Before making any measurements, an absolute dose calibration that converts the MapCHECK 2 relative dose values to absolute dose values, was performed. This calibration was done by irradiating the central diode of the MapCHECK 2 device with a rotational plan of 1 Gy. To perform the calibration, the MapPHAN phantom had to be placed on the couch, whereupon the MapCHECK 2 detector was inserted and attached to the phantom. The phantom was then aligned with the help of laser beams and a MVCT scan of the arrangement.

The control measurement of the DQA-plan could then be performed from the TomoTherapy Operator Station, by choosing the correct relative array calibration file from the drop-down menu in the MapCHECK program and setting the MAPCHECK 2 device in measurement mode by pushing the start button (figure 10).

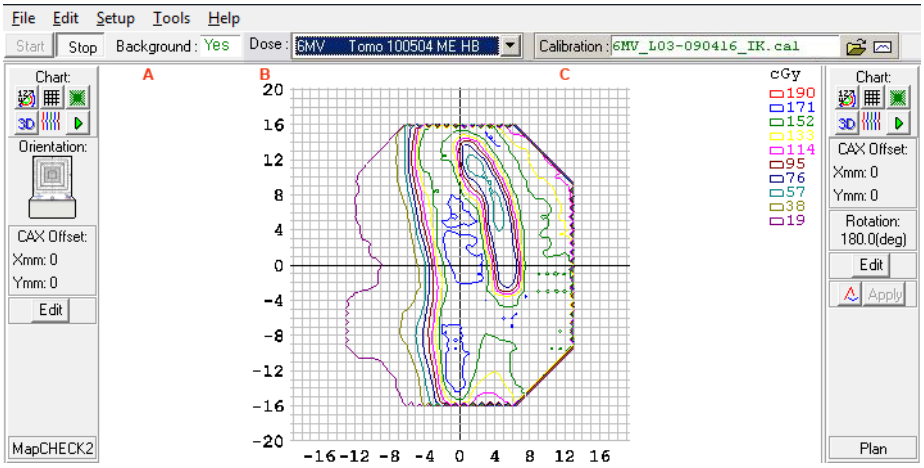

Figure 10: The choice of calibration file, in the MapCHECK program (version 5.00.00). A. Showing that a background measurement has been done. B. The 1 Gy calibration performed prior to measurement. C. The relative array calibration (performed a couple of times a year).

Relative Array Calibration The array calibration measured relative sensitivity differences between the detectors (diodes) in the MapCHECK 2. These differences were stored as individual correction factors and were applied to the raw measurements from each detector. The correction factors eliminated response differences between the individual detectors.[20] The array calibration was performed on a linear accelerator with the collimator set to a 37 x 37 cm field at an SSD of 100 cm. [20]

Absolute Dose Calibration The absolute dose calibration converted the MapCHECK 2 relative dose values to absolute dose values by applying a single calibration factor to all detectors.[20] This calibration resembled very much of the deriving of a daily correction factor for the Delta^4 and was performed before each measurement session started, so that daily variations in the output of the machine could be accounted for. The absolute dose calibration process consisted of an irradiation of the central diode of the MapCHECK 2 whereupon the calibration factor was derived. The detector was irradiated with an uniform 1 Gy rotational plan (planned on the MapCHECK 2 phantom) that consisted of a cylinder, with a diameter of 12 cm and a length of 8 cm, so that the central area of the detector was irradiated. To assure that the planned dose of 1 Gy indeed corresponded to 1 Gy a separate DQA measurement with an ionization chamber (A1SL Exradin Miniature Shonka Thimble Chamber, absolute calibrated at at the University of Wisconsin ADCL) in the TomoTherapy Commissioning Phantom was done (see figure 11). The dose was measured in a point located 2 cm from the centre of the phantom and compared with the corresponding dose value from the dose calculation in the TPS system.

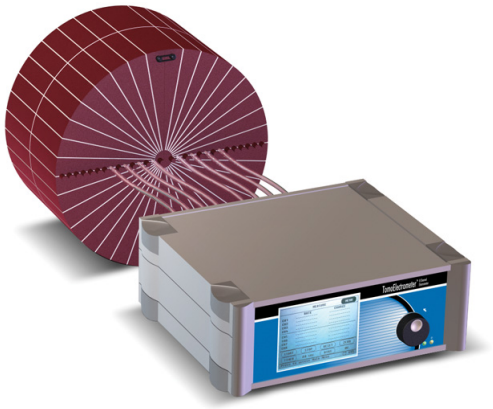

Figure 11: TomoTherapy Commissioning Phantom (Solid Water cylindrical "Cheese" phantom) shown with TomoElectrometer and eight Model A1SL Exradin Miniature Shonka Thimble Chambers. (Courtesy of TomoTherapy Incorporated, Madison, Wisconsin).

3.4 Evaluation Method

The measured patient plans were evaluated by examining the amount of diodes that passed the gamma evaluation. This was done for the acceptance criteria of; 3%/2mm, 3%/3mm, 3%/4mm and 3%/5mm with the different threshold, TH, settings; TH 5, TH 10, TH 20, TH 70, TH 90 and TH 95 applied. The threshold means that points with a dose less than the threshold value will be discriminated and not taken into account in the gamma evaluation. For example TH 10 means that points exposed to a dose less than 10 % of the maximum dose will not be taken into account. For the gamma calculation the algorithms of the associated software of the Delta^4 and the MapCHECK 2 was used respectively.

For the MapCHECK 2 the gamma evaluation offered two ways to calculate percent dose difference; Van Dyk percent difference and MapCHECK 2 percent difference. In the evaluation the MapCHECK 2 percent difference was used.

A DQA-plan is considered to be accepted if 90 % of the points passes the gamma evaluation, with the gamma criterion set to 3%/3mm at a threshold level of 10 %, as suggested by M. Geurts et al. [22]

4 Results and Discussion

4.1 Control Measurement of Plans for Daily Output Variation Correction Factor

Table 1 represents the results from the ionization chamber measurements of the plan used to derive a daily output variation correction factor for the Delta^4 device, as described in paragraph 3.3.1 Daily Correction Factor. The average deviation between the planned dose, from the TPS, and the measured dose was +2.2%. The readings of the ionization chamber could be regarded as stable and reproducible even though it should be noted that a quite small difference in collected charge of 0.02 nC (for the third and fourth measurement) causes a large deviation between planned- and measured dose. A possible explanation to the measured over-dosage of 2.2 % could be derived from the Image Value-to-Density Calibration Table, as it has not been revised since the installation of the Hi-Art system. Several components has broken since then and been replaced. It is not implausible that this might have changed the output slightly (and consequently the Image Value-to-Density Calibration Table), and thereby largely impact the measurements of the uniform plan. The plan has been re-measured and re-planned but the deviation remains constant (around 2.2 %), with some small variations depending on the daily output. The over-dosage cannot be addressed to sharp dose gradients since the planned dose has a very homogeneous behaviour in the measured area. The over-dosage can not be explained by differing dose rate either since it was rather low than high. In summary no solid explanation to the measured over-dosage has been found and further examinations of this issue are considered to be beyond the scope of this project. However the deviation of +2.2 % is considered to be acceptable, in its context, and the delivery of the cylindrical 1 Gy plan is found to be ascertained.

Table 1: Control measurement, with ionization chamber, of cylindrical plan for Delta^4 calculated on the "Cheese" phantom.

Plan: Delta^4
Dose from TPS: 0.88 Gy

Measurement	Charge [nC]	Dose [Gy]	Difference [%]
1	-1.62	0.900	2.2
2	-1.62	0.900	2.2
3	-1.61	0.894	1.6
4	-1.63	0.905	2.9
5	-1.62	0.900	2.2
	Average difference:		2.2

Table 2 represents the results from the ionization chamber measurements of the plan used for absolute dose calibration of the MapCHECK 2 device, as described in paragraph 3.3.2 Absolute Dose Calibration. The average deviation between the planned dose, from the TPS, and the measured dose was -1.7%. The readings of the ionization chamber were stable, disregarding the first measurement, and displayed constancy when the measurements were reproduced. A

possible explanation to this under-dosage could be that the tomotherapy machine, at the moment of measurement, delivered the radiation with an average dose rate of 874 MU/min instead of the expected 888 MU/min. Since the Hi-Art system solely works with a previously defined exposure time, given a constant dose rate of 888 MU/min, the total delivered dose to any site will be the time-integral of the dose-rate at that site. The fact that the dose rate was lower than expected by the TPS could therefore very well be a feasible explanation to the somewhat low readings of the ionization chamber.

Another explanation that might explain the measured under-dosage is that the shape and geometry of the MapCHECK 2 device and the "Cheese" phantom differ significantly, since the MapCHECK 2 resembles a flat rectangle whereas the "Cheese" phantom is cylindrical. This could affect the recalculation of the dose for the DQA plan, especially when contribution from side scatter is considered.

It is very hard to deduce that the measured under-dosage is due to a single reason, since there are several possible explanations. However the same measurements were done at several occasions, separated by a few days (not represented in the table), all of them showing that the average deviation between the planned dose and the measured dose did not differ much from the -1.7 %.

With the possible explanations, to the measured under-dosage stated above, in mind the plan was finally considered to be reasonably in control *i.e.* the deviation of -1.7 % is considered to be acceptable and the delivery of the cylindrical 1 Gy plan is found to be ascertained.

Table 2: Control measurement of the cylindrical MapCHECK 2 plan calculated on the Cheese phantom, performed with an ionization chamber.

Plan: MapCHECK 2
Dose from TPS: 1.01 Gy

Measurement	Charge [nC]	Dose [Gy]	Difference [%]
1	-1.78	0.988	-2.1
2	-1.79	0.994	-1.6
3	-1.79	0.994	-1.6
4	-1.79	0.994	-1.6
5	-1.79	0.994	-1.6
	Average difference:		-1.7

4.2 Delta^4 Calibration Results

When the relative calibration of the Delta^4 was completed the calibration summary (figure 12) was examined. The calibration factor-distribution showed a Gaussian distribution well centred around 0 %, as was expected.

After the directional calibration had been performed for all energies of interest, a frequency histogram was shown consisting of the number of detectors with a specific directional calibration factor. The function represented a Gaussian distribution, with a mean of 0.3 % and a standard deviation of 2 %. A result that was in line with what was expected from the theory.

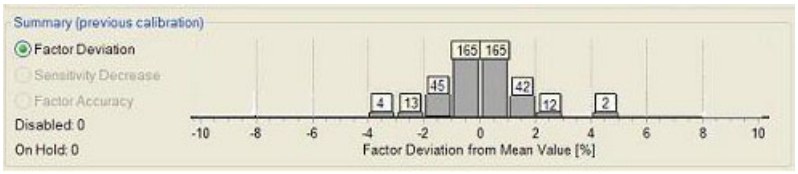

Figure 12: Delta^4 calibration summary showing a Gaussian distribution.

4.3 Evaluated DQA Plans

In total 25 randomly selected clinical cases for a variety of disease sites were recalculated and measured as DQA plans for the Delta^4- and the MapCHECK 2 detector. The measurements were evaluated with the associative software of the two systems and the results can be found as a whole in the appendix A. Since each of the DQA plans had to be created individually for the Delta^4- and the MapCHECK 2 device, the plans might differ somewhat from each other. However every DQA plan has been planned so that the detector encapsulates as much as possible of the planning treatment volume. The two detector systems were evaluated and compared by studying the gamma pass rate of the measured DQA plans for different gamma- and threshold criteria. Relevant graphs were created from the corresponding results, to make the results more illustrative and easier to review. These graphs are presented and explained below.

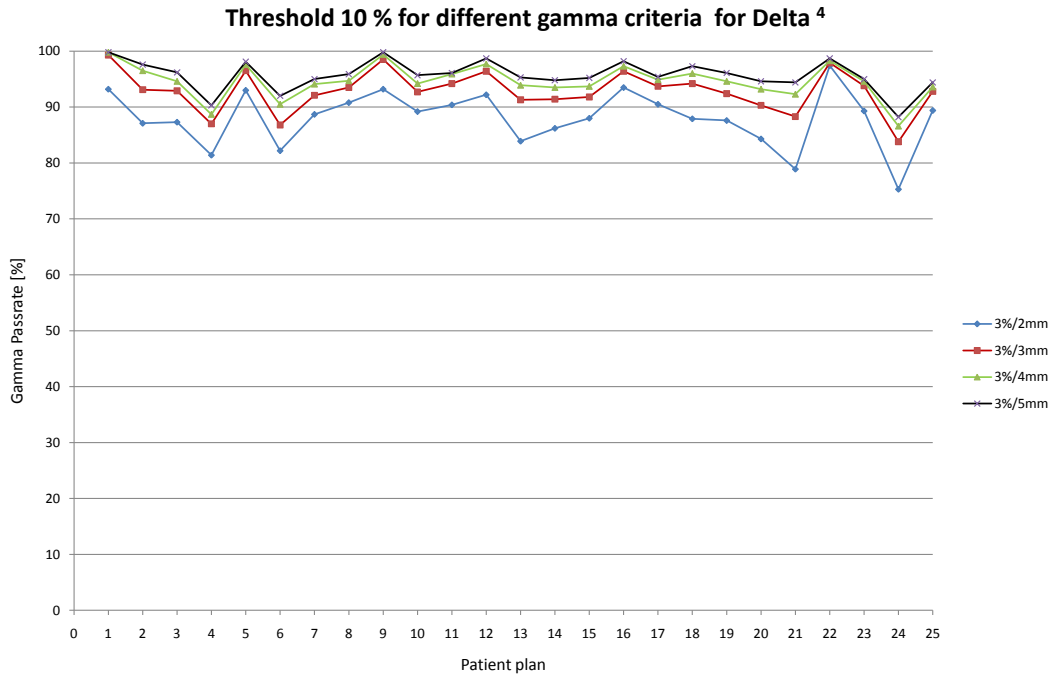

Figure 13: The gamma pass rate for the Delta^4, with a threshold level of 10 %, for different dose criteria.

Figure 13 and 14 illustrates the gamma pass rate for the Delta^4 and the Map-

CHECK 2, with a threshold level of 10 %, for different dose criteria. The dose criterion is kept at 3 % whereas the DTA is varied between 2-5 mm. The graphs are presented to give an opinion of how much the gamma pass rate is affected by varying the DTA.

From the graph in figure 13 it can be seen that the curves, representing the different criteria, are gathered tight together and seem to follow each other very well which implies that the choice of a specific gamma criterion is not that important as long as a proper gamma pass rate level for acceptance of a plan, is chosen. For example the gamma pass rate level used for acceptance of a tomotherapy DQA plan at Skåne University Hospital in Lund is 90 % at a threshold level of 10 %, with the gamma criterion 3%/3mm. Applying these criterion to the graph in figure 13 would result in 21 plans that are considered accepted and 4 plans that are considered unaccepted. However when looking at the 3%/2mm criterion only 9 out of 25 plans are accepted with a 90 % pass rate level. It is thus obvious that the acceptance level has to be adjusted to correspond to the chosen gamma criterion. However, it should be kept in mind that the proper choice of the DTA criterion depends very much on the pixel spacing, which needs to be sufficiently small to provide an accurate calculation of γ at regions of steep steep dose gradients in the evaluated distribution. As a general rule, the spacing should be less than or equal to 1/3 of Δd or expressed in another way; Δd has to be three times the pixel spacing. [14] The Hi-Art System offers three different calculation grids for optimisation: coarse=0.936x0.936 cm, normal=0.468x0.468 cm and fine=0.234x0.234 cm. And three different calculation grids for DQA plans: coarse=0.612x0.612 cm, normal=0.306x0.306 and fine=0.153x0.153. The calculation grid titled "normal" has been used both in the optimisation process and when creating DQA plans. The criterion that pixel spacing should be less than or equal to 1/3 of Δd is thus not fulfilled for the examined DTA criteria. The use of the "normal" calculation grid was justified with that it is not practicable to use the grid named "fine" since it is very time-wasting. When comparing two different grids "fine" and "normal" the difference was very small or nonexistent.

The biggest drawback of the Delta^4 system can be seen when looking at the gamma angle (not illustrated in the results) where the dose difference is the parameter that mostly influences the gamma value. In general the DTA is highly conformable, often with an agreement above 97 % (within 3 mm). The agreement for the dose difference is however around 85 % at best (within 3 %). The same effect is seen when measuring IMRT plans with a conventional linac, however no explanation to this has been found.

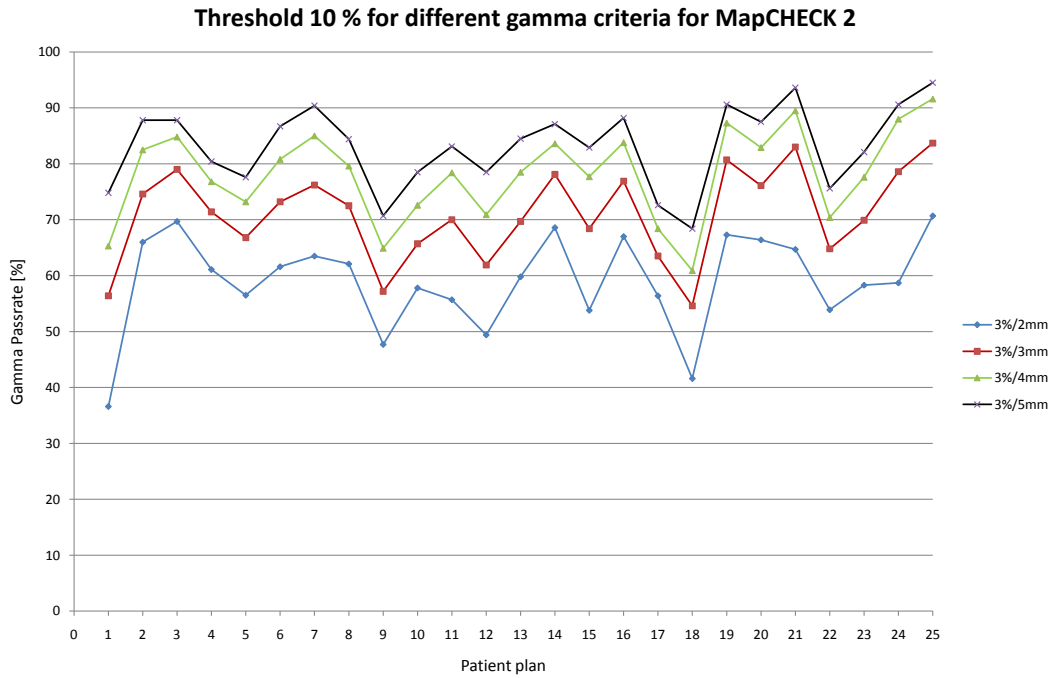

Figure 14: The gamma pass rate for the MapCHECK 2, with a threshold level of 10 %, for different dose criteria.

The graph in figure 14 shows the same appearance of the curves, as could be seen in figure 13 *i.e.* the curves seem to follow each other, despite the fact that different DTA criteria are used in the gamma evaluation. However the gamma pass rate for the MapCHECK 2 is greater affected by different DTA criteria than seen for the Delta^4 system. From the graph it is obvious that the acceptance criterion of 90 % in pass rate for the gamma evaluation is not suitable or even possible to use with the MapCHECK 2. It is possible that other acceptance criteria, that are more suitable for the MapCHECK 2, could be defined but that has not been further investigated in this thesis.

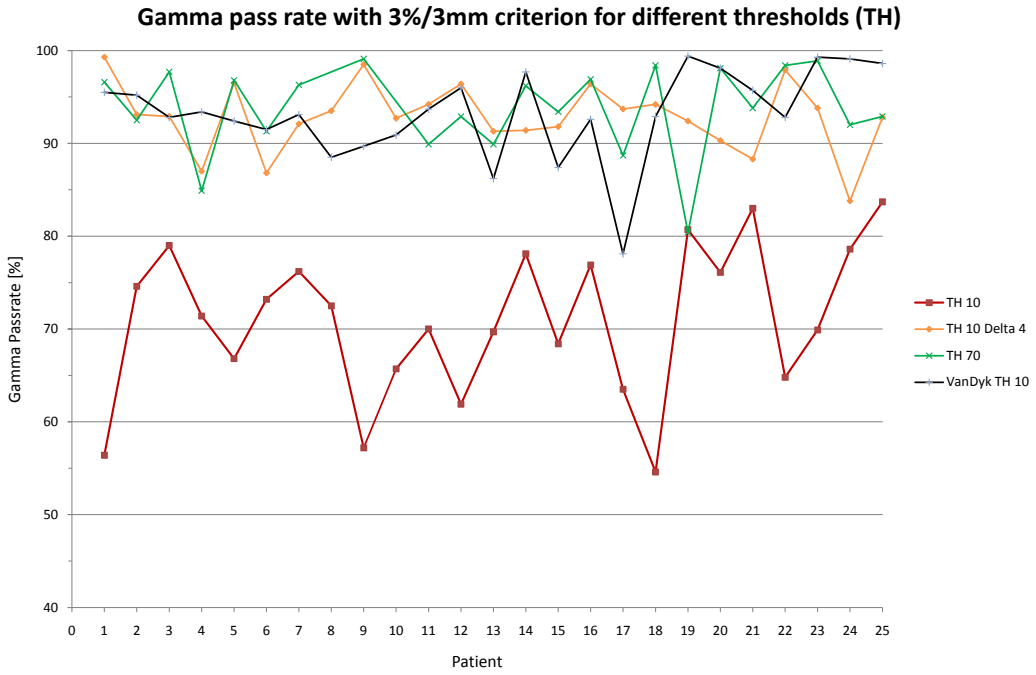

Figure 15: The gamma pass rate with 3%/3mm criterion for the threshold of 10 %, with (black curve) and without (red curve) the Van Dyk percentage difference, and the threshold of 70 % (green curve) for the MapCHECK 2 compared to the pass rate for the Delta^4 at a threshold level of 10 % (orange curve). Different pass rates are compared to illustrate the differences in gamma pass rate for the MapCHECK 2 and the Delta^4.

Figure 15 illustrates the differences in gamma pass rate for the MapCHECK 2 and the Delta^4. First of all a comparison can be done when the clinical settings are used, *i.e.* 3%/3mm at TH 10. The Delta^4 yields 21 accepted plans out of 25, whereas none of the plans measured with the MapCHECK 2 is even close to the acceptance level. However when applying the Van Dyk percent difference, in the gamma evaluation for the MapCHECK 2, the black curve is achieved with 21 accepted plans out of 25 *i.e.* the same amount as for the Delta^4. It should however be noted that the plans that are not accepted are in fact accepted with the Delta^4 and the plans that are not accepted by the Delta^4 are accepted when using the Van Dyk percent difference in the gamma evaluation for the MapCHECK 2. Using the Van Dyk percent difference might be a tempting way to achieve good results in the gamma evaluation of measured DQA plans. But the physicist should be aware that a normalisation to the maximum dose is performed, thus extending the dose difference criteria to 30 % at an isodose level of 10 % (if a 3 % dose difference criterion is used). In other words the Van Dyk percent difference does not reflect local dose differences which might be more relevant for organs at risk.

To get the same level in gamma pass rate for the MapCHECK2 as with the Delta^4, without using the Van Dyk percent difference, a threshold level of 70 percent is needed. This is of course not a good way to deal with the problem

since such a high threshold level is not suitable with regards to organs at risk. However, high thresholds are useful when investigating the dose to the planning treatment volume. This problem might be due to a directional dependence of the diodes and CT metal artifacts in combination with the fact that the 2D dose-distribution information available when the beam direction is perpendicular to the detector plane is gradually reduced to 1D, as the incident angle approaches 90°.[23][6][8]. This means that when a substantially amount of the irradiation is incident from the sides a deviation between planned and measured dose will arise. The effect is especially apparent if the planning treatment volume is placed in the middle of the detector. This deviation will have a larger impact at lower dose levels and might thus explain why a higher threshold level gives a better consistency in gamma pass rate. However, as stated above, a threshold level of 70 % is not at all useful since it does not tell anything about how organs at risk are effected.

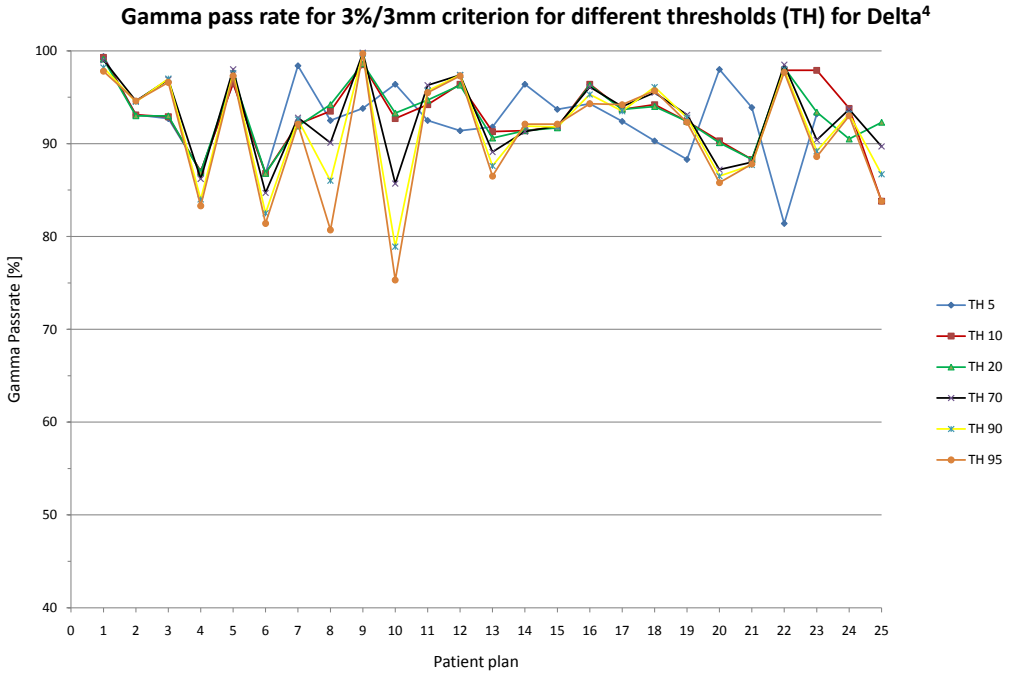

Figure 16: Gamma pass rate for 3%/3mm criterion for different thresholds for the Delta^4.

Figure 16 and 17 depict the curves for all the examined threshold levels for the Delta^4 and the MapCHECK 2 respectively, with a 3%/3mm gamma criterion. Figure 16 illustrates that the gamma pass rate does not show a notable variation between different thresholds for the Delta^4. However some drops at high thresholds are seen for patient 8 and 10. This effect is also seen for the same patients for the MapCHECK 2, see appendix A, where it can be attributed to the fact that no diodes seem to receive a dose higher than 70 % of the maximum dose. This is an effect that can emerge, especially for long targets, when the maximum dose point is not embedded by the QA phantom. In general one

should be extra careful when evaluating gamma pass rates for higher thresholds since the gamma values, at higher dose levels, are very sensitive to the number of points in agreement because of the reduced number of data points available.[24] For some cases, ten or less data points have been detected in regions of isodoses of 90 % or higher.

Figure 17 confirm, what was also indicated in figure 15, that the MapCHECK 2 dosimetry system has some shortcomings when low threshold levels are used. The drop at the higher threshold levels for patient 19 is probably attributed to the effect described above.

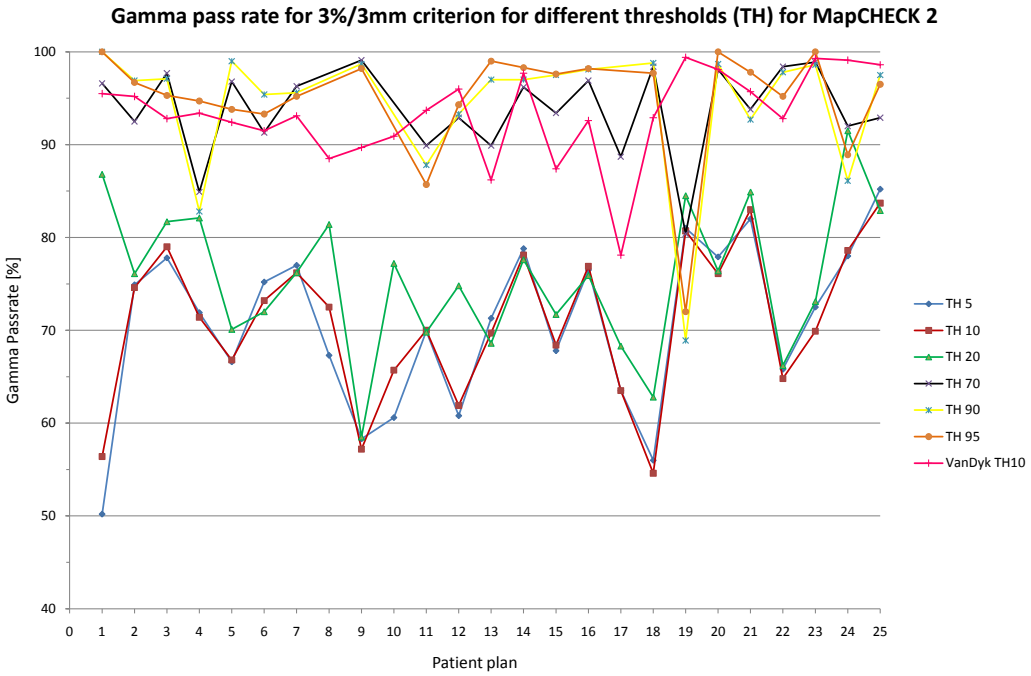

Figure 17: Gamma pass rate for 3%/3mm criterion for different thresholds levels for the MapCHECK 2.

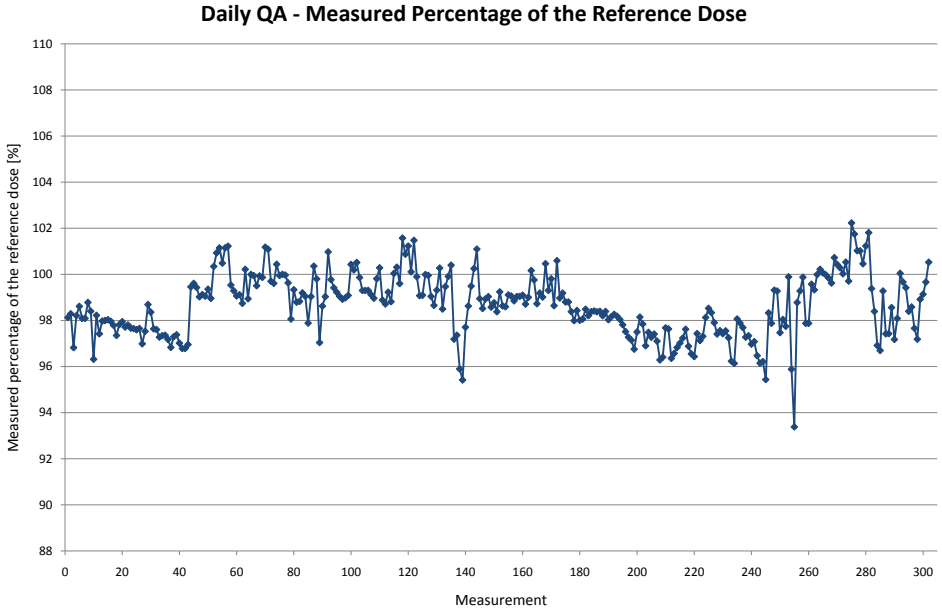

Figure 18: Diagram showing the results, normalised to the reference dose of 5 Gy, of the daily QA measurement covering well over 300 days of the tomotherapy unit.

Figure 18 shows the results of the Daily QA that is done every morning prior to the first treatment of the day. As can be seen from the diagram the machine is quite unstable with differences in output of up to 5 % from one day to another, for a reference dose of 5 Gy. This effect is accounted for by deriving a daily correction factor (see 3.3.1 Daily Correction Factor and 3.3.2 Absolute Dose Calibration).

4.4 Comparison With Other Studies

Several articles have proposed that it is reasonable to maintain a tolerance of 90 % of the gamma map achieving 3 % and 3 mm agreement, for the Delta^4 system i.e. requiring at least 90 % of the diodes to pass the gamma statistic correctly. [6, 22, 18] For the MapCHECK 2 a tolerance of 95 % of the gamma map achieving 3 % and 3 mm agreement is proposed by Winningham et al. [8]

In the article written by Feygelman et al. [6] 9 clinical helical Tomotherapy plans and 2 uniform cylindrical fields were measured and evaluated with the Delta^4 system. The average (3%/3mm) passing rate for absolute dose was 97 $\pm$ 2.7 %. The range was 90.7-99.8 %, with only one case $<$ 96%.

The corresponding results from this thesis, which covered 25 clinical plans, was an average (3%/3mm) passing rate for absolute dose of 92.8$\pm$3.7 %. The range was 83.8-99.3 % with a median value of 92.9 %, see Appendix B for more details.

For the MapCHECK 2 Winningham et al. reports that less than 5% of more than 100 patient treatment plans verified with the 2D array had pass rates less than 95% when using 3%/3mm Van Dyk criteria (TH 10).[8]

The corresponding results acquired in this thesis was an average (3%/3mm)

passing rate for absolute dose of 93.2±4.8 % when using the Van Dyk percent difference in the gamma evaluation (TH 10). The range was 78.1-99.4 % with a median value of 93.1 %.

As stated many times before in this thesis the Van Dyk percent difference has clear drawbacks and the DQA plans has therefore not been subject for a deeper analyse regarding the Van Dyk criteria. Without the Van Dyk percent difference the average (3%/3mm) passing rate for absolute dose was 70.9±8.1 %. The range was 54.6 - 83.7 % with a median value of 71.4 %.

5 Conclusions

In the present work a great effort has been dedicated to get the Delta^{4}® dosimetry system working along with the TomoTherapy Hi-Art System®. The MapCHECK 2™ dosimetry system has not received the same attention, since it was already implemented clinically, both for verification of DQA plans for tomotherapy as well as for QA of IMRT plans.

During the measurements it has become clear that the dose delivery of the tomotherapy unit is not as stable as the vendor claims. This is seen both on a day-to-day basis, where the output of the machine can vary up to 5 %, but also during a single measurement, where the dose rate can vary up to 10 MU/min between the start and the end of the irradiation. A daily output variation correction can be derived from the Delta^{4} measurement in a standard helical beam, producing a uniform cylindrical dose distribution. Thus the day-to-day output variation can be accounted for. The variations in dose rate during irradiation is on the other hand not that easy to account for and might be a potential source of error, since the total delivered dose to any site will be the time-integral of the dose-rate at that site.

The Delta^{4} system were found satisfactory in terms of fulfilling the high demands that are expected from a competent system for patient-specific QA. A conclusion that can be drawn from comparing the results from the MapCHECK 2 with the results from the Delta ^{4} is that the Delta^{4} system is superior to the MapCHECK 2 system. In fact the MapCHECK 2, when used without the Van Dyk percent difference, is not at all usable for patient-specific QA on helical tomotherapy.

21 of the 25 clinical plans, measured with the Delta^{4} system, passed the (3%, 3 mm) gamma evaluation above 90 %. The corresponding result for the MapCHECK 2 was zero accepted plans of 25, when not using the Van Dyk percent difference in the calculation. With the Van Dyk criteria 21 clinical plans were approved however the remaining 4 plans were not the same as for the Delta ^{4}.

The gamma evaluation for the MapCHECK 2 and the Delta^{4} has been calculated according to the same principle, described in the subsection 2.2.3 Gamma Evaluation *i.e.* the Van Dyk percent difference has generally not been used in the calculation. The only occasion when the Van Dyk percent difference in fact has been used in the gamma evaluation is as a complement to the threshold level of 10 %, for the gamma criterion of 3%/3mm, since that is the settings used clinically at Skåne University Hospital in Lund.

Even though the Delta^{4} system seem to be promising for patient-specific QA of helical tomotherapy beams, there are still further improvements to make before optimal performance is reached. For example one thing that is needed to be improved is the agreement in dose difference which is quite low *i.e.* around 85 % (3 % dose difference criterion). The somewhat acceptable gamma pass rate, in this thesis, for the Delta^{4} is almost solely due to the almost perfect agreement in DTA.

The principal conclusion of this MSc thesis is thus that the Delta^{4} system is superior to the MapCHECK 2 system and should be put in clinical use as soon as possible. At last in can be concluded that with this thesis the essential preparatory work in order to implement the Delta^{4} system for clinical usage has been performed.

References

[1] Socialstyrelsen. Dödsorsaker 2007. *Sveriges officiella statistik: Hälso- och sjukvård*, 2009.

[2] World Health Organisation (WHO). Web page: http://www.who.int/.

[3] U. Ringborg, D. Bergqvist, B. Brorsson, E. Cavallin-ståhl, J. Ceberg, N. Einhorn, J. Frödin, J. Järhult, G. Lamnevik, and C. Lindholm et al. The Swedish Council on Technology Assessment in Health Care (SBU) Systematic Overview of Radiotherapy for Cancer including a Prospective Survey of Radiotherapy Practice in Sweden 2001 - Summary and Conclusions. *Acta Oncologica*, 42(5-6):357–365, 2003.

[4] R. Alfonso-Laguardia and M. Vega-Hernández. In vivo dosimetry for patient specific quality assurance in aperture based IMRT. *World Congress on Medical Physics and Biomedical Engineering*, 25/1:859–862, 2009.

[5] M Alber, S Broggi, C. De Wagter, I. Eichwurzel, P. Engström, C.Fiorino, D. Georg, G.Hartmann, and T. Knöös et al. *Guidelines for the verification of IMRT*. ESTRO, 2008.

[6] V. Feygelman, K. Javedan, A.J. Saini, and G. Zhang. Evaluation of a 3D diode array dosimeter for helical tomotherapy delivery QA. *Medical Dosimetry*, 2010: in press.

[7] Jonathan G. Li, Guanghua Yan, and Chihray Liu. Comparison of two commercial detector arrays for IMRT quality assurance. *Journal of applied clinical medical physics*, 10, No 2:62 – 74, 2009.

[8] T. Winningham, R. Staton, and S. Meeks. SU-FF-T-209: Implementation of 2D Arrays for TomoTherapy Patient-Specific QA. *Medical Physics*, 36 Issue 6:2568, 2009.

[9] P. Metcalfe, T. Kron, and P. Hoban. *The Physics of Radiotherapy X-Rays and Electrons*. Madison: Medical Physics Publishing, 2007.

[10] J.D. Fenwick, W.A Tomé, H.A. Jaradat, S. K. Hui, J.A. James, J.P. Balog, C.N. DeSouza, D.B. Lucas, G.H. Oliviera, T.R. Mackie, and B.R. Paliwal. Quality assurance of a helical tomotherapy machine. *Physics in Medicine and Biology*, 49:2933–2953, 2004.

[11] I. Van de Vondel, K. Tournel, D. Verellen, M. Duchateau, S. Lelie, and G. Storme. A diagnostic tool for basic daily quality assurance of a tomotherapy Hi-Art machine. *Journal of applied clinical medical physics*, 10 Number 4:151–164, 2009.

[12] W.B. Harms Sr, D.A. Low, J.W. Wong, and J.A.Purdy. A software tool for the quantitative evaluation of 3D dose calculation algorithms. *Medical Physics*, 25(10):1830–1836, 1998.

[13] D.A. Low, W.B.Harms, S.Mutic, and J.A.Purdy. A technique for the quantitative evaluation of dose distributions. *Medical Physics*, 25(5):656–661, 1998.

[14] D.A. Low and J.F. Dempsey. Evaluation of the gamma dose distribution comparison method. *Medical Physics*, 30(9):2455–2464, 2003.

[15] T. Depuydt, A. Van Esch, and D.P. Huyskens. A quantitative evaluation of IMRT dose distributions: refinement and clinical assessment of the gamma evaluation. *Radiotherapy and Oncology*, 62:309–319, 2002.

[16] Tomotherapy Incorporated, 1240 Deming Way Madison, WI 53717 USA. *Planned Adaptive-Handbok Version 3.X*, 2007.

[17] Commissioning of Radiotherapy Treatment Planning Systems: Testing for Typical External Beam Treatment Techniques. *IAEA-TECDOC-1583*, 2008.

[18] J.L. Bedford, Y.K. Lee, P. Wai, C.P. South, and A.P. Warington. Evaluation of the Delta^4 phantom for IMRT and VMAT verification. *Physics in Medicine and Biology*, 54:N167–N176, 2009.

[19] *Delta^4®: Getting Started*. ScandiDos AB, 2009.

[20] *MapCHECK 2 , Reference Guide (Document 1177011, Rev D)*. Sun Nuclear Corporation, 2009.

[21] Delta^4® Help File, April 2009.

[22] M. Geurts, J. Gonzalez, and P. Serrano-Ojeda. Longitudinal study using a diode phantom for helical tomotherapy IMRT QA. *Medical Physics*, 36:4977–4983, 2009.

[23] J. Zhang. SU-GG-T-247: A New Method to Compensate Angular Dependency of MapCHECK Device in Intensity Modulated Arc Therapy. *Medical Physics*, 37 Issue 6:3242, 2010.

[24] C. Andenna, M. Benassi, B. Caccia, S. Marzi, M.Pedrini, and C. Zicari. Comparison of Dose Distribution in IMRT Planning Using the Gamma Function. *Journal of Experimental and Clinical Cancer Research*, 25, 2:229–234, 2006.

A Measurement Results

Results of the Measured DQA Plans

CS is an abbreviation for clinical settings and means that the Van Dyk percent difference has been used in the gamma evaluation. The cells in light blue is the values that has been measured, by the MapCHECK 2, when accepting the plans for treatment. For some cells, for the MapCHECK 2, data is missing and the reason to this is that no diodes seems to receive a dose higher than 70 % of the maximum dose which makes the gamma evaluation not available.

Gamma Pass Ratio [%] for MapCHECK 2							
Patient 1	CS	TH 5	TH 10	TH 20	TH 70	TH 90	TH 95
3%/2mm	78.2	34.3	36.6	53.8	69.0	81.3	84.6
3%/3mm	95.5	50.2	56.4	86.8	96.6	100	100
3%/4mm	100	59.0	65.3	94.5	100	100	100
3%/5mm	100	68.8	74.8	96.7	100	100	100

Gamma Pass Ratio [%] for Delta^4						
Patient 1	TH 5	TH 10	TH 20	TH 70	TH 90	TH 95
3%/2mm	94.0	93.2	90.5	85.7	85.7	91.1
3%/3mm	99.4	99.3	99.1	99.0	98.2	97.8
3%/4mm	99.8	99.8	99.7	99.0	98.2	97.8
3%/5mm	99.8	99.8	99.7	99.0	98.2	97.8

Gamma Pass Ratio [%] for MapCHECK 2							
Patient 2	CS	TH 5	TH 10	TH 20	TH 70	TH 90	TH 95
3%/2mm	91.7	65.8	66.0	67.6	86.1	92.0	91.8
3%/3mm	95.2	74.9	74.6	76.1	92.5	96.9	96.7
3%/4mm	97.0	82.8	82.5	84.2	95.5	98.1	96.7
3%/5mm	98.3	88.2	87.8	89.7	97.0	98.1	96.7

Gamma Pass Ratio [%] for Delta^4						
Patient 2	TH 5	TH 10	TH 20	TH 70	TH 90	TH 95
3%/2mm	87.2	87.1	86.8	89.4	90.7	91.9
3%/3mm	93.1	93.1	93.0	94.6	94.5	94.6
3%/4mm	96.4	96.5	96.6	96.8	96.4	96.4
3%/5mm	97.6	97.6	97.7	97.7	97.2	96.6

Gamma Pass Ratio [%] for MapCHECK 2							
Patient 3	CS	TH 5	TH 10	TH 20	TH 70	TH 90	TH 95
3%/2mm	89.2	67.3	69.7	73.1	93.3	94.6	92.2
3%/3mm	92.8	77.8	79.0	81.7	97.7	97.1	95.3
3%/4mm	94.0	84.8	84.8	87.1	98.6	98.1	96.9
3%/5mm	94.6	88.4	87.8	89.8	99.1	98.7	98.4

Gamma Pass Ratio [%] for Delta^4						
Patient 3	TH 5	TH 10	TH 20	TH 70	TH 90	TH 95
3%/2mm	87.2	87.3	87.3	92.3	94.2	94.3
3%/3mm	92.7	92.9	93.0	96.9	97.0	96.6
3%/4mm	94.5	94.6	94.9	98.1	97.8	97.5
3%/5mm	96.1	96.2	96.2	98.6	98.4	98.2

Gamma Pass Ratio [%] for MapCHECK 2							
Patient 4	CS	TH 5	TH 10	TH 20	TH 70	TH 90	TH 95
3%/2mm	92.4	61.5	61.1	71.3	78.6	80.3	93.3
3%/3mm	93.4	71.9	71.4	82.1	84.9	82.8	94.7
3%/4mm	94.0	77.3	76.8	86.5	90.2	89.2	98.7
3%/5mm	94.2	80.9	80.4	89.2	91.6	90.4	98.7

Gamma Pass Ratio [%] for Delta^4						
Patient 4	TH 5	TH 10	TH 20	TH 70	TH 90	TH 95
3%/2mm	81.4	81.4	81.2	77.6	76.6	76.7
3%/3mm	87.0	87.0	86.9	86.2	84.0	83.3
3%/4mm	88.7	88.7	88.6	87.8	85.8	85.0
3%/5mm	90.3	90.3	90.2	89.5	87.5	86.4

Gamma Pass Ratio [%] for MapCHECK 2							
Patient 5	CS	TH 5	TH 10	TH 20	TH 70	TH 90	TH 95
3%/2mm	88.6	56.5	56.5	59.4	90.7	99.0	93.8
3%/3mm	92.4	66.6	66.8	70.1	96.8	99.0	93.8
3%/4mm	94.0	73.2	73.2	76.5	97.5	99.0	93.8
3%/5mm	95.7	77.6	77.6	80.9	98.6	99.0	93.8

Gamma Pass Ratio [%] for Delta^4						
Patient 5	TH 5	TH 10	TH 20	TH 70	TH 90	TH 95
3%/2mm	93.0	93.0	93.5	92.9	91.3	90.7
3%/3mm	96.5	96.5	97.3	98.0	97.6	97.3
3%/4mm	97.6	97.6	98.2	98.7	98.3	98.1
3%/5mm	98.1	98.1	98.7	99.2	99.0	98.8

Gamma Pass Ratio [%] for MapCHECK 2							
Patient 6	CS	TH 5	TH 10	TH 20	TH 70	TH 90	TH 95
3%/2mm	87.7	63.7	61.6	61.0	85.7	94.1	91.0
3%/3mm	91.5	75.2	73.2	72.0	91.3	95.4	93.3
3%/4mm	94.8	82.2	80.8	79.9	95.4	96.8	94.9
3%/5mm	96.7	87.8	86.7	85.9	97.0	97.9	97.2

Gamma Pass Ratio [%] for Delta^4						
Patient 6	TH 5	TH 10	TH 20	TH 70	TH 90	TH 95
3%/2mm	82.2	82.2	82.2	80.5	79.1	77.9
3%/3mm	86.8	86.8	86.8	84.7	82.5	81.4
3%/4mm	90.5	90.5	90.5	88.5	86.5	85.4
3%/5mm	92.0	92.0	92.0	90.2	88.5	87.6

Gamma Pass Ratio [%] for MapCHECK 2							
Patient 7	CS	TH 5	TH 10	TH 20	TH 70	TH 90	TH 95
3%/2mm	85.2	63.7	63.5	65.0	89.5	93.2	91.9
3%/3mm	93.1	77.0	76.2	76.2	96.3	95.6	95.2
3%/4mm	96.0	86.4	85.0	84.9	97.9	97.1	97.1
3%/5mm	97.6	91.4	90.4	90.2	98.6	98.2	98.2

Gamma Pass Ratio [%] for Delta^4						
Patient 7	TH 5	TH 10	TH 20	TH 70	TH 90	TH 95
3%/2mm	88.4	88.7	88.6	89.5	89.5	89.1
3%/3mm	91.8	92.1	91.9	92.8	92.6	92.1
3%/4mm	93.7	94.1	93.9	94.6	94.4	94.1
3%/5mm	94.9	95.0	94.9	95.5	95.5	95.2

Gamma Pass Ratio [%] for MapCHECK 2							
Patient 8	CS	TH 5	TH 10	TH 20	TH 70	TH 90	TH 95
3%/2mm	81.8	57.5	62.1	71.8	-	-	-
3%/3mm	88.5	67.3	72.5	81.4	-	-	-
3%/4mm	92.4	74.1	79.6	86.6	-	-	-
3%/5mm	94.3	79.0	84.4	89.7	-	-	-

Gamma Pass Ratio [%] for Delta^4						
Patient 8	TH 5	TH 10	TH 20	TH 70	TH 90	TH 95
3%/2mm	90.8	90.8	91.1	83.2	76.7	70.3
3%/3mm	93.5	93.5	94.2	90.1	86.0	80.7
3%/4mm	94.8	94.7	95.6	92.4	89.2	85.1
3%/5mm	95.9	95.9	96.5	94.2	91.8	88.6

Gamma Pass Ratio [%] for MapCHECK 2							
Patient 9	CS	TH 5	TH 10	TH 20	TH 70	TH 90	TH 95
3%/2mm	85.5	47.3	47.7	50.0	95.9	96.7	97.2
3%/3mm	89.7	58.3	57.2	58.5	99.1	98.7	98.2
3%/4mm	91.7	67.2	64.9	65.4	99.1	98.7	98.2
3%/5mm	93.2	72.6	70.7	70.6	99.1	98.7	98.2

Gamma Pass Ratio [%] for Delta^4						
Patient 9	TH 5	TH 10	TH 20	TH 70	TH 90	TH 95
3%/2mm	93.6	93.2	93.1	97.6	98.5	99.0
3%/3mm	98.4	98.5	98.6	99.8	99.7	99.7
3%/4mm	99.3	99.4	99.5	99.8	99.7	99.7
3%/5mm	99.6	99.8	99.9	100.0	100.0	100.0

Gamma Pass Ratio [%] for MapCHECK 2							
Patient 10	CS	TH 5	TH 10	TH 20	TH 70	TH 90	TH 95
3%/2mm		53.0	57.8	69.7	-	-	-
3%/3mm		60.6	65.7	77.2	-	-	-
3%/4mm		67.1	72.6	82.9	-	-	-
3%/5mm		72.8	78.5	87.1	-	-	-

Gamma Pass Ratio [%] for Delta^4						
Patient 10	TH 5	TH 10	TH 20	TH 70	TH 90	TH 95
3%/2mm	89.1	89.2	89.3	78.2	67.6	61.8
3%/3mm	92.5	92.7	93.3	85.7	78.9	75.3
3%/4mm	93.9	94.2	94.9	88.8	83.8	81.2
3%/5mm	95.4	95.7	96.7	93.1	89.2	87.1

Gamma Pass Ratio [%] for MapCHECK 2							
Patient 11	CS	TH 5	TH 10	TH 20	TH 70	TH 90	TH 95
3%/2mm	86.1	54.6	55.7	56.4	80.3	84.4	82.1
3%/3mm	93.7	69.9	70.0	69.8	89.9	87.8	85.7
3%/4mm	96.7	79.1	78.4	77.5	96.8	95.6	100.0
3%/5mm	97.3	84.4	83.1	82.1	98.2	97.8	100.0

Gamma Pass Ratio [%] for Delta^4						
Patient 11	TH 5	TH 10	TH 20	TH 70	TH 90	TH 95
3%/2mm	90.2	90.4	91.0	93.2	93.8	93.5
3%/3mm	93.8	94.2	94.7	96.3	95.7	95.5
3%/4mm	95.3	95.9	96.3	97.6	97.3	97.2
3%/5mm	95.6	96.1	96.4	97.6	97.3	97.2

Gamma Pass Ratio [%] for MapCHECK 2							
Patient 12	CS	TH 5	TH 10	TH 20	TH 70	TH 90	TH 95
3%/2mm	91.9	48.1	49.4	61.4	87.3	90.0	94.3
3%/3mm	96.0	60.8	61.9	74.8	92.9	93.3	94.3
3%/4mm	97.6	69.7	70.9	83.2	94.9	95.0	97.1
3%/5mm	98.4	77.7	78.5	88.9	97.2	96.7	97.1

Gamma Pass Ratio [%] for Delta^4						
Patient 12	TH 5	TH 10	TH 20	TH 70	TH 90	TH 95
3%/2mm	92.2	92.2	92.2	93.3	94.1	94.1
3%/3mm	96.4	96.4	96.3	97.4	97.4	97.3
3%/4mm	97.7	97.7	97.6	98.5	98.3	98.2
3%/5mm	98.7	98.7	98.7	98.9	98.9	98.8

Gamma Pass Ratio [%] for MapCHECK 2							
Patient 13	CS	TH 5	TH 10	TH 20	TH 70	TH 90	TH 95
3%/2mm	77.7	59.3	59.8	61.7	86.6	95.8	98.0
3%/3mm	86.2	71.3	69.7	68.6	89.9	97.0	99.0
3%/4mm	91.9	80.6	78.5	76.7	95.0	97.0	99.0
3%/5mm	95.2	86.3	84.5	82.8	98.3	98.2	99.0

Gamma Pass Ratio [%] for Delta^4						
Patient 13	TH 5	TH 10	TH 20	TH 70	TH 90	TH 95
3%/2mm	86.1	83.9	82.8	82.3	82.8	82.5
3%/3mm	92.5	91.3	90.6	89.1	87.6	86.5
3%/4mm	94.8	93.9	93.4	91.5	90.1	89.3
3%/5mm	96.0	95.3	94.9	92.7	91.5	90.8

Gamma Pass Ratio [%] for MapCHECK 2							
Patient 14	CS	TH 5	TH 10	TH 20	TH 70	TH 90	TH 95
3%/2mm	93.5	68.6	68.6	68.2	87.7	92.6	93.1
3%/3mm	97.7	78.8	78.1	77.6	96.2	97.0	98.3
3%/4mm	98.6	84.5	83.6	82.9	98.7	98.5	98.3
3%/5mm	98.9	88.0	87.1	86.6	99.1	98.5	98.3

Gamma Pass Ratio [%] for Delta^4						
Patient 14	TH 5	TH 10	TH 20	TH 70	TH 90	TH 95
3%/2mm	86.2	86.2	86.2	86.5	86.9	87.3
3%/3mm	91.4	91.4	91.4	91.3	91.9	92.1
3%/4mm	93.5	93.5	93.5	93.5	93.9	94.2
3%/5mm	94.8	94.8	94.8	94.9	94.8	95.1

Gamma Pass Ratio [%] for MapCHECK 2							
Patient 15	CS	TH 5	TH 10	TH 20	TH 70	TH 90	TH 95
3%/2mm	75.9	52.9	53.8	57.9	86.4	93.2	95.1
3%/3mm	87.4	67.8	68.4	71.7	93.4	97.5	97.6
3%/4mm	92.8	77.6	77.7	80.5	97.7	98.3	97.6
3%/5mm	95.4	84.0	82.9	85.9	99.2	99.2	100.0

Gamma Pass Ratio [%] for Delta^4						
Patient 15	TH 5	TH 10	TH 20	TH 70	TH 90	TH 95
3%/2mm	88.0	88.0	88.0	87.7	87.7	87.8
3%/3mm	91.8	91.8	91.7	91.9	91.8	92.1
3%/4mm	93.6	93.7	93.7	94.0	94.2	94.4
3%/5mm	95.1	95.2	95.2	95.3	95.3	95.4

Gamma Pass Ratio [%] for MapCHECK 2							
Patient 16	CS	TH 5	TH 10	TH 20	TH 70	TH 90	TH 95
3%/2mm	88.2	65.7	67.0	66.5	94.2	98.1	98.2
3%/3mm	92.6	76.5	76.9	75.9	96.9	98.1	98.2
3%/4mm	95.8	83.9	83.8	83.0	98.7	98.1	98.2
3%/5mm	96.7	88.8	88.2	97.6	99.0	98.4	98.2

Gamma Pass Ratio [%] for Delta^4						
Patient 16	TH 5	TH 10	TH 20	TH 70	TH 90	TH 95
3%/2mm	93.5	93.5	93.5	93.6	92.8	91.3
3%/3mm	96.4	96.4	96.3	96.1	95.3	94.3
3%/4mm	97.3	97.3	97.3	97.0	96.4	95.6
3%/5mm	98.2	98.2	98.2	98.0	97.5	97.0

Gamma Pass Ratio [%] for MapCHECK 2							
Patient 17	CS	TH 5	TH 10	TH 20	TH 70	TH 90	TH 95
3%/2mm	74.9	56.4	56.4	60.7	85.0	-	-
3%/3mm	78.1	63.5	63.5	68.3	88.7	-	-
3%/4mm	80.7	68.4	68.4	73.3	91.1	-	-
3%/5mm	82.4	72.6	72.6	77.6	93.0	-	-

Gamma Pass Ratio [%] for Delta^4						
Patient 17	TH 5	TH 10	TH 20	TH 70	TH 90	TH 95
3%/2mm	90.5	90.5	90.5	91.5	91.3	91.9
3%/3mm	93.7	93.7	93.7	94.0	93.5	94.2
3%/4mm	94.9	94.9	94.9	95.0	94.4	95.0
3%/5mm	95.4	95.4	95.4	95.6	94.9	95.6

Gamma Pass Ratio [%] for MapCHECK 2							
Patient 18	CS	TH 5	TH 10	TH 20	TH 70	TH 90	TH 95
3%/2mm	83.9	40.6	41.6	50.7	91.3	98.8	97.7
3%/3mm	92.9	56.0	54.6	62.8	98.4	98.8	97.7
3%/4mm	95.9	64.1	60.9	67.8	99.2	98.8	97.7
3%/5mm	97.5	71.9	68.4	72.2	99.2	98.8	97.7

Gamma Pass Ratio [%] for Delta^4						
Patient 18	TH 5	TH 10	TH 20	TH 70	TH 90	TH 95
3%/2mm	88.7	87.9	87.1	89.6	93.2	93.2
3%/3mm	94.3	94.2	94	95.5	96.1	95.7
3%/4mm	96.2	96	95.9	96.3	96.4	96.1
3%/5mm	97.3	97.3	97.3	97.3	97.4	97.2

Gamma Pass Ratio [%] for MapCHECK 2							
Patient 19	CS	TH 5	TH 10	TH 20	TH 70	TH 90	TH 95
3%/2mm	95.9	67.7	67.3	71.4	69.8	53.8	58.0
3%/3mm	99.4	81.0	80.7	84.5	80.4	68.9	72.0
3%/4mm	99.6	87.3	87.3	90.4	87.1	76.4	82.0
3%/5mm	99.6	90.7	90.6	93.5	89.1	79.2	86.0

Gamma Pass Ratio [%] for Delta^4						
Patient 19	TH 5	TH 10	TH 20	TH 70	TH 90	TH 95
3%/2mm	87.6	87.6	87.4	87.5	86.4	86.0
3%/3mm	92.4	92.4	92.4	93.1	92.8	92.3
3%/4mm	94.6	94.6	94.6	95.2	95.0	94.7
3%/5mm	96.1	96.1	96.1	97.0	96.7	96.5

Gamma Pass Ratio [%] for MapCHECK 2							
Patient 20	CS	TH 5	TH 10	TH 20	TH 70	TH 90	TH 95
3%/2mm	95.4	67.0	66.4	66.9	94.2	96.4	100.0
3%/3mm	98.1	77.9	76.1	76.4	98.1	98.7	100.0
3%/4mm	98.4	84.6	82.9	83.6	99.5	99.6	100.0
3%/5mm	99.2	88.8	87.5	88.7	99.7	99.6	100.0

Gamma Pass Ratio [%] for Delta^4						
Patient 20	TH 5	TH 10	TH 20	TH 70	TH 90	TH 95
3%/2mm	84.3	84.3	83.9	79.6	77.9	77.6
3%/3mm	90.3	90.3	90.1	87.2	86.5	85.8
3%/4mm	93.2	93.2	93.1	90.5	88.5	87.7
3%/5mm	94.6	94.6	94.6	92.3	90.7	89.9

Gamma Pass Ratio [%] for MapCHECK 2							
Patient 21	CS	TH 5	TH 10	TH 20	TH 70	TH 90	TH 95
3%/2mm	85.3	62.9	64.7	68.1	82.4	84.5	93.5
3%/3mm	95.7	82.0	83.0	84.9	93.8	92.7	97.8
3%/4mm	97.9	89.4	89.5	90.9	96.9	96.4	100.0
3%/5mm	98.6	94.0	93.6	94.6	98.3	97.3	100.0

Gamma Pass Ratio [%] for Delta^4						
Patient 21	TH 5	TH 10	TH 20	TH 70	TH 90	TH 95
3%/2mm	79.0	78.9	78.8	77.8	77.0	76.9
3%/3mm	88.3	88.3	88.3	88.0	87.7	87.8
3%/4mm	92.3	92.3	92.4	92.3	92.2	92.2
3%/5mm	94.4	94.4	94.5	94.5	94.5	94.6

Gamma Pass Ratio [%] for MapCHECK 2							
Patient 22	CS	TH 5	TH 10	TH 20	TH 70	TH 90	TH 95
3%/2mm	85.5	53.1	53.9	56.4	94.8	97.0	95.2
3%/3mm	92.8	65.8	64.8	66.2	98.4	97.8	95.2
3%/4mm	96.0	73.3	70.4	69.7	98.4	97.8	95.2
3%/5mm	96.7	78.7	75.6	73.5	98.4	97.8	95.2

Gamma Pass Ratio [%] for Delta^4						
Patient 22	TH 5	TH 10	TH 20	TH 70	TH 90	TH 95
3%/2mm	97.5	97.4	97.6	97.3	96.5	96.5
3%/3mm	98.0	97.9	98.2	98.5	98.0	97.7
3%/4mm	98.4	98.3	98.7	99.2	99.0	98.8
3%/5mm	98.7	98.7	99.0	99.2	99.0	98.8

Gamma Pass Ratio [%] for MapCHECK 2							
Patient 23	CS	TH 5	TH 10	TH 20	TH 70	TH 90	TH 95
3%/2mm		59.8	58.3	63.3	97.4	97.9	100.0
3%/3mm		72.5	69.9	73.1	98.9	98.6	100.0
3%/4mm		80.4	77.6	80.6	99.5	99.3	100.0
3%/5mm		84.5	82.1	84.8	99.5	99.3	100.0

Gamma Pass Ratio [%] for Delta^4						
Patient 23	TH 5	TH 10	TH 20	TH 70	TH 90	TH 95
3%/2mm	89.8	89.3	88.9	87.7	87.0	86.4
3%/3mm	93.9	93.8	93.4	90.4	89.2	88.6
3%/4mm	94.7	94.6	94.2	90.8	89.7	89.1
3%/5mm	95.2	95.0	94.6	91.3	90.2	89.6

Gamma Pass Ratio [%] for MapCHECK 2							
Patient 24	**CS**	**TH 5**	**TH 10**	**TH 20**	**TH 70**	**TH 90**	**TH 95**
3%/2mm		56.8	58.7	71.0	77.0	75.0	72.2
3%/3mm		78.0	78.6	91.5	92.0	86.1	88.9
3%/4mm		88.0	88.0	95.8	97.7	94.4	94.4
3%/5mm		91.1	90.6	97.3	98.9	97.2	100.0

Gamma Pass Ratio [%] for Delta^4						
Patient 24	**TH 5**	**TH 10**	**TH 20**	**TH 70**	**TH 90**	**TH 95**
3%/2mm	74.2	75.3	80.5	83.0	81.6	81.0
3%/3mm	81.4	83.8	90.5	93.7	93.2	93.0
3%/4mm	83.7	86.6	93.1	96.0	95.7	95.5
3%/5mm	85.1	88.2	94.5	96.9	96.6	96.5

Gamma Pass Ratio [%] for MapCHECK 2							
Patient 25	**CS**	**TH 5**	**TH 10**	**TH 20**	**TH 70**	**TH 90**	**TH 95**
3%/2mm		69.9	70.7	70.7	85.3	95.1	96.5
3%/3mm		85.2	83.7	82.9	92.9	97.5	96.5
3%/4mm		92.5	91.6	91.0	97.8	97.5	96.5
3%/5mm		95.2	94.5	94.0	98.7	98.4	96.5

Gamma Pass Ratio [%] for Delta^4						
Patient 25	**TH 5**	**TH 10**	**TH 20**	**TH 70**	**TH 90**	**TH 95**
3%/2mm	90.2	89.4	88.7	85.5	83.5	81.4
3%/3mm	93.2	92.8	92.3	89.7	86.7	83.8
3%/4mm	93.9	93.6	93.1	90.4	87.6	84.7
3%/5mm	94.5	94.4	93.9	91.5	89.1	86.5

B Statistics

For all the DQA plans for the different gamma criteria and for the different threshold levels, some basic statistics for the gamma pass rate were compiled. The parameters looked upon were the average value, the standard deviation, the minimum value, the maximum value and the median value. *E.g.* the average percentage of passed data points for all the 25 plans for the MapCHECK 2, TH 5, and the gamma criteria 3%/2mm was 58.2 %. The standard deviation of the pass rates were 8.9 and the plan with the least amount of data points passing the gamma evaluation had a pass rate of 34.3 %. In the same way the plan with most data points passing the gamma evaluation had a pass rate of 69.9 %. The median pass rate were 59.3 %.

3%2mm MapCHECK 2					
	Average	Std.Dev.	Min	Max	Median
TH5	58.2	8.9	34.3	69.9	59.3
TH10	59.0	8.6	36.6	70.7	59.8
TH20	63.8	6.8	50.0	73.1	65.0
TH70	86.0	7.5	69.0	95.9	86.6
TH90	90.2	10.5	53.8	99.0	93.7
TH95	91.4	9.8	58.0	100.0	93.7

3%2mm Delta^4					
	Average	Std.Dev.	Min	Max	Median
TH5	88.2	5.1	74.2	97.5	88.7
TH10	88.0	5.0	75.3	97.4	88.7
TH20	88.0	4.6	78.8	97.6	88.6
TH70	87.1	6.0	77.6	97.6	87.7
TH90	87.6	6.5	76.6	98.5	87.4
TH95	87.6	6.8	76.7	99.0	88.5

3%3mm MapCHECK 2					
	Average	Std.Dev.	Min	Max	Median
TH5	70.7	8.9	50.2	85.2	71.9
TH10	70.9	8.1	54.6	83.7	71.4
TH20	75.6	7.8	58.5	91.5	76.1
TH70	93.7	4.8	80.4	99.1	93.8
TH90	94.4	7.3	68.9	100.0	97.1
TH95	94.9	6.2	72.0	100.0	96.6
CS	93.2	4.8	78.1	99.4	93.1

3%3mm Delta^4					
	Average	Std.Dev.	Min	Max	Median
TH5	92.8	4.0	81.4	99.4	93.1
TH10	92.8	3.7	83.8	99.3	92.9
TH20	93.1	3.3	86.8	99.1	93.0
TH70	93.2	4.3	84.7	99.8	93.7
TH90	92.6	5.0	82.5	99.7	93.0
TH95	92.1	5.3	81.4	99.7	92.7

3%4mm MapCHECK 2					
	Average	Std.Dev.	Min	Max	Median
TH5	78.3	8.7	59.0	92.5	80.4
TH10	78.2	8.2	60.9	91.6	78.5
TH20	82.2	7.8	65.4	95.8	83.0
TH70	96.7	3.3	87.1	100.0	97.7
TH90	96.4	5.0	76.4	100.0	98.0
TH95	96.9	3.8	82.0	100.0	97.7

3%4mm Delta^4					
	Average	Std.Dev.	Min	Max	Median
TH5	94.5	3.4	83.7	99.8	94.7
TH10	94.6	3.1	86.6	99.8	94.6
TH20	95.0	2.7	88.6	99.7	94.9
TH70	94.8	3.6	87.8	99.8	95.2
TH90	94.2	4.3	85.8	99.7	95.4
TH95	93.7	4.7	84.7	99.7	95.1

3%4mm MapCHECK 2					
	Average	Std.Dev.	Min	Max	Median
TH5	83.4	7.6	68.8	95.2	84.5
TH10	83.2	7.1	68.4	94.5	84.4
TH20	86.8	7.5	70.6	97.6	88.7
TH70	97.7	2.7	89.1	100.0	98.6
TH90	97.2	4.4	79.2	100.0	98.3
TH95	97.7	3.1	86.0	100.0	98.3

3%4mm Delta^4					
	Average	Std.Dev.	Min	Max	Median
TH5	95.6	3.1	85.1	99.8	95.6
TH10	95.7	2.7	88.2	99.8	95.7
TH20	96.0	2.3	90.2	99.9	96.1
TH70	95.8	3.2	89.5	100.0	96.9
TH90	95.2	3.8	87.5	100.0	96.7
TH95	94.8	4.3	86.4	100.0	96.5

C Delta$^{4®}$ Software

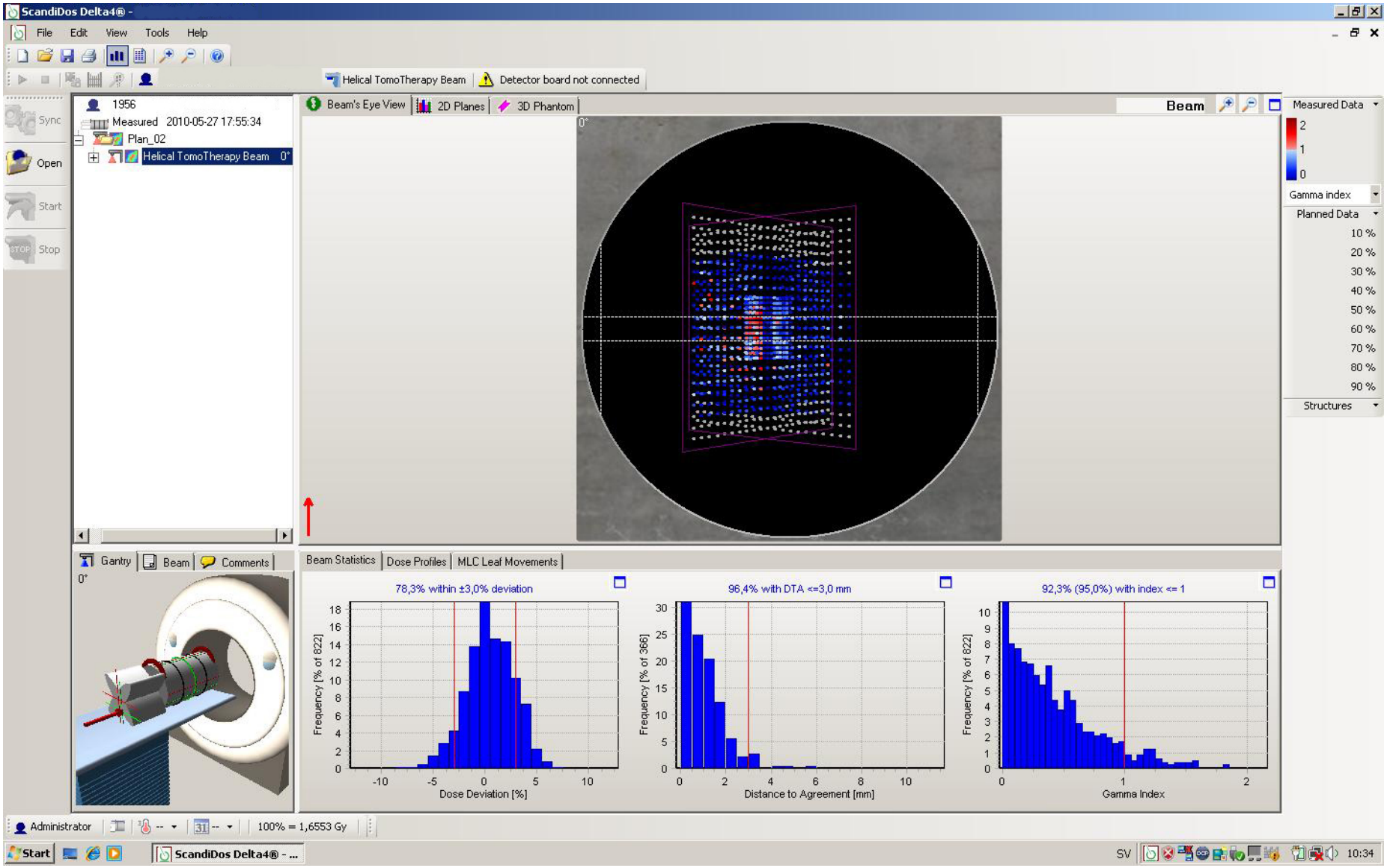

Figure 19: Delta$^{4®}$ Software

D MapCHECK™ Software

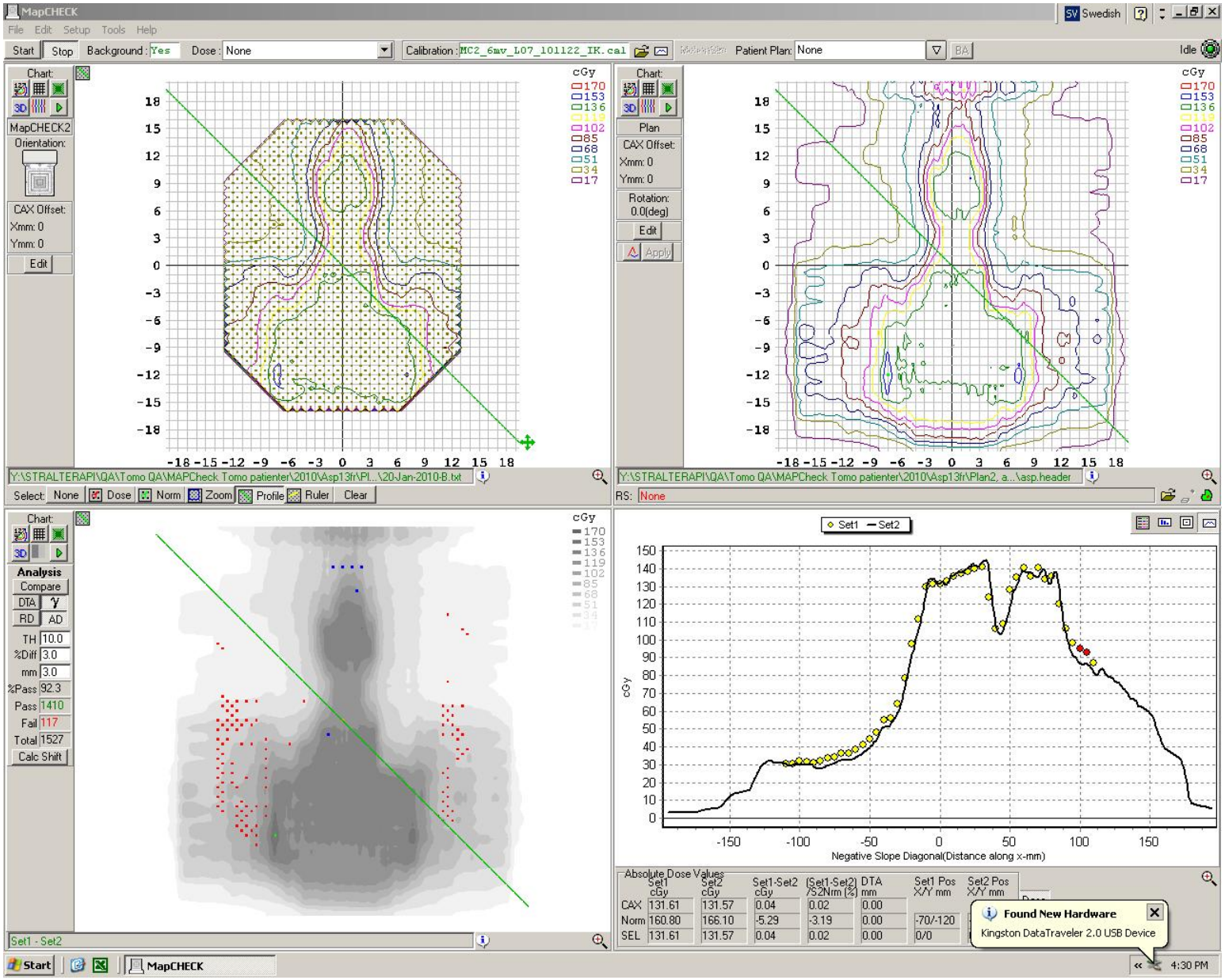

Figure 20: MapCHECK™ Software