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Petersson, Kristoffer; Ceberg, Crister; Engström, Per; Knöös, Tommy

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Beam commissioning and measurements validating the beam model in a new TPS that converts helical tomotherapy plans to step-and-shoot IMRT plans

Kristoffer Petersson,[a)] and Crister Ceberg

Medical Radiation Physics, Clinical Sciences, Lund University, Lund, Sweden

Per Engström, and Tommy Knöös

Medical Radiation Physics, Clinical Sciences, Lund University, Lund, Sweden

and Radiation Physics, Skåne University Hospital, Lund, Sweden

Author of correspondence:
Kristoffer Petersson

Medical Radiation Physics,

Clinical Sciences, Lund,

Skåne University Hospital Lund

Barngatan 2:1

221 85 Lund

Sweden

Kristoffer.petersson@skane.se +46 70 597 39 84

Validation of a New TPS

Abstract

Purpose: A new type of treatment planning system called SharePlan has been studied that enables transfer of treatment plans generated for helical tomotherapy delivery to plans that can be delivered on C-arm linacs. The purpose is to ensure continuous patient treatment during periods of un-scheduled downtime for the TomoTherapy unit, particularly in clinics without a backup unit.

The purpose of this work was to verify that the plans generated in this novel planning system are deliverable and accurate. The work consists primarily of beam commissioning, verification of the beam-model and measurements verifying that generated plans are deliverable with sufficient accuracy.

Methods: The beam commissioning process involves input of general geometric properties of the modeled linac, profiles and depth dose curves for a specific photon nominal energy (6 MV) and the automated modeling of other beam properties. Some manual tuning of the beam model is required. To evaluate its accuracy, the confidence limit concept (J. Venselaar, H. Welleweerd and B. Mijnheer, "Tolerances for the accuracy of photon beam dose calculations of treatment planning systems," Radiother Oncol **60**, 191-201 (2001)) was used which is a method supported by ESTRO. Measurements were conducted with a 2D diode array at the commissioned linac as a final check of the beam model, and to evaluate whether the generated plans were deliverable and accurate.

Results: The comparison and evaluation of calculated data points and measured data according to the method applied confirmed the accuracy of the beam model. The profiles had a confidence limit of 1.1% and the depth dose curves had a confidence limit of 1.7%, both of which were well below the tolerance limit of 2%. Plan specific QC measurements and evaluation verified that different plans generated in the TPS were deliverable with sufficient accuracy at the commissioned linac, as none of the 160 beams for the 20 different plans evaluated had a fraction of approved data points below 90%, the local clinical approval criterion for delivery QA measurements.

Conclusions: This study is a validation of the new TPS as it verifies that generated plans are deliverable at a commissioned linac with adequate accuracy. A thorough investigation of the treatment plan quality will require a separate study. The TPS is proving to be a useful and time-saving complement, especially for clinics having a single unit for helical delivery among its conventional linacs.

I. Introduction

When a single TomoTherapy® Hi·Art® treatment system (TomoTherapy Incorporated, WI, USA) is installed in a clinical environment alongside conventional linear accelerators (linacs), a backup plan for treatment with a conventional linac for every patient being treated on the TomoTherapy unit is desirable. This is a precaution to ensure continuous patient treatment in case of unintended, as well as planned downtime of the TomoTherapy unit. Creating backup Intensity-Modulated Radiation Therapy (IMRT) plans for conventional linear accelerators for every patient is a time consuming task and the resulting plan might differ substantially from the prescribed TomoTherapy plan. In order to simplify the creation of these plans Raysearch Laboratories AB (Stockholm, Sweden) has developed a software solution called "SharePlan™" for TomoTherapy.[1] This software enables transfer of treatment plans generated with the TomoTherapy system to plans deliverable on conventional linacs with step-and-shoot (SS) delivery. The result is a selection of optimal plans with DVHs and dose distributions as similar to the TomoTherapy plan as possible.

Proper beam commissioning and QA procedures are important for the implementation and use of any radiation treatment planning system.[2] The purpose of this work was to verify that the plans generated in this novel treatment planning

system are deliverable and accurate. The work presented here consists of a short description of the software, beam commissioning, verification of the beam model and results from measurements verifying that generated plans are deliverable and accurate.

II. Material and Methods

Though SharePlan got FDA (U.S. Food and Drug Administration) approval on 2009-01-27, it has just recently been released to customers (May 2010). The work presented here was performed with a pre-release evaluation version of the software.

II.A. A Description of the Software

In order to generate conventional IMRT plans in SharePlan, a TomoTherapy plan has to be imported and used as a starting point. RT plan containing dose objectives, RT dose, the structure set and the CT images are consequently exported from the TomoTherapy system and imported into SharePlan. The optimizer in SharePlan utilizes a sequential quadratic programming algorithm.[b)] The dose is calculated using a collapsed cone convolution (CC) algorithm. The optimizer uses objective functions based on the differences between the DVH of the plan under optimization and DVH of a reference plan *i.e.* a TomoTherapy plan. Based on the original plan and additional restrictions *i.e.* linac limitations, number of beams, maximum number of segments, etc. the optimizer tries to find the optimal plan.

The optimizer variables are based on a beam's eye view of the targets. During the optimization phase, the incident fluence is approximated by the relative direct fluence for an uncollimated open field and the approximated modulation taking into account collimator leakage (MLC and/or jaws). A second term in the fluence expression adds the head scatter contribution modeled as a convolution of a head scatter source over the field. The head scatter kernel includes scatter contributions from the flattening filter and other sources. The segmentation tries to reconstruct the individual fluence distribution for each beam as closely as possible by determining the MLC segments with corresponding monitor units. To determine the shape of the MLC segments, the optimized fluence values are constrained to a number of equidistant fluence levels (intensity levels), distributed among the beams in proportion to the maximum fluence values. Each level is decomposed into smaller elements, representing openings for individual leaf pairs. The number of openings created depends on the complexity of the fluence in the level. [b)] The openings are combined to segments, taking into account leaf position requirements *i.e.* leaf position bounds, minimum gaps for opposing and opposing adjacent leaves, leaf overtravel, etc. Identical segments are merged into one segment. The segmentation settings *i.e.* minimum allowed segment area, minimum number of open MLC leaves per segment and minimum number of allowed monitor units per segment are taken into account and segments not fulfilling these requirements are discarded. If the number of generated segments is larger (or much less) than the maximum number of segments allowed, the segmentation process is repeated with an adjusted number of initial fluence levels.

This process is repeated until the number of segments is close to, but not exceeding, the user specified maximum number of segments or until the result can not be improved further. When segments have been determined, they are sorted to minimize the delivery time for each beam by minimizing the needed leaf travelling distance between segments. The monitor units per generated segment are determined with an accurate fluence calculation, performed for each individual segment. The individual fluence distributions from all segments are used to optimize the segment weights. This is performed to minimize the differences between the desired fluence distributions after optimization and the fluence from the actual segments. More details about the system can be found in the optimization and segmentation manual.[b)]

Modeling of conventional linacs in the beam commissioning part of the software consists of the beam model (which describes the beam from the linac *e.g.* sources, collimators, *etc*), the "fluence engine" (which calculates energy fluence distributions for both electrons and photons from the beam model) and the "dose engine" (which calculates dose).[c)] In the software, the linac beam is assumed to have four sources:[c)]

1. The primary photon source positioned approximately where the target is situated.
2. A secondary photon source positioned approximately at the level of the flattening filter.

3. The primary electron source positioned approximately where the target is situated.
4. An additional secondary electron source modeling electrons originating from the flattening filter position.

The four sources of energy fluence are modeled by 2D Gaussians using weighting parameters describing the distribution widths in the x- and y-direction described by σ_x and σ_y respectively. The position of the sources along the beam axis (z-position) can also be adjusted/optimized.[c)] The jaws and the MLC positions along the beam, the transmission through them and their offset (the amount of misalignment of the MLC with respect to the radiologic leaf position and the geometric leaf position) are taken into account in the model. Additional MLC parameters that are taken into account in the model are "tongue and groove" and "leaf tip width".[c)] The "dose engine" uses a collapsed cone convolution (CC) type of algorithm based on the work by Ahnesjö.[3] The method considers the primary photon transport and the transport of secondary particles set in motion via primary photon interaction, separately. First a trace of the primary beam through the patient and a calculation of the spatial distribution of energy deposited in the patient, taking into account inhomogeneities, is performed by calculation of the total energy released per unit mass (TERMA). The transport and energy absorption of the secondary particles are taken into account by superposition of point kernels calculated in advance by the use of Monte Carlo technique.[3] The CC algorithm has proven to be a very accurate way of calculating dose in different media.[4-8]

II.B. Beam Commissioning

The beam commissioning of a 6 MV beam (Elekta Synergy® linac by Elekta AB, Stockholm, Sweden) started with the creation of a treatment unit model based on an existing template model in the machine database. The input of configuration parameters of the linac being modeled was done before importing the algorithm input data needed.[2] The imported profiles and depth dose curves were measured for the following field sizes 3x3 cm^2, 5x5 cm^2, 10x10 cm^2, 15x15 cm^2 and 20x20 cm^2. Relative output factors in water for every imported field size and an absolute dose measurement at 10 cm depth for a 10x10 cm^2 field were also part of the required algorithm input data. All measurements were performed at a source to surface distance (SSD) of 90 cm. Profiles and output factors were measured at a depth of 10 cm. The distances from collimators and flattening filter to the target were corrected from the template to the parameters of the specific linac being commissioned. After the input of all the above data and the input of chosen computation setting, the auto-modeling process could begin. This process involved modeling of the photon energy spectrum, electron energy spectrum, off axis softening, beam profile corrections, output factor corrections and both primary and scatter (secondary) sources widths and weights. Some of the auto-modeling steps had to be repeated once or twice. Some manual tuning of the model was required after the auto-modeling in order to obtain a good fit between measured curves and curves calculated from the model. When these first steps of the beam commissioning were done the modeling of the MLC parameters

(mentioned above) could begin. To model the MLC parameters three different IMRT plans (two head and neck cases and one prostate case, see Figure 1) were generated with the system. The plans were exported, delivered and measured at the commissioned linac with a 2D diode array (MapCheck^{TM}, Sun Nuclear Corporation, USA). CT images and a structure set of a large water tank phantom were imported into the built-in QA tool in the TPS. The three different plans were setup on the phantom and calculations were performed for each single beam and for a composite of all beams for each plan. Calculated dose distributions (DD) were imported to the diode array software and compared with measured distributions. The MLC parameters in the model were then optimized in an iterative way by changing one of the parameters. Though not all MLC leaves were involved in the delivery of these plans, the MLC parameters were global, affecting the model of all MLC leaves. The QA-plans were then recalculated and compared with the measurements. The results were evaluated for the plans to see if the dose distributions were more alike *i.e.* if the model was improved. The parameter was changed again, the plans recalculated, compared and evaluated, and so on until the best fit between measured and calculated distributions was found. This procedure was carried out for all MLC parameters. The whole procedure was then repeated once for all parameters.

II.C. Verification of the beam model

Calculated dose points along depth doses and profiles were compared with the algorithm input data (see Figure 2) and evaluated using the method proposed by

Venselaar *et al*.,[9] as recommended by ESTRO.[10] Relative dose differences were calculated locally, except for the low dose regions (the regions outside the outer penumbra regions of the profile) of a field, where dose differences were calculated relative to the dose on the central axis at the same depth. For the regions with a large dose gradient (>3% / mm), *i.e.* the beam fringes (the regions of the profile between the 50% and 90% dose relative to the maximum of the profile), the outer penumbra (the regions of the profile below the 50% dose relative to the maximum with a large dose gradient), and the build-up part (the part of the depth dose curve between the phantom surface and the depth of the 90% isodose surface), differences in position of isodose lines were calculated (distance to agreement (DTA)). The mean and standard deviations were calculated together with the confidence limits. The confidence limit (Δ) was defined by Venselaar *et al*. as:[9]

$$\Delta = |\text{Mean Deviation}| + 1.5 \times \text{Standard Deviation} \qquad (1)$$

The calculated confidence limits for different regions of the fields were compared to the tolerance levels set by Venselaar *et al*. (see Table 1).[9]

II.D. Clinical Measurements

Twenty clinical plans from the TomoTherapy unit were transferred to SharePlan and converted to plans deliverable with SS IMRT. These plans cover treatment for

various cancers in various anatomical regions (see Table 2). The plans were exported, delivered and measured with the 2D diode array on the commissioned treatment unit. The plan restrictions used for the conversion of the plans are shown in Table 3. Seven-beam plans (minimum number of beams allowed by SharePlan) were generated for half of the cases with plan restrictions similar to those used for our clinical IMRT plans. Though seven beams are used for generating IMRT plans at our clinic nine beams are commonly used for generating IMRT plans at other departments. Therefore nine-beam plans were generated for half of the cases with the same plan restrictions except for a maximum allowed number of segments which was increased to 150 (see Table 2). QA-plans were generated and calculated in the new TPS. The resulting dose distributions were exported to the 2D diode array software for comparison with measurements. To assure the quality of the dose calculations in the new TPS, the plans were exported via DICOM RT and recalculated on the same QA-phantom in our clinical TPS (Oncentra MasterPlan by Nucletron B.V., Veenendaal, The Netherlands) using the Collapsed Cone (CC) algorithm as well as the Pencil Beam (PB) algorithm, which is used clinically. Both algorithms work well for calculating doses in large homogenous water phantoms[6, 11] as the one used for calculation of QA-plans. The CC algorithm is used since SharePlan uses a CC algorithm and the PB algorithm is used since the CC algorithm has not been verified for clinical use at our department. These calculated dose distributions were also compared with measured distributions. The evaluation involved using two dimensional γ-analysis [12-14] of a single transverse slice with the criteria set to 3% and 3 mm and with a threshold level of 10% (data points with a dose lower

than this threshold were excluded from the comparison). The analysis was performed for each beam as well as for the pooled data (composites) for all the beams in a plan. The evaluation of the measurement was treated as a final check of the beam model for the commissioned linac as well as an End-to-End test as it involved plans passing each step of the chain, which a plan generated in SharePlan would have to go through in a clinical setting.

III. Results

III.A. Verification of the beam model

The comparison of the calculated data and the algorithm input data using the methods proposed by Venselaar *et al.*[9] confirmed the accuracy of the beam-model. In general, the calculated confidence limits were well within the tolerance levels , set by Venselaar *et al.*,[9] see Table 1. For the profiles, the calculated confidence limit (not including the beam fringe, the outer penumbra or the low dose regions) was 1.1%, which was well within the tolerance level of 2.0%. The confidence limit in the beam fringe regions was 2.0%, equal to the tolerance level. The confidence limit for the outer penumbra regions was 2.7 mm, which was outside the tolerance level of 2.0 mm. The confidence limit for the low dose regions was 1.7%, which was well within the tolerance level of 3%. For depth dose curves (not including the build up region) the confidence limit was1.7%, also within the tolerance level of 2.0%. For the build up part of the depth dose curves

the confidence limit was 1.2 mm, well within the tolerance level of 2.0 mm, see Table 1.

III.B. Clinical Measurements

The analysis of the measurements performed for the twenty clinical cases showed that plans converted to SS IMRT plans (for treatment in various anatomical regions) were deliverable with high accuracy at the commissioned linac. The results from the gamma analysis are shown in Figure 3 (averages of all beams per plan) and Figure 4 (composites of each plan), in (a) for the seven-beam plans, and in (b) for the nine-beam plans.

The fractions of approved data points per beam ranged from 95.2% to 100% with a mean of 98.8 % for the seven-beam plans, and from 90.9% to 100% with a mean of 97.5% for the nine-beam plans (see Figures 3(a) and 3(b)). For the seven-beam plans recalculated in the clinical TPS with the PB and the CC algorithms, the fractions of approved data points per beam ranged from 93.5% to 100% with a mean of 98.7% and from 94.1% to 100% with a mean of 99.2% respectively, which is about the same as for plans calculated in SharePlan. For the nine-beam plans, the results ranged from 87.0% to 100% with a mean of 95.9% and from 89.9 to 100% with a mean of 97.6% respectively. For the nine-beam plans the results were similar for plans calculated in SharePlan and in the clinical TPS with the CC algorithm, but somewhat lower for plans calculated with the PB algorithm

(see Figures 3(a) and 3(b)). The fractions of approved data points for the composites (pooled data from all beams in a plan) of the seven-beam plans ranged from 96.2% to 100% with a mean of 99.3% and from 95.3% to 100% with a mean of 98.0% for the nine-beam plans calculated in SharePlan (see Figures 4(a) and 4(b)). The results for the composites of the plans calculated in the clinical TPS with the PB and the CC algorithms ranged from 97.4% to 100% with a mean of 99.4% and from 98.3% to 100% with a mean of 99.6%, respectively, for the seven-beam plans, and ranged from 93.7% to 100% with a mean of 97.3% and from 95.5 to 100% with mean of 98.3%, respectively, for the nine-beam plans (see Figures 4(a) and 4(b)). The results show that seven-beam plans in general have a higher fraction of approved data points than nine-beam plans. The results also show less variation for seven-beam plans than nine-beam plans (see Figures 3 and 4). Other trends that are visible in the results are that the CC algorithm in the clinical TPS has in general a somewhat higher fraction of approved data points than the other algorithms, which is especially visible for the seven-beam plans (see Figures 3(a) and 3(b)). For the nine-beam plans the PB algorithm in the clinical TPS has in general a somewhat lower fraction of approved data points than the other algorithms (see Figures 4(a) and 4(b)). None of the 160 beams evaluated had a fraction of approved data points below 90% (the local clinical approval criteria used for delivery QA measurements), neither when compared to calculations performed in SharePlan, nor when compared to calculations performed in the clinical TPS with the CC algorithm. Three beams had a fraction of approved data points of about 87% when compared to calculations performed in the clinical TPS with the PB algorithm, see Figure 5. These results are just

below the local clinical criteria for delivery QA approval. There are numerous plausible reasons for these results. The dose distribution maps used for the γ-analysis (Figure 5) show that some of the failed data points are in low dose areas. The results from the measurements were considered as a successful final check of the SharePlan beam model, as well as a completed End-to-End test.

VI. Discussion

The commissioning resulted in a beam-model which could produce output data which were in good agreement with the algorithm input data, as can be seen in Figure 2. In most cases the calculated confidence limits were well within the adopted tolerance levels, which indicate that the beam model accurately describes the beam from the commissioned linac. However, the model does not depict the dose in the outer penumbra regions and the beam fringe regions as well as in the other regions. The calculated confidence interval for the outer penumbra region was outside the tolerance level. An improvement of the auto-modeling of these regions would be useful in later versions of the program.

The measurements and evaluations of the 20 different plans confirmed the accuracy of both the beam-model and the in-patient dose engine for calculations of different treatment plans. The plan calculations were in good agreement with the 2D diode array measurements. The results for the different TPSs and

algorithms used were similar for each of the different plans measured. This indicates that the results depend more on the linacs capability to deliver the plan rather than the TPSs ability to calculate the dose. The validation performed in this study is somewhat limited to the plan restrictions used but should be valid also for more strict plan restrictions (lower maximum number of segments allowed, higher minimum MU/segment, larger minimum segment area, etc.). Complementary measurements might be warranted to verify plans generated with looser plan restrictions.

The verification has only been performed for step-and-shoot IMRT using 6 MV for an Elekta Synergy® linac. The new software's capability to produce accurate and deliverable plans should also be investigated for other modalities, *i.e.* dMLC and modulated arc therapy *e.g.* VMAT and Rapid Arc if and when these modalities become available in the software. How the overall planning time is affected and how the software should be introduced into the clinical workflow is still to be determined, but as the planning is mostly automated, the overall planning time should not be too affected which should facilitate the introduction of SharePlan into the clinical workflow.

A preliminary report has been presented[15] regarding the quality of the plans generated. In the study the quality of plans generated for three head and necks cases were compared to the quality of plans generated with the TomoTherapy

system as well as plans generated with Oncentra MasterPlan, with the use of optimal fronts. The results from the study showed that plans generated in SharePlan are of high quality well within clinical acceptability. Regardless of this preliminary report, a more complete investigation is required, especially concerning how much a plan deteriorates when being converted from a TomoTherapy plan to a linac deliverable IMRT plan, and also how the generated plans compare to IMRT plans generated in a conventional way *i.e.* with TPS directly designed for this.

Plan specific QA measurements and evaluation of IMRT plans with the 2D diode array system is widely used and well established.[16] The uncertainties for these measurements are small (1 SD of about 0.06% to 0.15%) as have been documented in the literature.[17]

V. Conclusion

This study shows that plans generated in the new TPS SharePlan™ are deliverable with sufficient accuracy for the commissioned linac and for the plan restrictions used in this work. The results presented in this study should also be valid for stricter plan restrictions. To further validate the software, measurements similar to the ones performed in this study need to be performed for different types of linac, and for other modalities than the ones investigated here. SharePlan could prove to

be an important complement for clinics with both TomoTherapy units and C-arm linacs and especially for clinics with only a single TomoTherapy unit, as they are more affected by its down time.

[a)]Author to whom correspondence should be addressed. Electronic mail: kristoffer.petersson@skane.se

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Table 1: Calculated confidence limits and tolerance levels set by Venselaar *et al.* [9]

Regions	Confidence limits	Tolerance levels
Profiles (not including the beam fringe, the outer penumbra or the low dose regions)	1.1%	2%
Beam fringes	2.0 mm	2 mm
Outer penumbra regions	2.7 mm	2 mm
Low dose regions	1.7%	3%
Depth dose curves (not including the build-up part)	1.7%	2%
Build-up part	1.2 mm	2 mm

Table 2: The anatomical sites and number of patients in each category, for the twenty clinical plans used.

Anatomical region	Number of patients
Intracranial	4
Head and neck	7
Thorax	3
Abdomen	3
Pelvis	3

Table 3: Plan restrictions used when generating 7-beam/9-beam IMRT plans in the new TPS.

Plan restrictions	**Settings**
Energy (MV)	6
Delivery technique	SS
Number of beams	7/9
Maximum allowed number of segments	100/150
Minimum number of Monitor Units/segment	3
Minimum segment area (cm^2)	5
Minimum equivalent square (cm^2)	5
Leaf jaw overlap (cm)	0

Figure Legends

Figure 1: Transversal and sagittal slices with ROIs visualized for the three different clinical cases used to model the MLC parameters, a) and b) are the two head and neck cases and c) is the prostate case used.

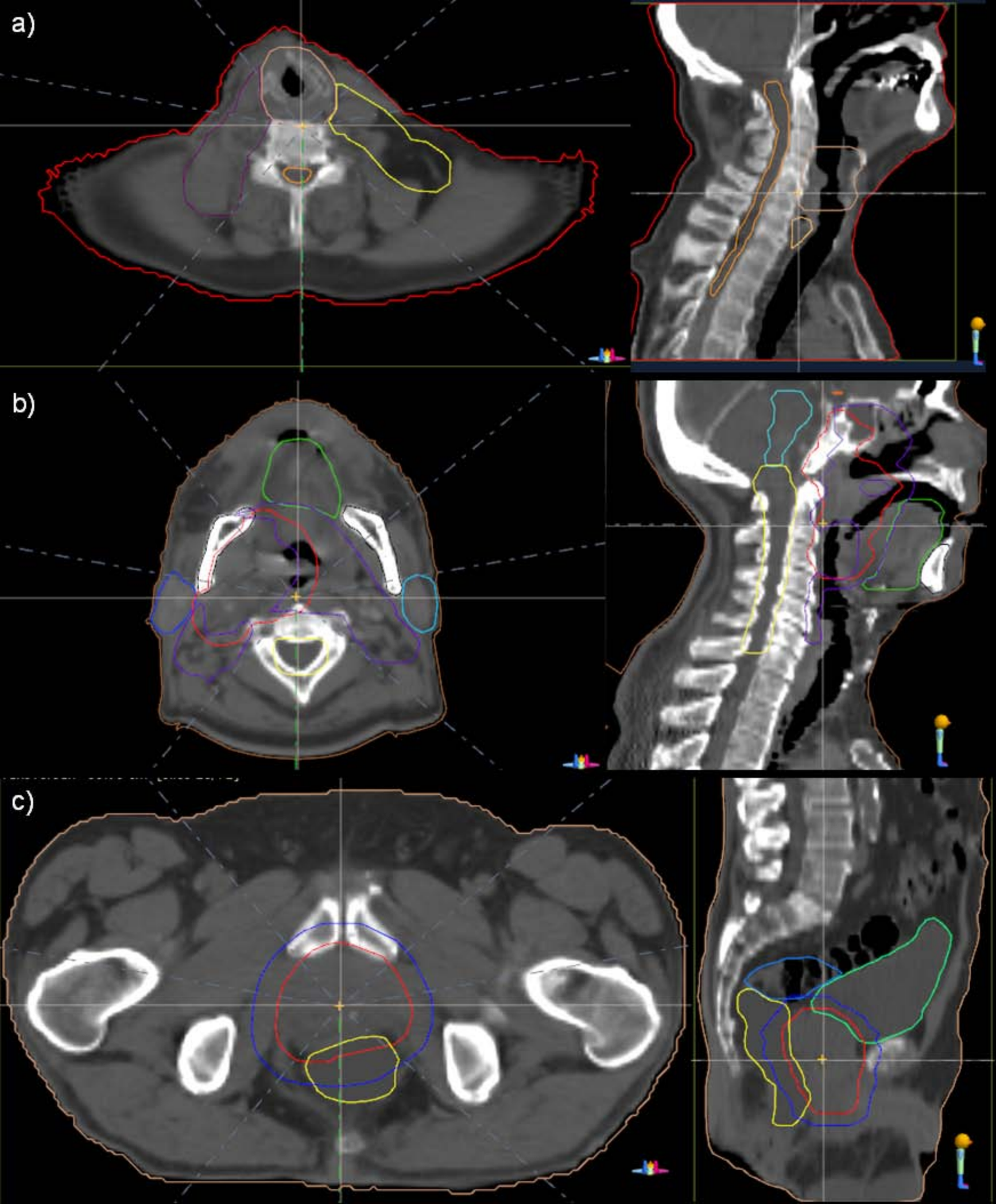
Figure 2: The algorithm input data and dose points calculated from the beam model along the depth doses c) and the profiles a) and b), which were compared as part of the verification of the beam model. The Figure visualizes the curves in the same manner as they are visualized in the beam commissioning part of the software.

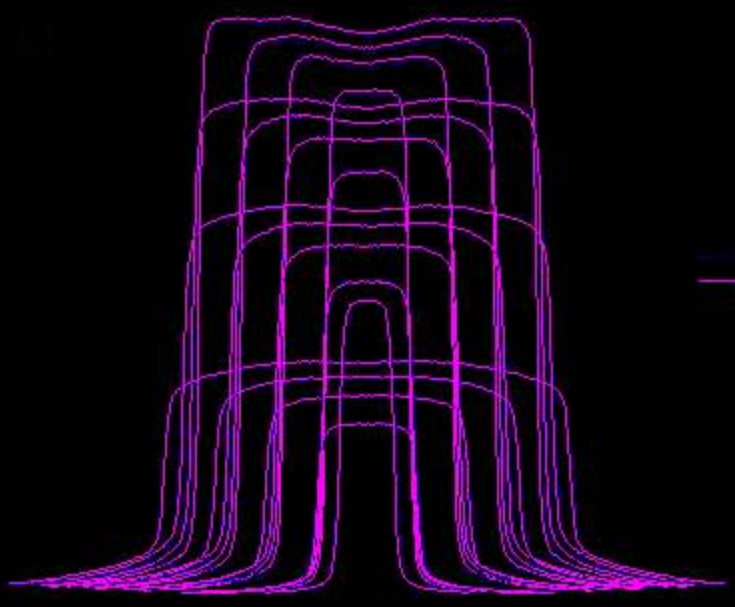
Figure 3: Averages of fractions of approved data points per beam for the ten different (a) seven-beam plans, and (b) nine-beam plans, from γ-analysis of the 2D diode array measurements and the calculations performed in the new TPS (SharePlan) and in our clinical TPS (OMP) (with the PB and the CC algorithms).

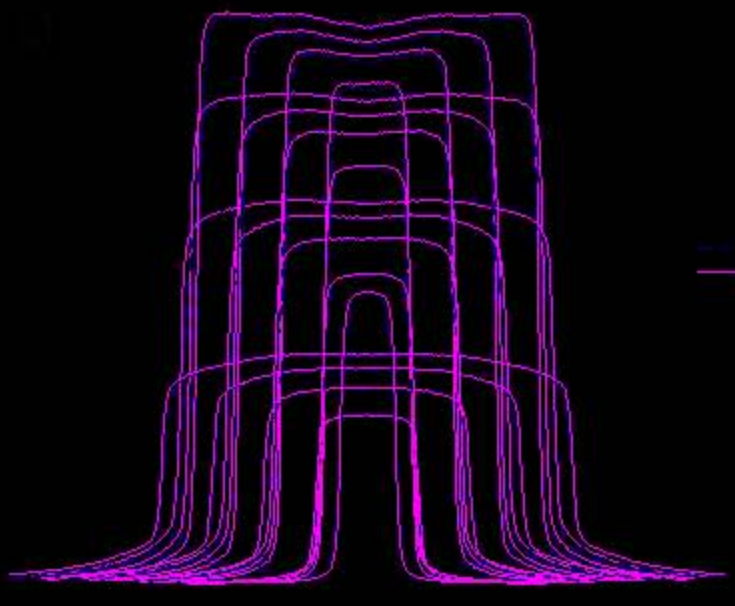
Figure 4: Fractions of approved data points for composites of the ten different (a) seven-beam plans, and (b) nine-beam plans, from γ-analysis of the 2D diode array measurements and the calculations performed in the new TPS (SharePlan) and in our clinical TPS (OMP) (with the PB and the CC algorithms).

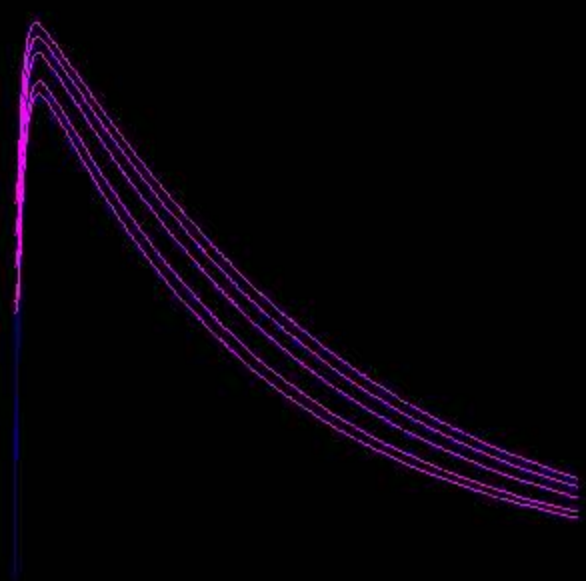
Figure 5: Dose distribution maps used for the γ-analysis of the three beams that had a fraction of approved data points below the local criteria for delivery QA

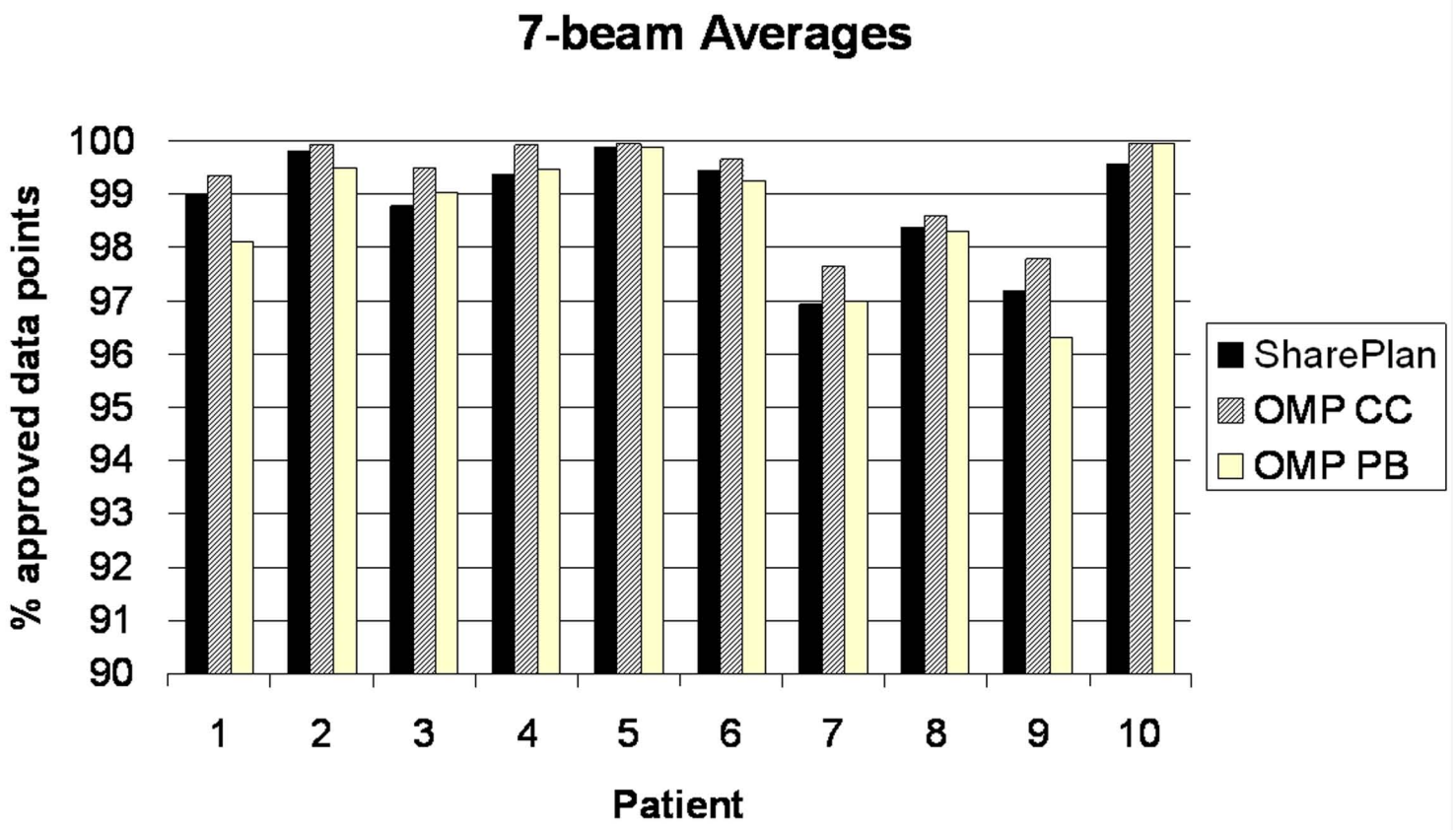
approval. The beams were calculated in the clinical TPS with the PB algorithm. All the failed data points are visualized. Blue means that a too low a dose has been measured, red means too high.

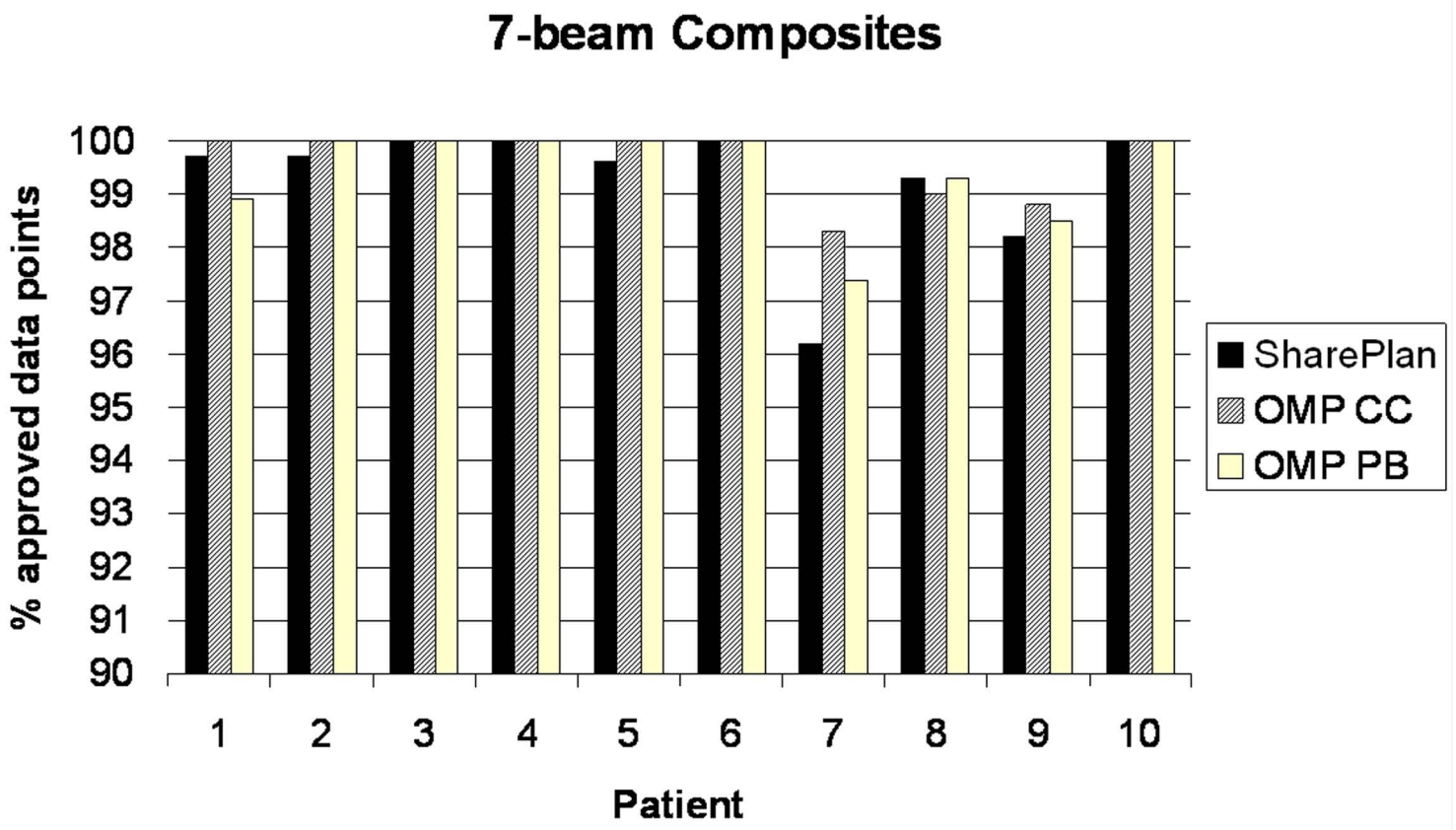

a)
b)
c)

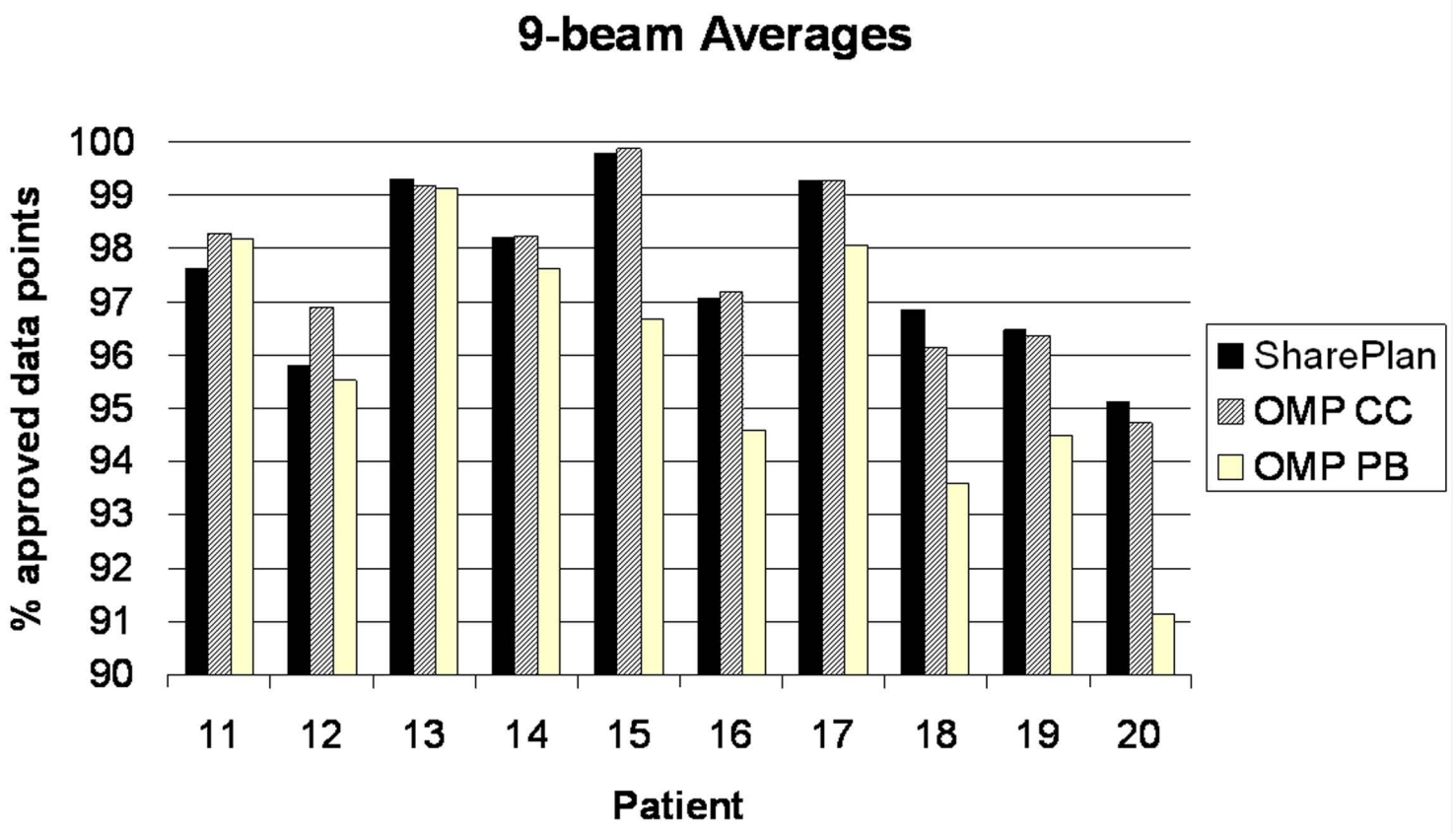


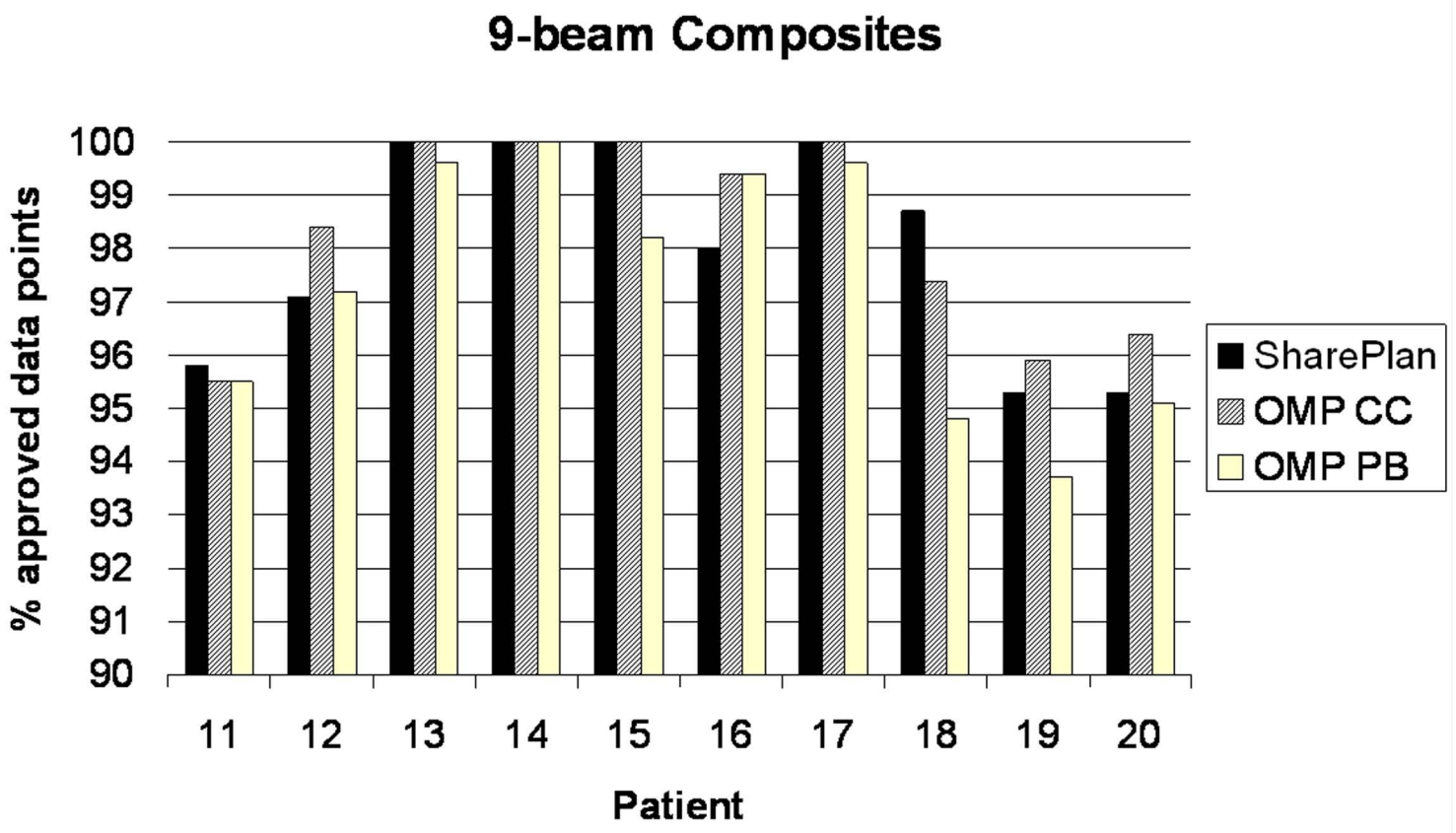





7-beam Averages
% approved data points
100
99
98
97
96
95
94
93
92
91
90
1
2
3
4
5
6
7
8
9
10
Patient
SharePlan
OMP CC
OMP PB


7-beam Composites
% approved data points
100
99
98
97
96
95
94
93
92
91
90
1
2
3
4
5
6
7
8
9
10
Patient
SharePlan
OMP CC
OMP PB


9-beam Averages
% approved data points
100
99
98
97
96
95
94
93
92
91
90
11
12
13
14
15
16
17
18
19
20
Patient
SharePlan
OMP CC
OMP PB


9-beam Composites
% approved data points
100
99
98
97
96
95
94
93
92
91
90
11
12
13
14
15
16
17
18
19
20
Patient
SharePlan
OMP CC
OMP PB

a)

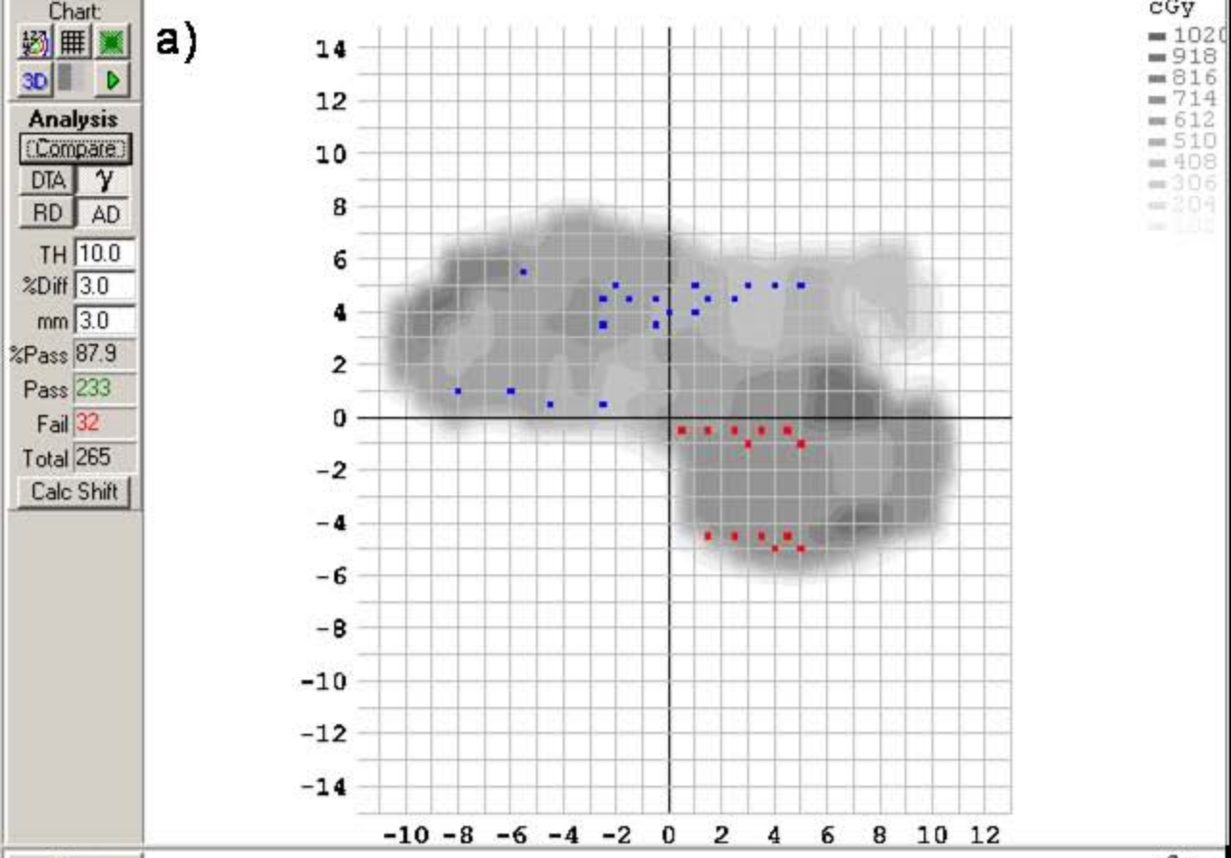


Chart:
3D
Analysis
Compare
DTA
γ
RD
AD
TH 10.0
%Diff 3.0
mm 3.0
%Pass 87.9
Pass 233
Fail 32
Total 265
Calc Shift
cGy
1020
918
816
714
612
510
408
306
204
14 12 10 8 6 4 2 0 -2 -4 -6 -8 -10 -12 -14
-10 -8 -6 -4 -2 0 2 4 6 8 10 12

b)

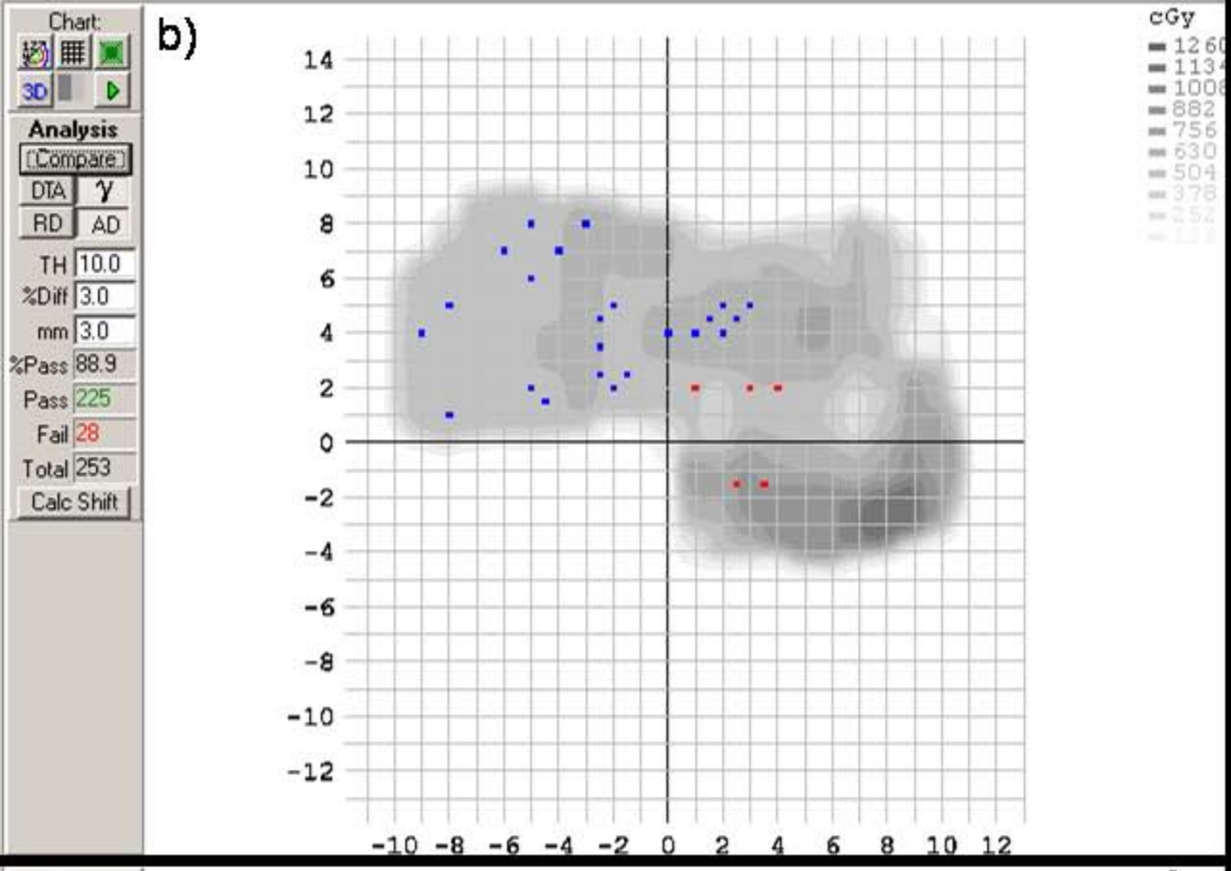


Chart:
3D
Analysis
Compare
DTA
γ
RD
AD
TH 10.0
%Diff 3.0
mm 3.0
%Pass 88.9
Pass 225
Fail 28
Total 253
Calc Shift
cGy
1260
1134
1008
882
756
630
504
378
252
14 12 10 8 6 4 2 0 -2 -4 -6 -8 -10 -12
-10 -8 -6 -4 -2 0 2 4 6 8 10 12

c)

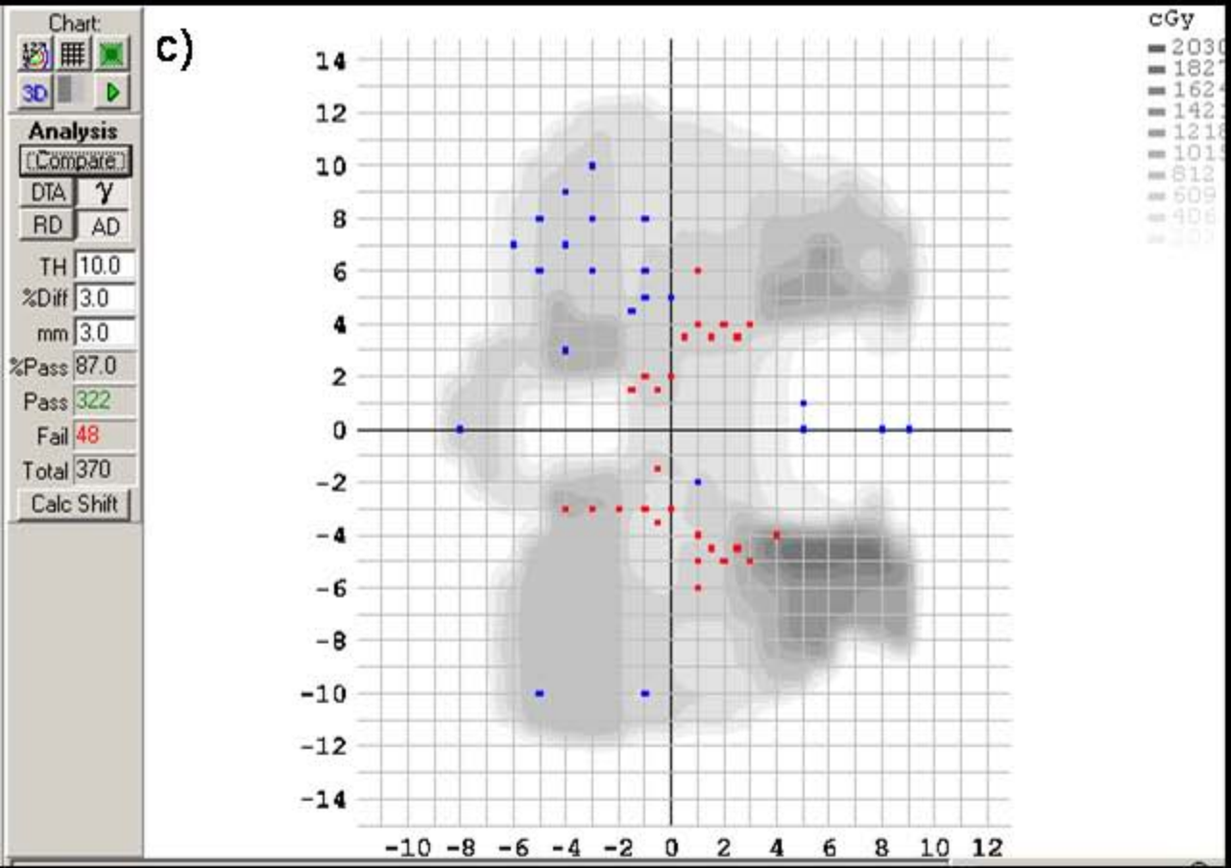


Chart:
3D
Analysis
Compare
DTA
γ
RD
AD
TH 10.0
%Diff 3.0
mm 3.0
%Pass 87.0
Pass 322
Fail 48
Total 370
Calc Shift
cGy
2030
1827
1624
1421
1218
1015
812
609
406
14 12 10 8 6 4 2 0 -2 -4 -6 -8 -10 -12 -14
-10 -8 -6 -4 -2 0 2 4 6 8 10 12