Narodowy Instytut Onkologii

The Abdus Salam International Centre for Theoretical Physics

# IMRT/VMAT Theory and definitions

PhD Anna Zawadzka

MSCNRIO - Medical Physic Department

ICTP/School on Medical Physics for Radiation Therapy: Dosimetry, Treatment Planning and Delivery for Advanced Applications
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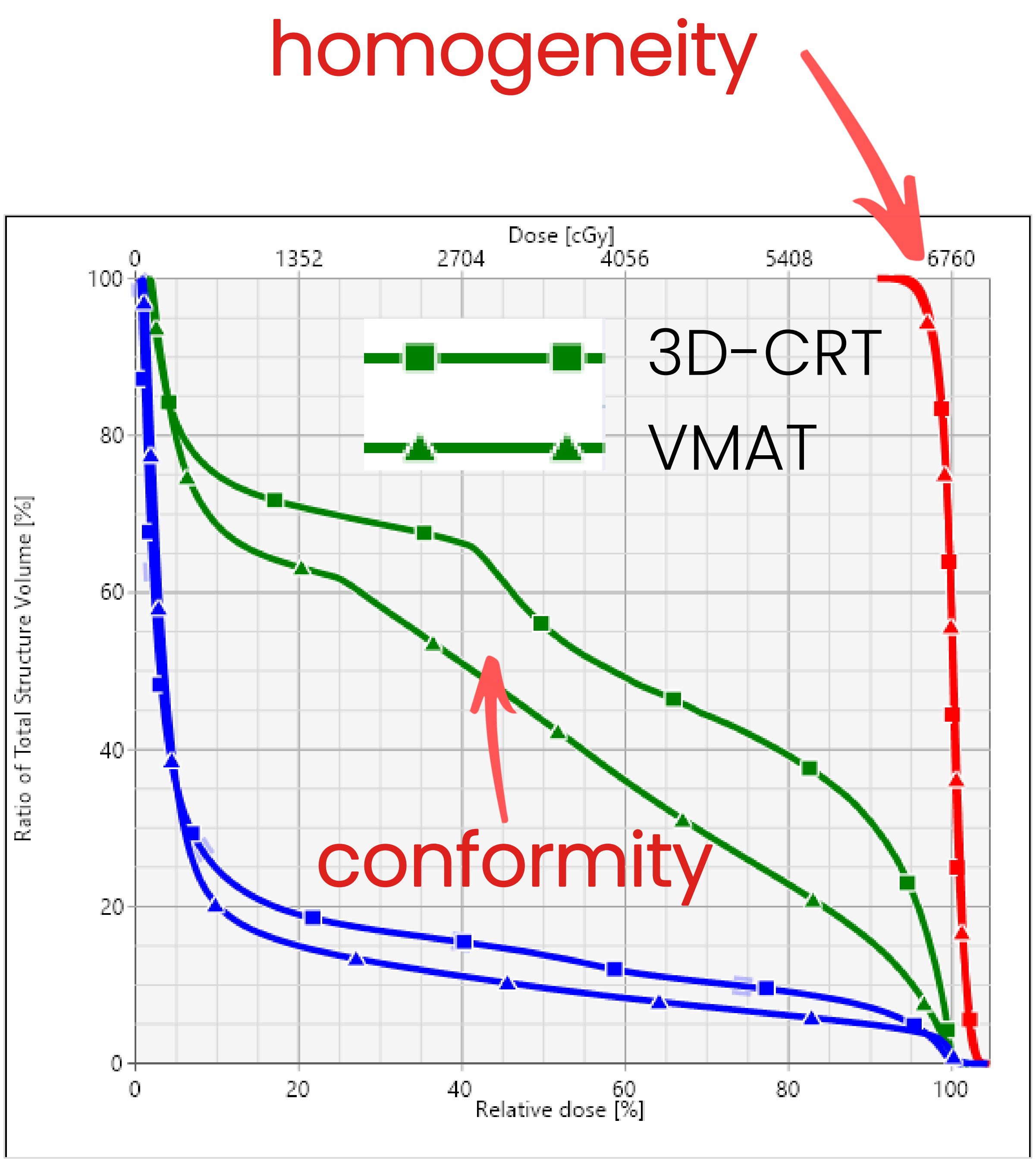
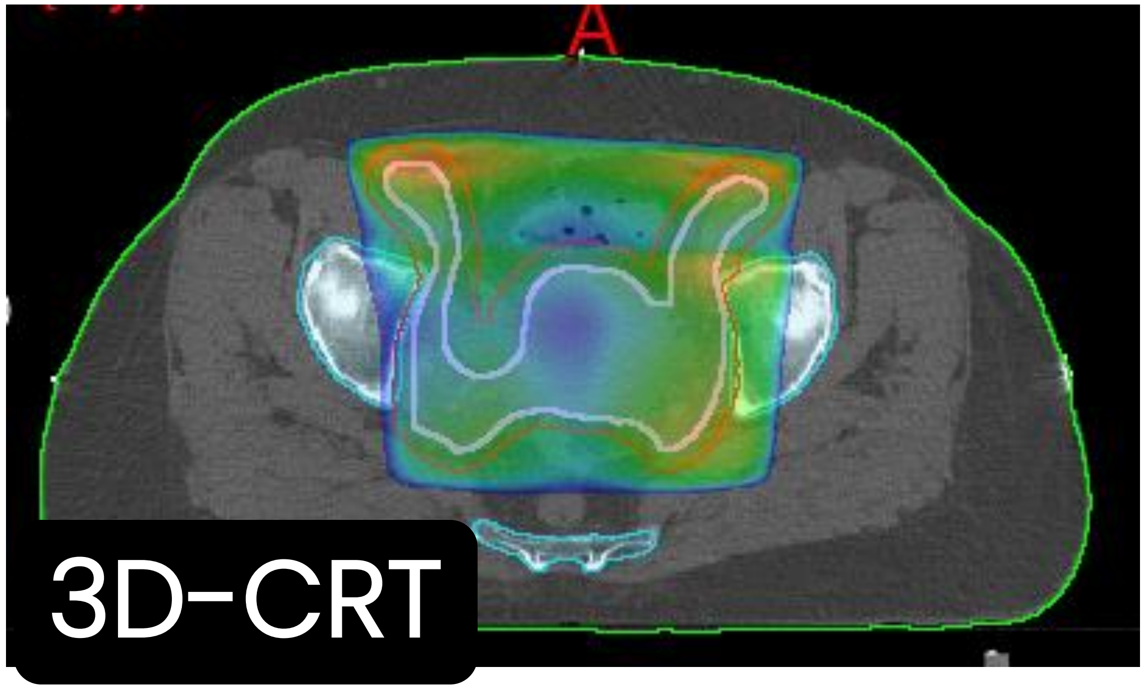
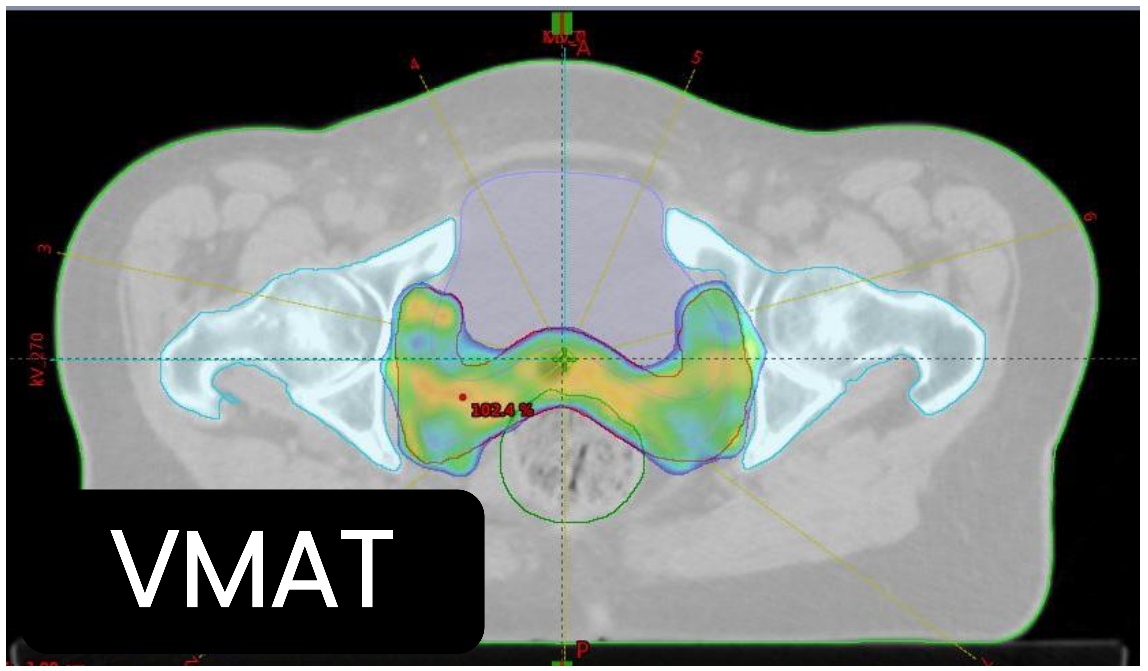
# **Why dynamic technique?**

Beams with non-uniform intensity from various directions can provide:

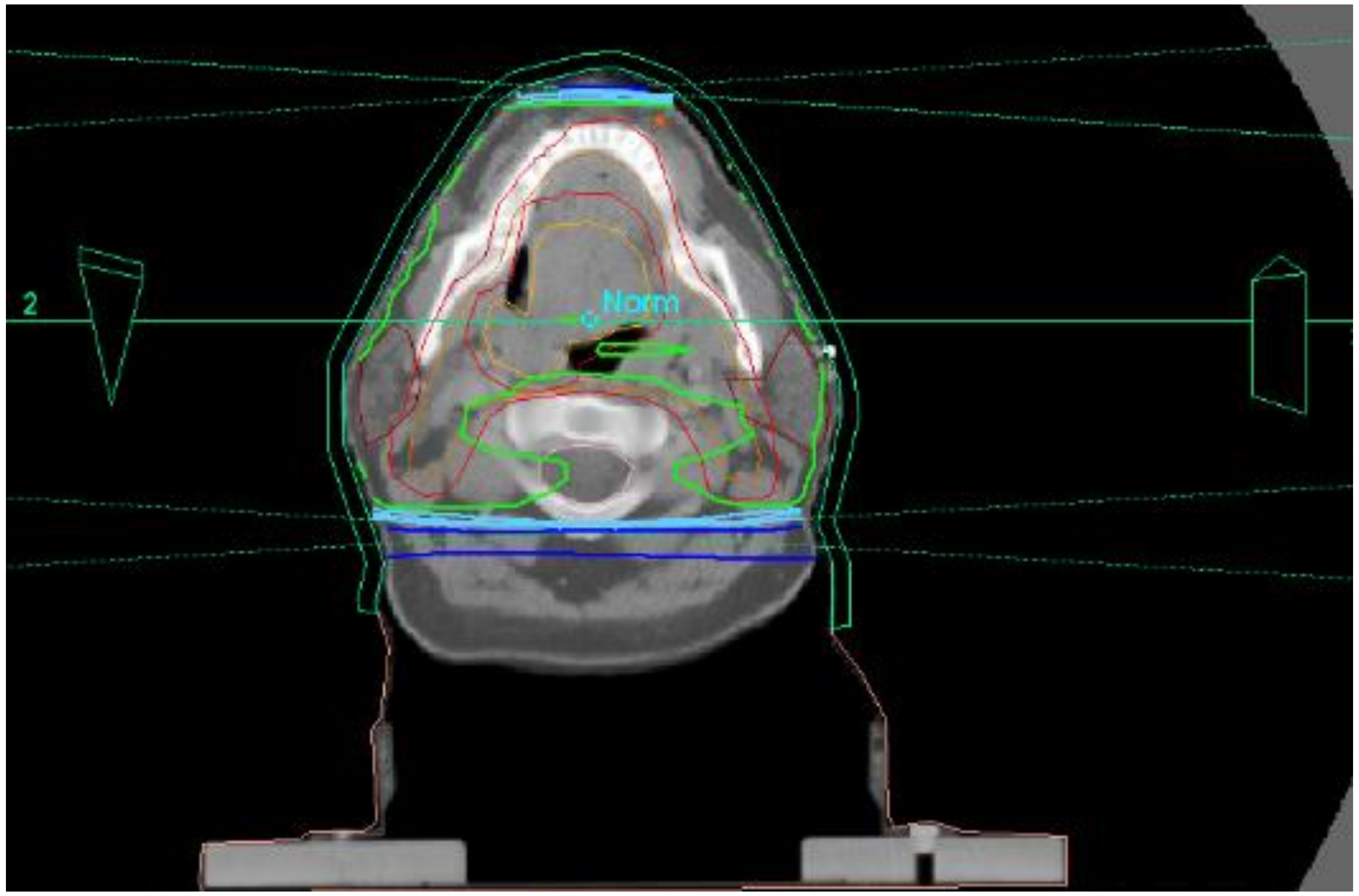
- dose homogeneity to PTV similar to conventional radiotherapy - homogeneity
- superior conformity (concave structures) -> better OAR's sparing - conformity

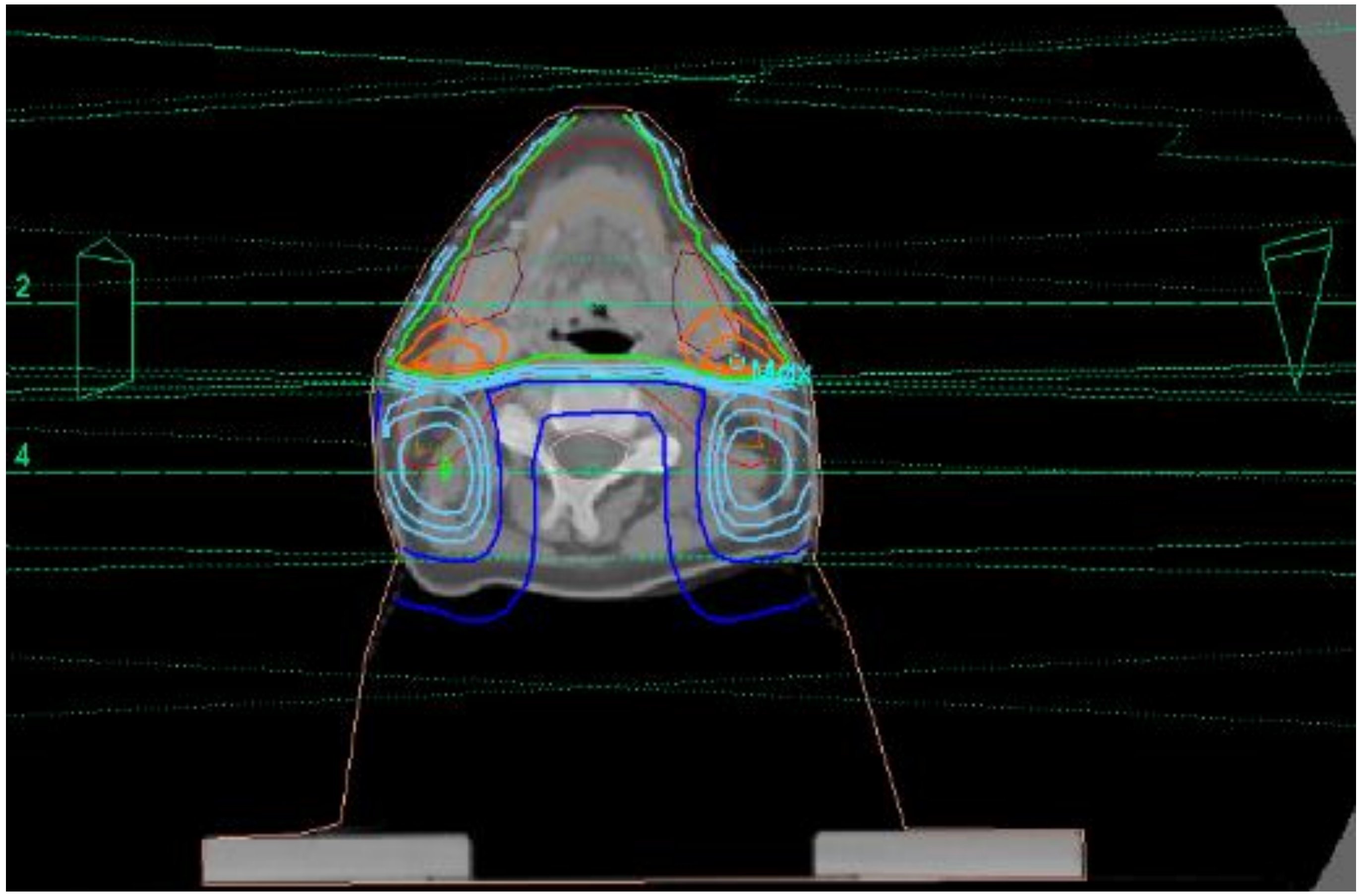


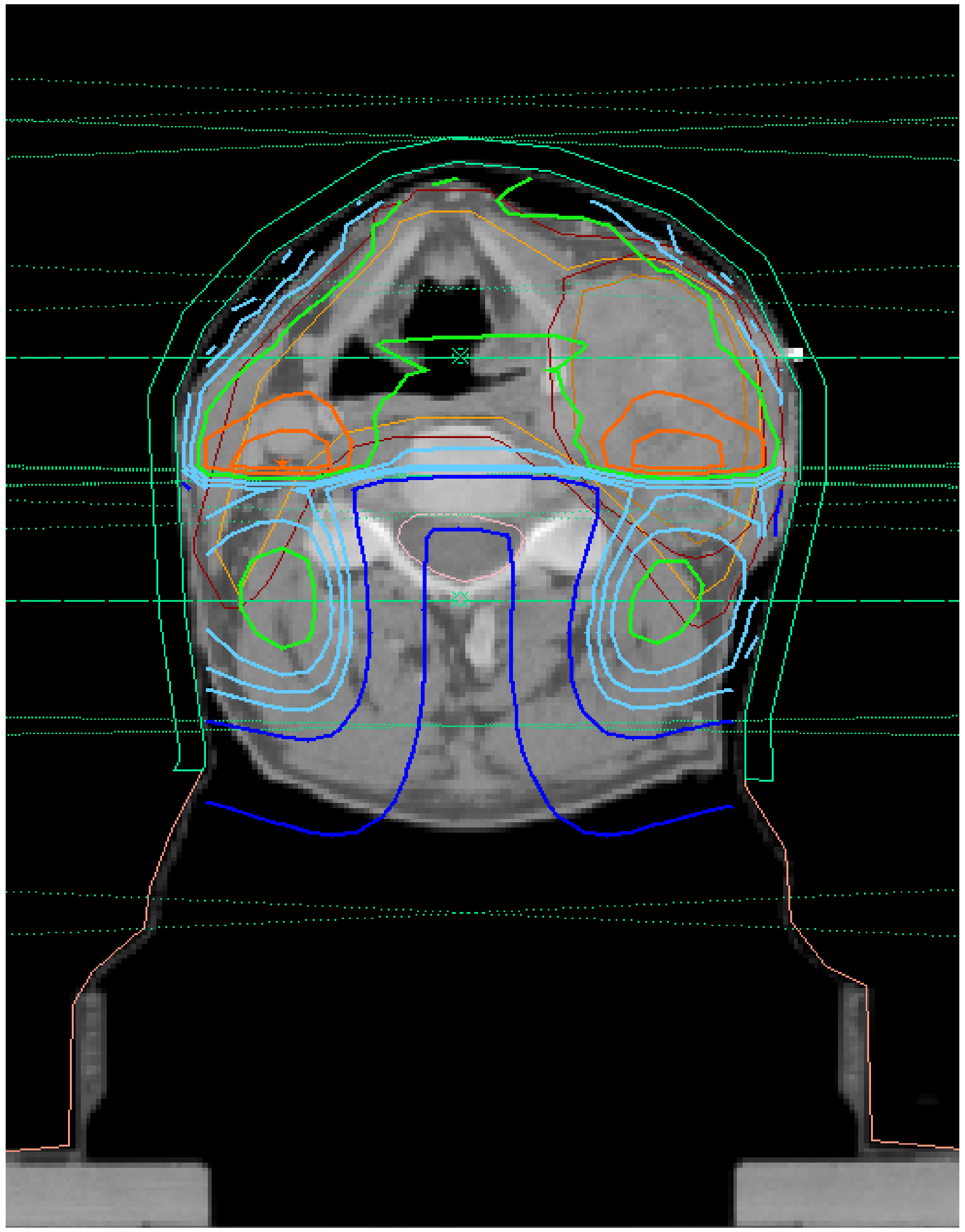



# 3D vs. IMRT



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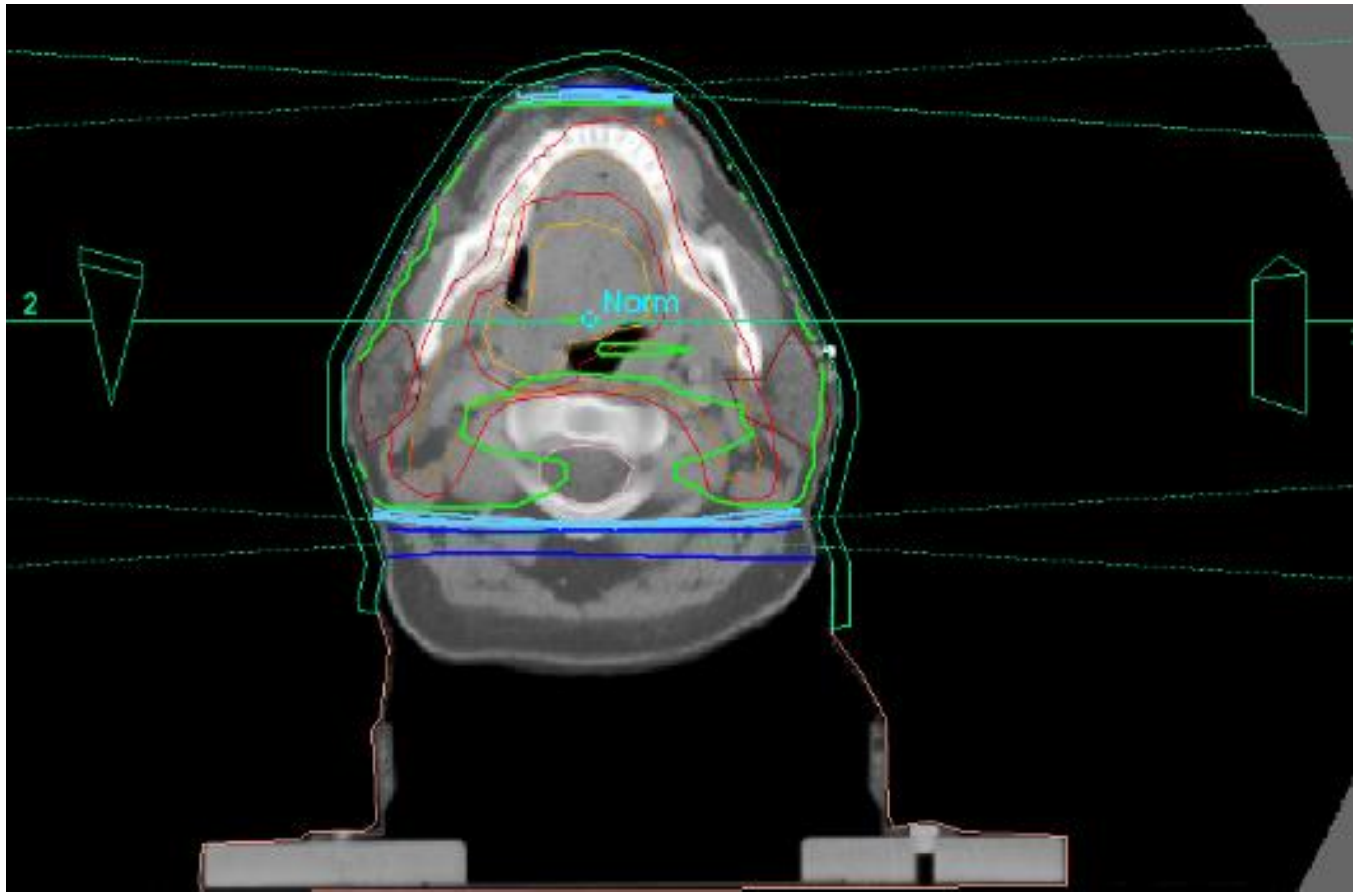
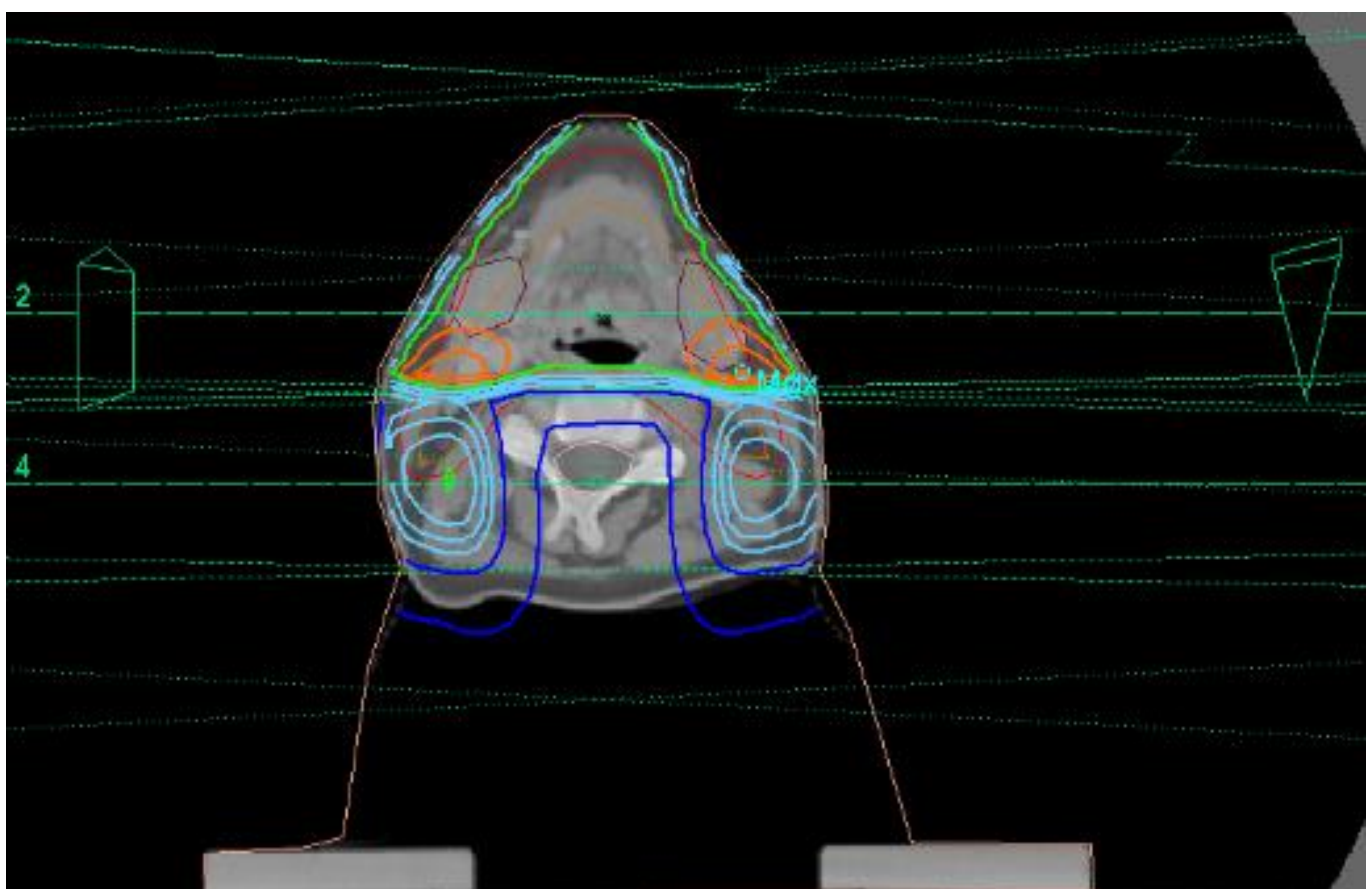
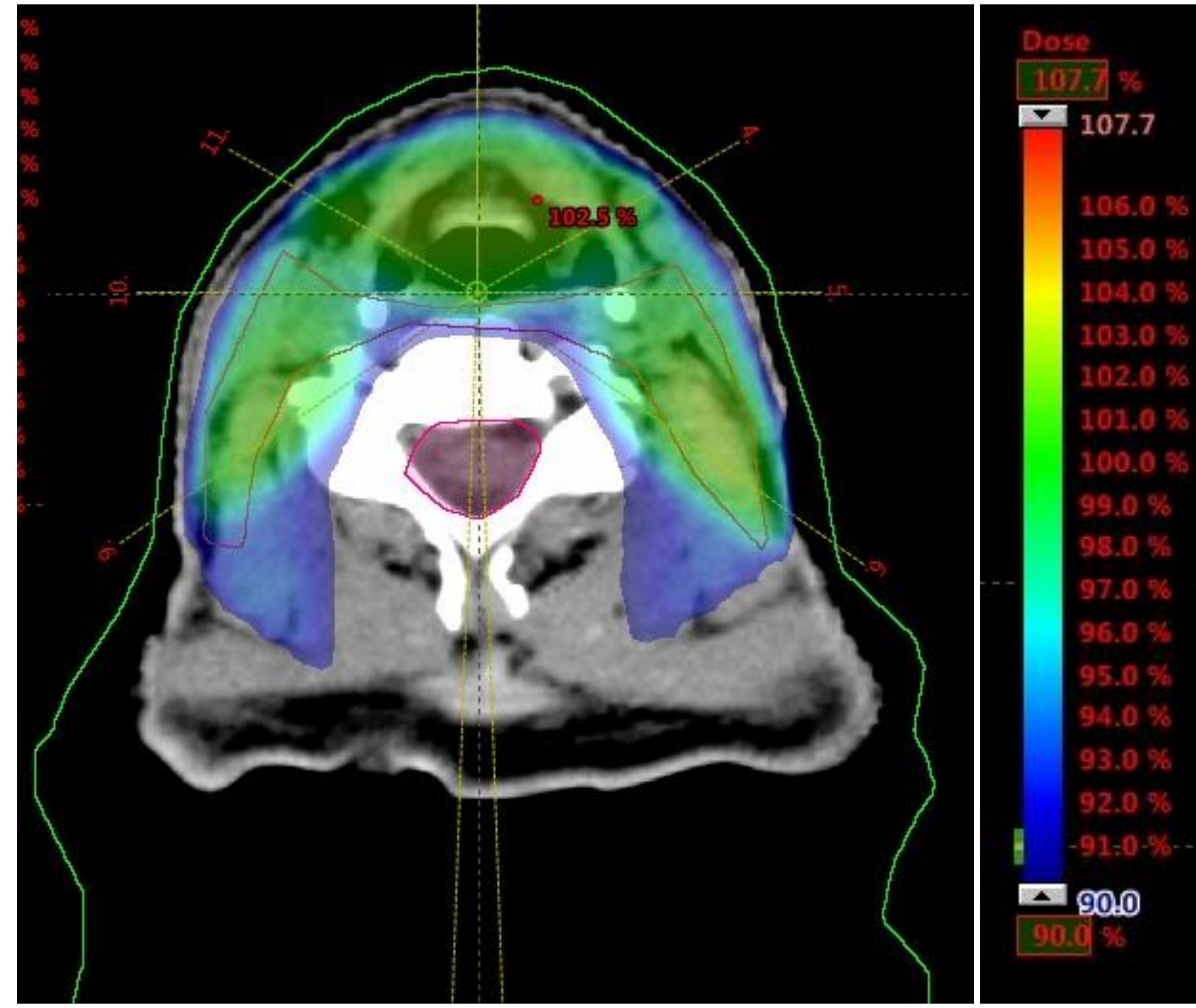

Norm




102.5 %
Dose
107.7 %
107.7
106.0 %
105.0 %
104.0 %
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102.0 %
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99.0 %
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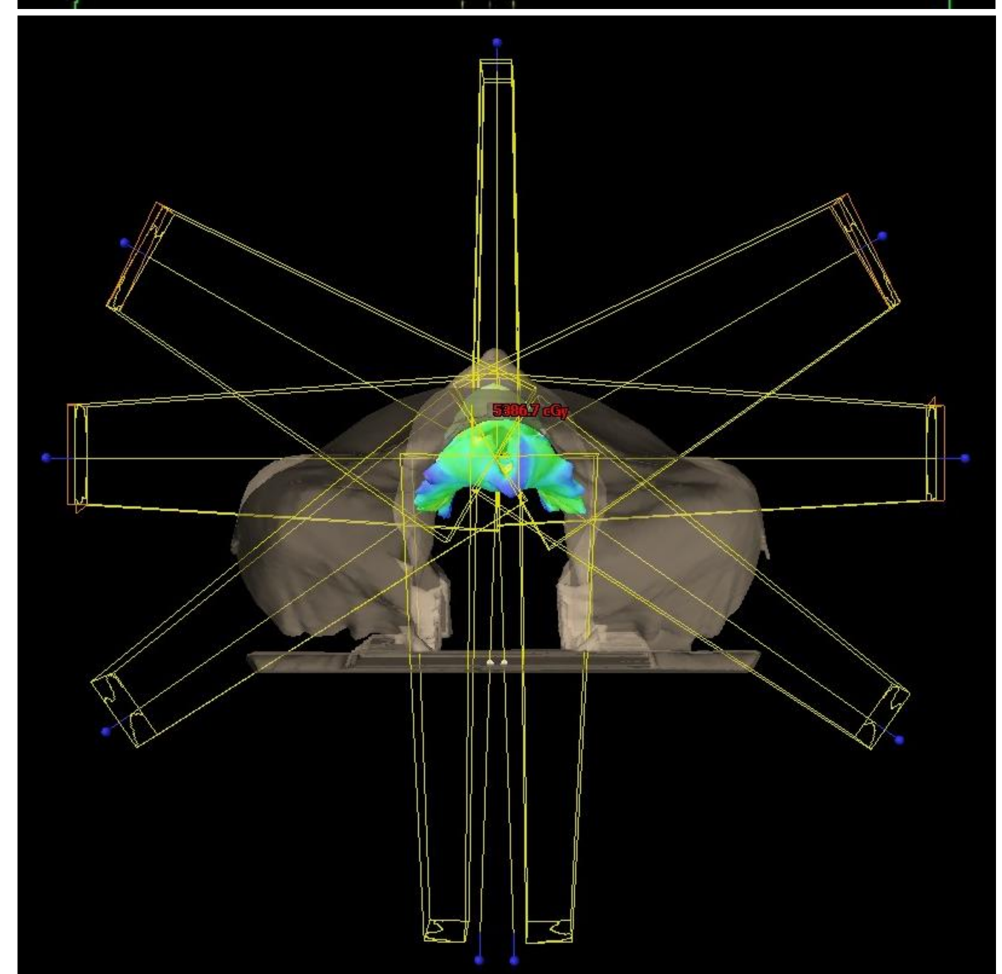
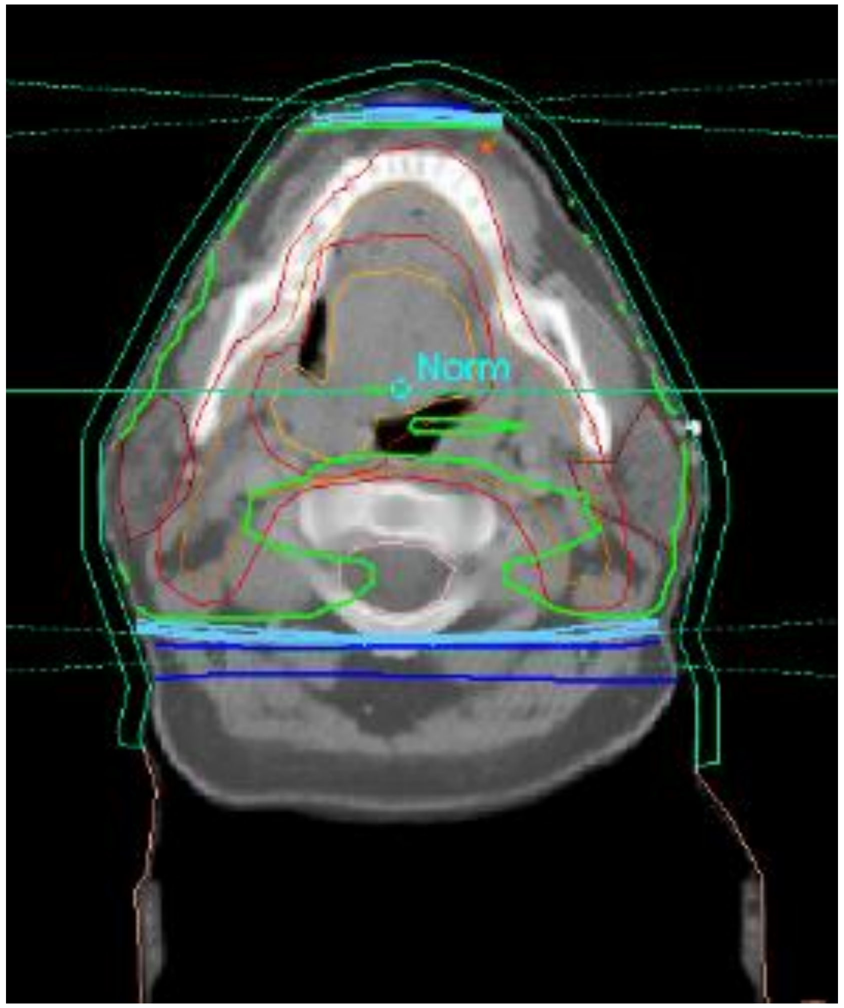
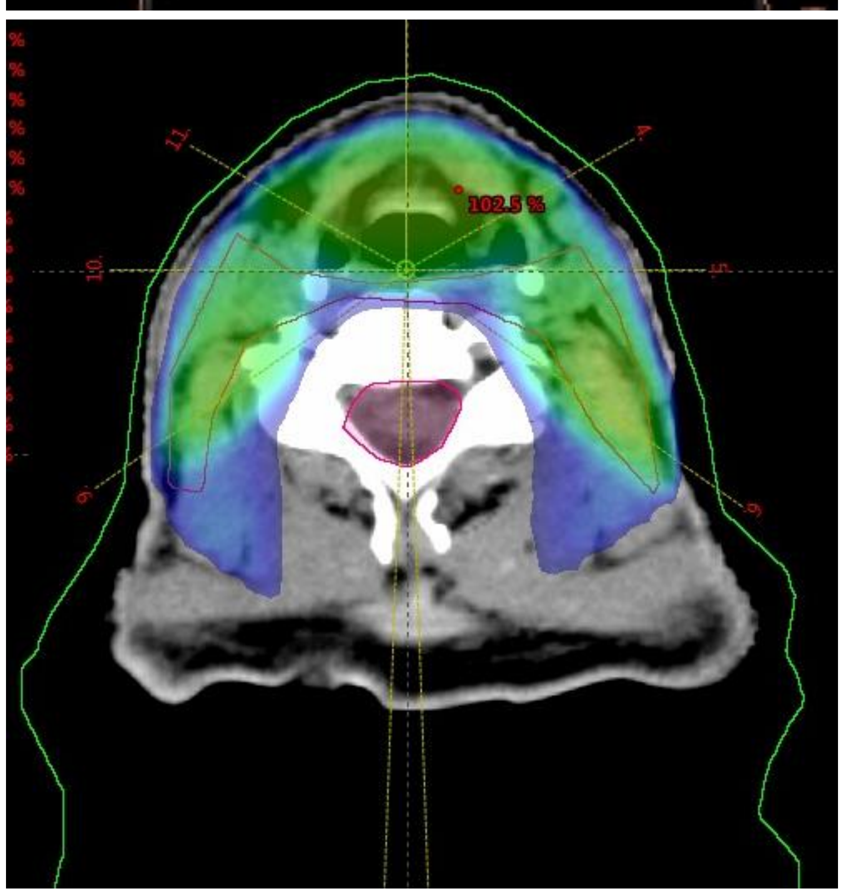
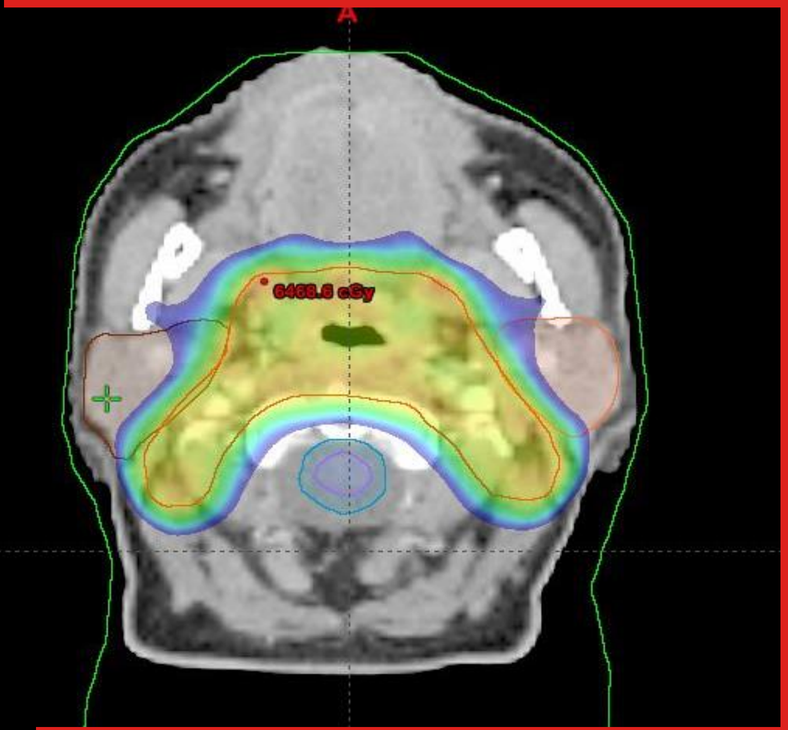


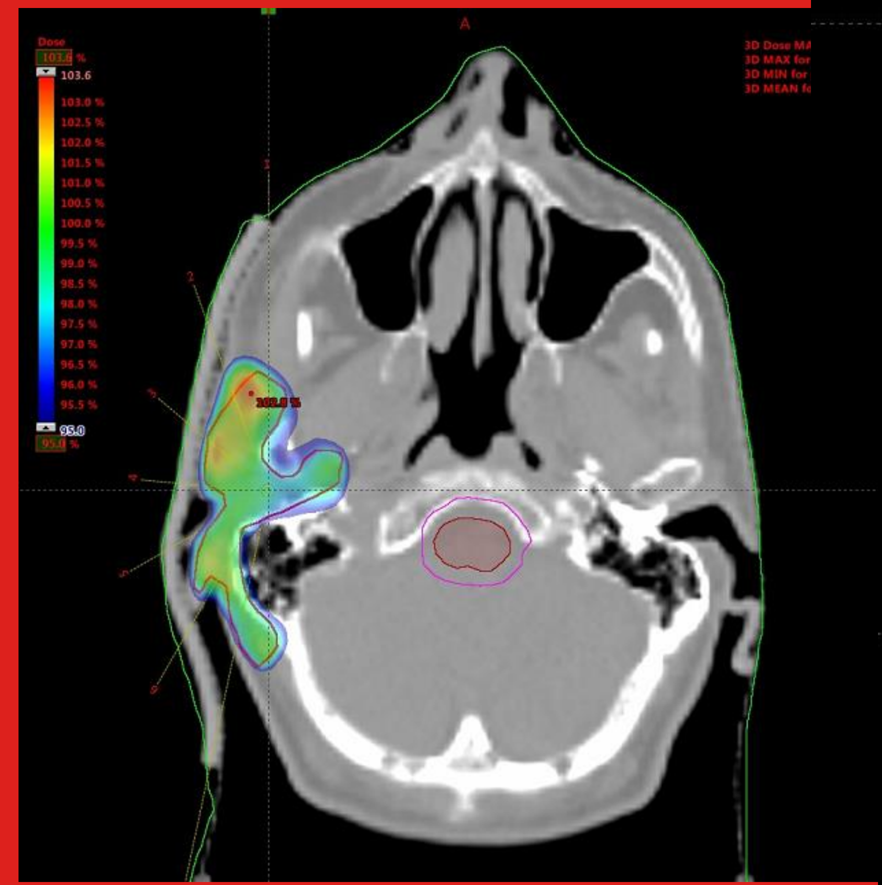
# 3D CRT

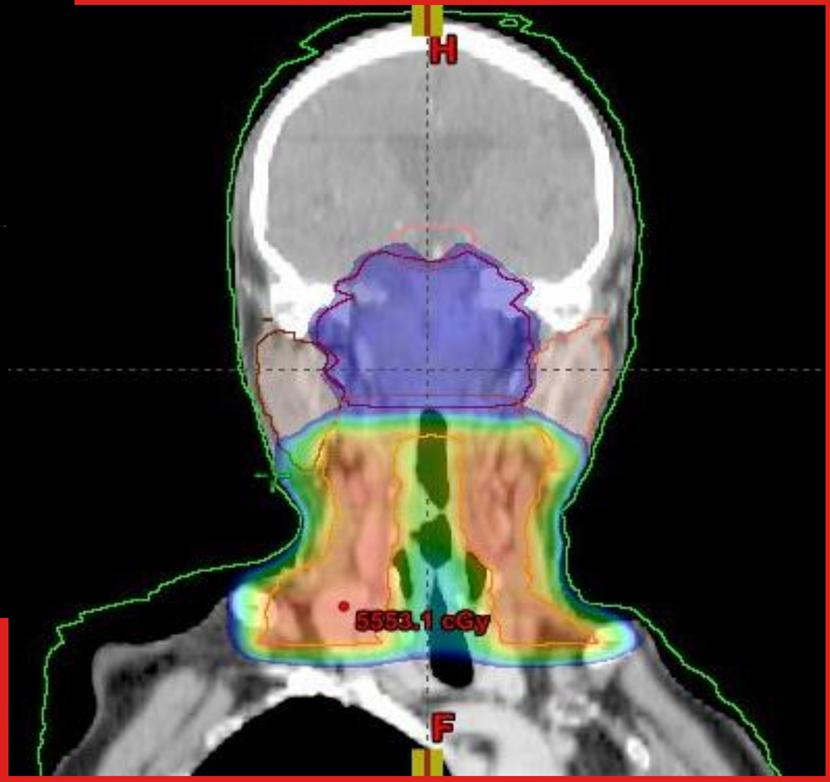



# VMAT/IMRT







# IMRT versus 2D/3D conformal RT in oropharyngeal cancer A review of the literature and meta-analysis

2020 - Oral Diseases - Alterio et el.

- IMRT high-precision RT, high dose gradient between PTV and OARs
- 2D/3D - high doses to PTV but potentially resulting in higher TC (lower rate of marginal failures)
- no differences in disease-related outcomes (OS and DFS)
- late and acute toxicity profile data were quantitative estimates – lack of complete and time-homogenous data reporting

# 3D vs IMRT

the most severe long-term toxicity of RT in HN

- Xerostomia
- Dysphagia
- Mucositis

| References | Technique of RT | Acute toxicity | | | | | | | | | |
|---|---|---|---|---|---|---|---|---|---|---|---|
| | | Time Threshold | Skin>3 | Mucositis>3 | Blood>3 | Nausea or vomiting>3 | Tube dependence | Dysphagia>3 | Pain>3 | Complication parotid gland | Xerostomia>2 |
| Lee et al. 2006 | IMRT | not specified | 10% | 66% | 27% | 17% | | | | | |
| | 2D-CRT | | 20% | 72% | 30% | 10% | | | | | |
| Lohia et al. 2014 | IMRT | not specified | 7% | 37% | | | 61% | | | | |
| | 3D-CRT | | 23% | 76% | | | 79% | | | | |
| Kerr et al. 2015 | IMRT | within 3 months | | | | | 21% | | | | |
| | 3D-CRT | | | | | | 46% | | | | |
| Hodge et al. 2007 | IMRT | not specified | | 58% | | | 46% | | | | |
| | 3D-CRT | | | 75% | | | 61% | | | | |
| Al-Mamgani et al. 2013 | IMRT | within 90 days | 45% | 68% | | | | 49% | 32% | | |
| | 3D-CRT | | 51% | 82% | | | | 72% | 52% | | |
| Rusthoven et al. 2008 | IMRT | not specified | 34% | 81% | | | | | | | |
| | 3D-CRT | | 52% | 78% | | | | | | | |
| Braam et al. 2006 | IMRT | within 6 weeks | | | | | | | | 55% | |
| | 3D-CRT | | | | | | | | | 87% | |
| Clifford Chao et al. 2001 | IMRT | | 25% | 42% | | | 25% | 25% | | | 75% |
| | CRT | | 8% | 25% | | | 18% | 18% | | | 79% |

# Late toxicity

| References | Time Threshold | Late toxicity | | | | | | | | | | | | | | | | |
|---|---|---|---|---|---|---|---|---|---|---|---|---|---|---|---|---|---|---|
| | | Xerostomia>2 | Xerostomia>2 at 6 months | Xerostomia>2 at 12 months | Xerostomia>2 at 18 months | Xerostomia>3 | Esophageal>2 | Trismus>2 | Tube dependence at 1 year | Tube dependence at 2 years | Weigth loss>10% | Osteoradionecrosis>3 | Mucositis>3 | Dysphagia>3 | Skin>3 | Pain>3 | Complication parotid gland | General toxicity G>3 |
| Lee et al. 2006 | after 20 months | 12%<br>67% | | | | | 7%<br>4% | 0%<br>7% | | | | | | | | | | |
| Lohia et al. 2014 | after 12 months | | | | | | | | 13%<br>35% | | 35%<br>34% | | | | | | | |
| Kerr et al. 2015 | after 24 months | | | | | | | | | | 5%<br>3% | | | | | | | |
| Hodge et al. 2007 | not specified | 56%<br>63% | | | | | | | | | | | | | | | | 2%<br>5% |
| Al-Mamgani et al. 2013 | after 90 days | | | | | 7%<br>23% | | | | | | | 2%<br>11% | 10%<br>20% | 2%<br>5% | 5%<br>3% | | |
| Rusthoven et al. 2008 | different time points | | 62%<br>100% | 15%<br>94% | 6%<br>93% | | | | 37%<br>51% | 25%<br>14% | | | | | | | | |
| Braam et al. 2006 | after 6 months | | | | | | | | | | | | | | | | 56%<br>81% | |
| Clifford Chao et al. 2001 | | 0%<br>4% | | | | | | 0%<br>2% | | | | 0%<br>7% | 0%<br>0% | | 0%<br>0% | | | |

## Side effects

- **Xerostomia**
  IMRT reduces patient-reported xerostomia, allows recovery of salivary flow, and improves QoL after treatment without any reduction in tumor outcome compared with conventional radiotherapy.

- **Mucositis**
  In IMRT a significantly lower incidence of grade 3 or greater acute toxic effects to skin and mucous membranes than 2D/3D-RT in almost all investigated studies.

- **PEG tube dependence**
  - IMRT was associated with a lower rate of PEG tube dependence
  - a shorter median duration of PEG tube use was reported

significant QoL benefits

# VMRT vs VMAT

## Volumetric intensity-modulated arc therapy vs. conventional IMRT in head-and-neck cancer: A comparative planning and dosimetric study

- VMAT single vs. double arc vs. IMRT (7 fields – sliding window)
- Mean reduction in the number of MU (by nearly 60%)
- similar sparing of all OAR
- Double arc provided the best dose homogeneity to PTV

Palma et al. 2008 (prostate)

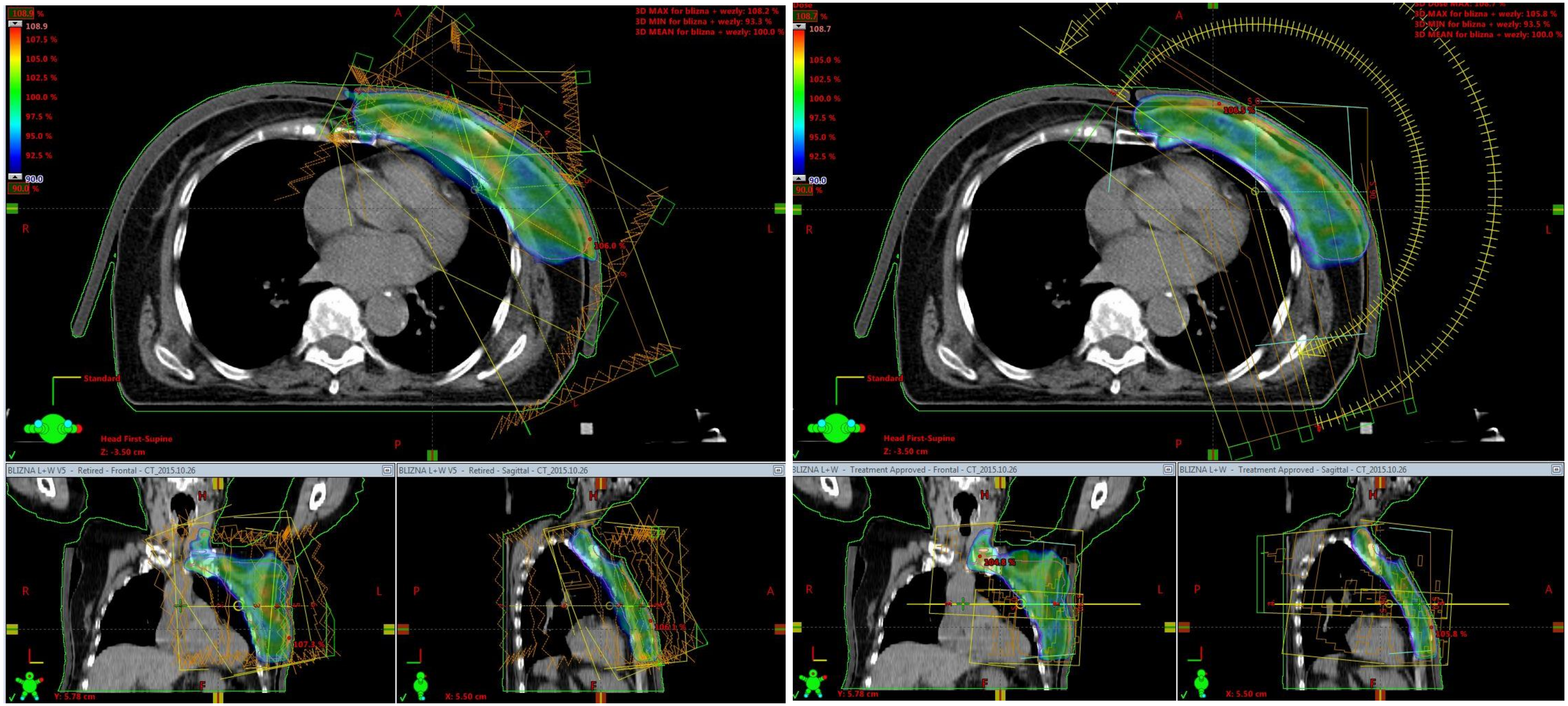
Cozzi et al. 2008 (gynec)

Fogliata et al. (brain)

W Verbakel et al., Int. J. Radiation Oncology Biol. Phys 74(1), 2009

# IMRT vs VMAT

7 fields IMRT vs. 4 arc VMAT



| parameters | Left side | | |
| --- | --- | --- | --- |
| | IMRT | VMAT | p-value |
| V90% CTV [%] | 100.0 ± 0.0 | 100.0 ± 0.0 | 1.00 |
| V95% CTV [%] | 99.3 ± 1.18 | 99.42 ± 0.99 | 0.73 |
| V90% PTV [%] | 99.84 ± 0.24 | 99.76 ± 0.44 | 0.75 |
| V95% PTV [%] | 96.4 ± 1.69 | 96.44 ± 3.41 | 0.86 |
| $V_{lung}$ 20 [%] | 25.49 ± 3.66 | 27.89 ± 5.9 | 0.41 |
| $D_{mean}$ lung R [Gy] | 15.21 ± 1.58 | 15.81 ± 2.26 | 0.52 |
| V20 heart [%] | 12.49 ± 2.83 | 13.61 ± 5.88 | 0.95 |
| $D_{mean}$ heart [Gy] | 12.44 ± 1.18 | 12.46 ± 2.21 | 0.86 |
| V20 lung L [%] | 11.71 ± 1.74 | 12.74 ± 2.5 | 0.24 |
| V30 lung L [%] | 5.88 ± 1.27 | 6.27 ± 1.61 | 0.68 |
| $D_{mean}$ lung [Gy] | 10.16 ± 1.14 | 10.5 ± 1.56 | 0.37 |
| $D_{max}$ cord [Gy] | 13.24 ± 3.48 | 14.43 ± 3.22 | 0.44 |
| MU | 1194.67 ± 274.11 | 776.89 ± 104.25 | 0.01 |

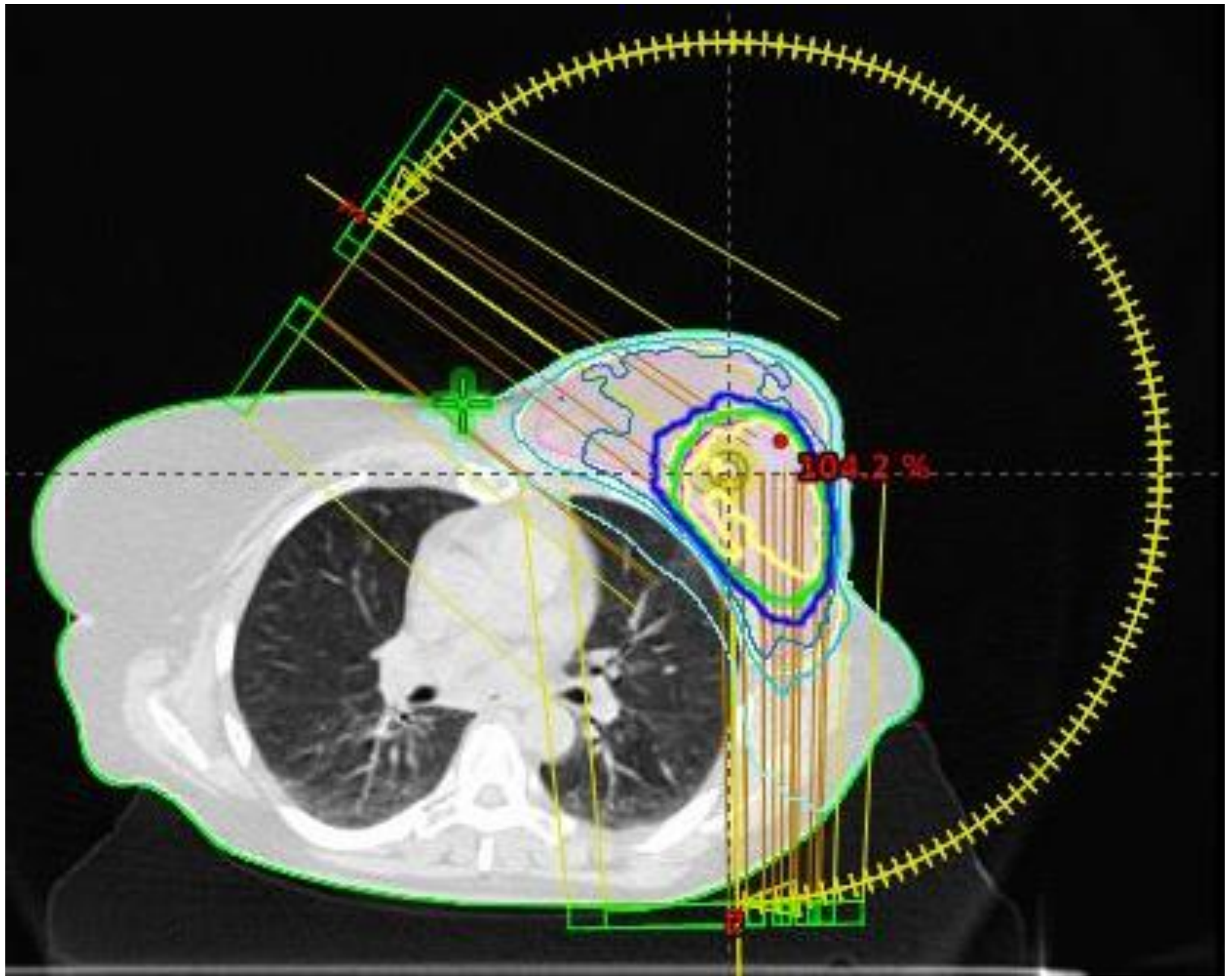
# IMRT vs VMAT

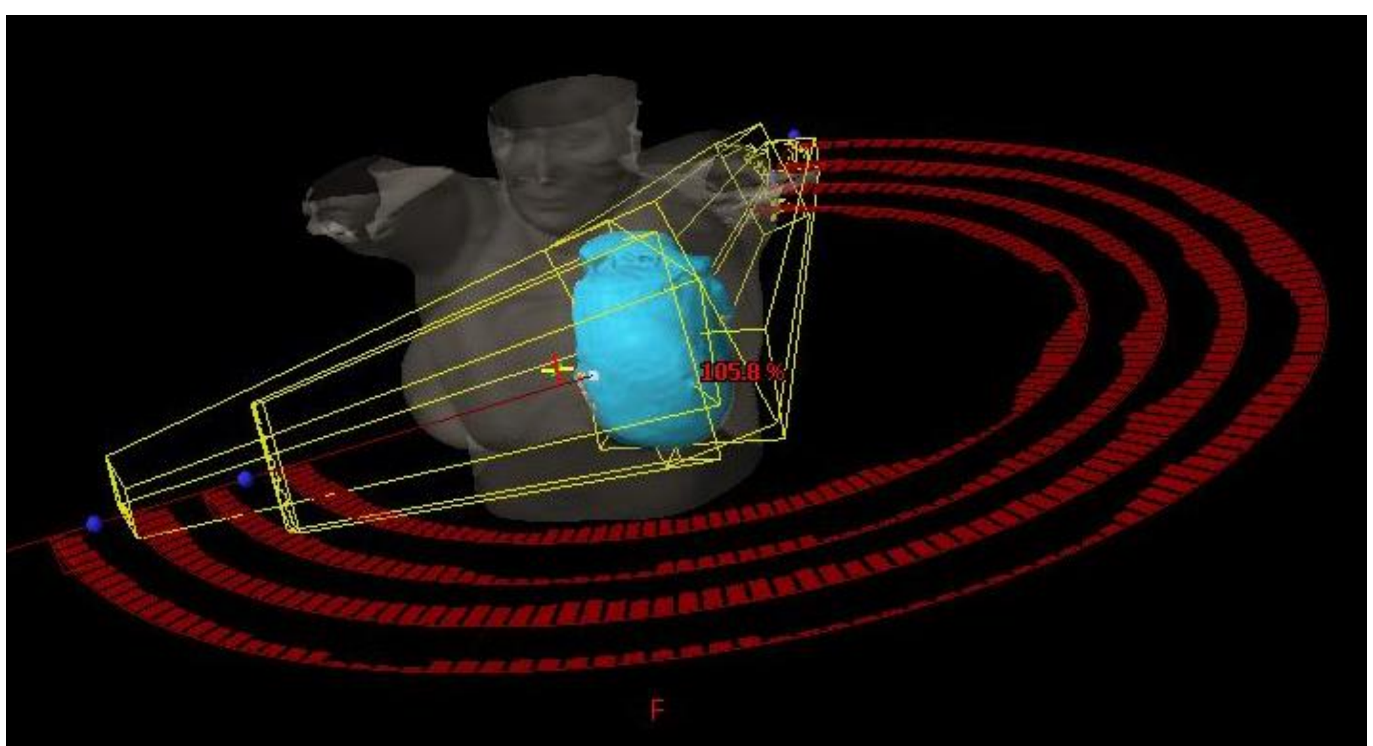
- no statistical differences in dose distribution between techniques
- fewer MU in VMAT
- shorter treatment time (measured for 9 IMRT and 13 VMAT plans) **7.51 min** (5.55 ÷ 11.43, IMRT) vs **3.38± min** (2.33 ÷ 4.52, VMAT)

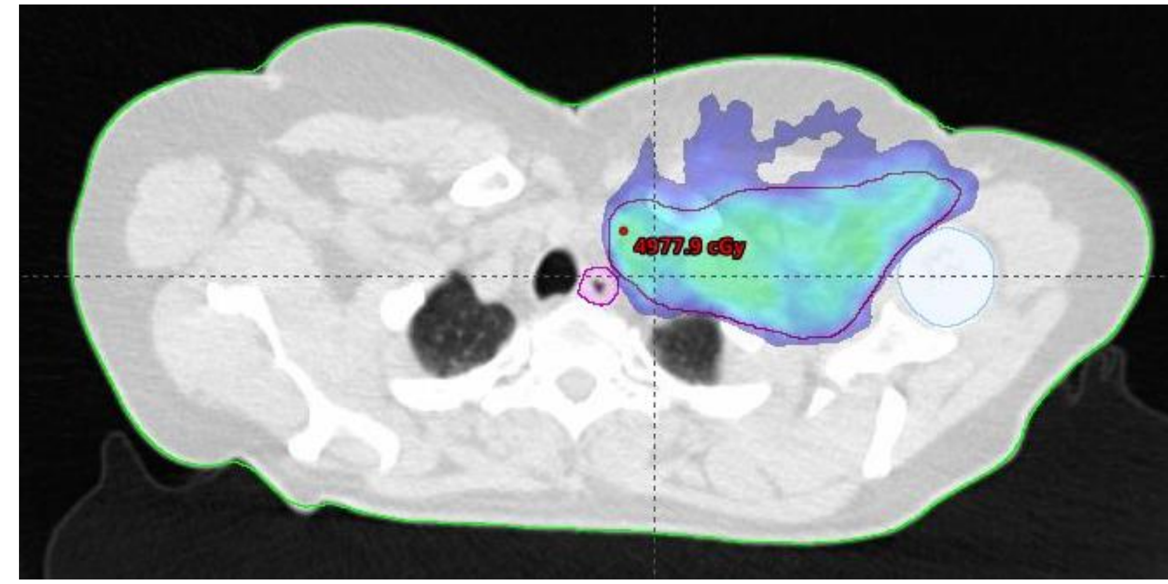
**Shorter time means greater comfort for the patient, greater reproducibility (intrafraction movement)**

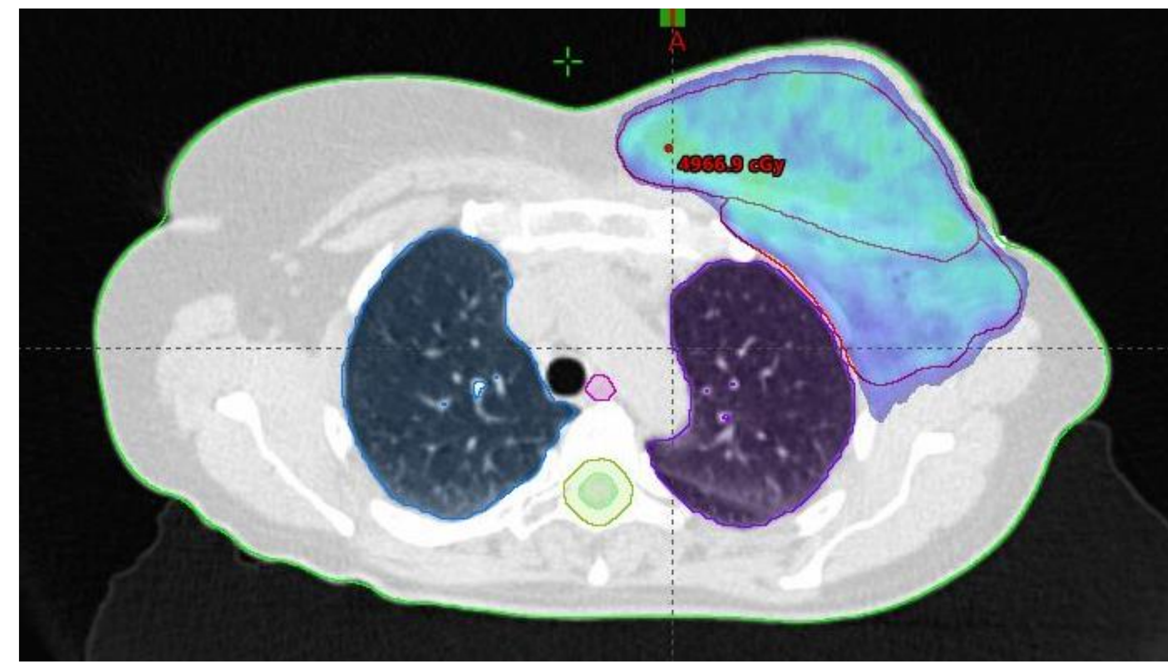
E. Dąbrowska, A. Zawadzka, P. Mężeński, J. Gałecki, P. Kukołowicz, M. Spałek. The comparison of TMRT and VMAT plan Quality for hypofractionated post-mastectomy chest Wall Irradiation. ESTRO 2016.

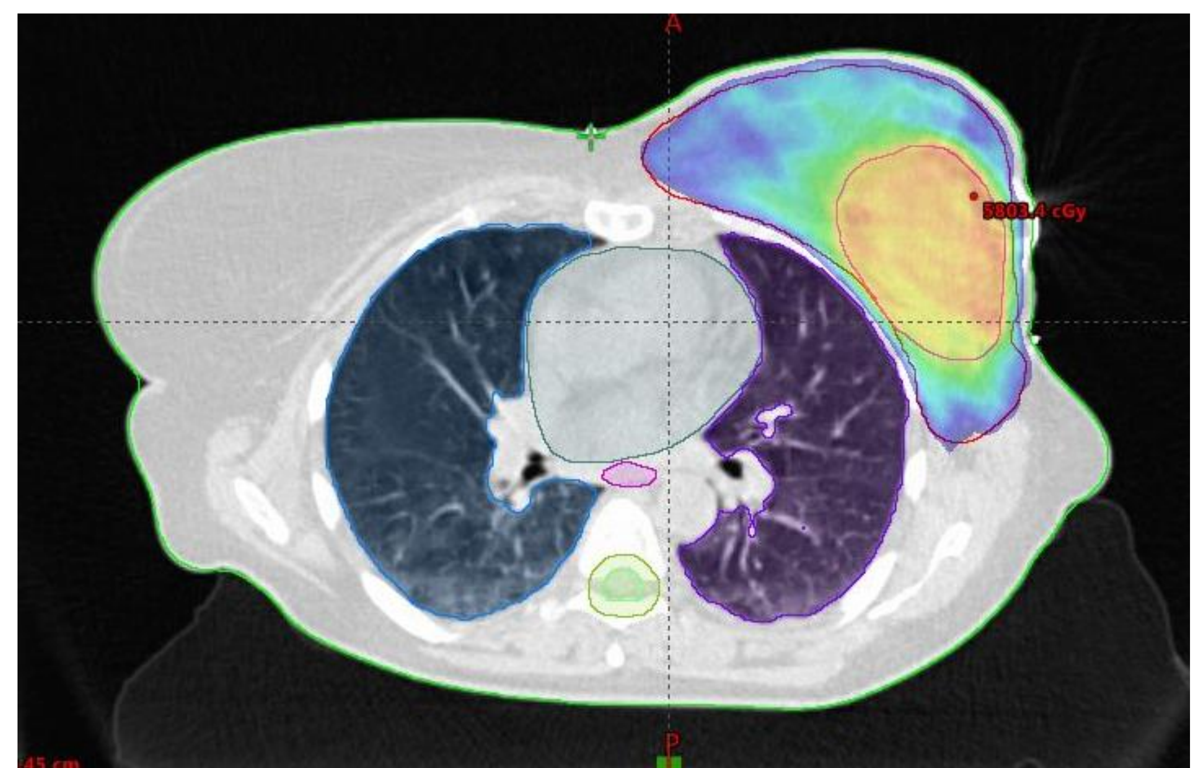
# Brest + regional nodes









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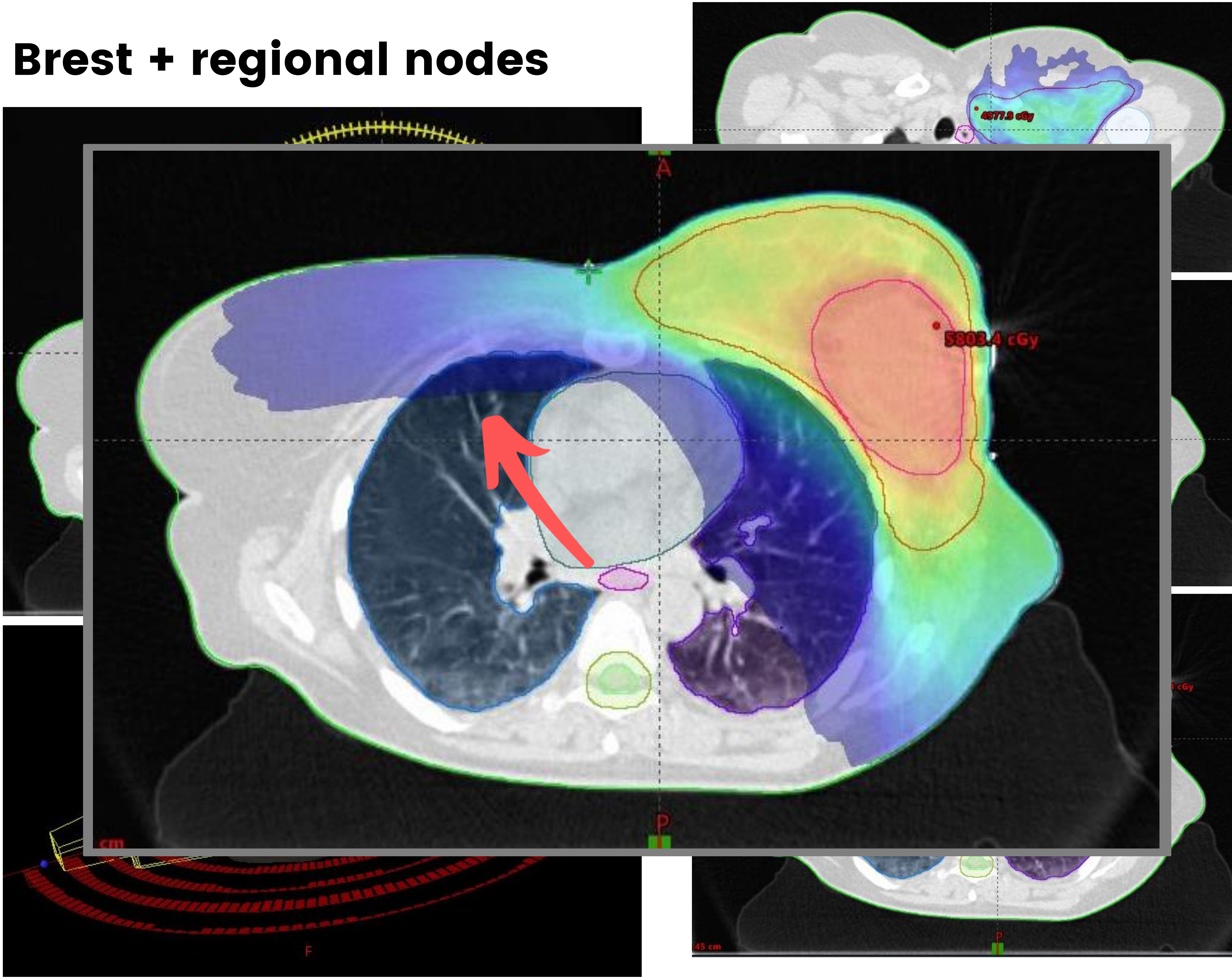

# IMRT vs VMAT – breast

Table 6. Comparative planning studies in breast cancer

| Paper [ref] *VMAT commercial system* | Number of patients | Site | Comparison | PTV | OAR | MU per fraction | Treatment time per fraction |
|---|---|---|---|---|---|---|---|
| Qiu et al [142] *Rapidarc* | 8 | Breast (partial breast radiotherapy) | 3D-CRT (non coplanar, 4–5F) *vs* VMAT (modified partial arc) | Similar PTV coverage. VMAT slightly better than 3D-CRT at conformity (not statistically significant) | VMAT better than 3D-CRT at sparing ipsilateral normal breast tissue, ipsilateral lung | VMAT, 488.6; 3D-CRT, 634.1 | VMAT, 1.21 min; 3D-CRT, 6.3 min |
| Popescu et al [143] *Predecessor to RapidArc* | 5 | Breast (+ regional nodes including internal mammary nodes) | 3D-CRT *vs* IMRT (9F,SW) *vs* VMAT (2 partial arcs) | Similar PTV coverage, homogeneity, conformity | VMAT better than IMRT and 3D-CRT at sparing heart and ipsilateral lung (low and intermediate doses), contralateral breast (mean dose). VMAT – lower mean dose to healthy tissue but higher V5Gy compared with 3D-CRT and IMRT | VMAT, 862; IMRT, 1254; 3D-CRT,489 | VMAT, 3.9 min; IMRT, 8.8 min; 3D-CRT, 5 min |
| Johansen et al [144] *RapidArc* | 8 | Breast (chest wall and nodes including internal mammary nodes) | CRT (4F) *vs* IMRT (7F,SW) *vs* VMAT | Similar PTV coverage. VMAT and IMRT better than CRT for conformity. VMAT better than IMRT and CRT for homogeneity | VMAT and IMRT better than CRT at sparing ipsilateral lung. CRT – lowest doses to contralateral lung. VMAT – lowest doses to contralateral breast | | |
| Nicolini et al [145] *RapidArc* | 10 | Breast (Bilateral, SIB to tumour bed) | IMRT (12F,SW) *vs* VMAT (DA) | Similar PTV coverage. VMAT better than IMRT at homogeneity | VMAT better than IMRT at sparing heart and lungs (medium-high dose level) (for lungs, IMRT better at sparing at low dose levels). VMAT – higher mean and integral dose to healthy tissue | VMAT, 796; IMRT, 1398 | VMAT, 3 min; IMRT, 11.5 min |

Teoh M, The British Journal of Radiology, 84 (2011)

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PTV

Similar PTV coverage. VMAT slightly better than 3D-CRT at conformity (not statistically significant)

Similar PTV coverage, homogeneity, conformity

Similar PTV coverage. VMAT and IMRT better than CRT for conformity. VMAT better than IMRT and CRT for homogeneity

Similar PTV coverage. VMAT better than IMRT at homogeneity


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# IMRT v

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OAR

VMAT better than 3D-CRT at sparing ipsilateral normal breast tissue, ipsilateral lung

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VMAT and IMRT better than CRT at sparing ipsilateral lung. CRT – lowest doses to contralateral lung. VMAT – lowest doses to contralateral breast

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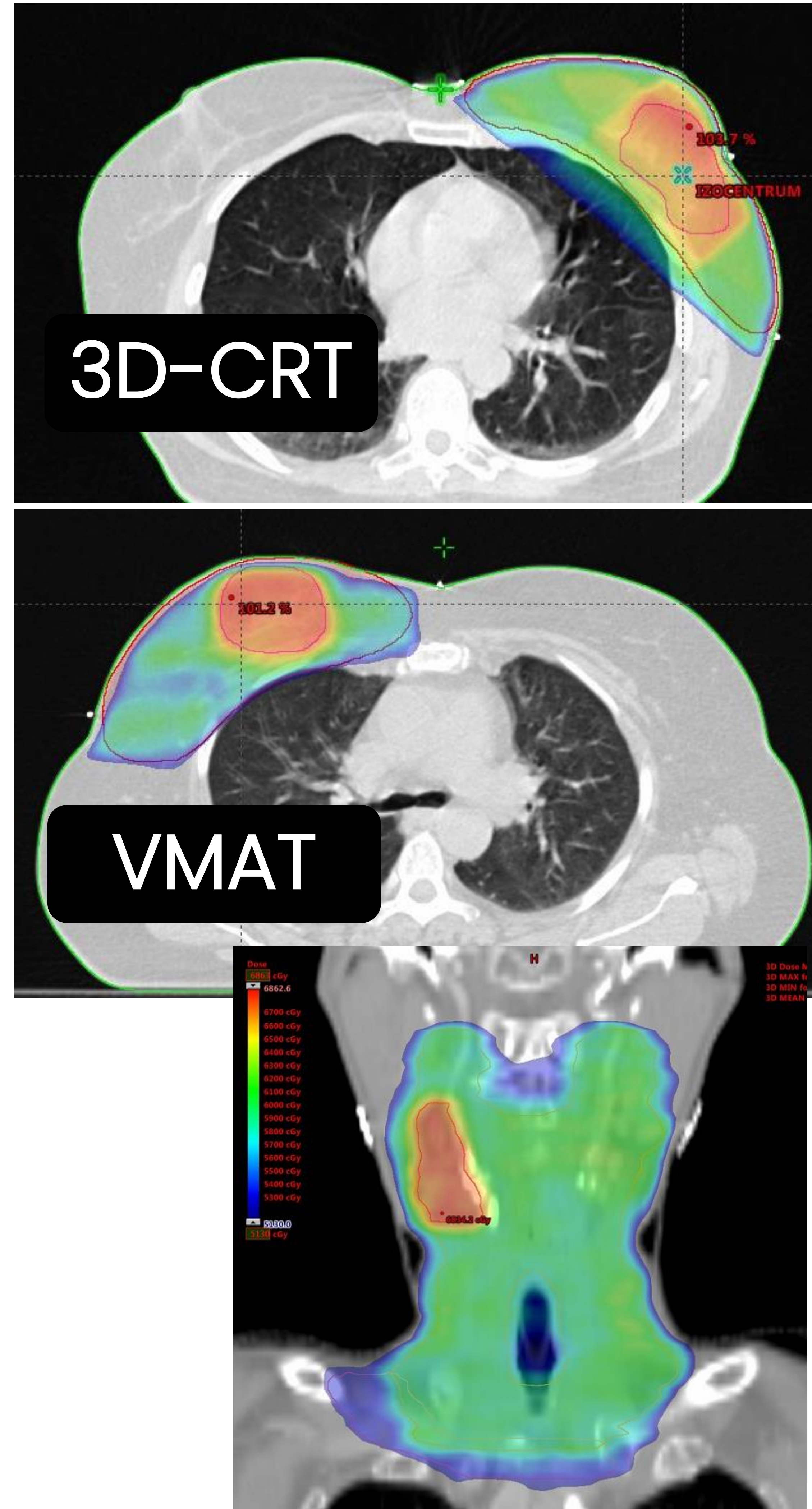
Teoh M, The British Journal of Radiolog

# Why dynamic technique?

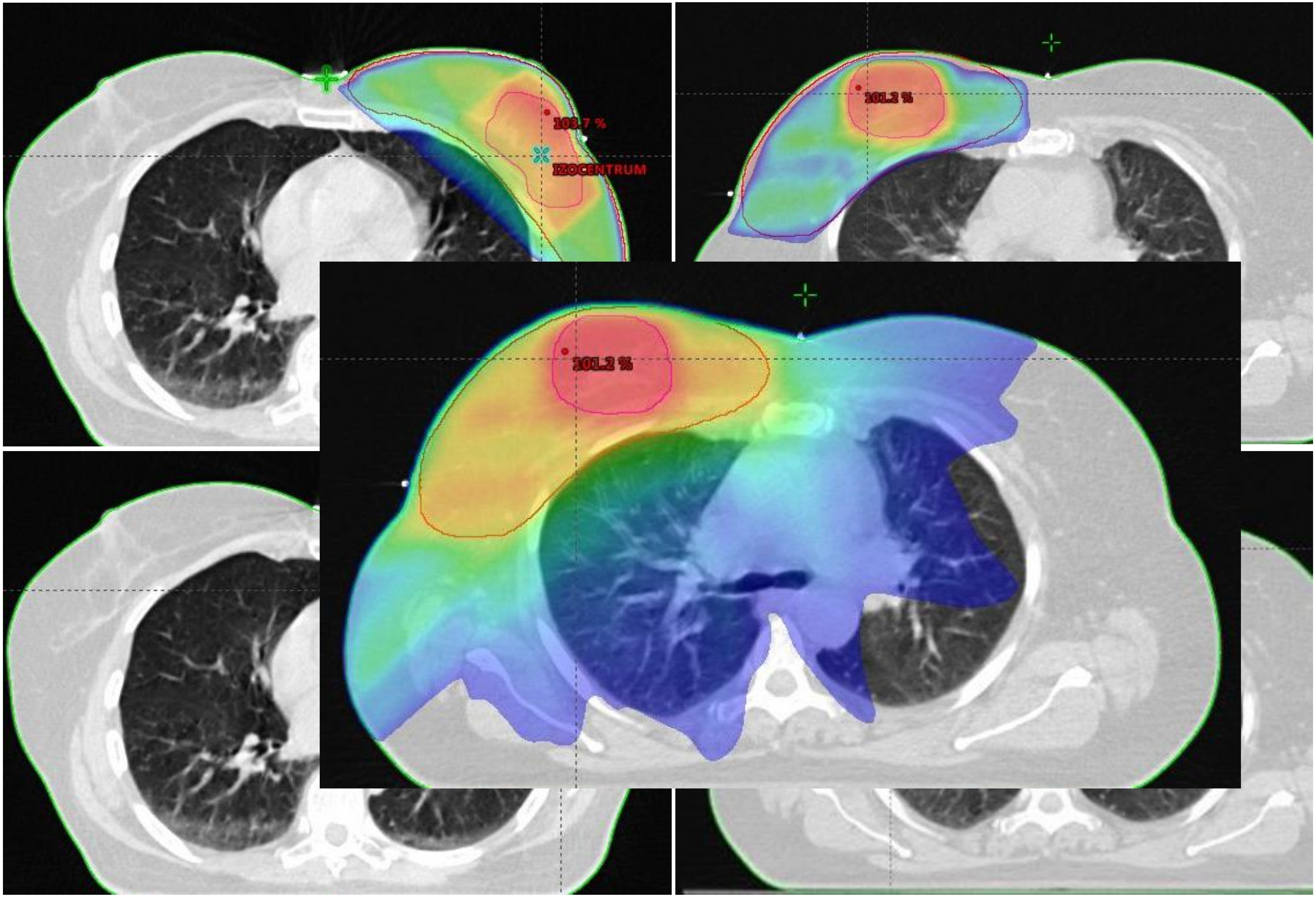
Beams with <u>non-uniform intensity</u> from various directions can provide:

- <u>dose homogeneity</u> to PTV <u>similar</u> to conventional radiotherapy - homogeneity
- <u>superior conformity</u> (concave structures) -> better OAR's sparing - conformity
- Non-uniform deposited dose distribution within various structures (<u>SIB - Simultaneously Integrated Boost</u>)

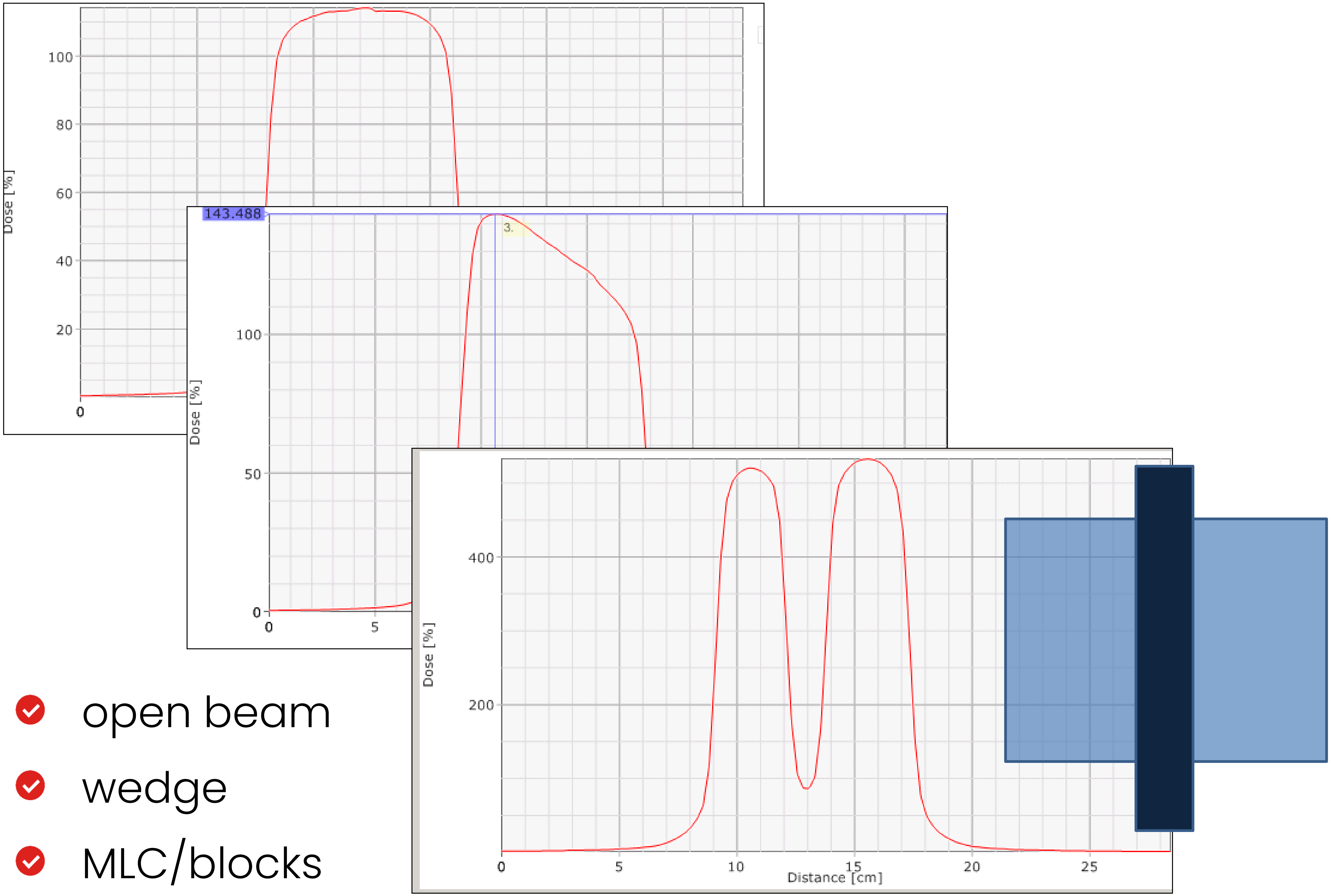



Brahme 1987, Cormack 1987

# 3D CRT vs. VMAT



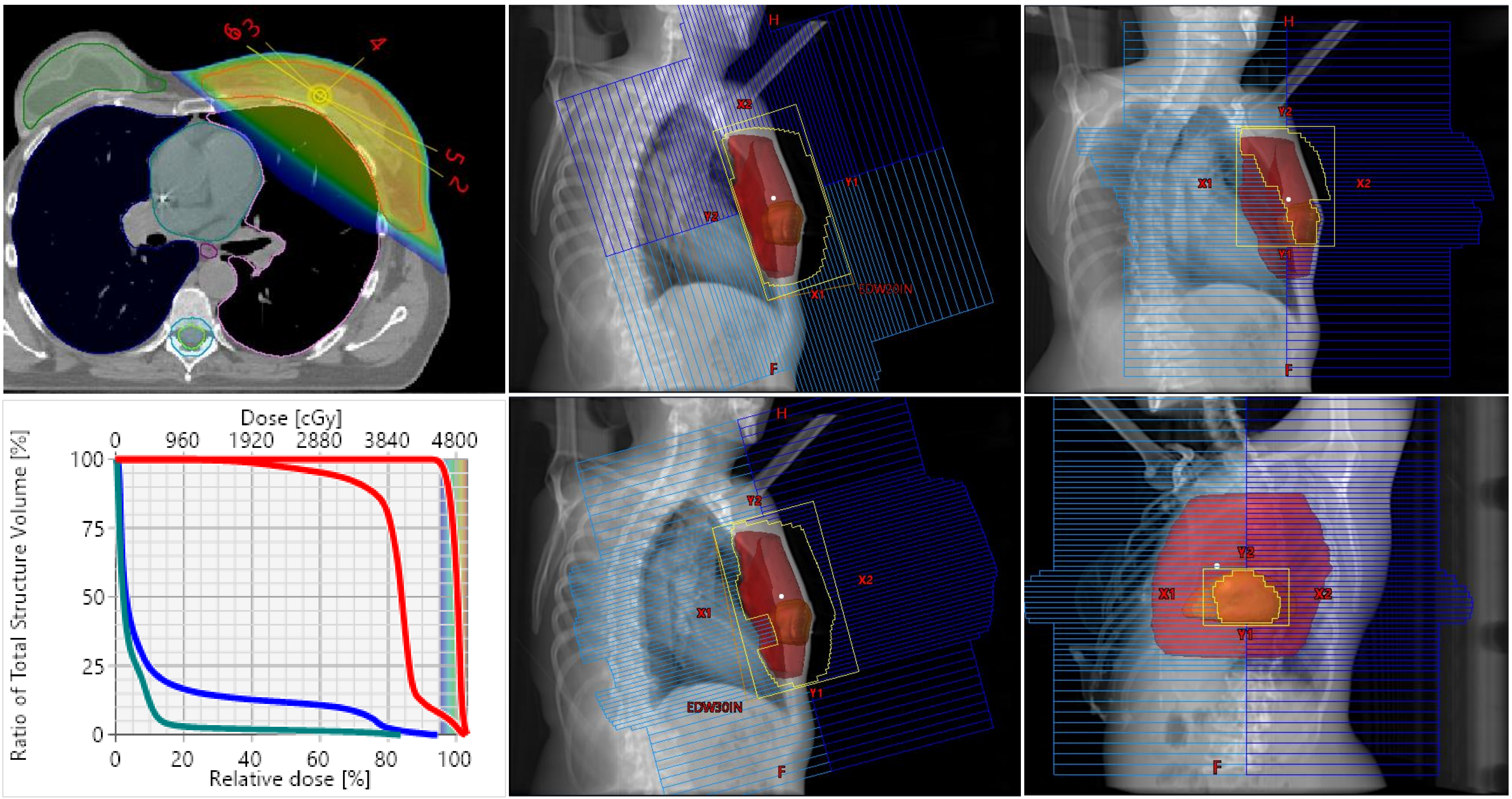
# Static field – what is achievable?



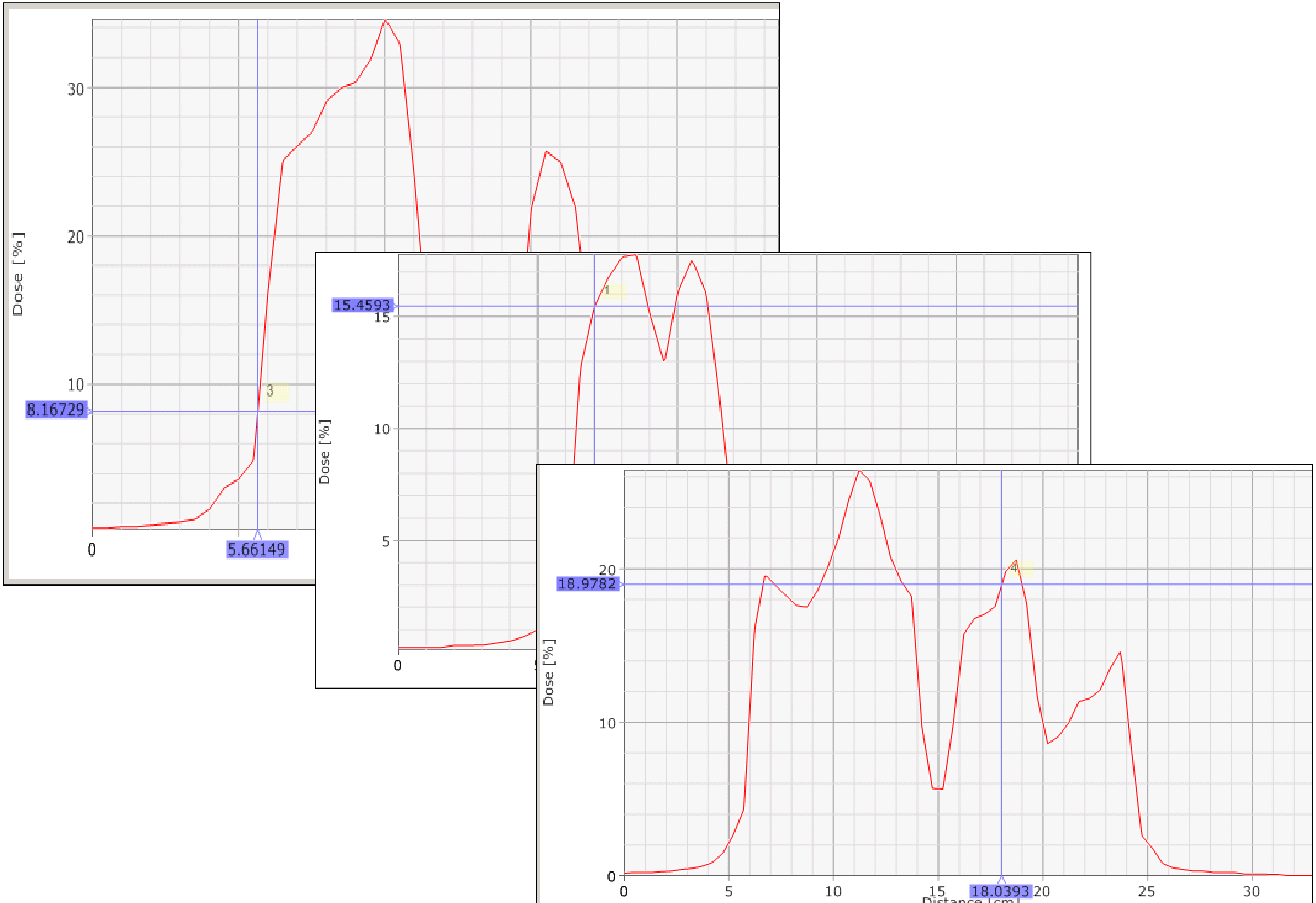

- open beam
- wedge
- MLC/blocks

# Static field - what is achievable?

Field-in-field technique (multisegment plan)



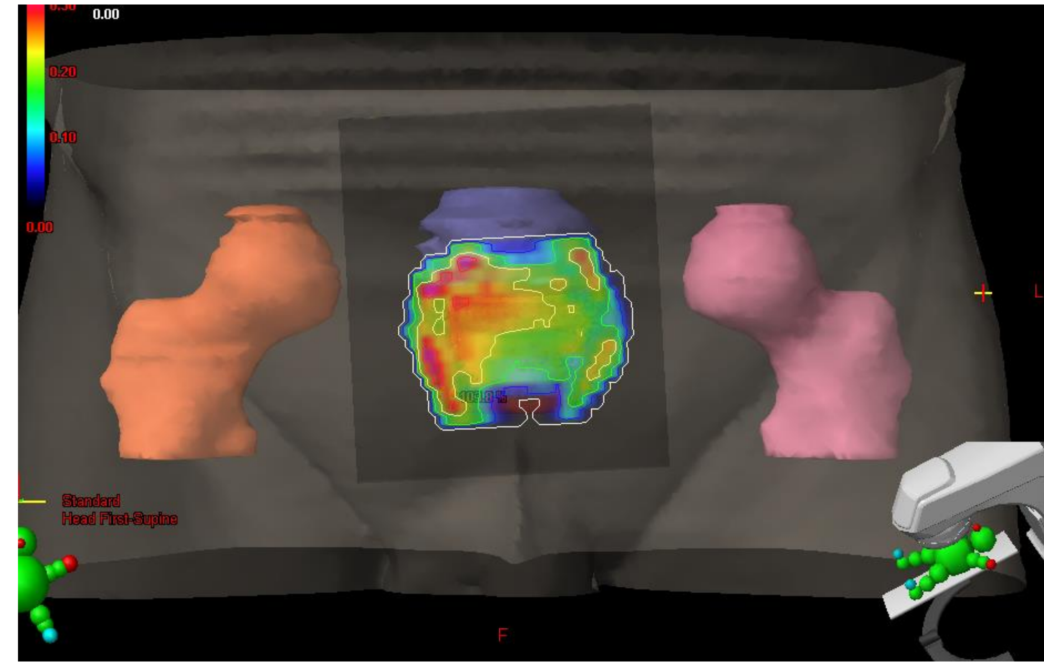
# Dynamic field – what is achievable?

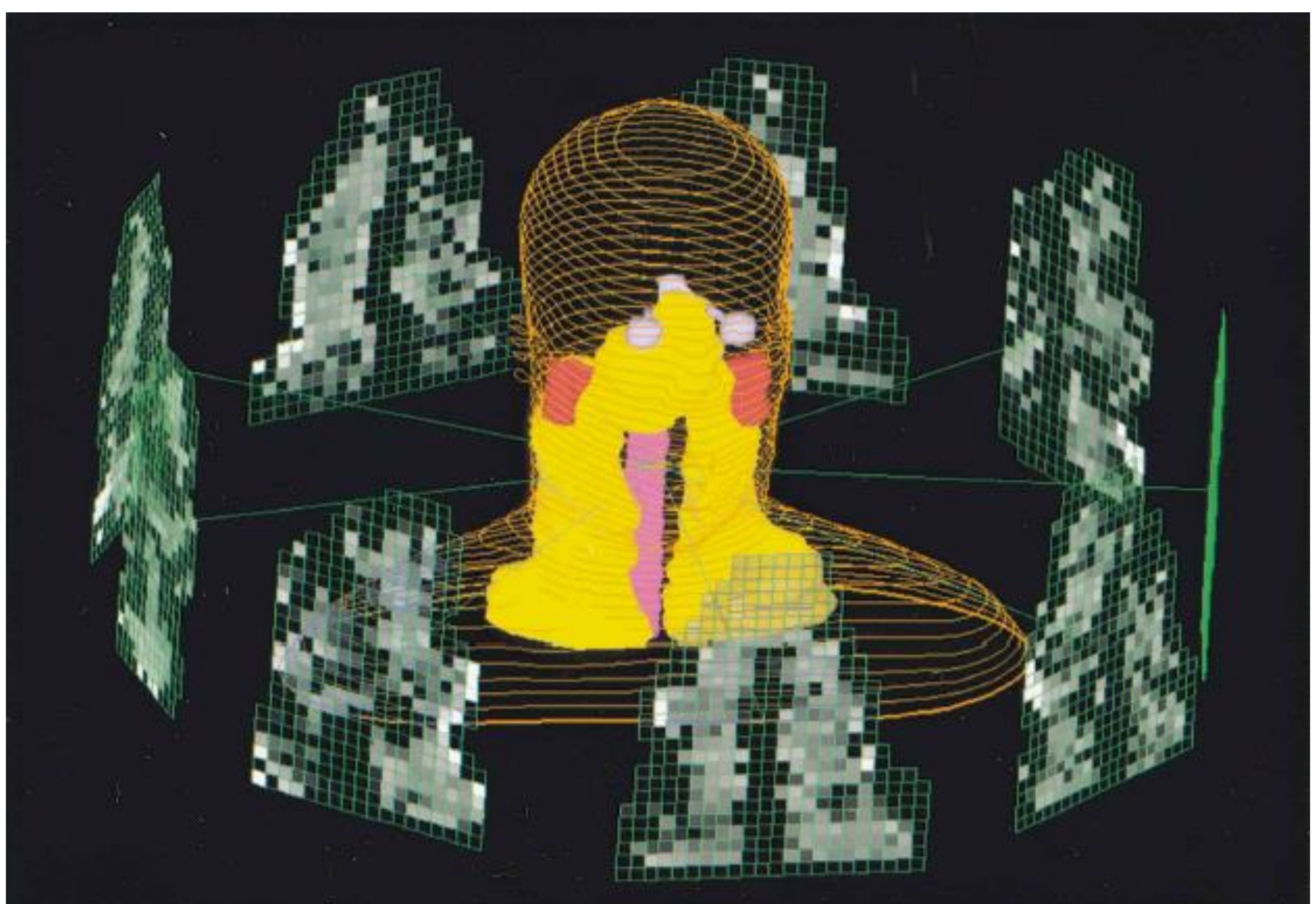


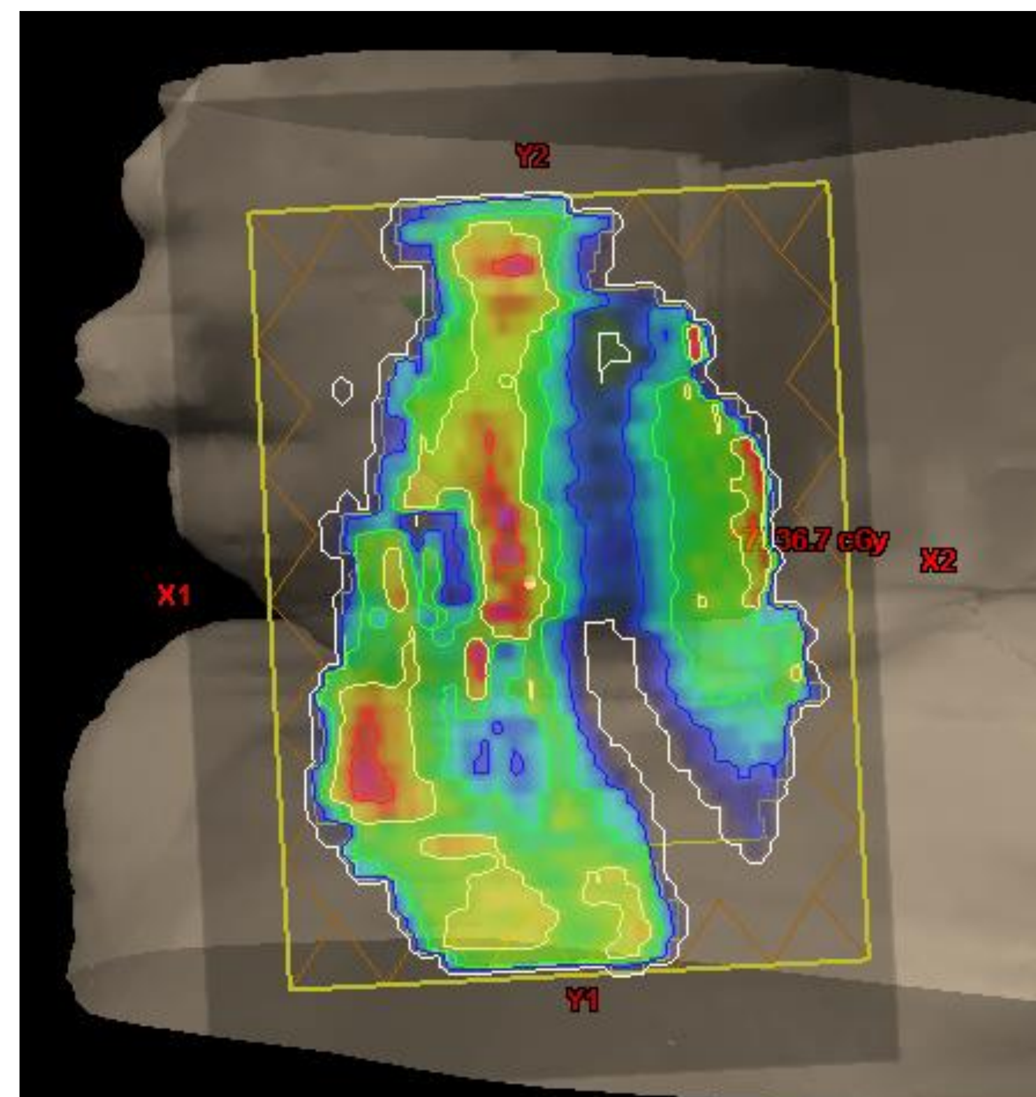
# Fluence maps

Subdividing beam into small segments (beamlets) with varying intensity

**fluence map**







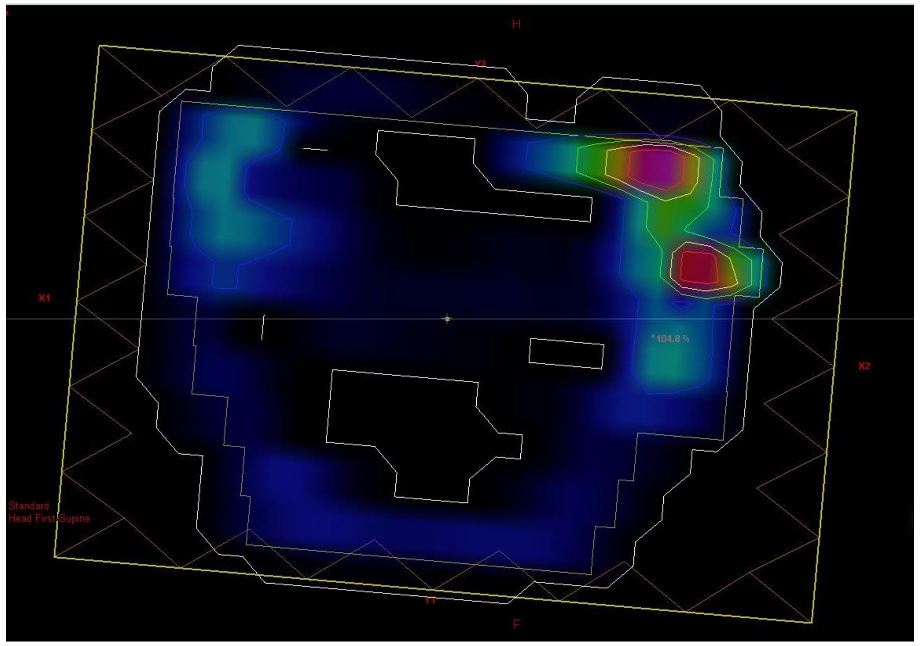
# Fluence maps

Subdividing beam into small segments (beamlets) with varying intensity

↓

**fluence map**

A mathematical solution is not always a clinically acceptable one.



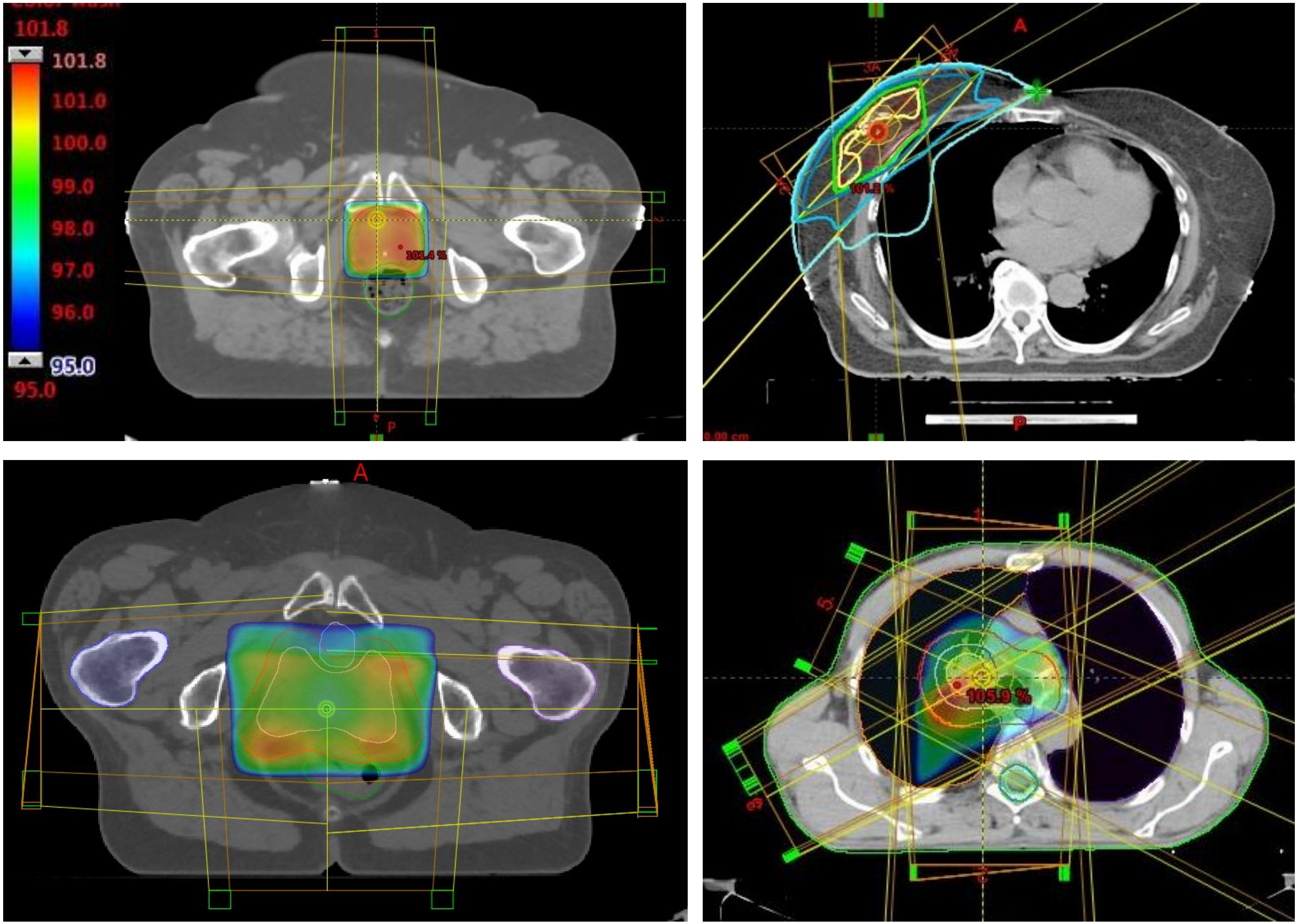

# Forward vs. Inverse

## Forward Planning – 3D Conformal static plan

- Defining the beam geometry
  (number of fields, beam angles, collimator angles …)
- Defining field shape and dose modifiers
  (MLC, wedges, bolus)
- Defining field weights
- Calculation of dose distribution
- Plan evaluation
  isodose distribution, DVH, TCP, NTCP
- If the distribution is not accepted - plan modification
  (modifiers, weight, geometry…)

If the dose distribution is unacceptable, we manually modify geometry/accessories/weights.

# Forward Planning – 3D Conformal static plan



# Forward vs. Inverse

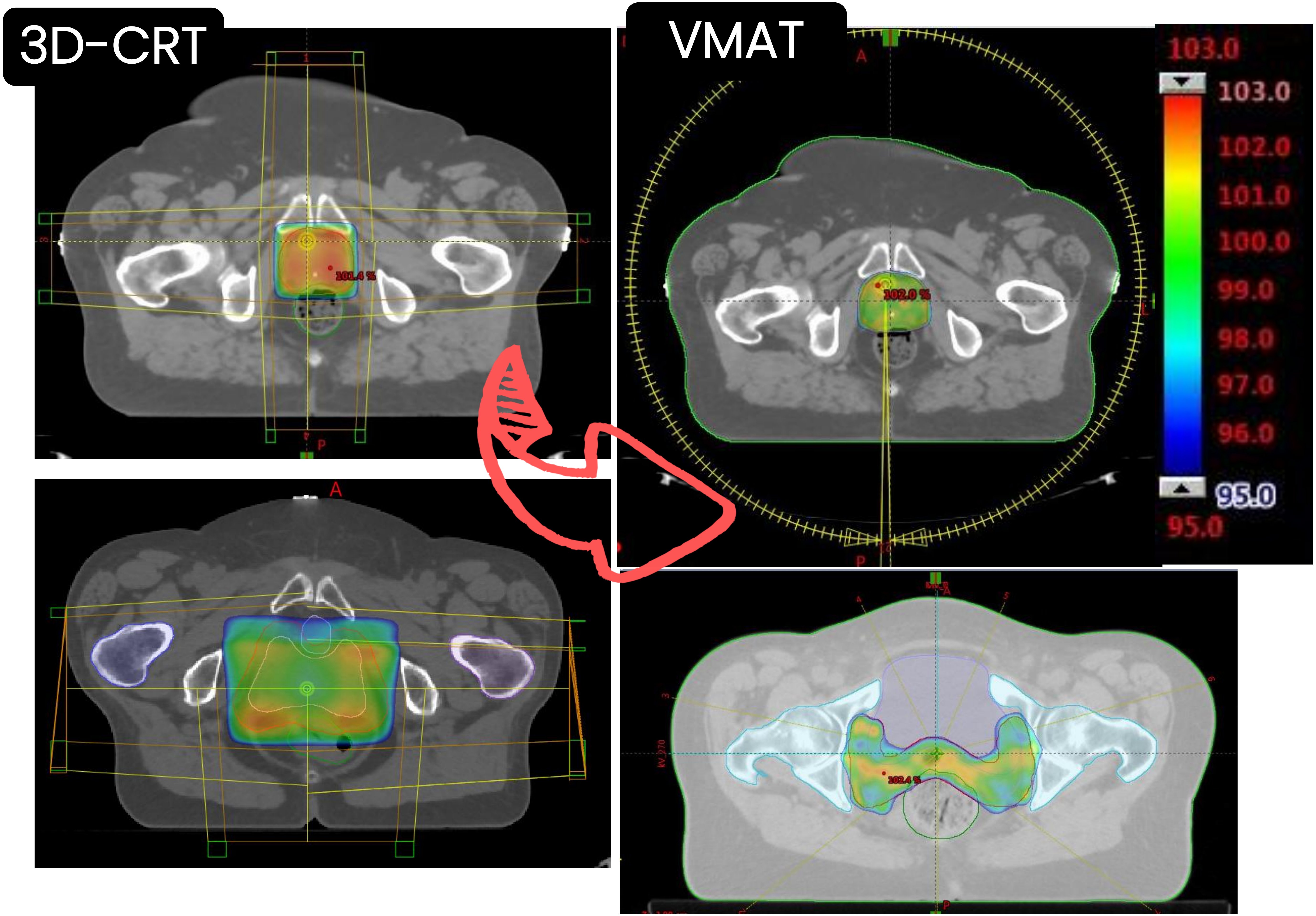
## Inverse Planning – dynamic plan

- Defining the beam geometry
  number of fields, beam angles, collimator angles, etc.…
- Defining the goal - the expected dose distribution
- Optimization – computer calculations adjust beam parameters (intensities of individual pixels) to the set requirements.
- Plan evaluation
  isodose distribution, DVH, TCP, NTCP
- If the distribution is unacceptable – <u>constraints modification</u>

**If the dose distribution is unacceptable, we modify the optimisation criteria.**

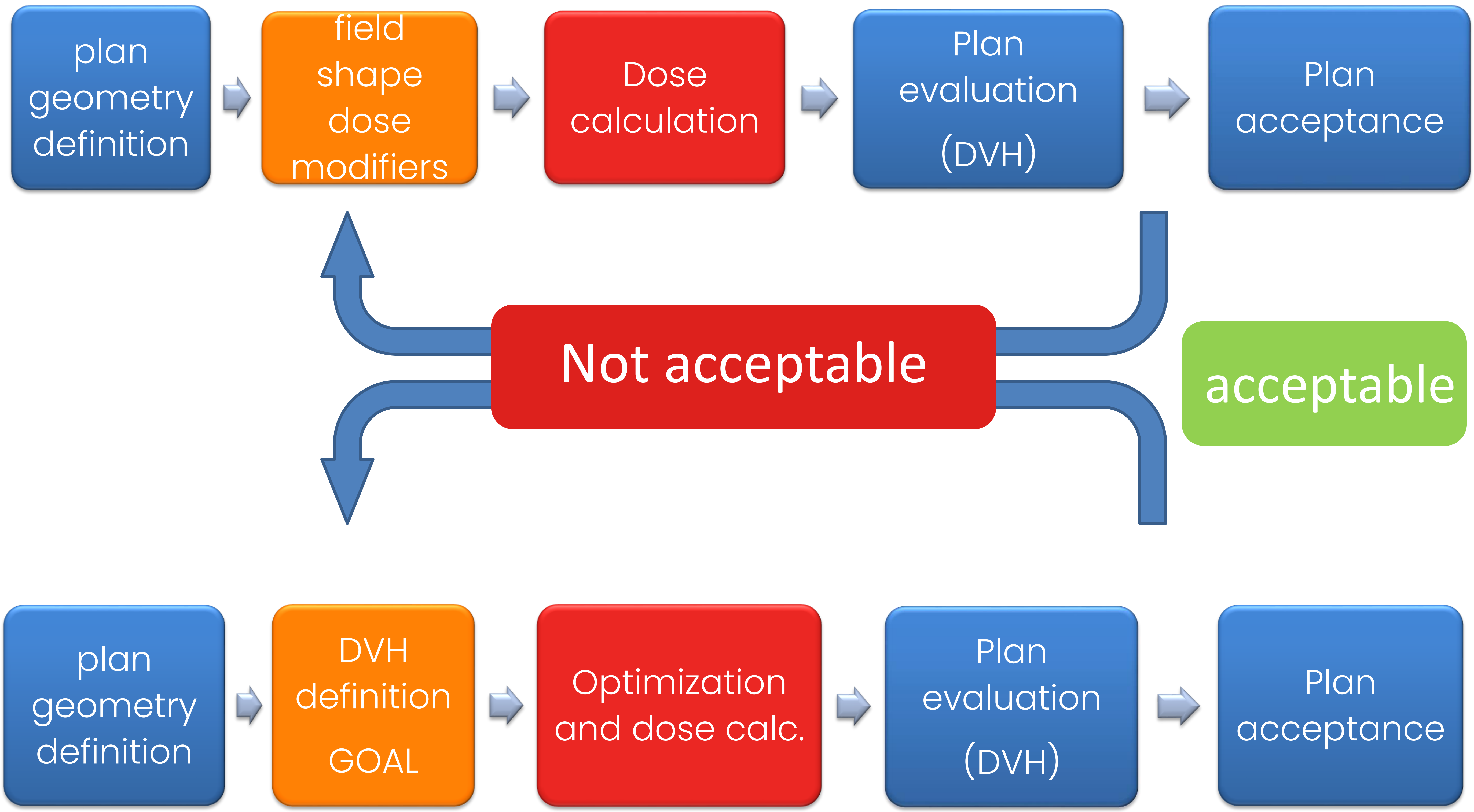
*A problem similar to image reconstruction, the signal process*

# **Inverse Planning** – dynamic plan



# Forward vs. Inverse planning

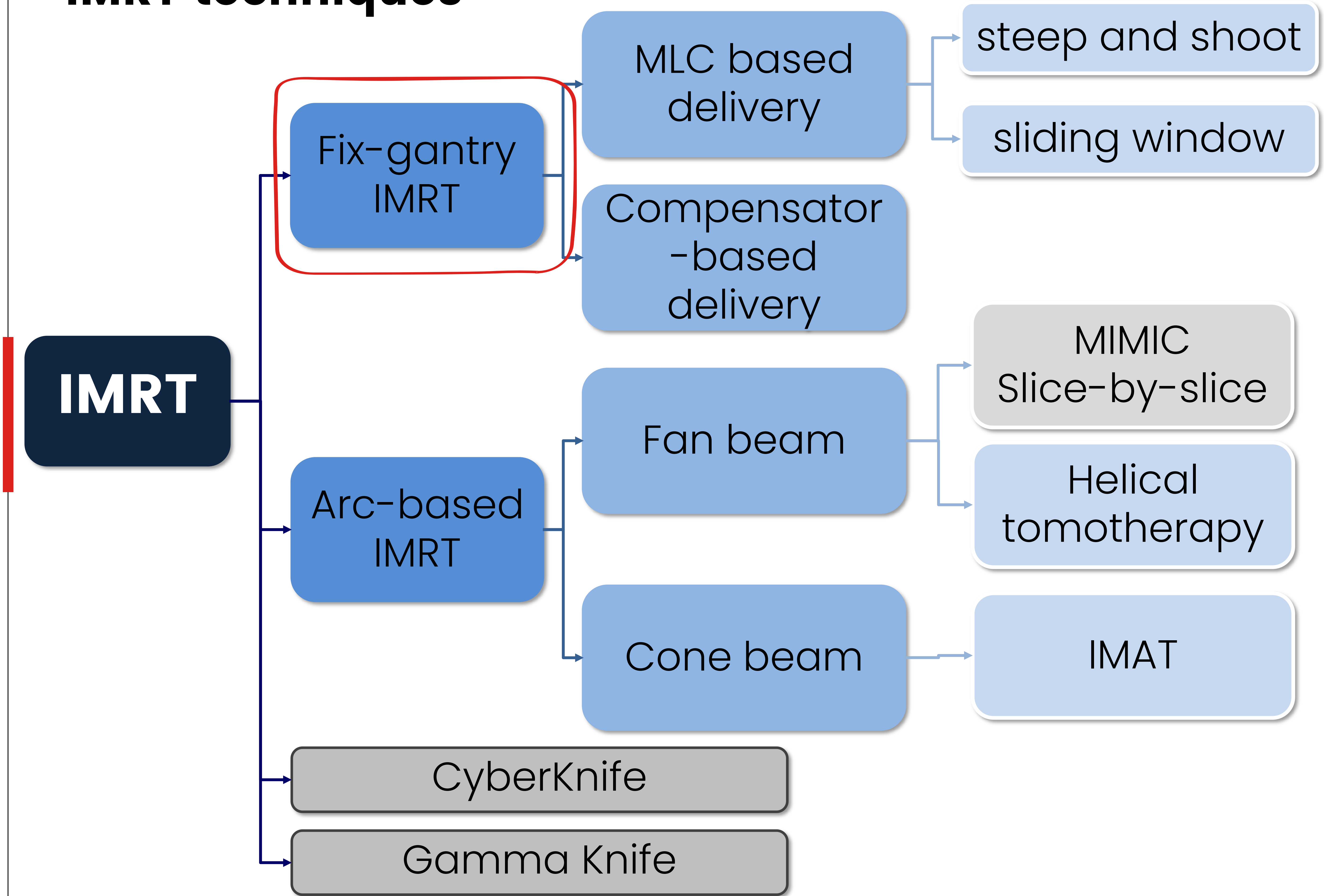
Looking for the desired dose distribution




Looking for the desired beam parameters

# DELIVERY METHODS

# IMRT techniques



# Fix-gantry IMRT - Physical modulators

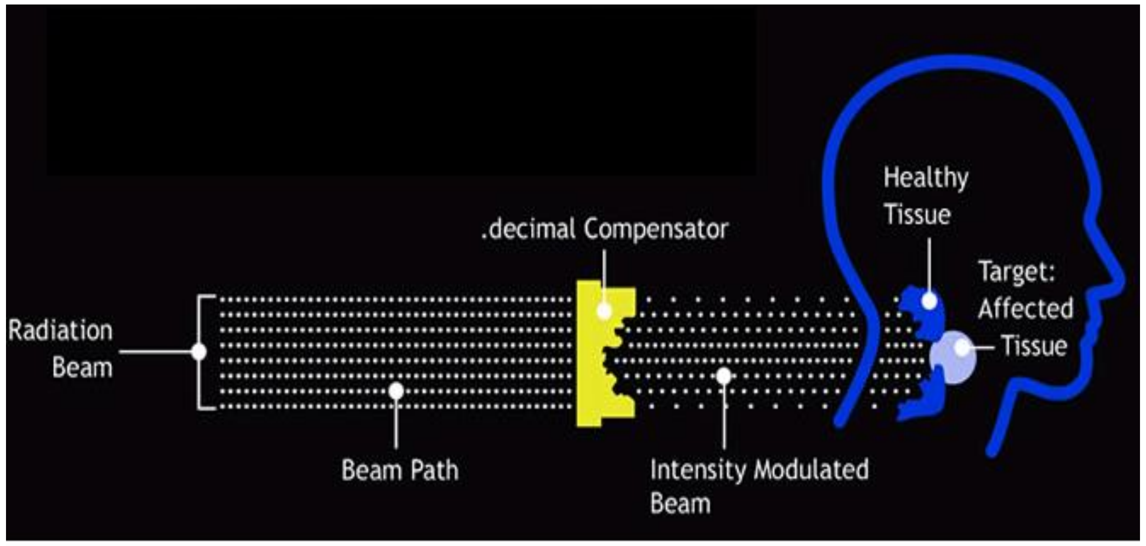
Made of brass or aluminium, often outside the centre.
Accuracy ± 0.25mm

<u>Advantages:</u>

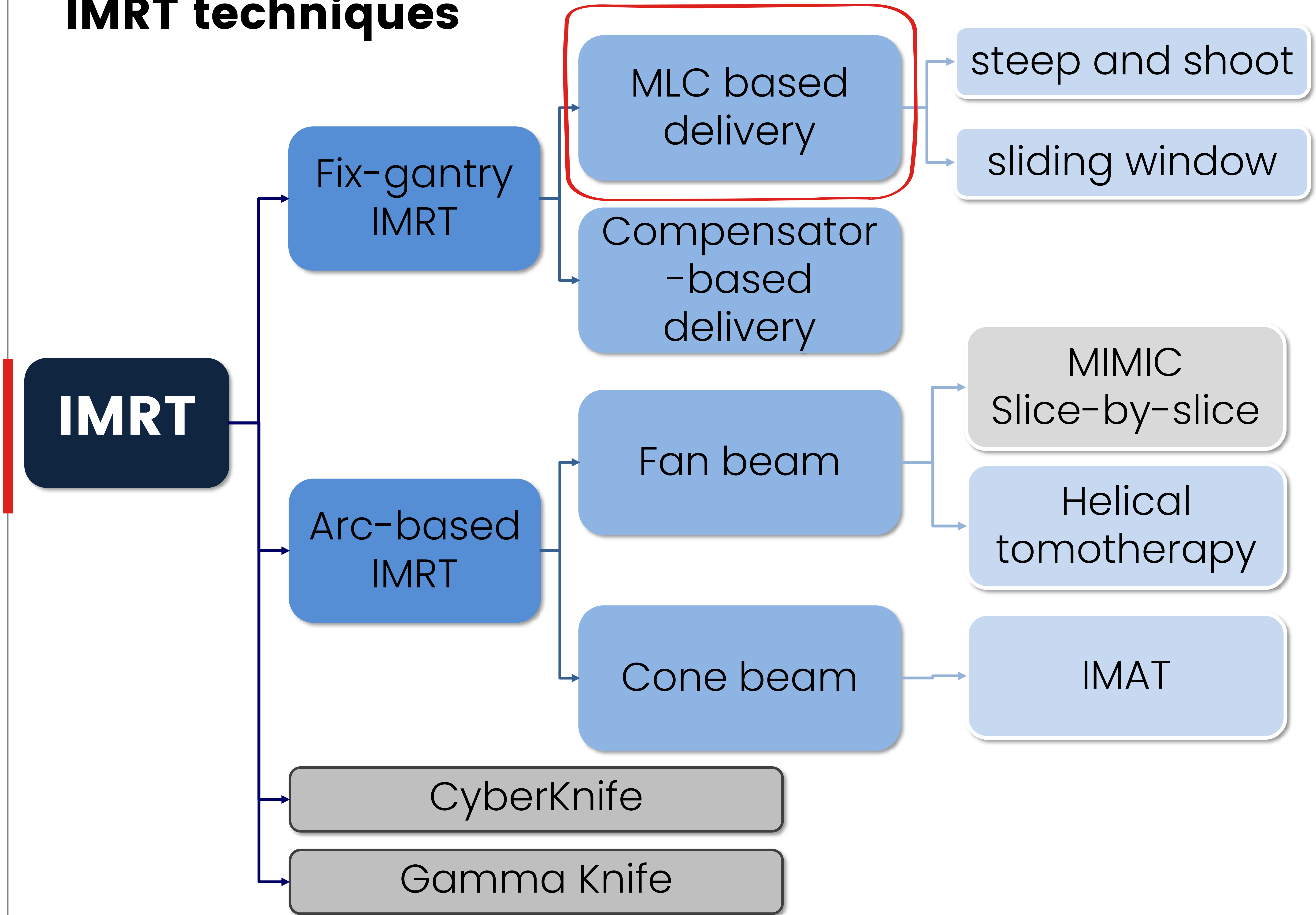
- high resolution;
- simple quality control;
- no connection problems (e.g. leaves transmission)
- fewer monitor units.

<u>Disadvantages:</u>

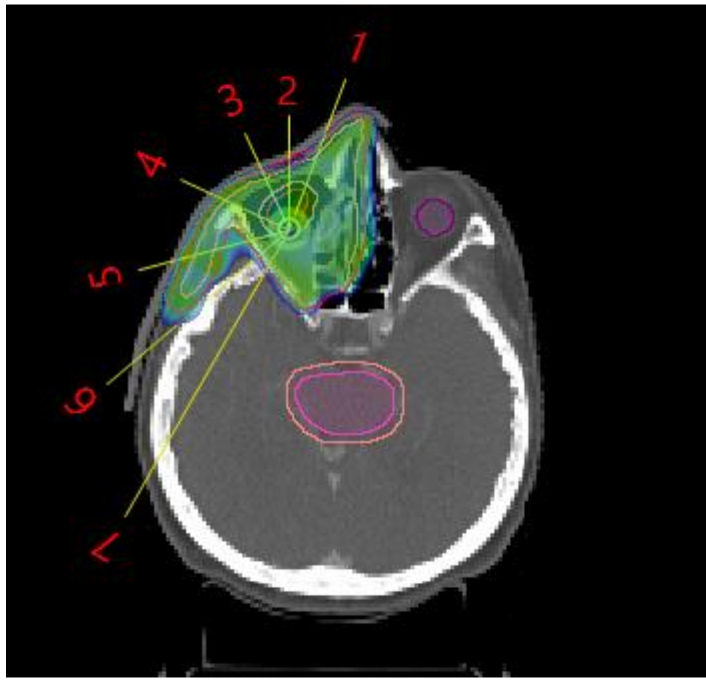
- preparing requires time
- entrance to the bunker is required between each field
- more radiation scattered outside the field.



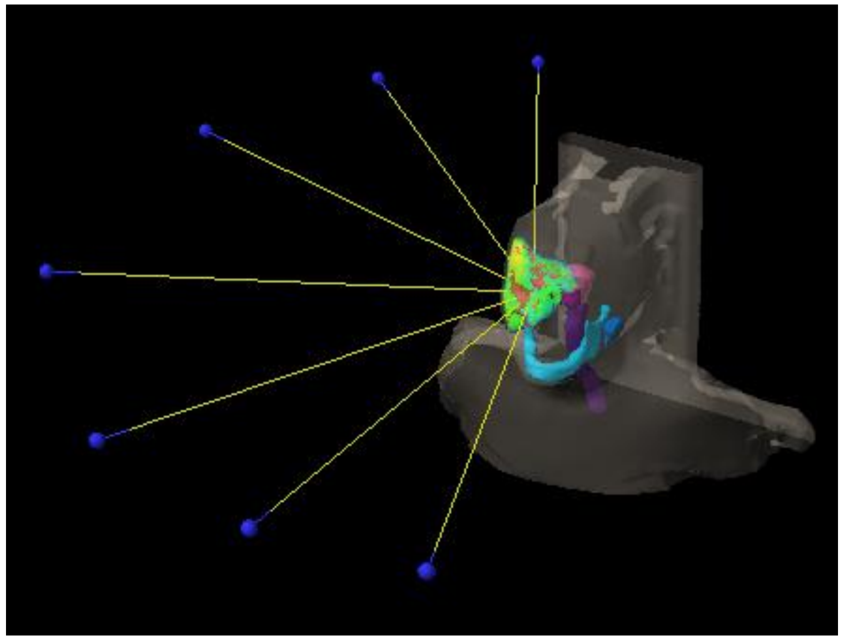
# IMRT techniques

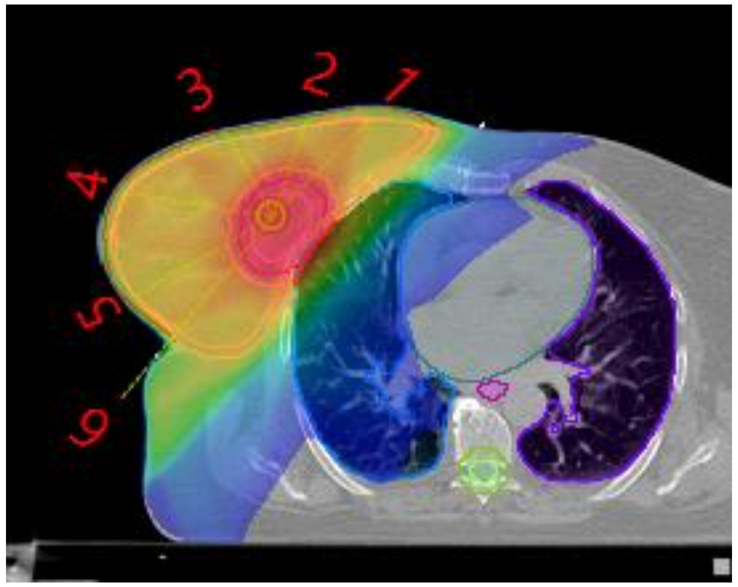


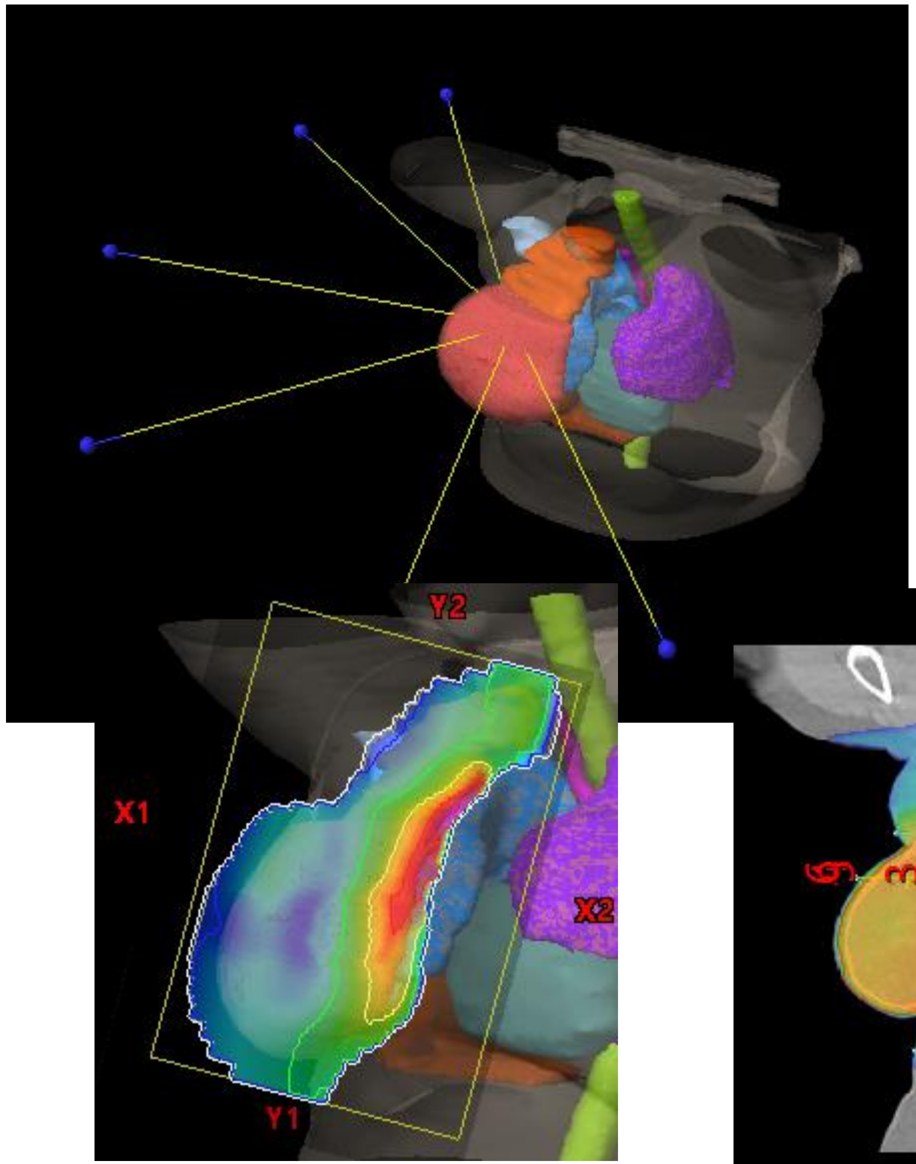
# Fix-gantry IMRT – MLC-based delivery

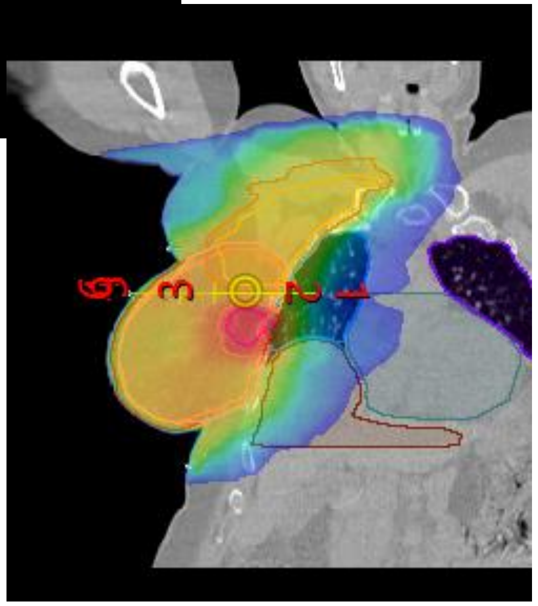


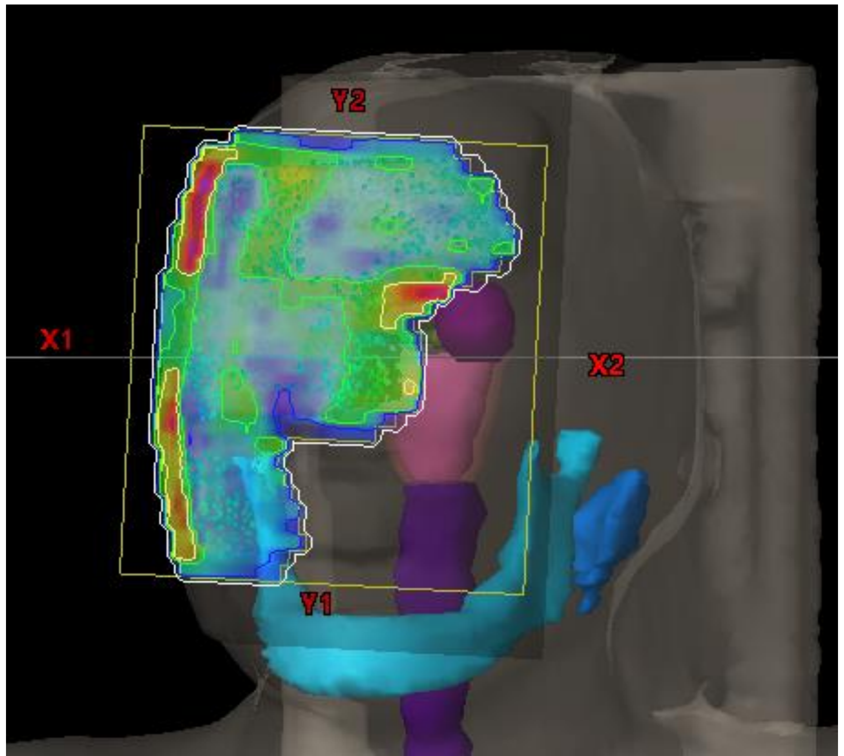

- Fixed beam directions
- Beams divided in a grid of beamlets – fluence maps (beamlet resolution: width = MLC leaf width, length – user-defined)







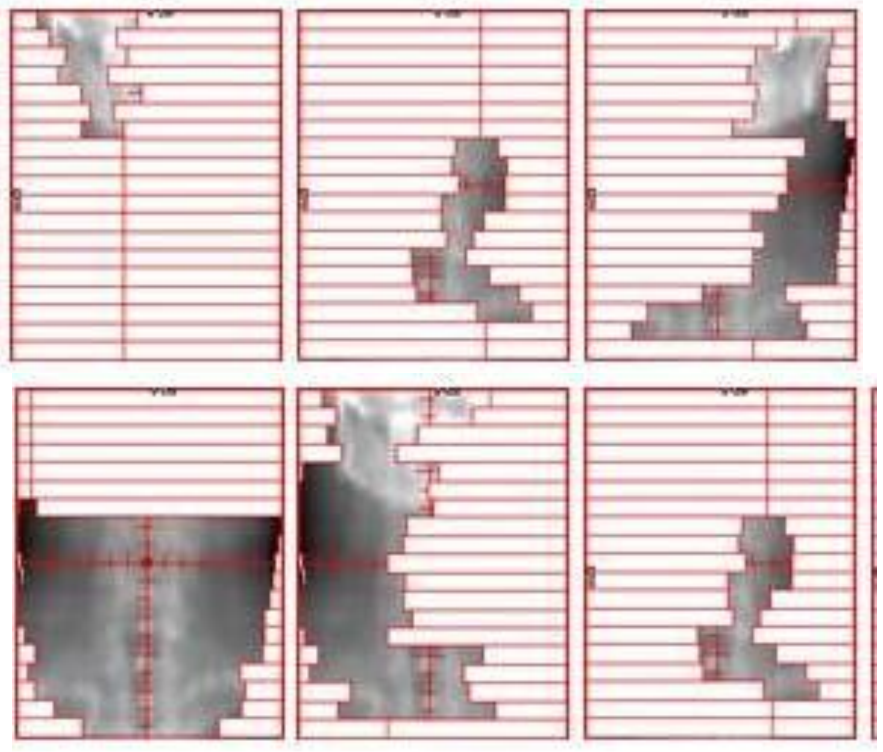
# Fix-gantry IMRT – MLC-based delivery

## Segmental MLC (Step-and-shoot)

**The shape** of the field remains **constant** at a fixed beam angle and when the beam is on.

Bema's shape changes when the beam is turned off.

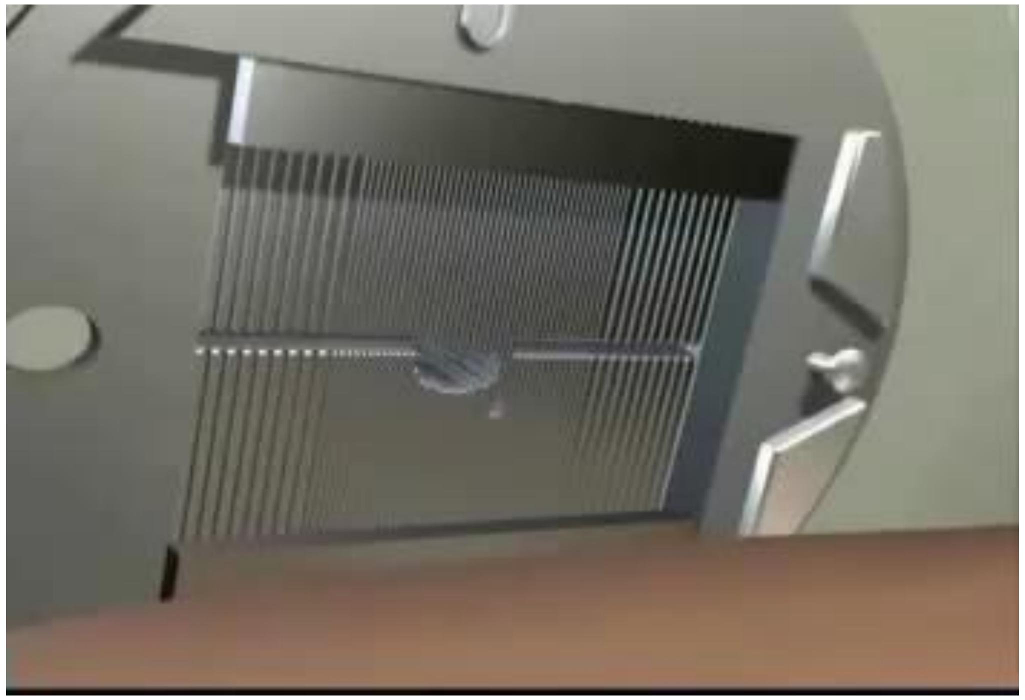
The planned fluence map is decomposed into a set of fields (segments, subfields).



## Dynamic MLC (sliding window)

Pairs of <u>opposing leaves move across the field</u> at a fixed beam angle and the beam is on.

Leaves move at a variable speed as a function of time.



# Fix-gantry IMRT – MLC-based delivery
## Optimization approaches

### Beamlet optimization

1. Field is divided into subfields of different intensity -> optimal fluence
2. Leaf segmentation -> actual fluence

convergence error – during optimization, simpler calculation algorithm (time) - not take into account physical limitations, MLC parameters (DLG, LT, tongue and groove effect, rounded leaf ends, penumbra, minimum MU …)

### Aperture-Based Optimization

1. The initial field shape (output aperture) PTV projection is defined
2. Modification - additional apertures added

no convergence error (if the same calculation algorithm is used during optimization). Missing leaf segmentation step. Physical MLC parameters are taken into account at the optimization stage.

# Beamlet vs Aperture-based optimization

Table 1.1. IMRT methods. The preferred optimization approaches for each IMRT method are described in Section 2.3.

| Type of method | Intensity modulation method | Preferred optimization approach |
|---|---|---|
| Compensators | A beam filter designed to provide a patient-specific intensity pattern designed by an optimization procedure | Optimized beamlets |
| Segmental MLC (step and shoot) | Multiple MLC segments delivered from each treatment direction | Direct-aperture optimization |
| Dynamic MLC (sliding window) | Leaves slide across the field at different rates | Optimized beamlets |
| Intensity-modulated arc therapy (IMAT) | Leaves move while the gantry is rotating. Can require multiple rotation arcs | Direct-aperture optimization |
| Serial tomotherapy | Gantry rotates around the patient with the couch fixed. Binary leaves modulate a fan beam. Upon completion of each rotation, the couch is moved in a step-wise fashion | Optimized beamlets |
| Helical tomotherapy | Gantry and couch move synchronously. Binary leaves modulate a fan beam | Optimized beamlets |
| Robotic radiotherapy | Multiple non-coplanar pencil beams delivered by a robot | Optimized beamlets |

# convergence error – calculations algorithms

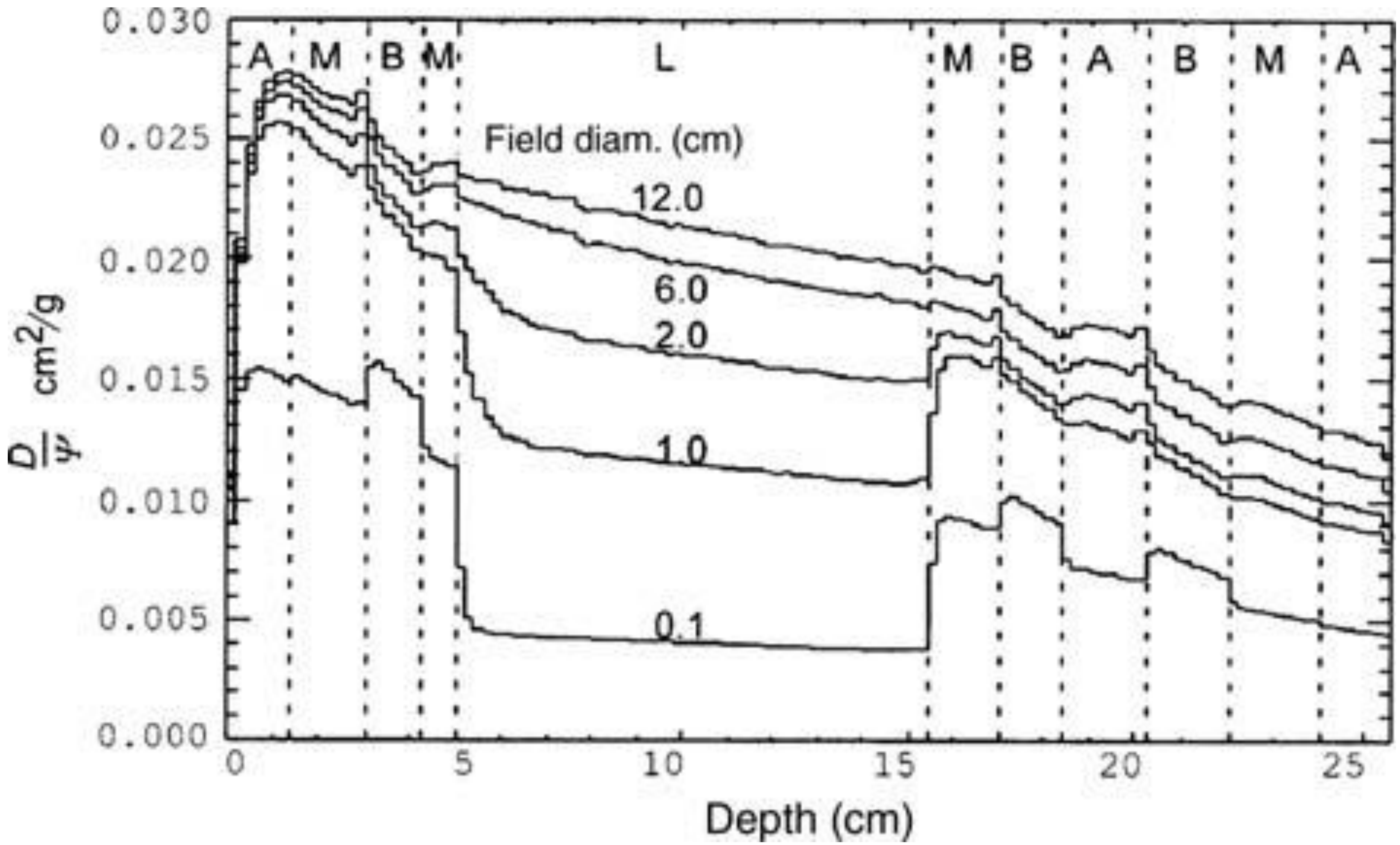
Perturbations in the absorbed dose

- Range of electrons in water from 0.3 – 2.5 cm, increases in areas with low density (lungs)
- IMRT fields comparable/smaller in relation to the electron range - impact on the absorbed dose
- Perturbations in the absorbed dose increase with energy
- Algorithms are required that can take into account the above effects

# convergence error – calculations alghoritms

Influence of field size on the depth-dose in a heterogeneous phantom

tissue (A), muscle (M), bone (B), and lung (L)




ICRU 83

## Dose calculations

1. Introducing simplifications and approximations when calculating the dose distribution (used to calculate the objective function value) during optimization.

2. Not taking into account realistic limitations related to the implementation of a given fluency

3. Limited number of calculation points

4. A way to take into account the heterogeneity, build-up area.

Attention should be paid to simplifications included in dose distribution calculations in optimization algorithms

# Leaf segmentation – I method

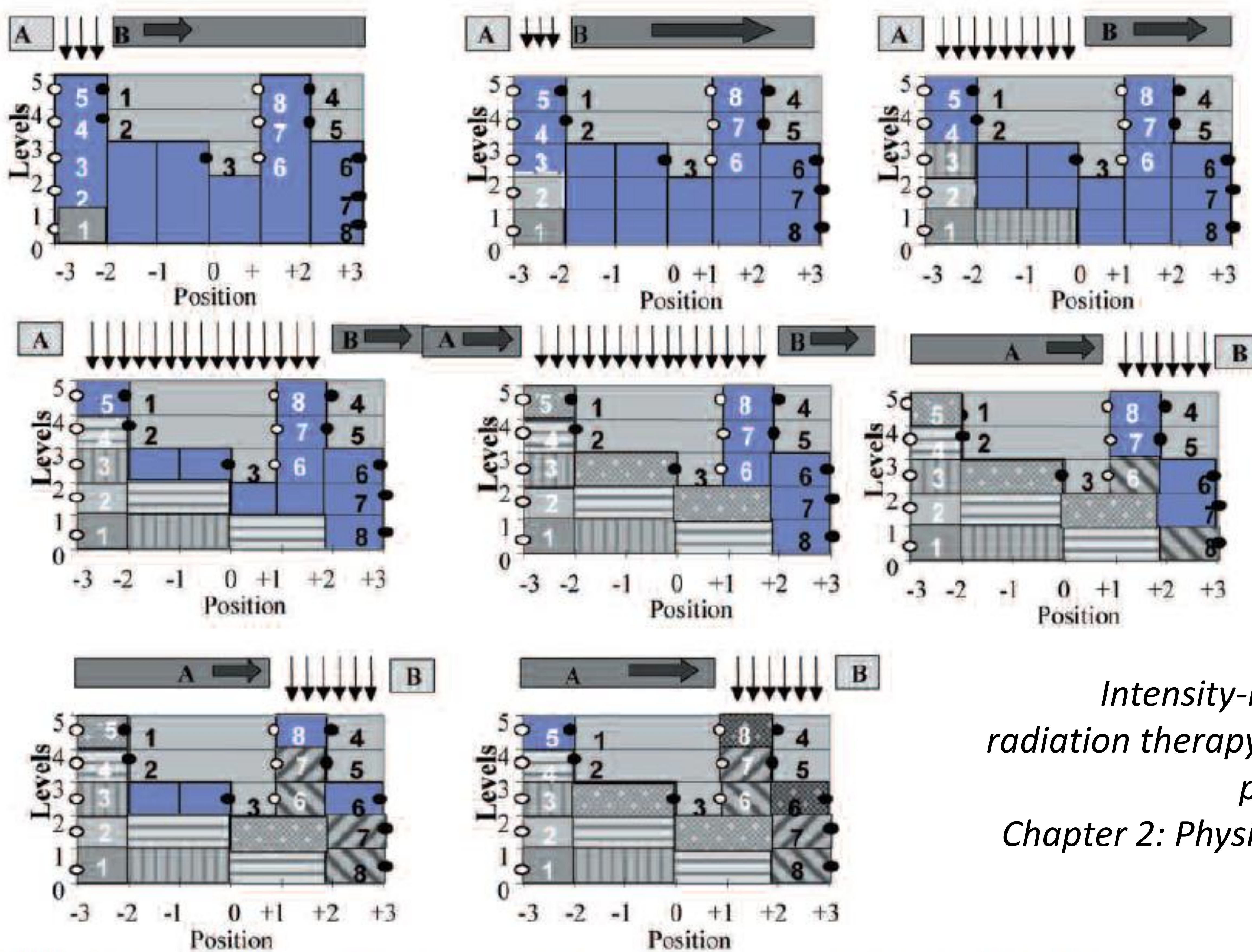


FIGURE 2-15. Intensity profile to be produced by leaf pair 12. Reproduced with permission from Van Dyk J and Purdy JA.[147]

*Intensity-modulated radiation therapy; a clinical perspective*
*Chapter 2: Physics of IMRT*
*Xing et al.*

# Leaf segmentation – II method

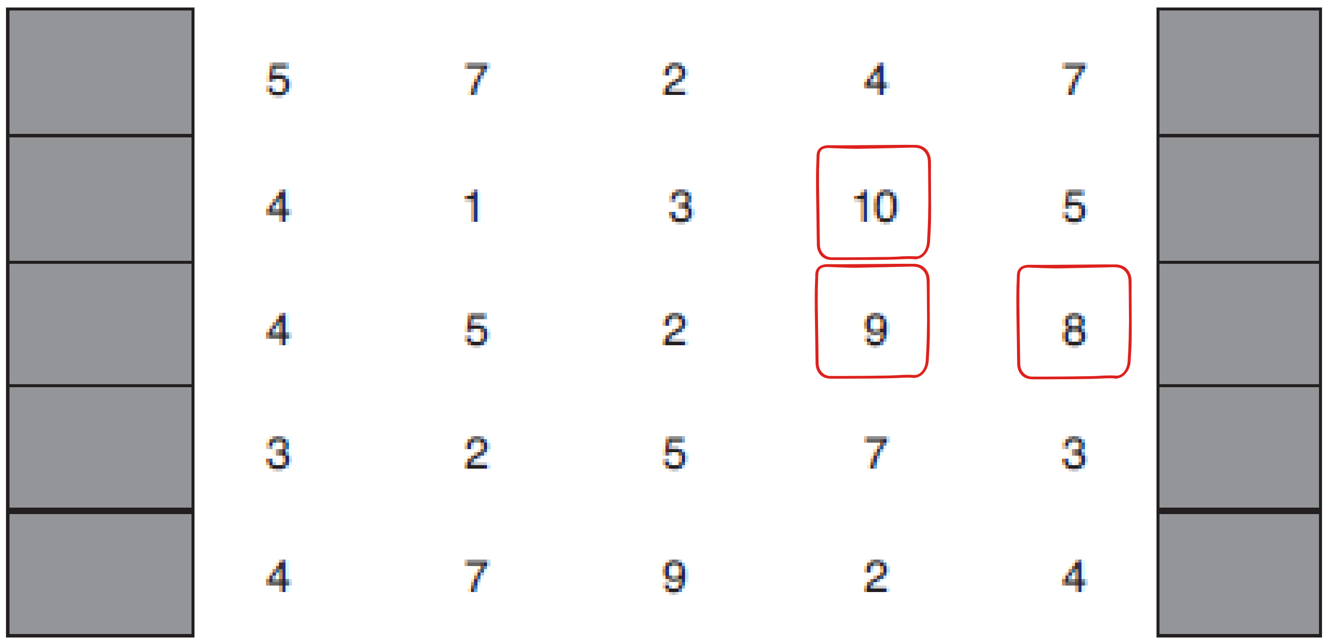
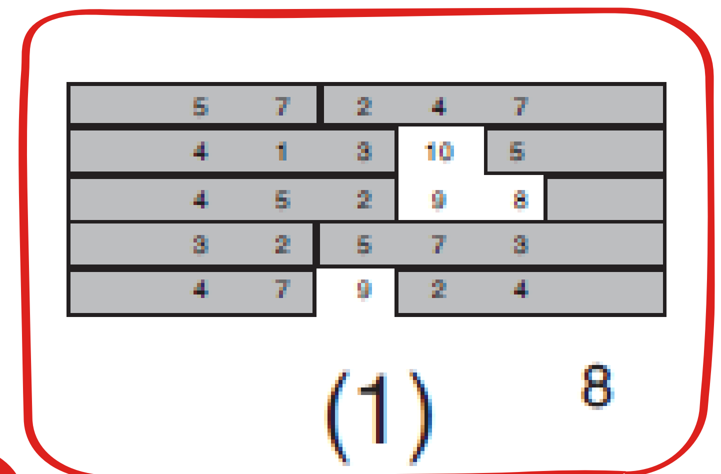
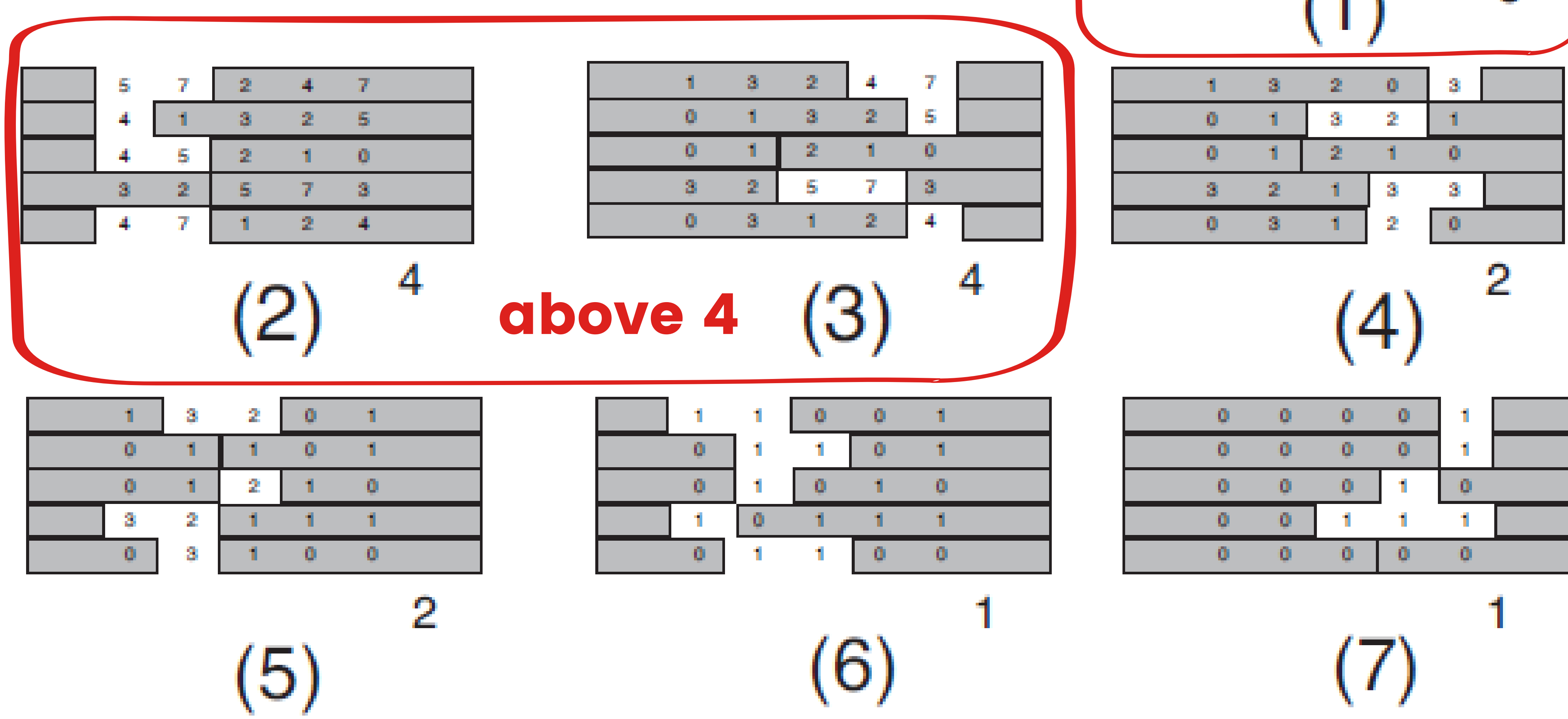



**FIGURE 2-16.** Example intensity map used for illustrating the "areal" leaf sequencing algorithm.

The sequence is to be delivered by increments that are powers of 2.
In this case, the increments are 8, 4, 2, and 1



# Leaf segmentation – sliding window

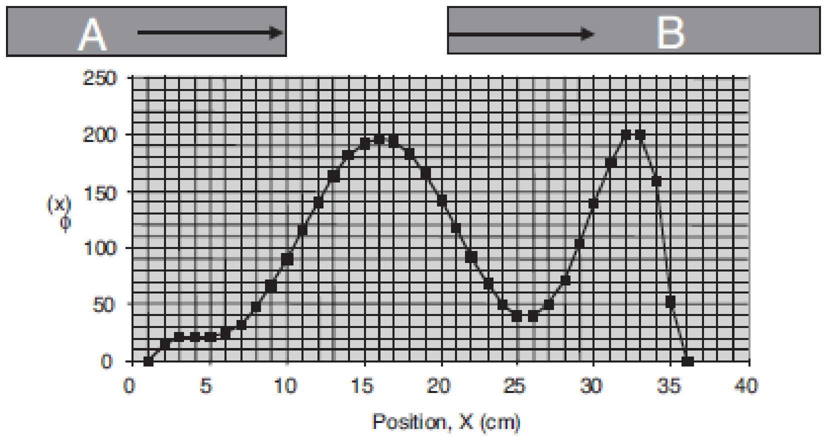



FIGURE 2-18. Example intensity map used for illustrating the dynamic leaf sequencing algorithm.

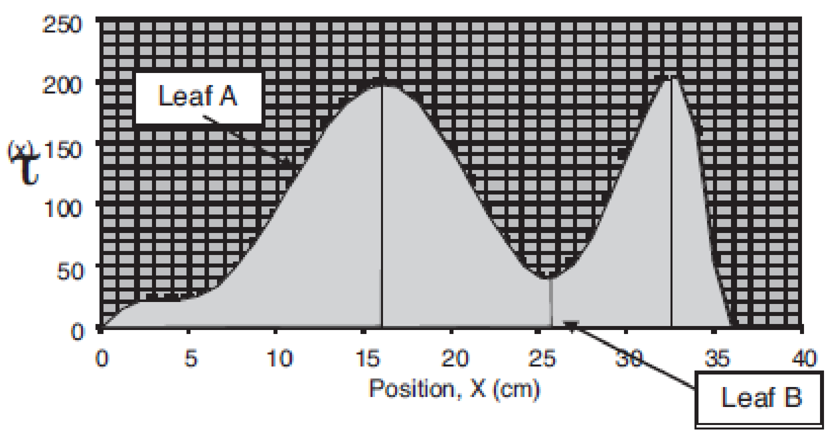



FIGURE 2-19. Time-position graph for the two leaves during the dynamic delivery process. Reproduced with permission from Van Dyk J and Purdy JA.[147]

$$\tau'''(x) = \tau''(x) + x/\upsilon_{max}$$

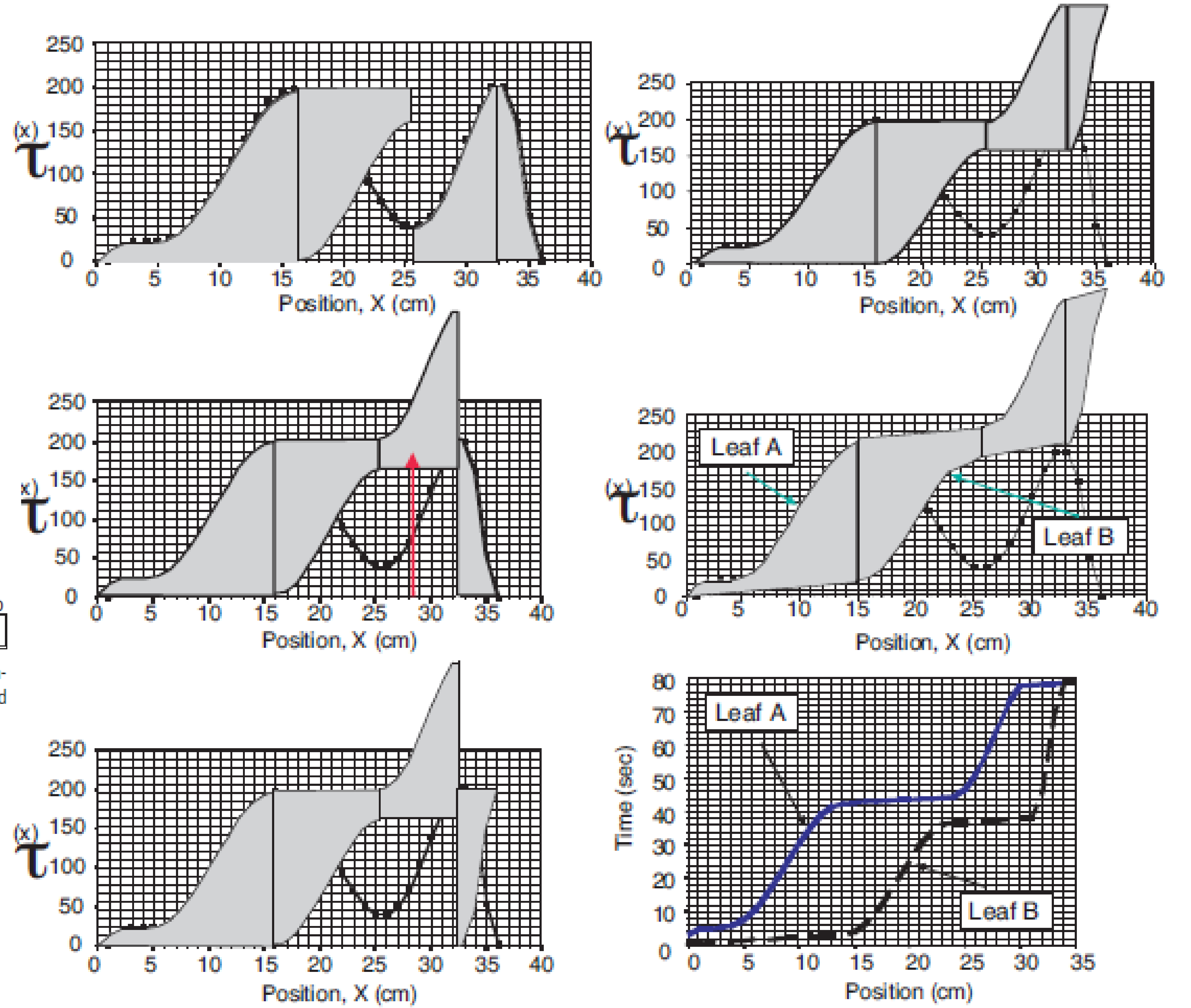
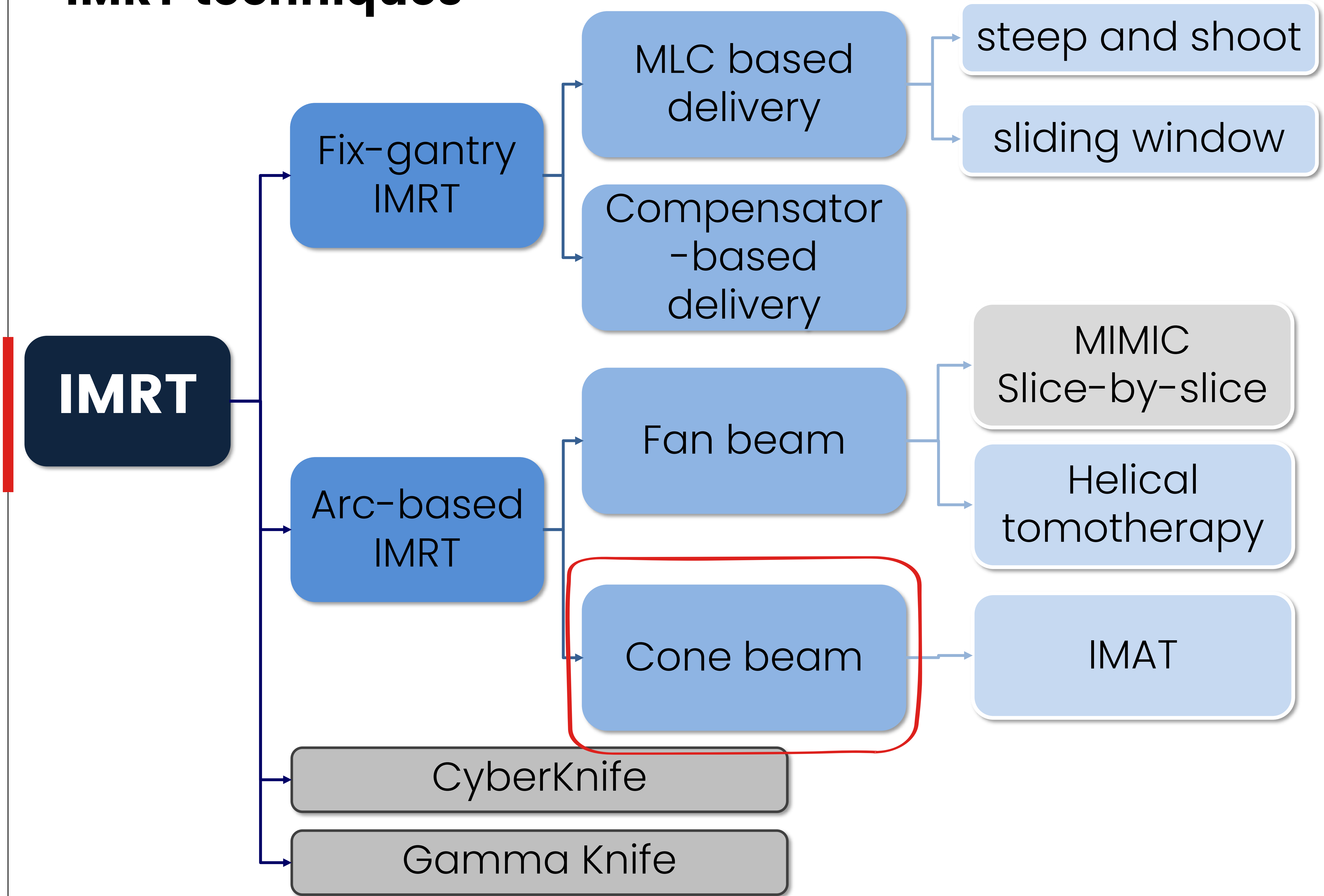



FIGURE 2-20. Trajectories of leaf A and leaf B during the dynamic delivery process. Reproduced with permission from Van Dyk J and Purdy JA.[147]

# IMRT techniques



# Intensity Modulated Arc Therapy

- Changing the shape of the field while rotating the gantry.
- Cone beam
- The modulation complexity is related to (limited by) gantry rotation speed and the leaf movement speed (the shape of the field cannot change infinitely quickly with the head rotation).

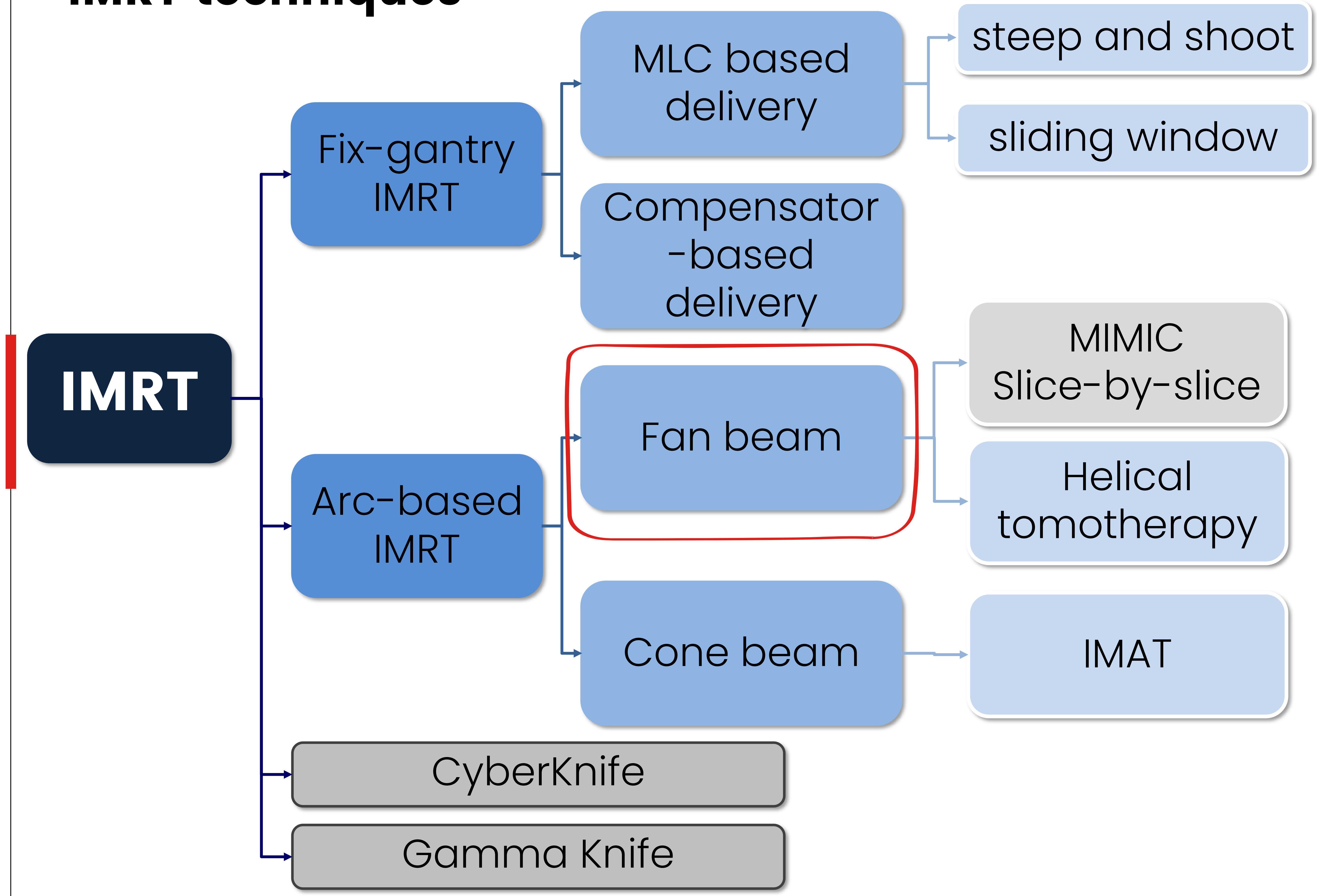
Compared to IMRT:

- better target coverage

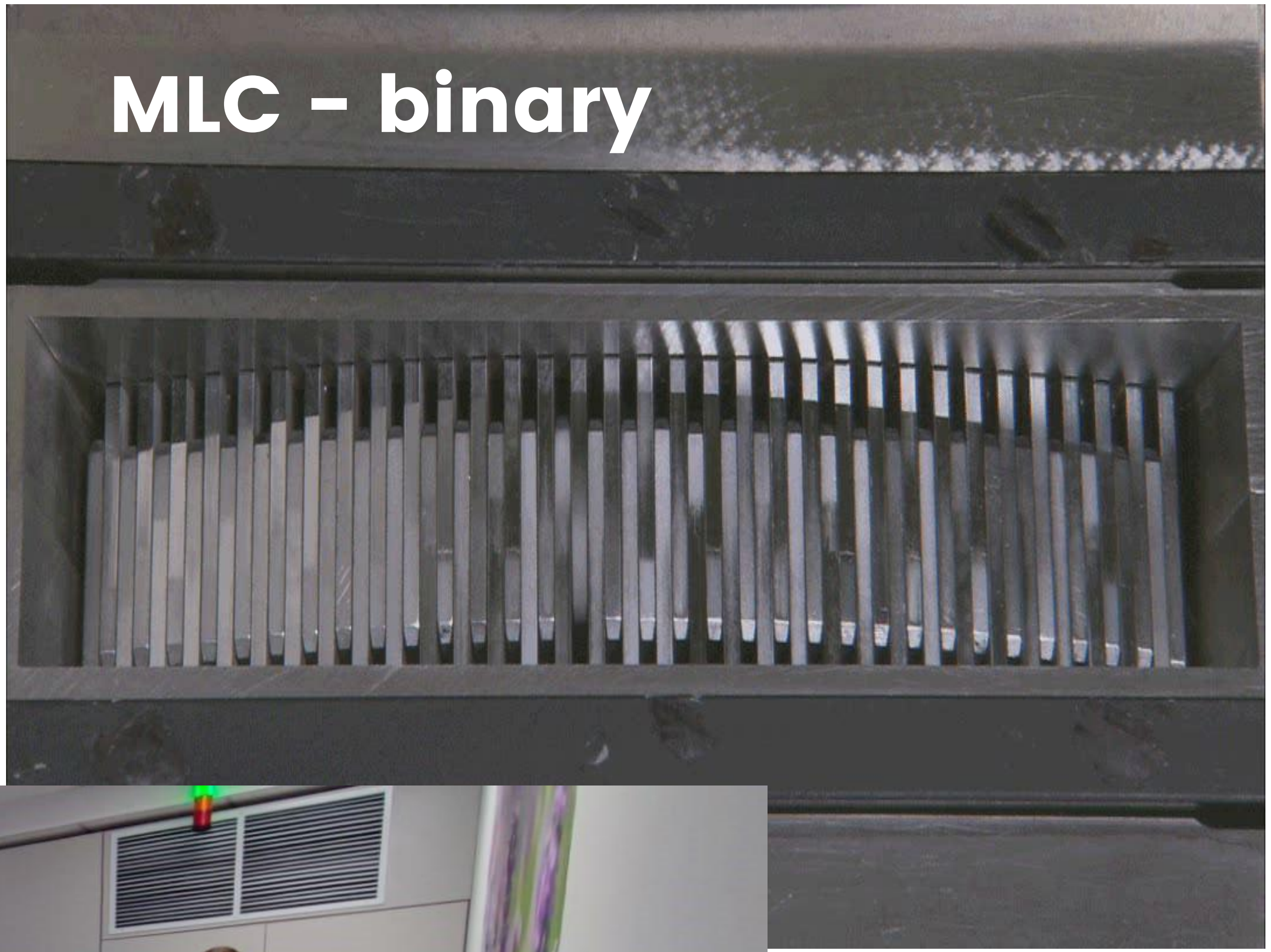
- better protection of critical organs.

- Shorter treatment time

# IMRT techniques



# Serial Tomotherapy


## Serial Tomotherapy

- Helical Fan-Beam, 6MV – FFF (max 850 cGy/min)
- Maximum treatment volume length - 135 cm
- Field size 1.0 cm, 2.5 cm, 5.0 cm x 40 cm (fix)
  1.0-5.0 cm x 40 cm (dynamic)
- 64 binary interlaced leaves
- 0.625 cm leaf widths at isocenter
- Daily 3D MVCT matched with 3D kVCT
- Precision TPS or RayStation TPS

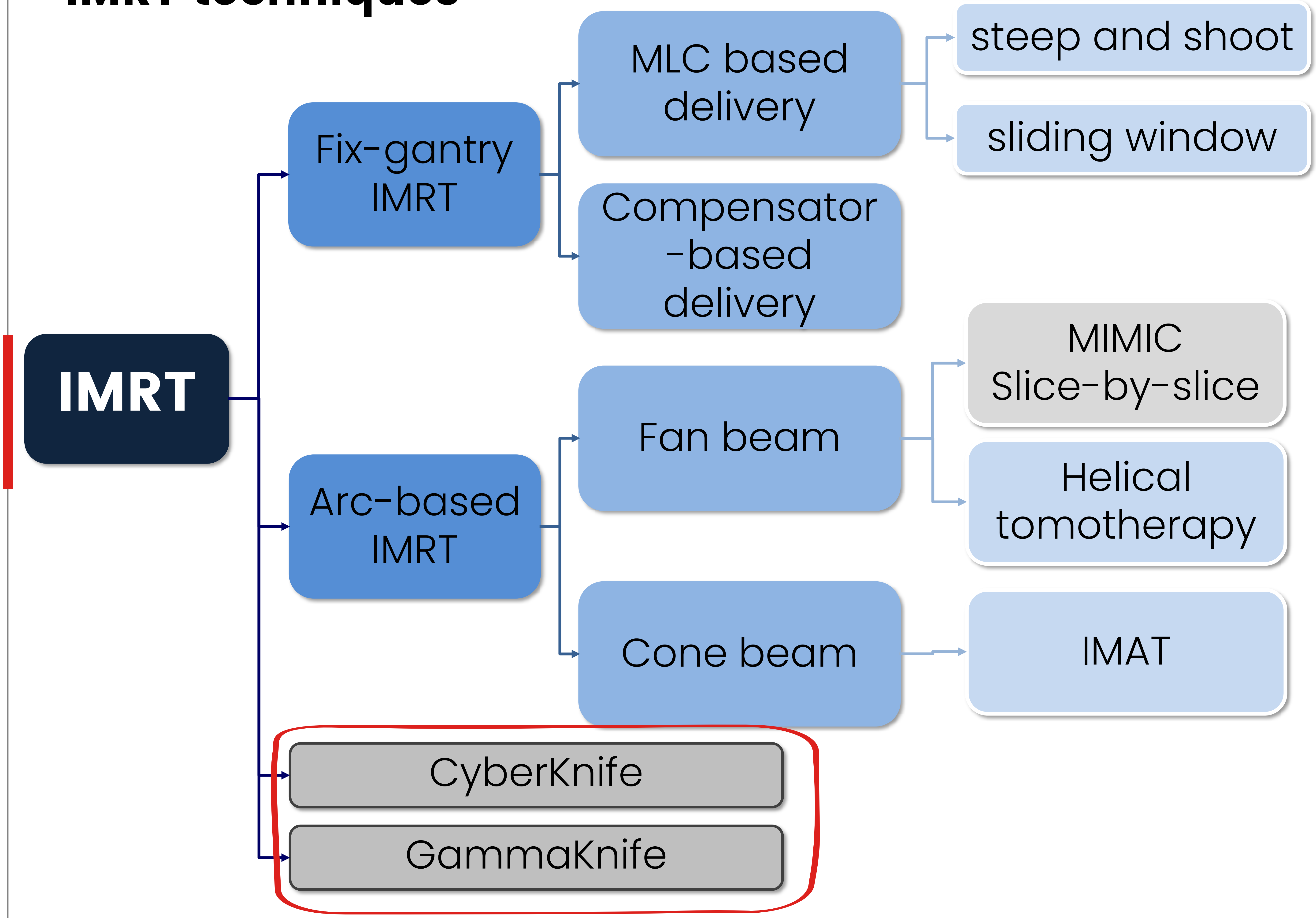
| *Vendor (Web Site)* | *Type* | *Delivery* | *No. of Leaves* | *Resolution, mm* | *Thickness, cm* | *Transmission, %* | *Focus* | *Maximum Field Size, cm* | *Overtravel, cm* | *Speed* |
|---|---|---|---|---|---|---|---|---|---|---|
| BrainLAB <www.brainlab.com> | Micro-MLC | Static or dynamic | 52 | 3–5 | 6.4 | 2 | Single; rounded ends | 10 × 10 | 5 | 1 cm/s |
| Elekta Inc. <www.elekta.com> | MLC | Static | 80 | 10 | 7.5 | 1.8–2.5 | Single; rounded ends | 40 × 40 | 12.5 | 2 cm/s |
| North American Scientific (Nomos) <www.nasmedical.com> | Binary | Tomotherapy | 40 | 4, 8, or 16 | 8 | 0.5 | Double | 20 × 30 | NA | 50 cm/s |
| Southeastern Radiation Products <www.seradiation.com> | Compensator | NA | NA | Based on planning system | 5.1 aluminum or brass | ~ 65 aluminum; ~ 84 brass | NA | 40 × 40 | NA | NA |
| Siemens Medical Systems <www.siemens.com> | MLC | Static | 82 | 10 | 7.5 | 0.9–1.25 | Double; flat ends | 40 × 40 | 10 | 2 cm/s |
| TomoTherapy Inc. <www.tomotherapy.com> | Binary | Tomotherapy | 64 | 6.25 | 10 | 0.4 | Double | 160 (long) × 40 (diameter) | NA | < 40 msec transit time |
| Varian Medical Systems <www.varian.com> | MLC | Static or dynamic | 120 | 5–10 | 6 | 1.6–1.9 | Single; rounded ends | 40 × 40 | 17 | 3 cm/s |

MLC = multileaf collimator; NA = not available.

*Intensity modulated radiation therapy; a clinical perspective*
*Chapter 12: Delivery Systems. Mundt, Arno J. and John C. Roeske.*

- O-arms: Tomotherapy, Halcyon (VARIAN)
- MNR-Linac (ELEKTA)

# IMRT techniques



# Cyber and gamma knifes

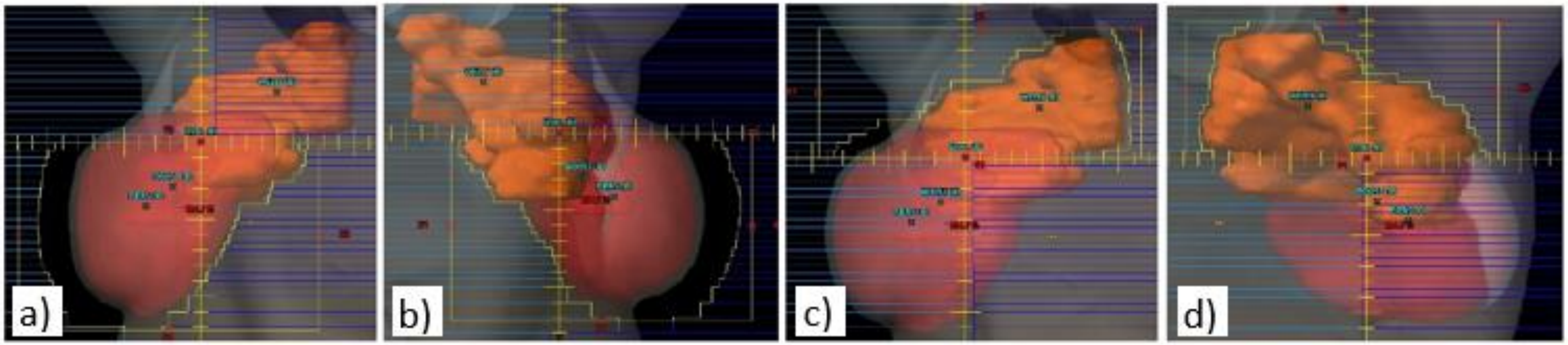
robotic arm (X6 FFF)

Co60 sources (≈ 200)

SBRT: liver, prostate H@N, lungs, spinal cord metastasis

SRT intracranial radiotherapy

# Hybrid technique




- Combination of 3D conformal fields with dynamic technique
- Removing bath wash of dynamic technique
- Removing lack of homogeneity in 3D CRT
- Reduce treatment time

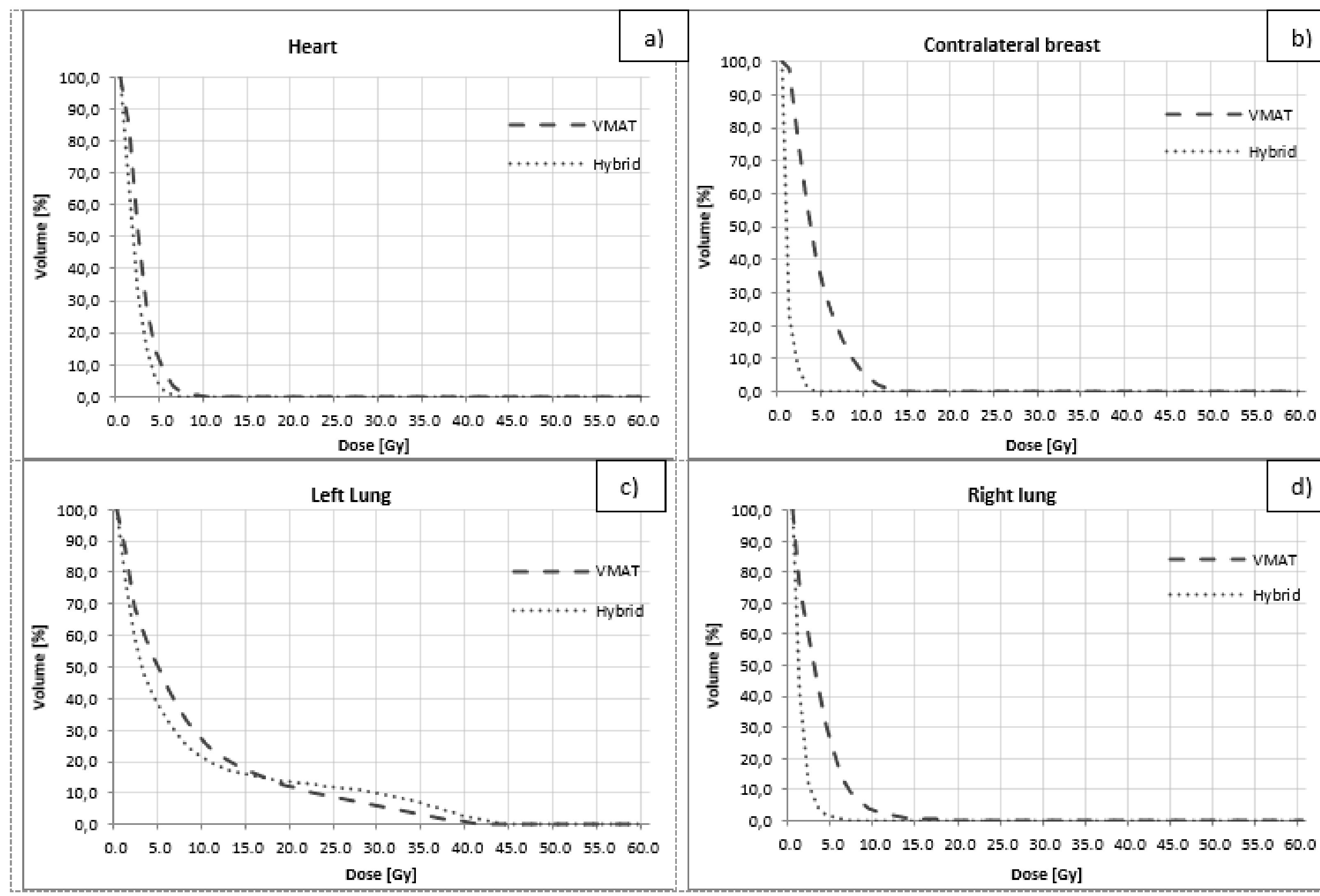
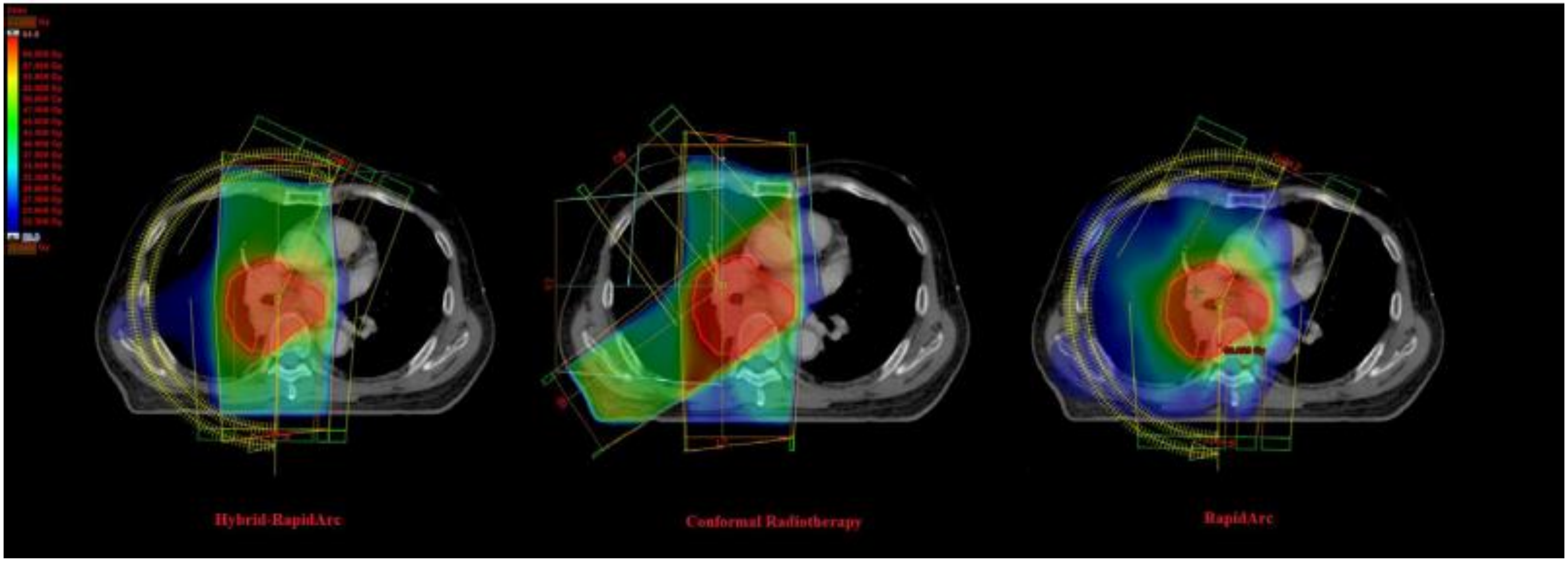
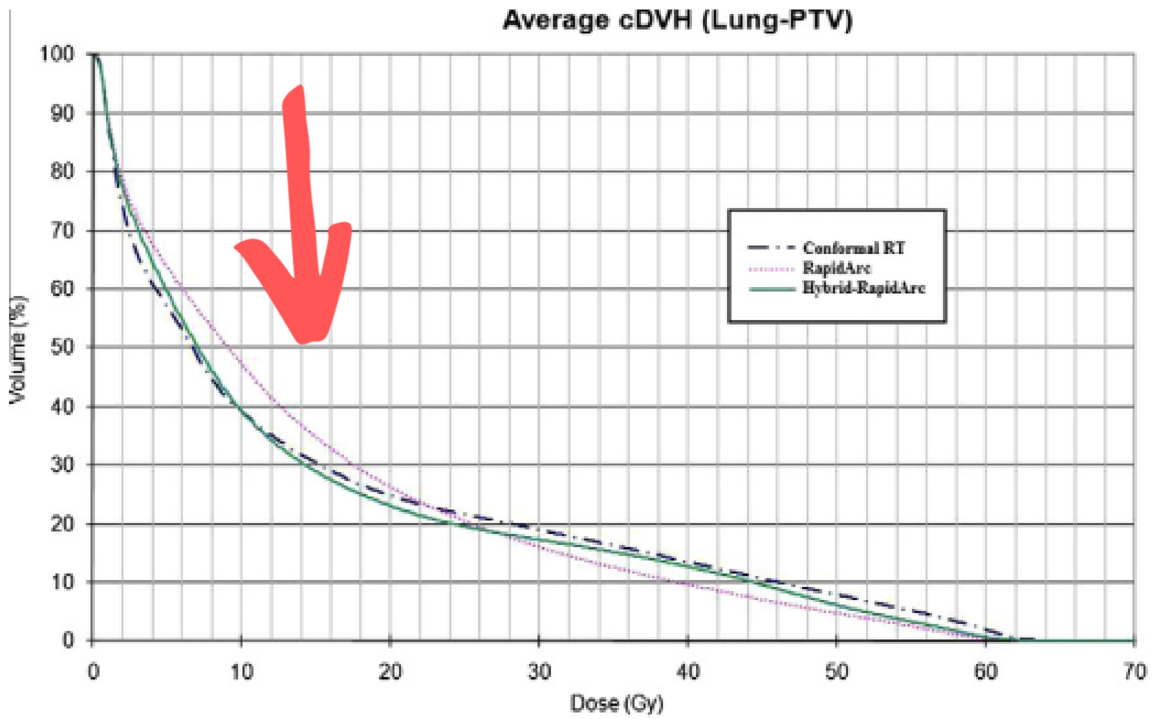


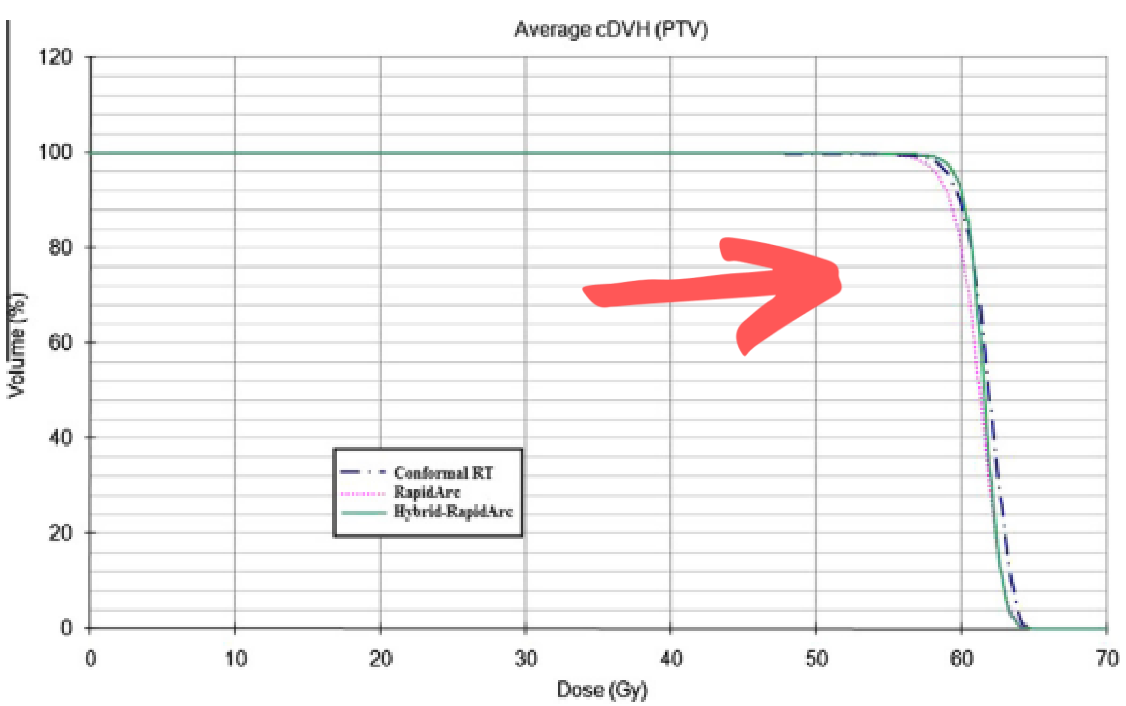

Figure 3 Cumulative histograms for VMAT and hybrid technique for a) heart, b) contralateral breast, c) left lung, d) right lung.

**Courtesy Dominika Bodzak**

# Hybrid technique



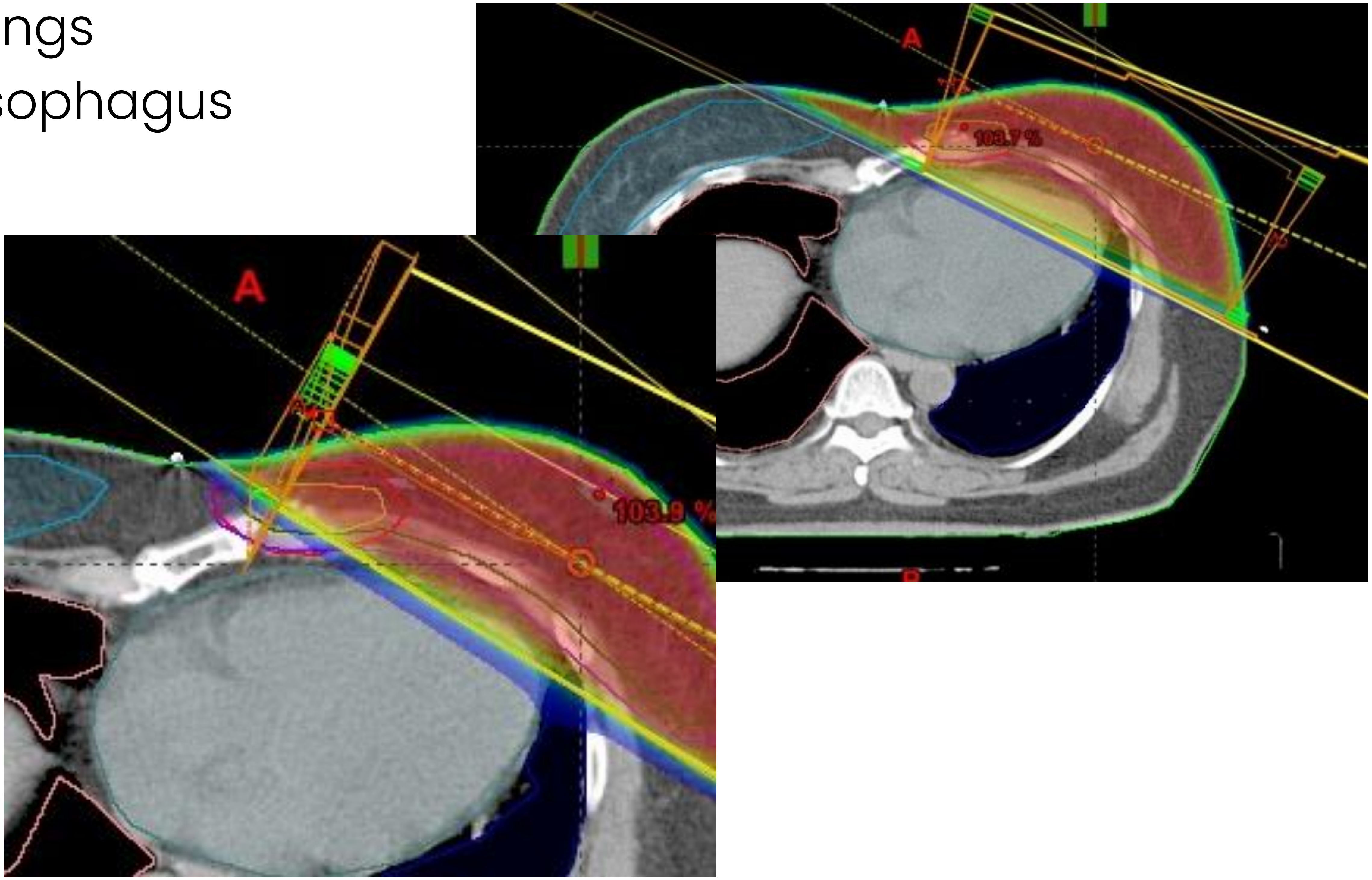

O.S.H. Chan et al. Radiotherapy and Oncology 101 (2011) 298–30

The superiority of hybrid-volumetric arc therapy (VMAT) technique over double arcs VMAT and 3D-conformal technique in the treatment of locally advanced non-small cell lung cancer – A planning study

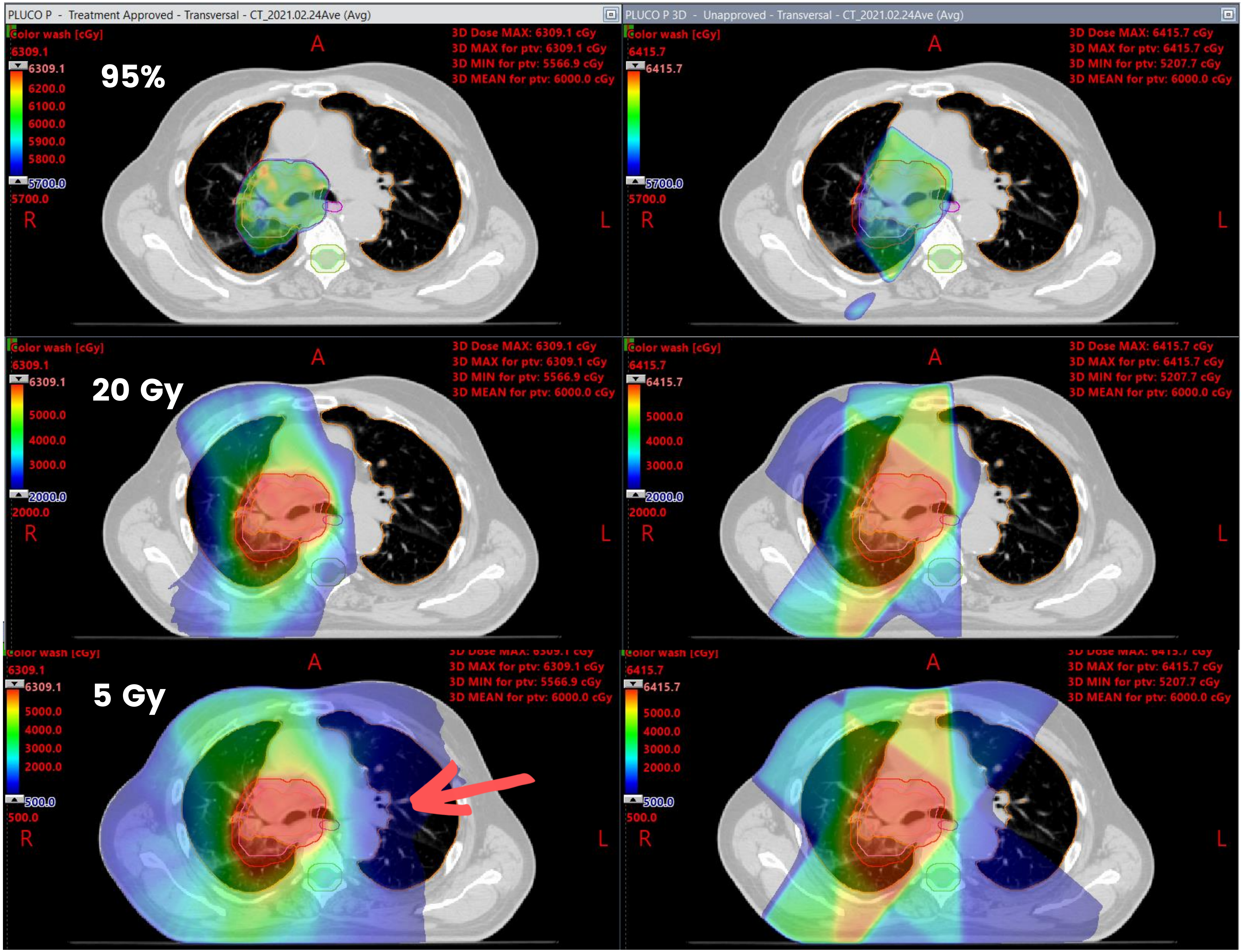
# Hybrid technique

All localization where low doses are unwanted

- Breast
- Lungs
- esophagus



# VMAT vs 3D

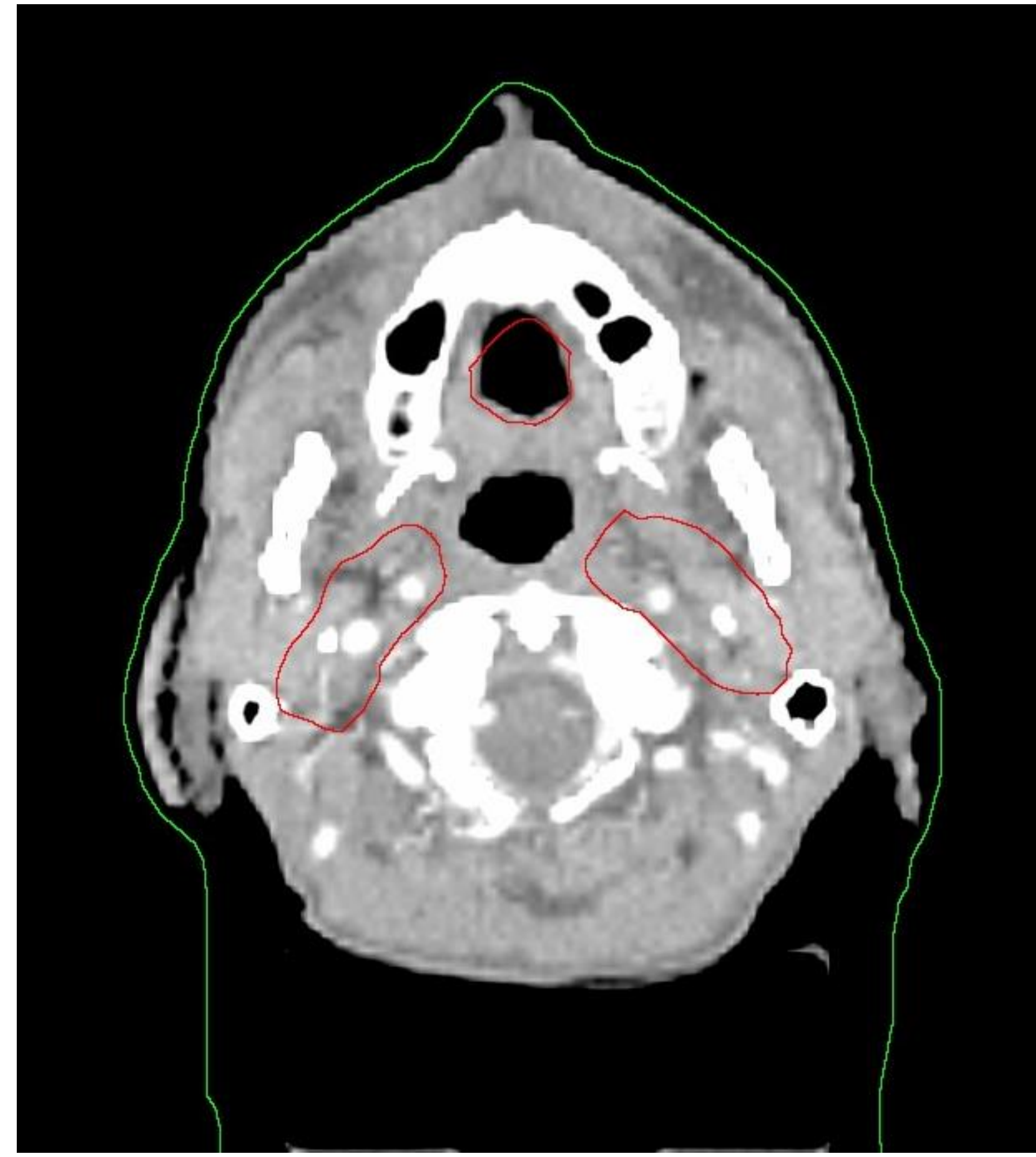


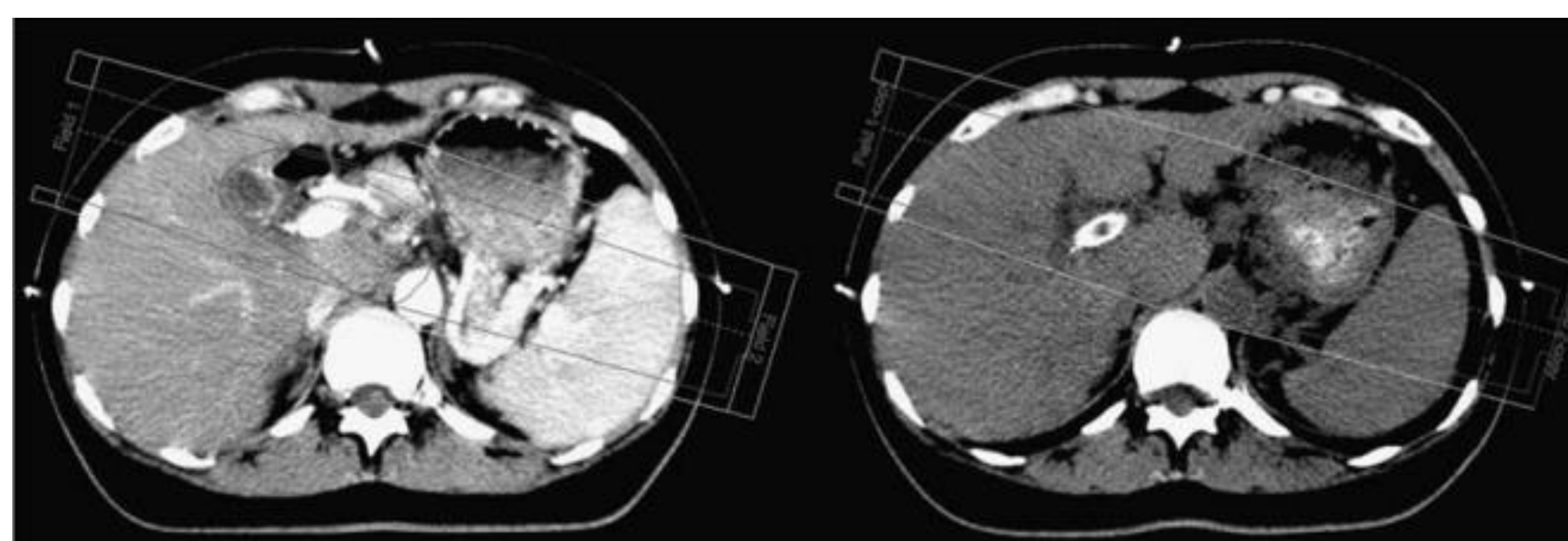
PLUCO P - Treatment Approved - Transversal - CT_2021.02.24Ave (Avg)
PLUCO P 3D - Unapproved - Transversal - CT_2021.02.24Ave (Avg)
95%
20 Gy
5 Gy
Color wash [cGy]
3D Dose MAX: 6309.1 cGy
3D MAX for ptv: 6309.1 cGy
3D MIN for ptv: 5566.9 cGy
3D MEAN for ptv: 6000.0 cGy
3D Dose MAX: 6415.7 cGy
3D MAX for ptv: 6415.7 cGy
3D MIN for ptv: 5207.7 cGy
3D MEAN for ptv: 6000.0 cGy
A
R
L

# IMRT TREATMENT PLANNING
## STEPS

# CT - contrast

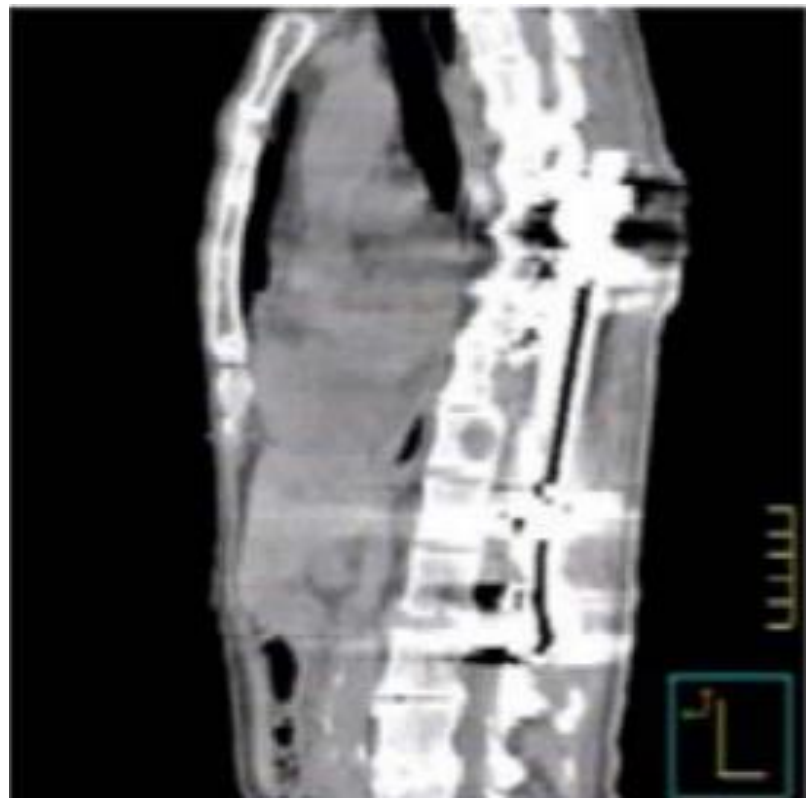
- 26 patients
- Changes in MU values due to the presence of contrast were investigated
- They found:
  - the impact to be negligible (less than 1%) for the HN, thorax, and pelvis
  - greater impact (>2%) in the upper abdominal region where there is usually more amount contrast

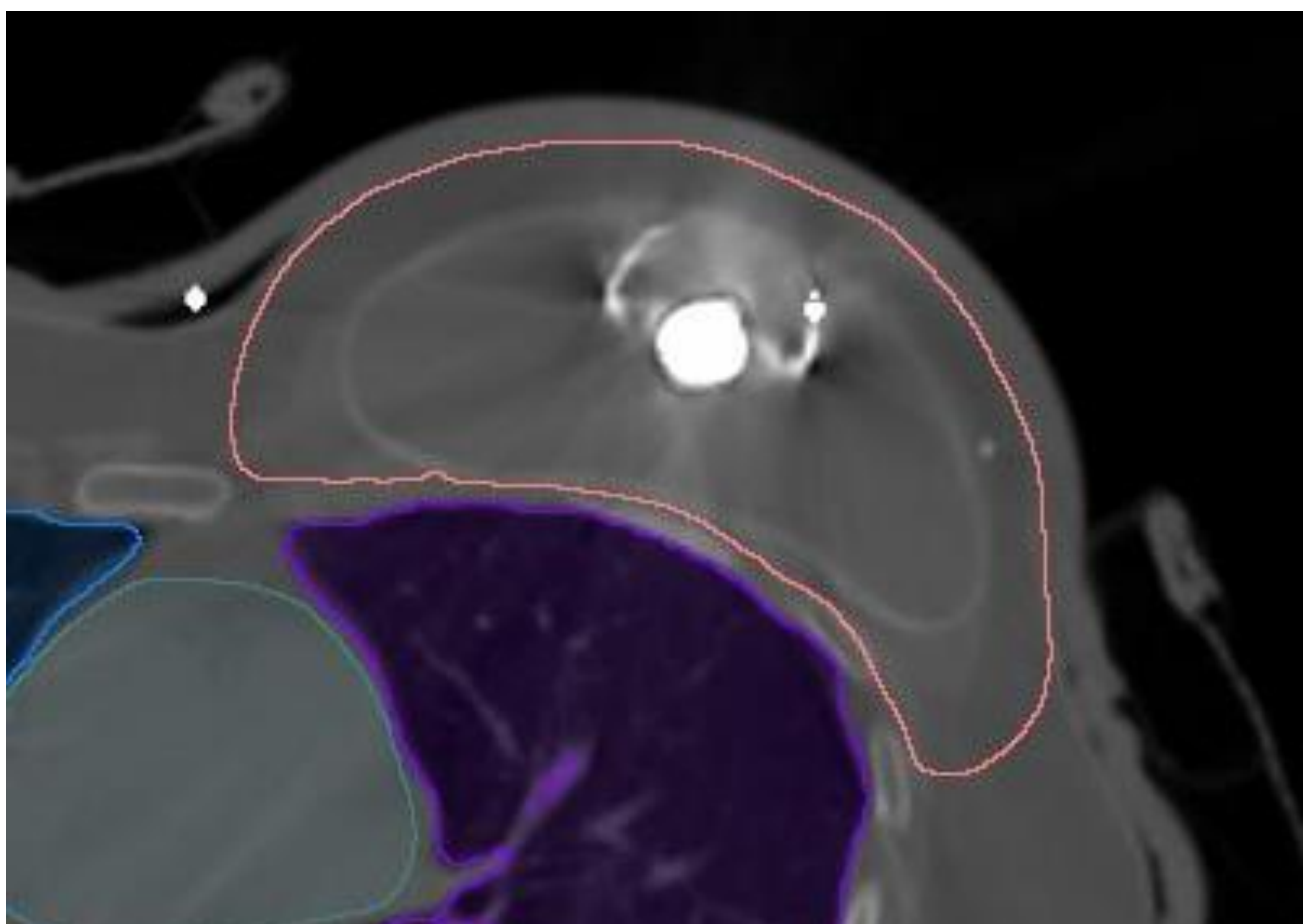


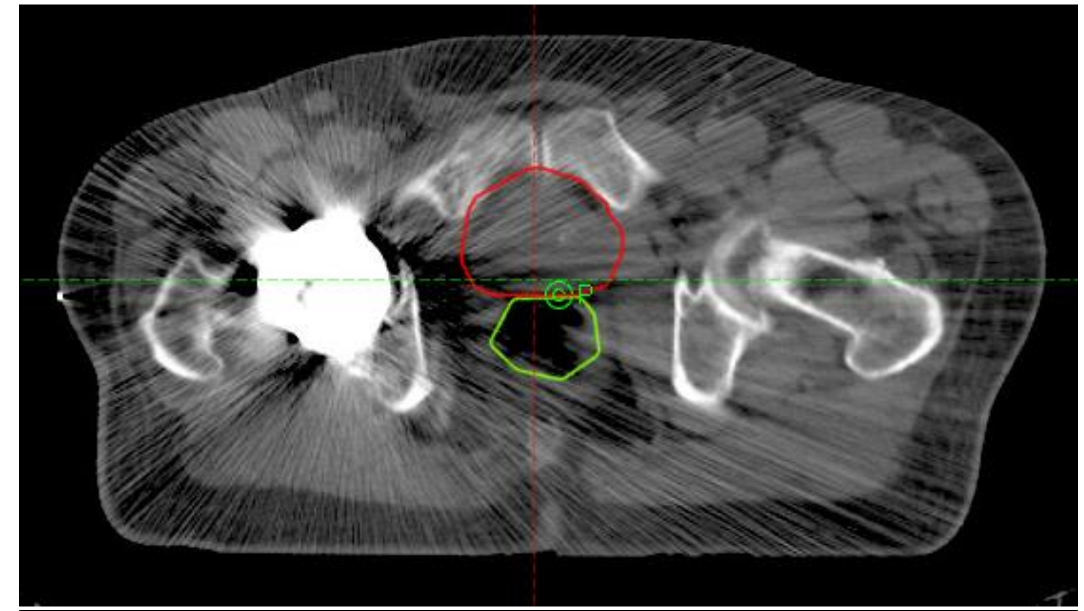


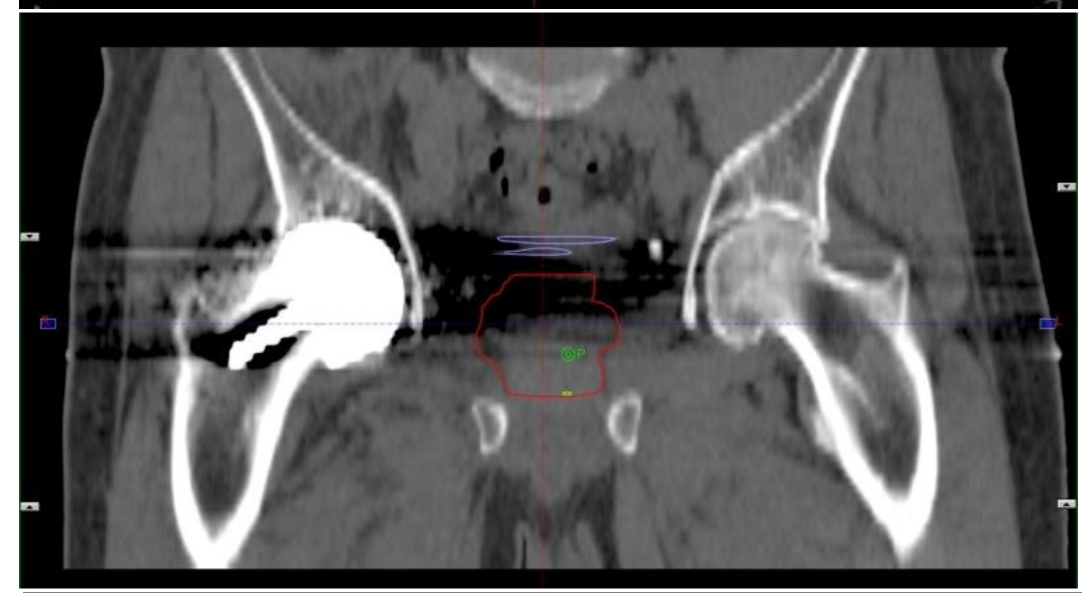
*Y. Shibamoto et al. Radiotherapy and Oncology 84 (2007)*

# CT – metal artefacts







# Dosimetric considerations for patients with HIP prostheses undergoing pelvic irradiation. Report of the AAPM Radiation Therapy Committee Task Group 63

Chester Reft
*University of Chicago, Chicago, Illinois 60637*

Rodica Alecu
*U.S. Oncology, Texas Cancer Center, Sherman, Texas*

Indra J. Das
*University of Pennsylvania, Philadelphia, Pennsylvania*

Bruce J. Gerbi
*University of Minnesota Medical School, Minneapolis, Minnesota*

Paul Keall
*Virginia Commonwealth University, Richmond, Virginia*

Eugene Lief
*New York University Medical Center, New York*

Ben J. Mijnheer
*Netherlands Cancer Institute, Amsterdam, Netherlands*

Nikos Papanikolaou
*University of Arkansas Medical Sciences*

Claudio Sibata
*East Carolina University School of Medicine, Greenville, North Carolina*

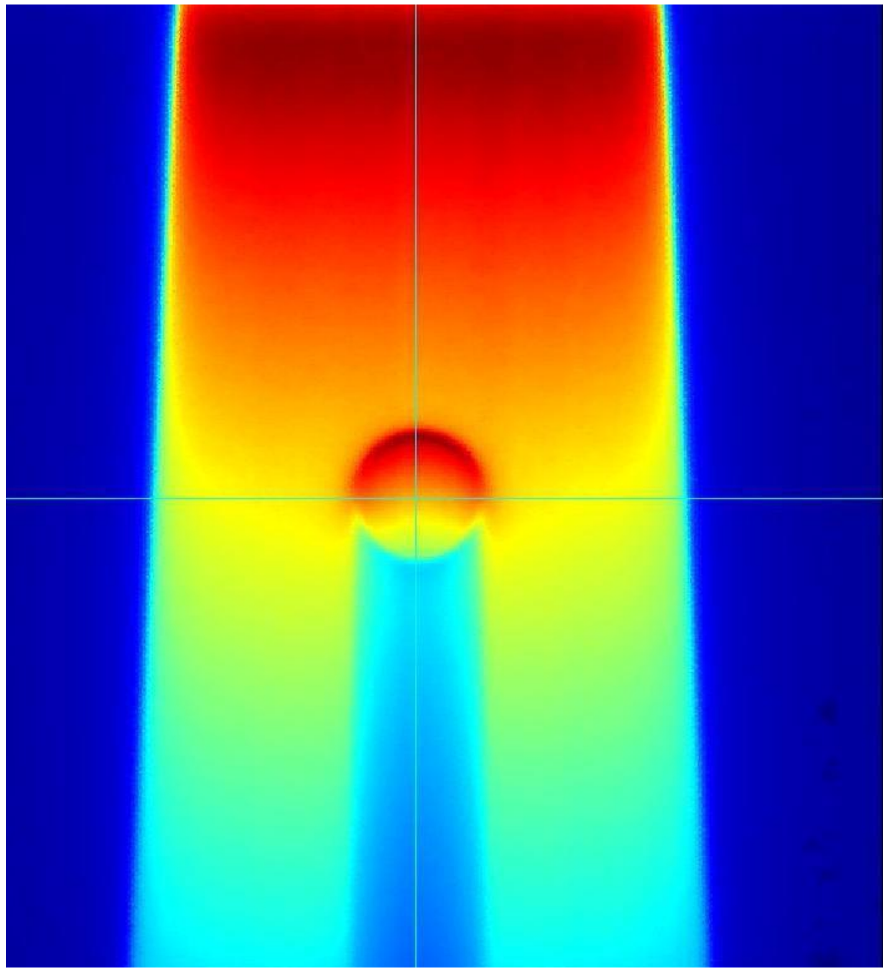
Jake Van Dyk
*London Regional Cancer Centre, Ontario, Canada*

(Received 25 July 2002; accepted for publication 10 February 2003; published 30 May 2003)

# The most frequently used prostheses

| | Co-Cr-Mo | titanium | steel |
|---|---|---|---|
| atomic composition | Co 60%<br>Cr 30%<br>Mo 5% | Ti 90%<br>Al 6%<br>Va 4% | Fe 65%<br>Cr 18%<br>Ni 12%<br>Mo 3% |
| ρ [g/cm³] | 7.9 | 4.3 | 8.1 |
| Relative electrons density | 6.8 | 3.6 | 6.7 |

| | water | titanium | steel |
|---|---|---|---|
| μ/ρ [cm²/g] | 0.0397 | 0.5361 | 0.0362 |
| ρ [g/cm³] | 1 | 4.3 | 8.1 |
| atten./1 cm [%] | 3.9 | 14.0 | 25.4 |



Gafchromic film, X6MV,, 10 x 10 cm, SSD = 90 cm, 200 MU
brass cylinder diam. = 25 mm, $RED_{brass}$ = 6.98

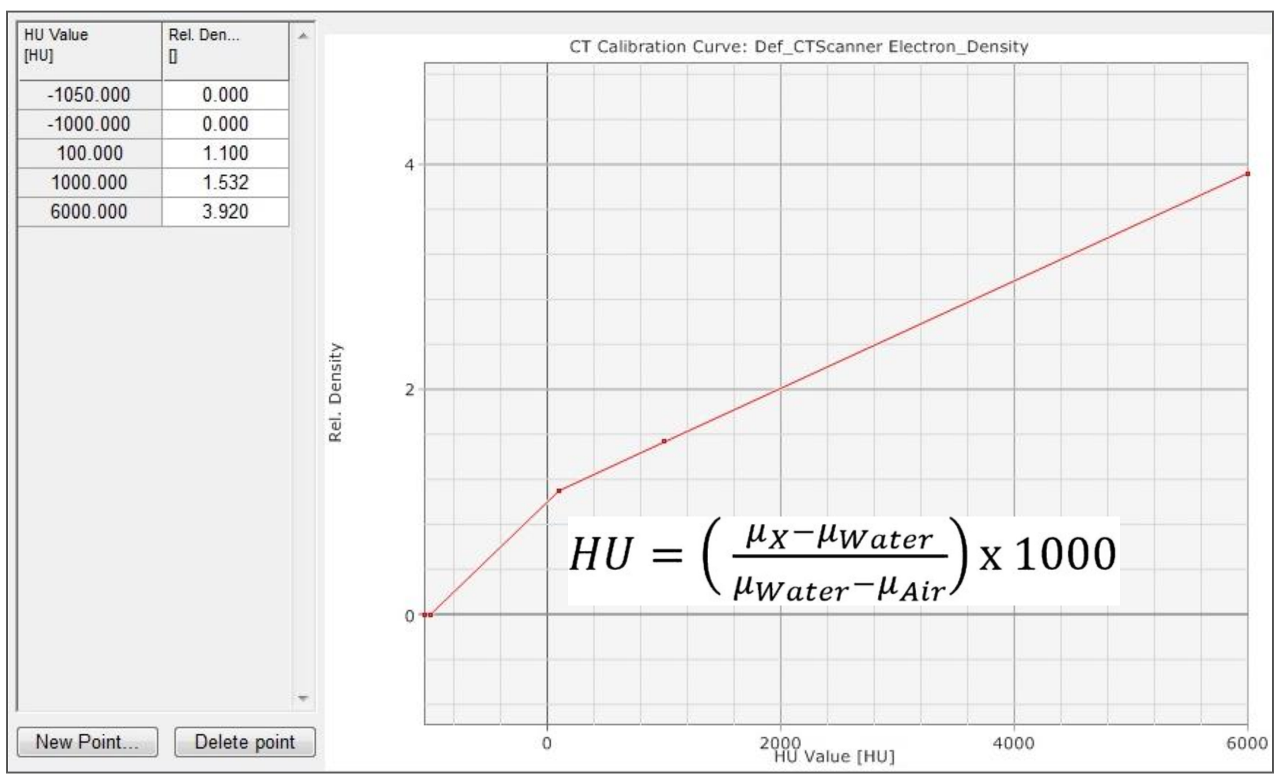
**Courtesy Ryszard Dąbrowski**

# Conversion curves

- CT **by default** saves HU values in the 12-bit mode - distinguish 4096 values ($2^{12}$). The typical HU range: **- 1024 to + 3071**
- **EXTENDED** mode enables 16-bit recording. It allows to create conversion curves (HU relative electron densities) covering high-density materials.

| HU Value [HU] | Density [g/cm3] |
|---|---|
| -1000.000 | 0.001 |
| -992.000 | 0.001 |
| -976.000 | 0.001 |
| -480.000 | 0.500 |
| -96.000 | 0.950 |
| 0.000 | 1.000 |
| 48.000 | 1.050 |
| 128.000 | 1.100 |
| 528.000 | 1.334 |
| 976.000 | 1.603 |
| 1488.000 | 1.850 |
| 1824.000 | 2.100 |
| 2224.000 | 2.400 |
| 2640.000 | 2.700 |
| 2832.000 | 2.830 |

Eclispe conversion curve

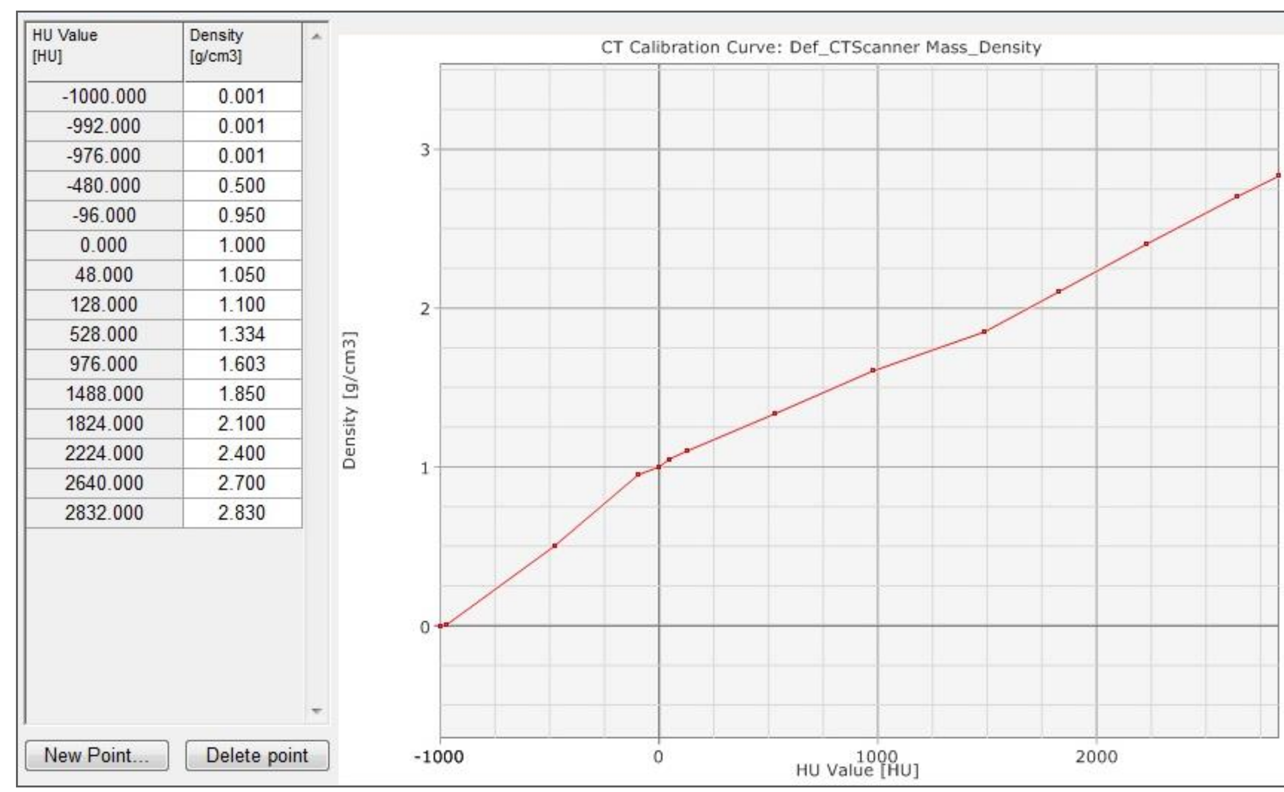


| HU Value [HU] | Rel. Den... [] |
|---|---|
| -1050.000 | 0.000 |
| -1000.000 | 0.000 |
| 100.000 | 1.100 |
| 1000.000 | 1.532 |
| 6000.000 | 3.920 |



$$HU = \left(\frac{\mu_X - \mu_{Water}}{\mu_{Water} - \mu_{Air}}\right) \text{x } 1000$$

Relative electron density – AAA

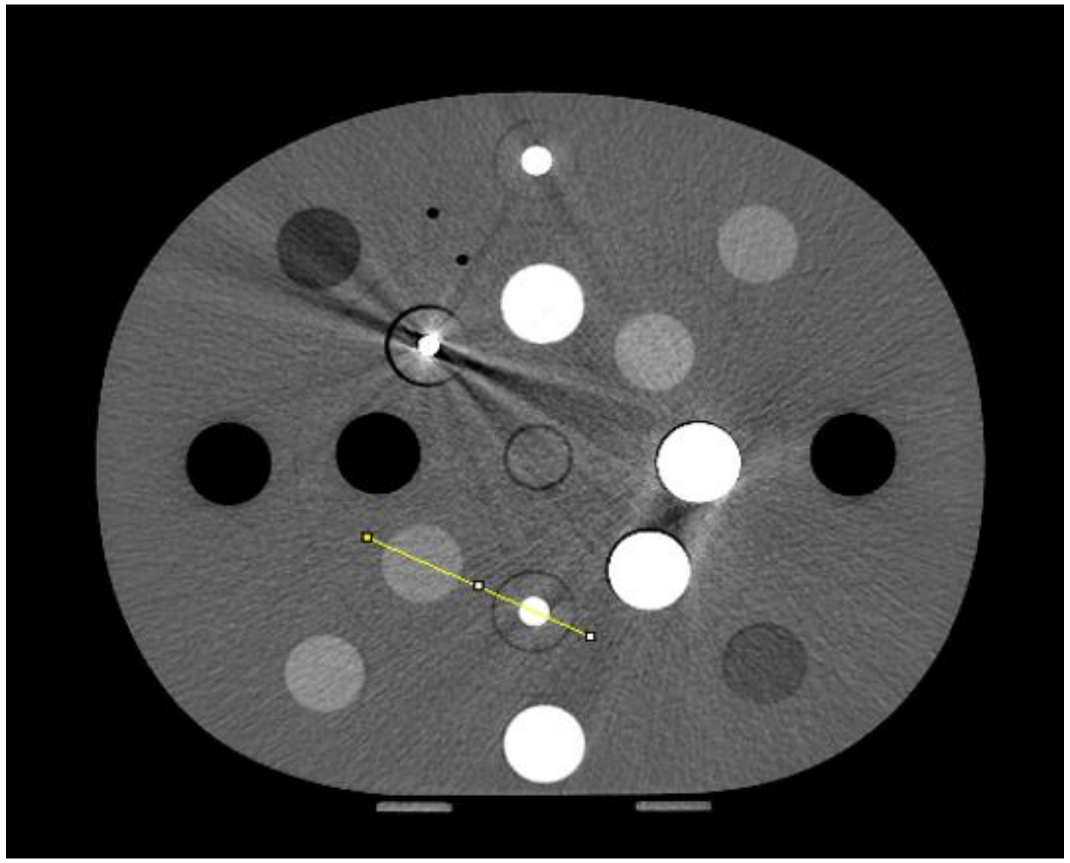


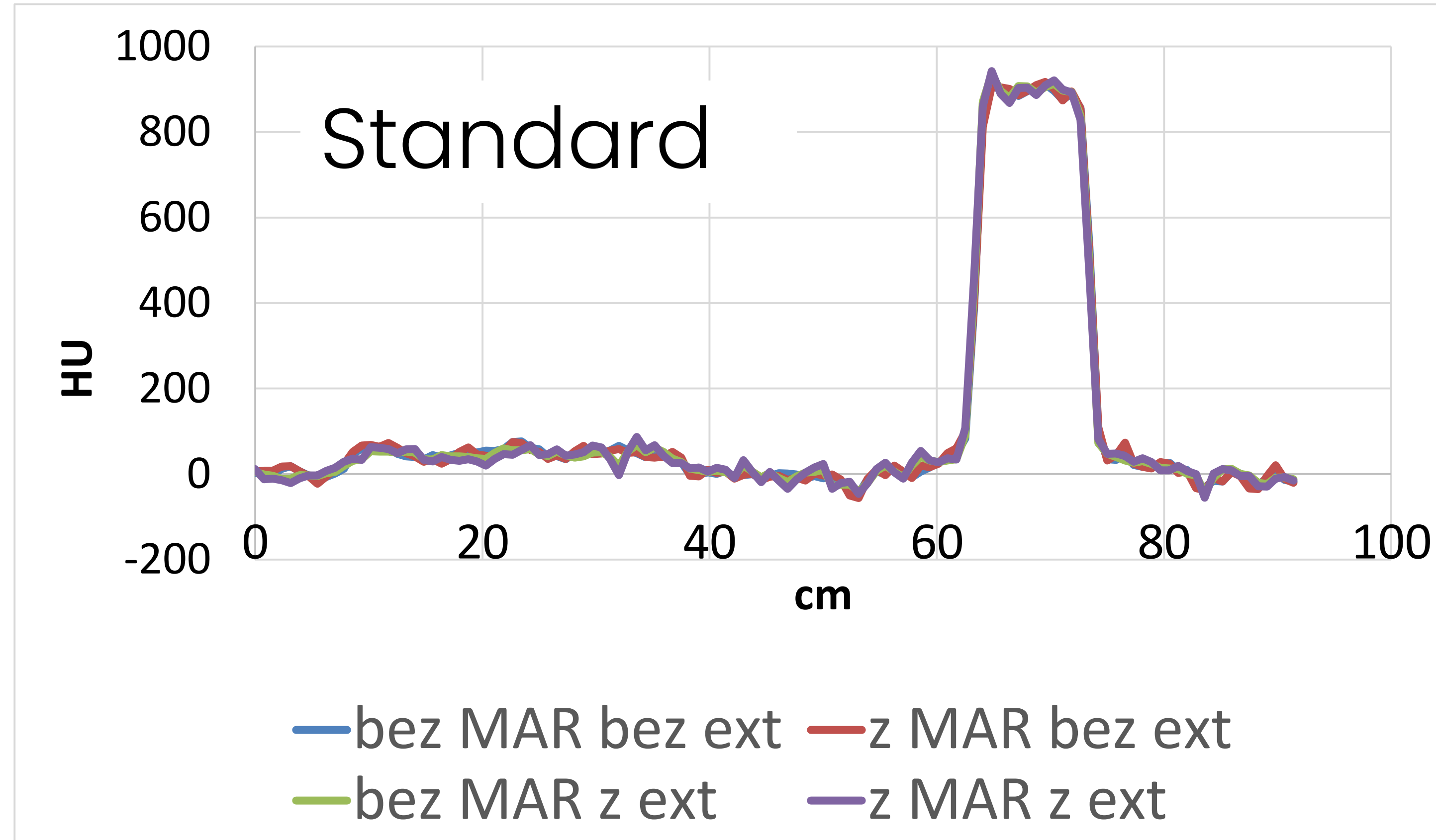
| HU Value [HU] | Density [g/cm3] |
|---|---|
| -1000.000 | 0.001 |
| -992.000 | 0.001 |
| -976.000 | 0.001 |
| -480.000 | 0.500 |
| -96.000 | 0.950 |
| 0.000 | 1.000 |
| 48.000 | 1.050 |
| 128.000 | 1.100 |
| 528.000 | 1.334 |
| 976.000 | 1.603 |
| 1488.000 | 1.850 |
| 1824.000 | 2.100 |
| 2224.000 | 2.400 |
| 2640.000 | 2.700 |
| 2832.000 | 2.830 |



Mass dencity [g/cm3] – Acuros, Colapse Cone

# EXTENDED






- AL

HU(12) = 2576 ± 56 (1SD)

HU(16) = 2484 ± **382 (1SD)**

(SD 382 HU -> SD 0.11 RED)

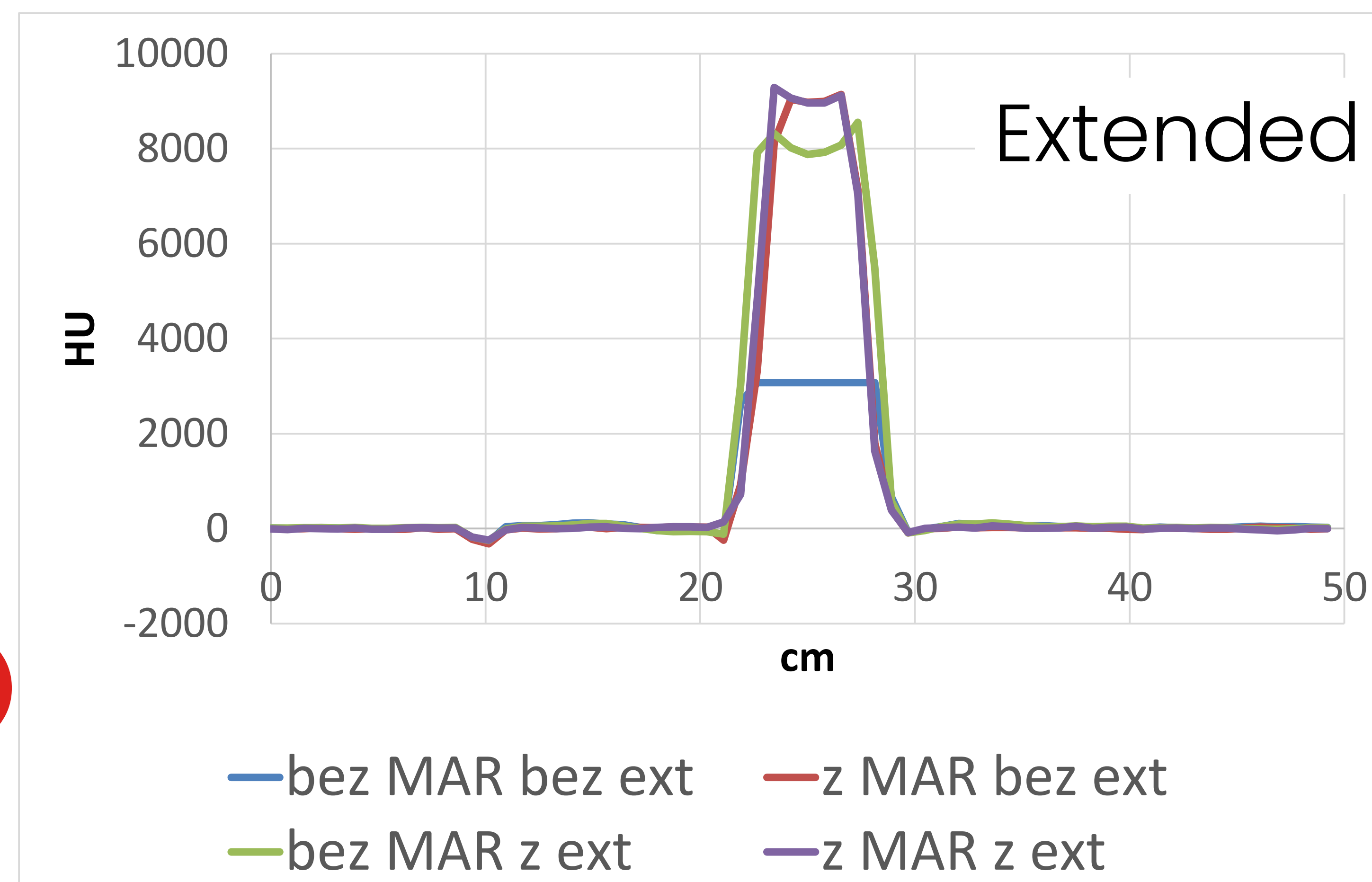
- Brass

HU(12) = 3071 ± 0 (1SD)

HU(16) = 9062 ± **2540 (1SD)**

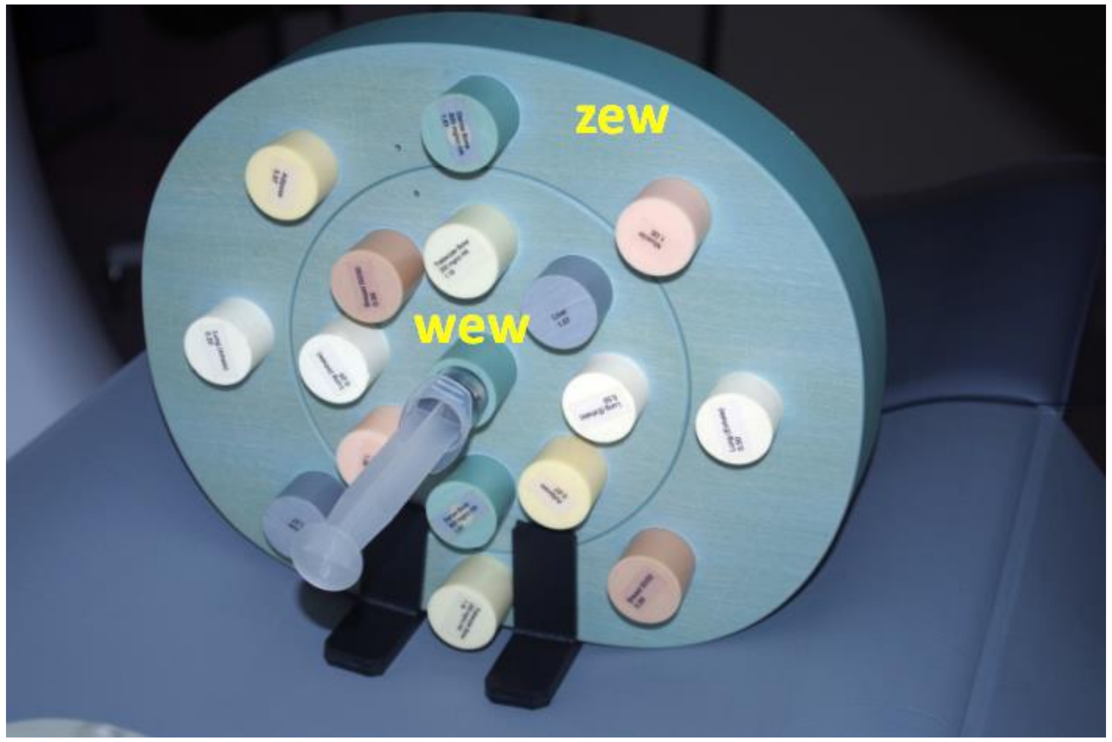
(SD 2540 HU -> SD 1.2 RED)

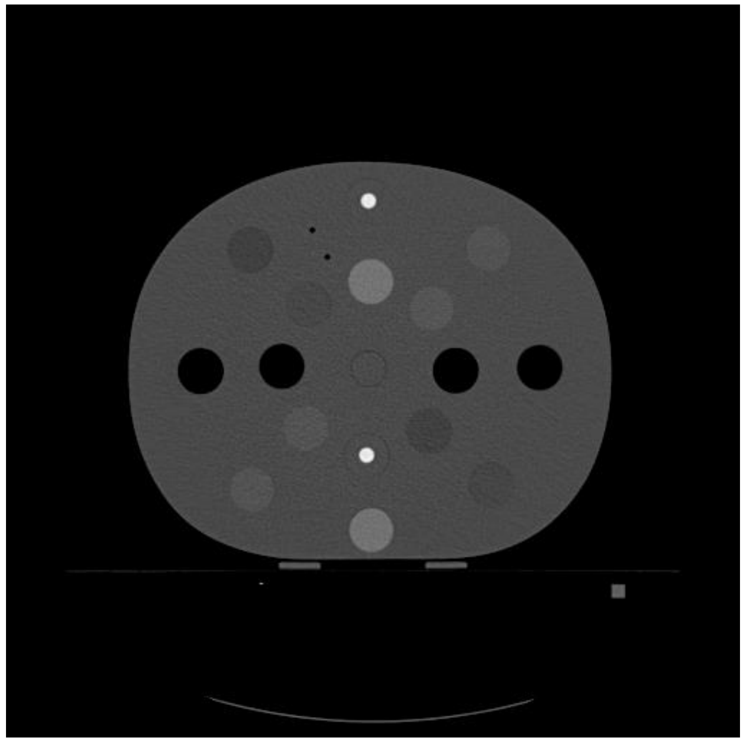
GE Discovery CT 590 RT)

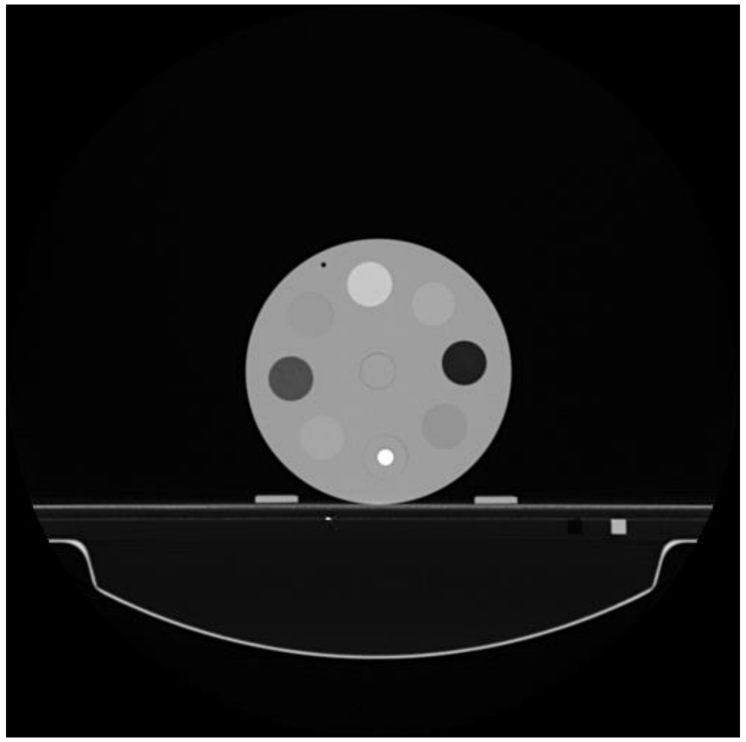


# Determination of CT-to-density conversion curve

Phantom CIRS 062 with inserts of different densities.



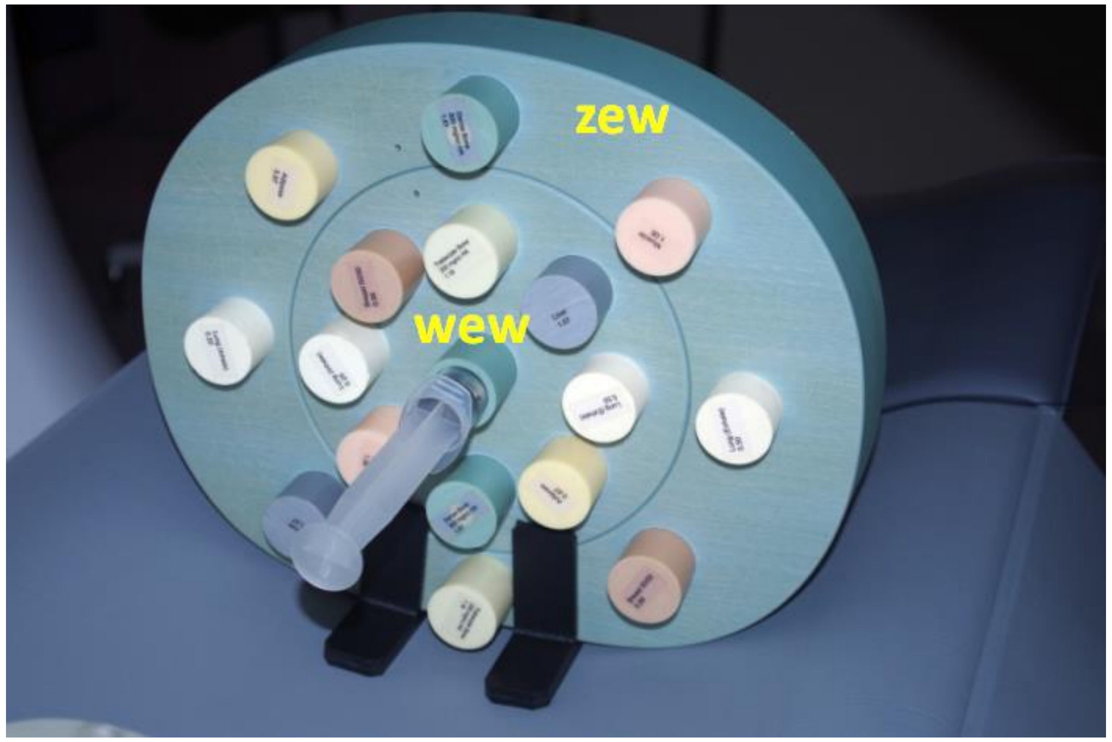



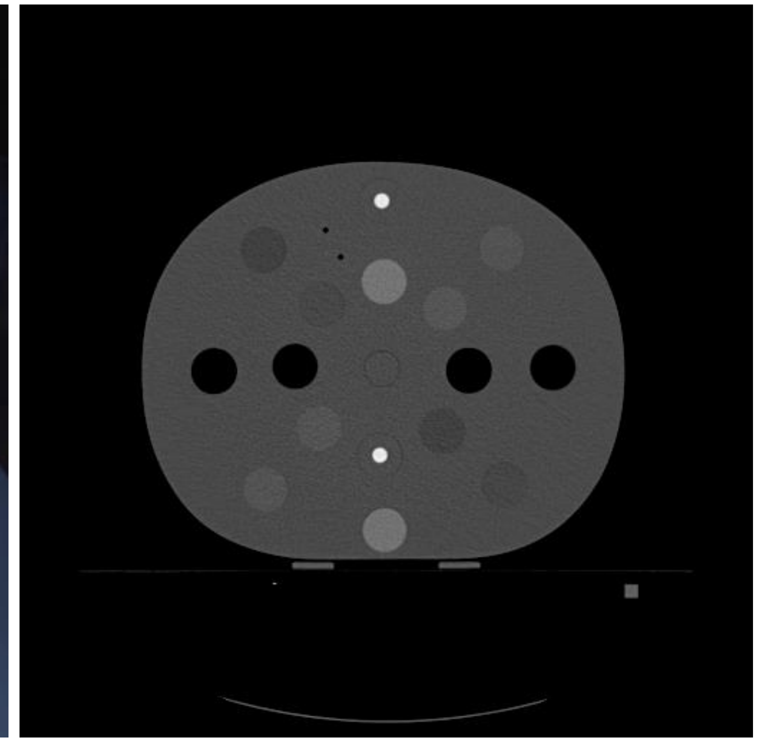


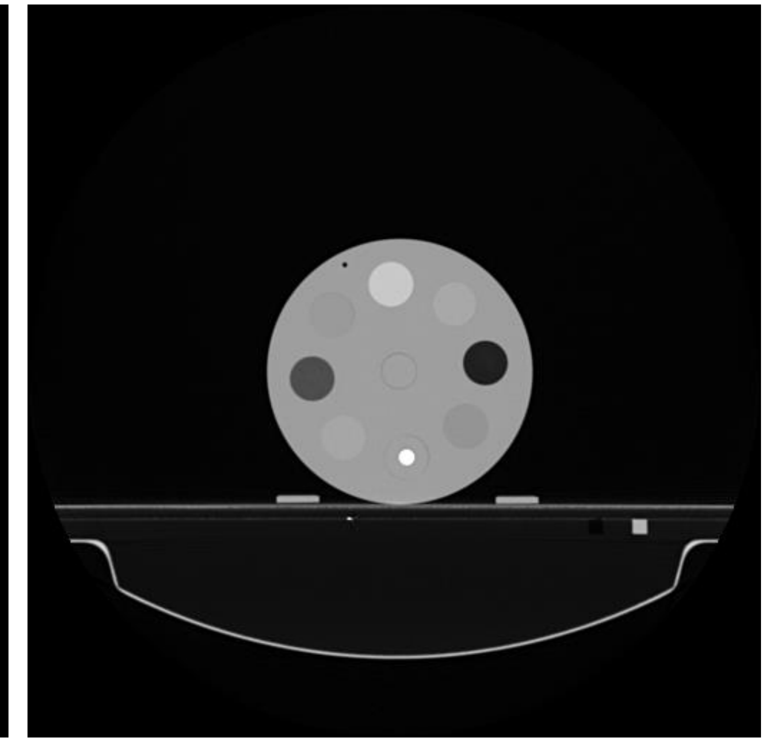
| tissues/materials | density [g/cm³] | Relative electron density |
|---|---|---|
| Lung (Inhale) | 0.2 | 0.19 |
| Lung (Exhale) | 0.5 | 0.489 |
| Adipose | 0.96 | 0.949 |
| Breast (50/50) | 0.99 | 0.976 |
| Water | 1 | 1 |
| Muscle | 1.06 | 1.043 |
| Liver | 1.07 | 1.052 |
| Trabecular Bone | 1.16 | 1.117 |
| Dense Bone | 1.61 | 1.512 |

# Determination of CT-to-density conversion curve

Phantom CIRS 062 with inserts of different densities.



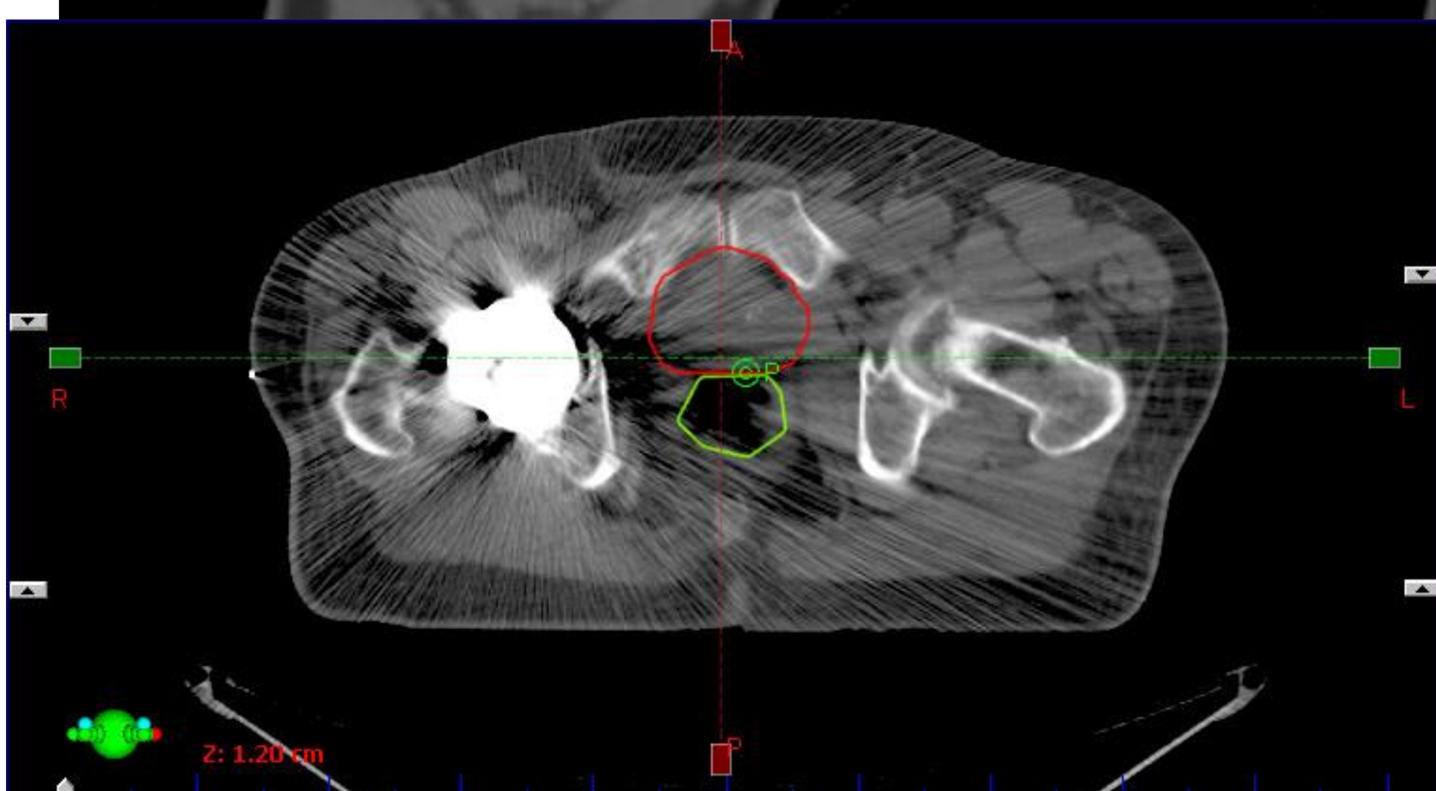



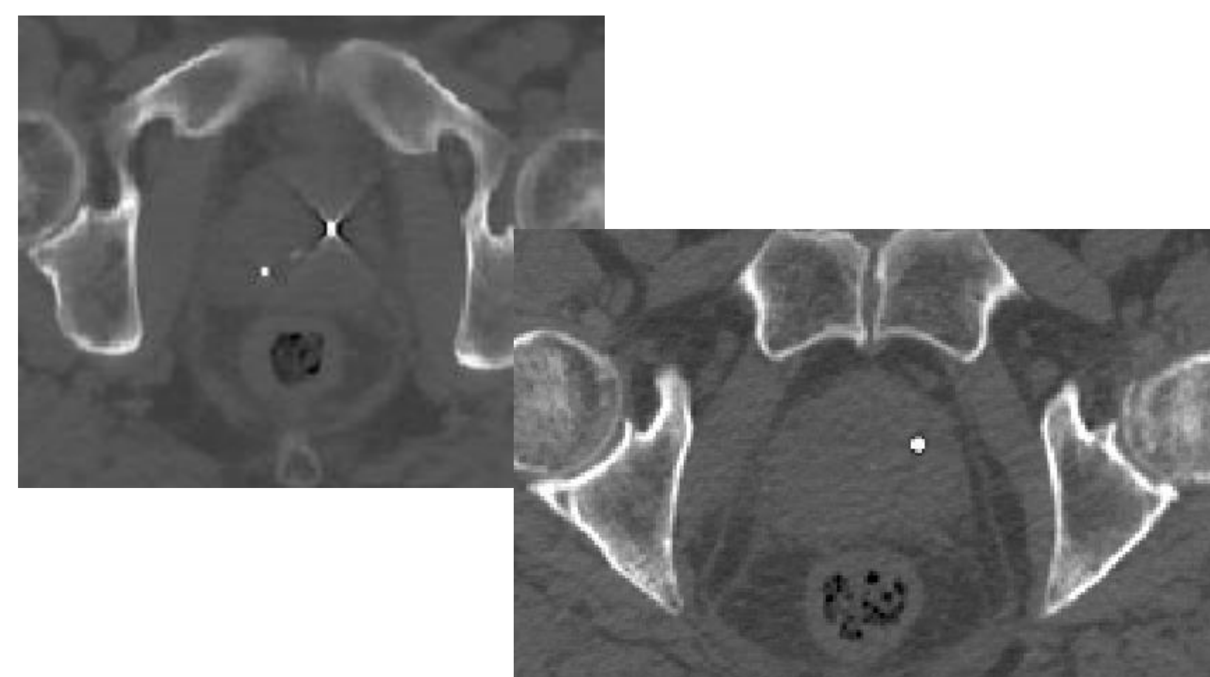


Additional inserts used in the CIRS phantom

| tissues/materials | density [g/cm$^3$] | Relative electron density |
|---|---|---|
| Bone 800/Dens Bone | 1.57 | 1.48 |
| Bone 1750 | 2.15 | 1.98 |
| Titanium | 4.51 | 3.74 |

# MAR - Metal Artifact Reduction



# MAR and EXTENDED mode

- cupping artefact
- inversely proportional to the size and proportional to the density of the metal
- For stainless steel implants the CT numbers can vary up to 43% from the mean value.
- For titanium implants maximum variation of 26% from the mean value.
- For both metals, the recorded CT values for the 10 mm metal implant were the most consistent with the mean value

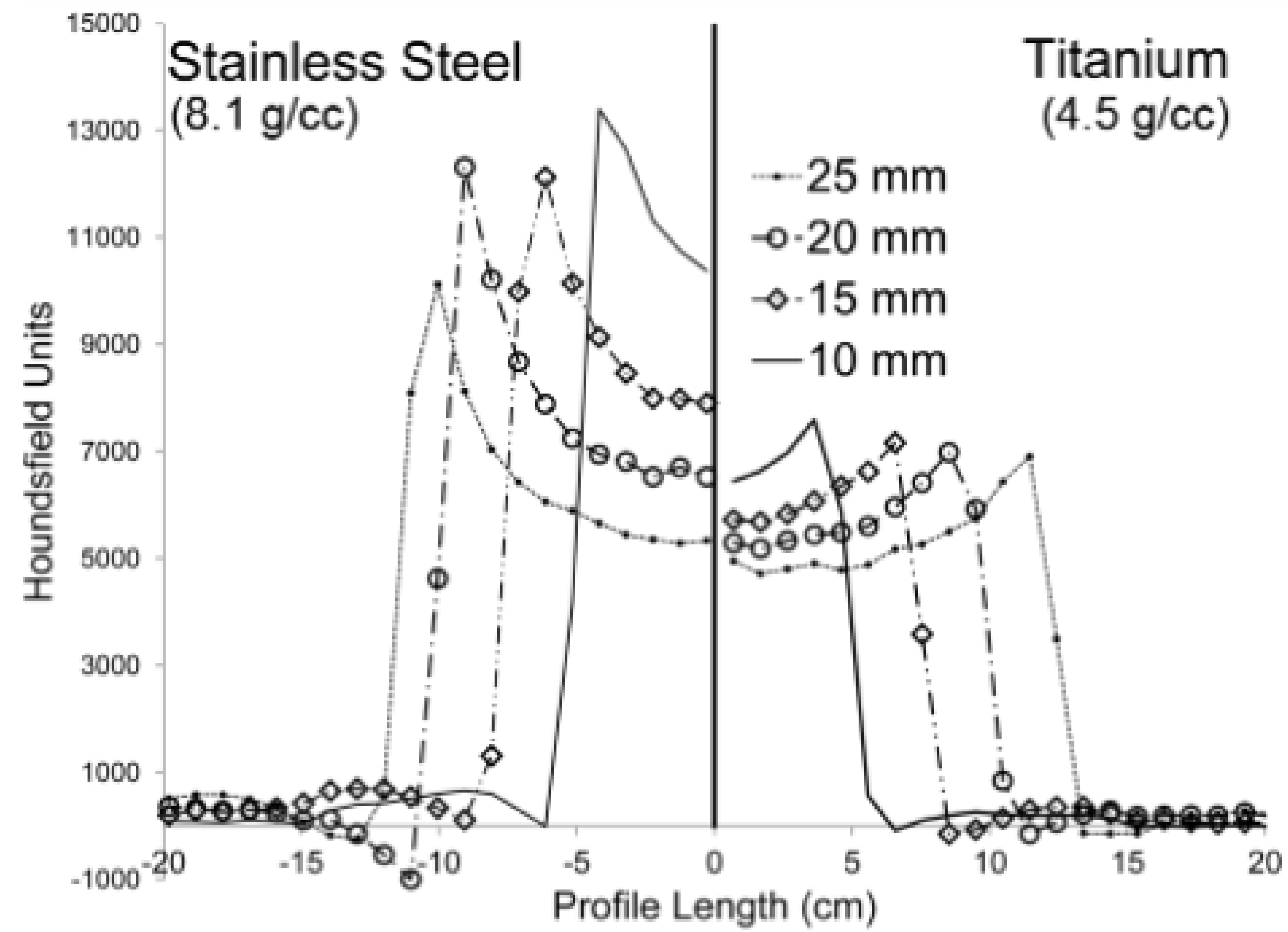
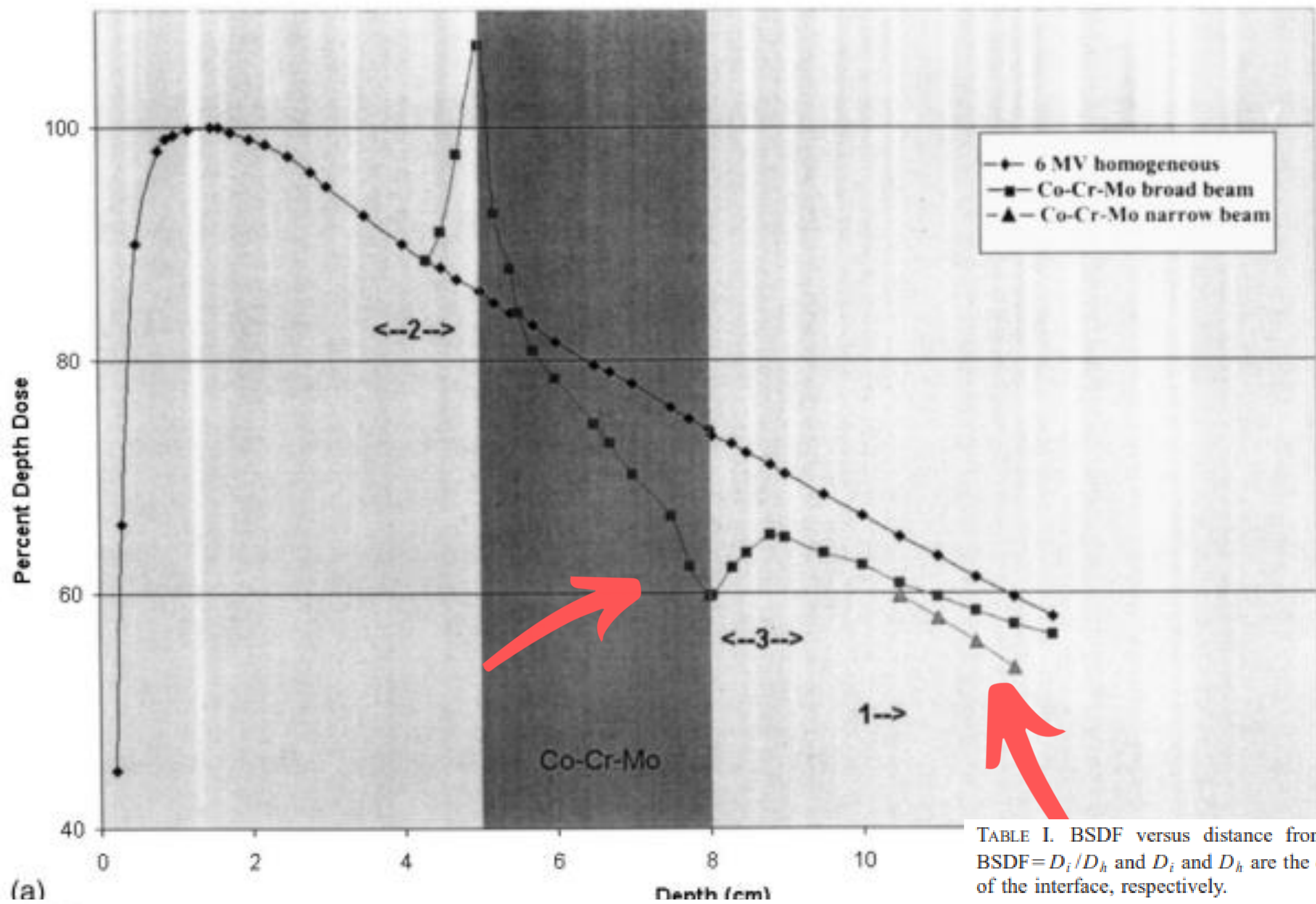



FIG. 7. (A) CT number to mass density data. (B) 2D profiles through metal inserts of different diameters, for both titanium and stainless steel.

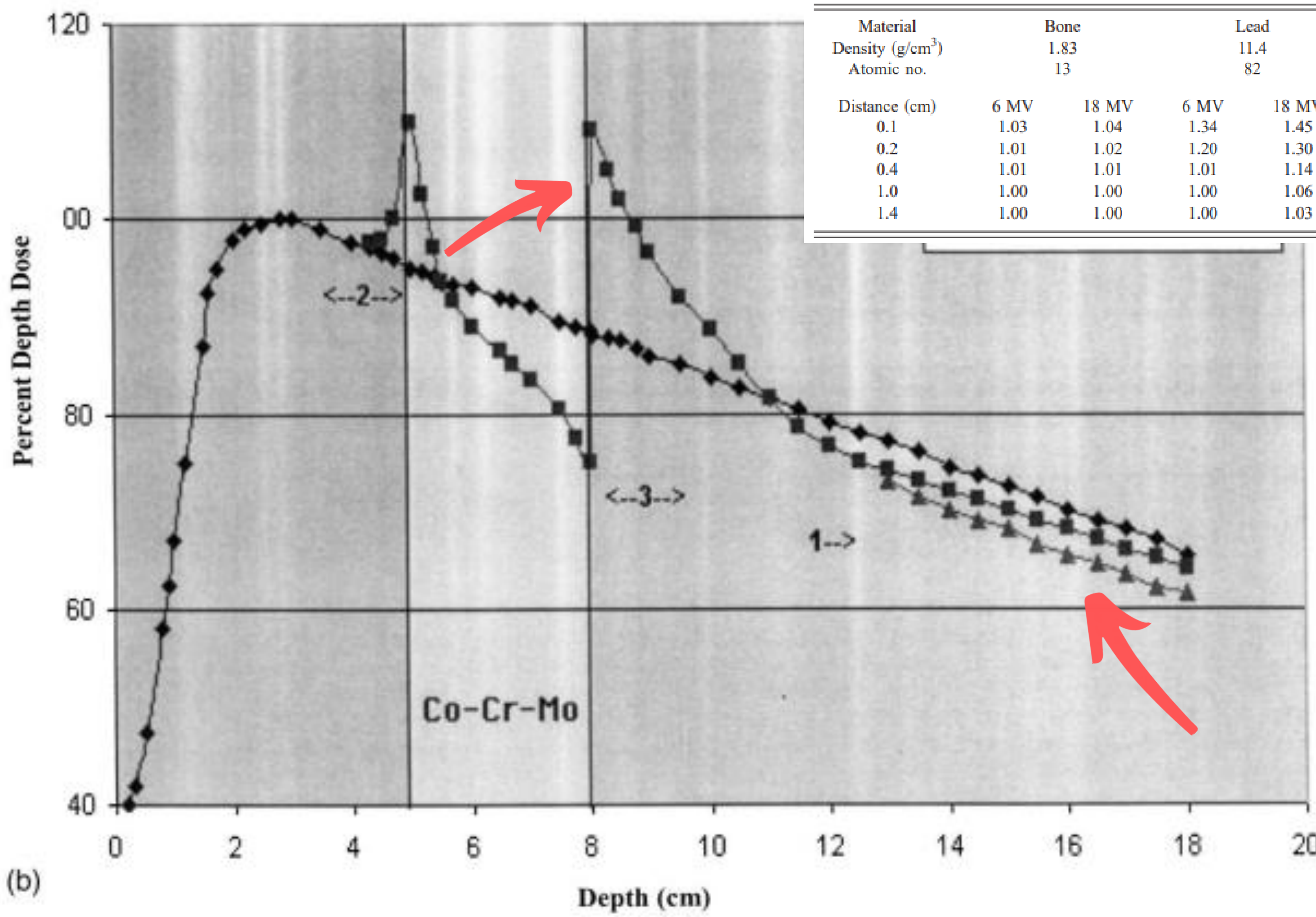
Axente et al.: IMAR clinical evaluation, Med. Phys. 42 (3), 2015




- 6MV
- Co-Cr-MO
- Homogenous vs. broad vs. narrow beam

TABLE I. BSDF versus distance from interface for $10 \times 10$ cm$^2$ where BSDF $= D_i / D_h$ and $D_i$ and $D_h$ are the doses with and without the presence of the interface, respectively.

| Material | Bone | | Lead | |
|---|---|---|---|---|
| Density (g/cm$^3$) | 1.83 | | 11.4 | |
| Atomic no. | 13 | | 82 | |
| Distance (cm) | 6 MV | 18 MV | 6 MV | 18 MV |
| 0.1 | 1.03 | 1.04 | 1.34 | 1.45 |
| 0.2 | 1.01 | 1.02 | 1.20 | 1.30 |
| 0.4 | 1.01 | 1.01 | 1.01 | 1.14 |
| 1.0 | 1.00 | 1.00 | 1.00 | 1.06 |
| 1.4 | 1.00 | 1.00 | 1.00 | 1.03 |




- 18MV
- Co-Cr-MO
- Homogenous vs. broad vs. narrow beam

TABLE I. BSDF versus distance from interface for $10\times10$ cm$^2$ where BSDF$=D_i/D_h$ and $D_i$ and $D_h$ are the doses with and without the presence of the interface, respectively.

| Material | Bone | | Lead | |
|---|---|---|---|---|
| Density (g/cm$^3$) | 1.83 | | 11.4 | |
| Atomic no. | 13 | | 82 | |
| Distance (cm) | 6 MV | 18 MV | 6 MV | 18 MV |
| 0.1 | 1.03 | 1.04 | 1.34 | 1.45 |
| 0.2 | 1.01 | 1.02 | 1.20 | 1.30 |
| 0.4 | 1.01 | 1.01 | 1.01 | 1.14 |
| 1.0 | 1.00 | 1.00 | 1.00 | 1.06 |
| 1.4 | 1.00 | 1.00 | 1.00 | 1.03 |

TABLE II. FDPF versus distance from the interface for $10\times10$ cm$^2$ where FDPF$=D_i/D_h$ and $D_i$ and $D_h$ are the doses with and without the presence of the interface, respectively.

| Material | Bone | | Steel | | Lead | |
|---|---|---|---|---|---|---|
| Density (g/cm$^3$) | 1.83 | | 7.76 | | 11.4 | |
| Thickness (g/cm$^3$) | 1.83 | | 2.56 | | 2.28 | |
| Atomic no. | 13 | | 26 | | 82 | |
| Distance (cm) | 6 MV | 18 MV | 6 MV | 18 MV | 6 MV | 18 MV |
| 0.05 | 0.94 | 1.05 | 0.85 | 1.20 | 0.84 | 1.41 |
| 0.1 | 0.95 | 0.04 | 0.87 | 1.19 | 0.85 | 1.40 |
| 0.5 | 0.98 | 1.03 | 0.92 | 1.15 | 0.88 | 1.29 |
| 1.0 | 0.99 | 1.02 | 0.94 | 1.11 | 0.91 | 1.21 |
| 2.0 | 0.99 | 1.01 | 0.95 | 1.05 | 0.93 | 1.10 |
| 4.0 | 0.99 | 1.00 | 0.94 | 0.98 | 0.93 | 0.98 |
| 6.0 | 0.99 | 0.99 | 0.94 | 0.96 | 0.93 | 0.94 |

# CT - metal artefacts

- Correct assignment of electron densities to the prosthesis (metal implants) is crucial for correct dose calculations

- This can be done in two ways:
  - to overwrite in the TPS the appropriate value of electron density in the area of the prosthesis
  - to define the correct conversion curve of HU to relative electron densities

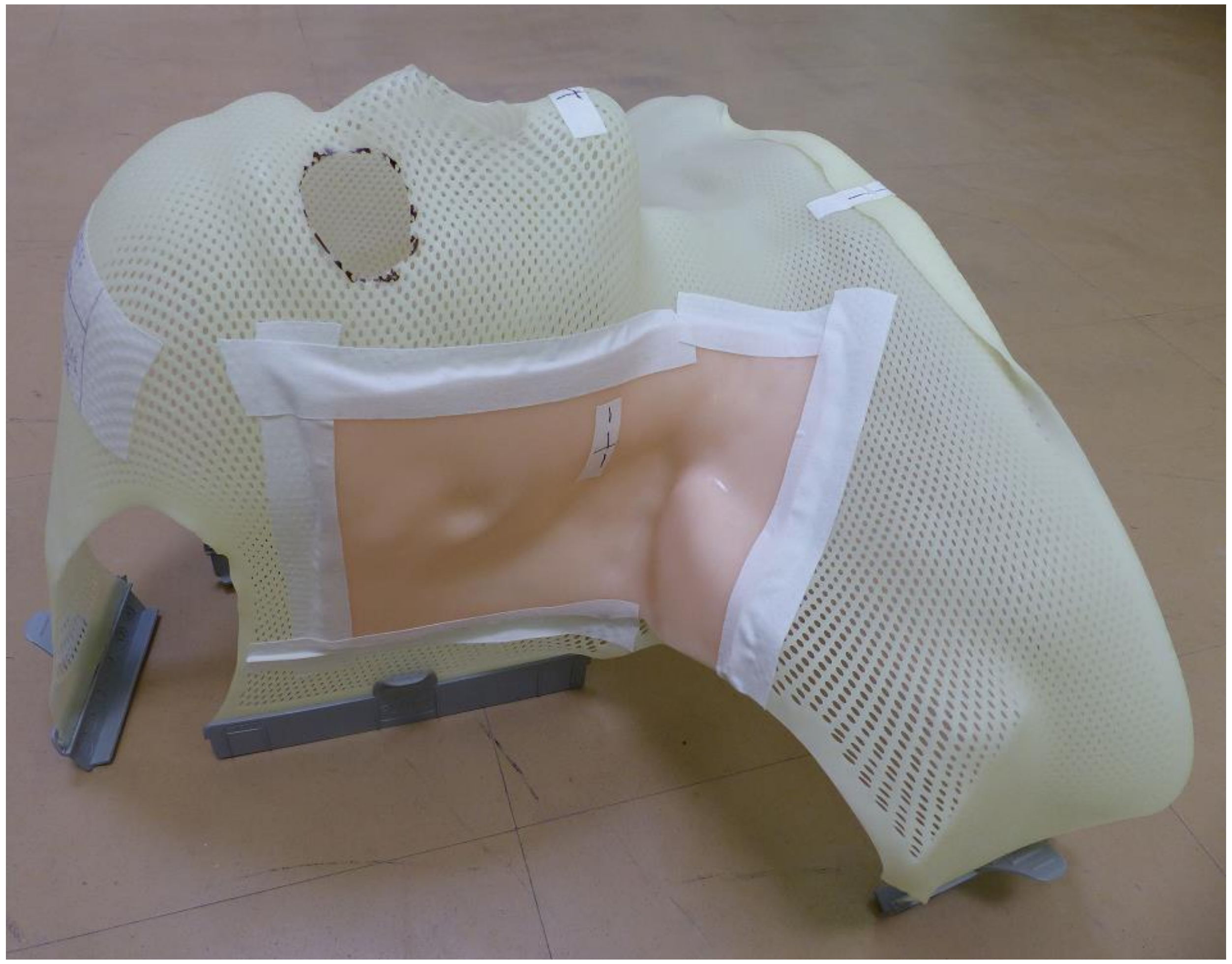

# CT - metal artefacts

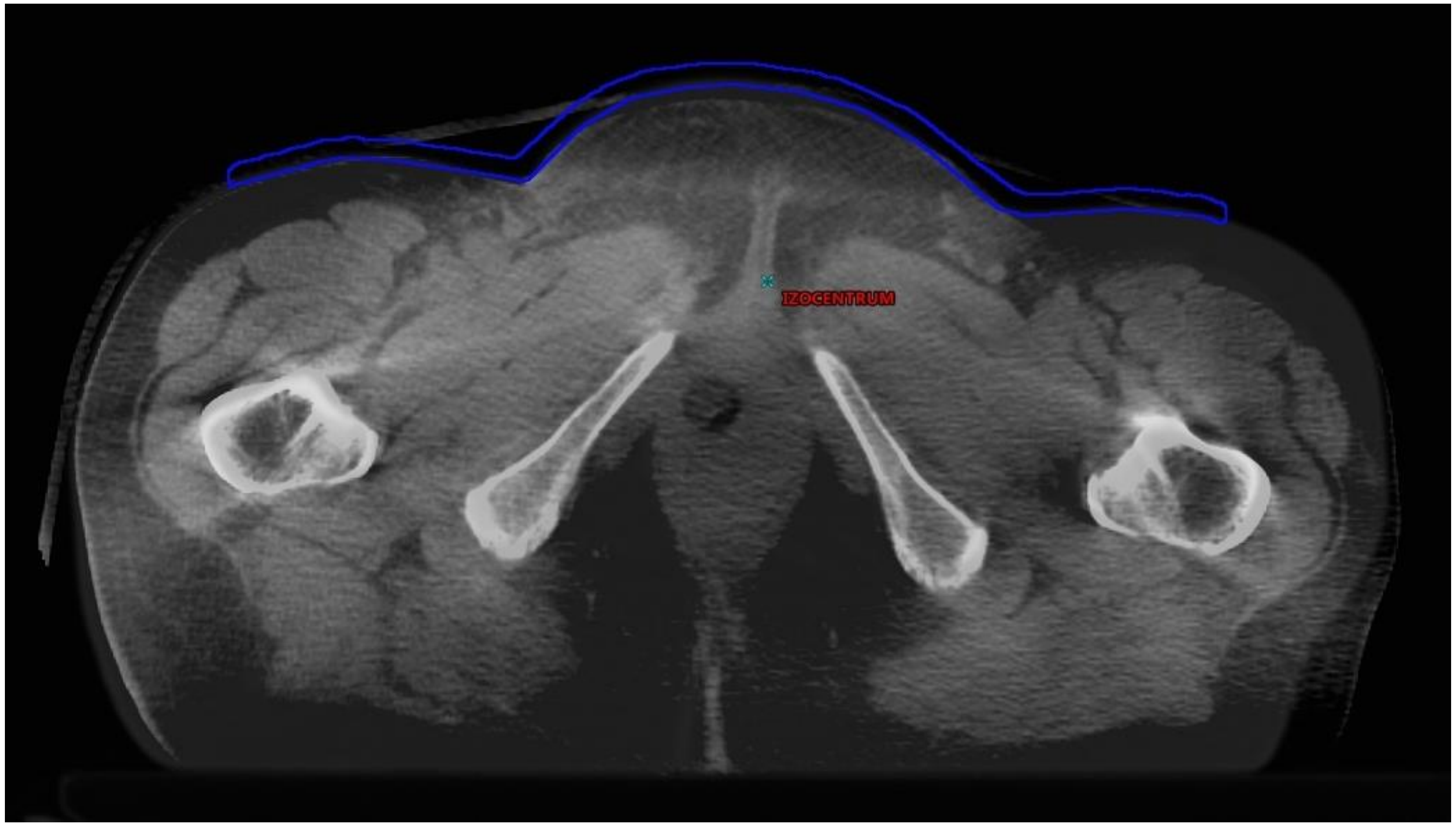
- PTV close to a prosthesis
  - avoid beams passing through the prosthesis
  - for VMAT delineate prosthesis as a dummy structure and minimize dose
  - block entrance dose

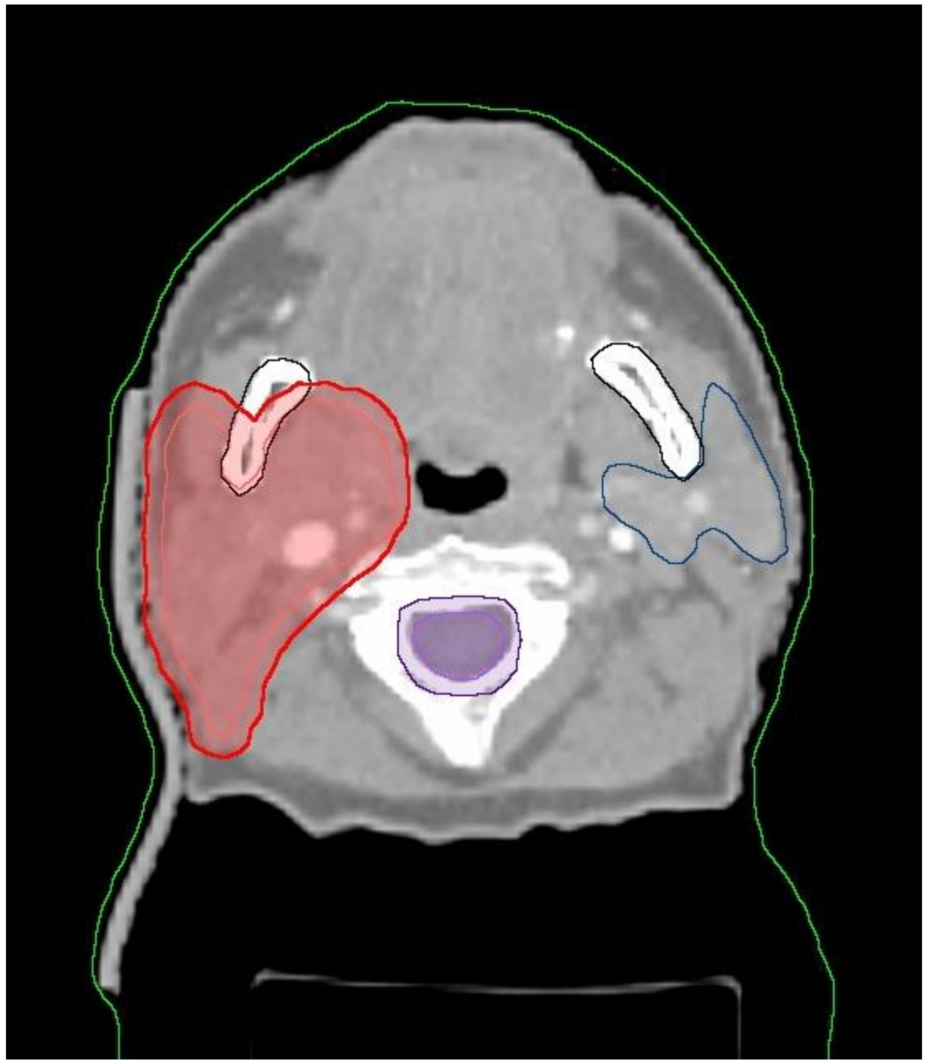

- PTV includes a prosthesis
  - use lower energy due to lower values and shorter ranges of dose perturbance on the border of high-density inhomogeneities
- Use multiple directions

# Bolus

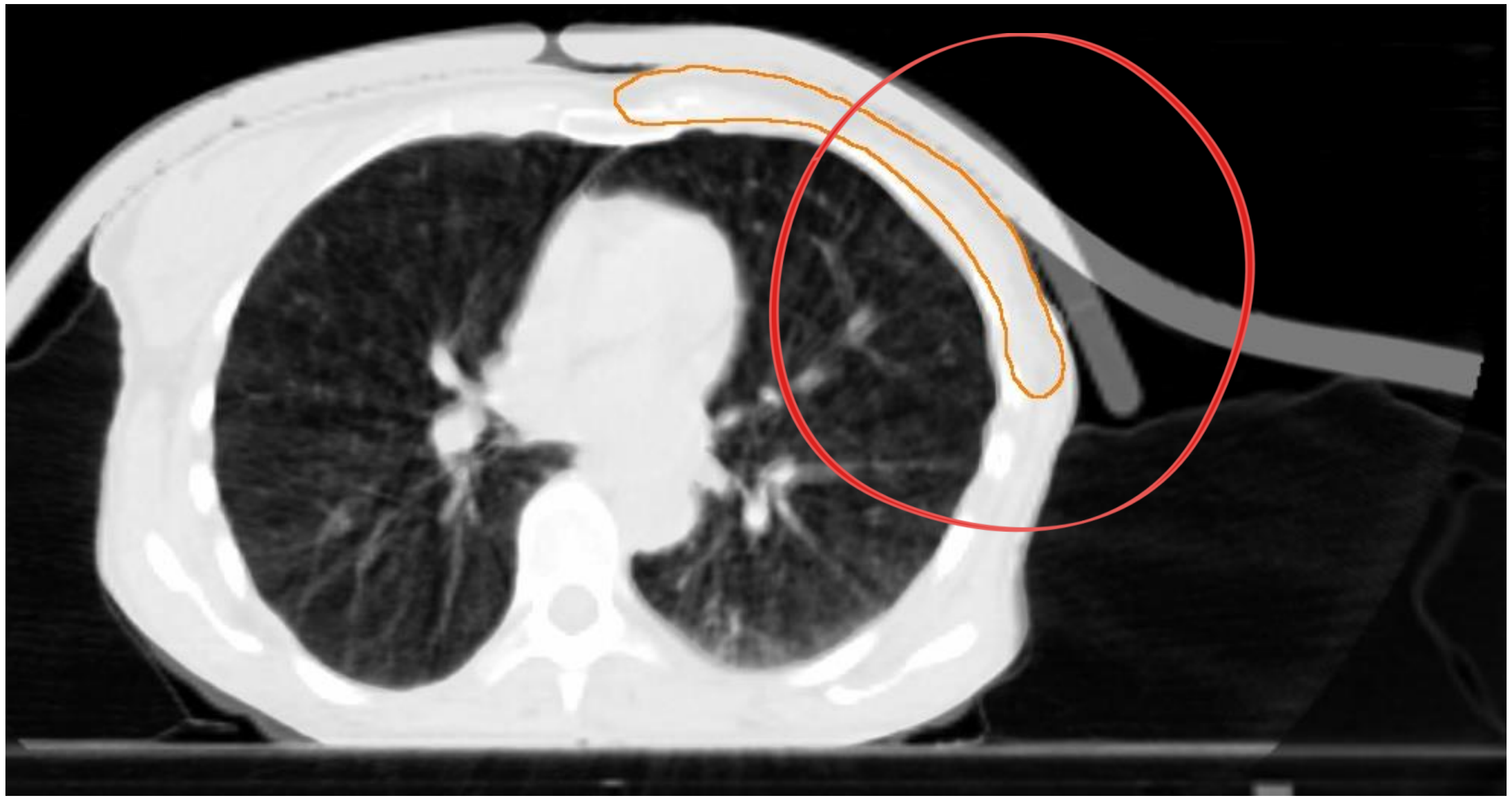
- Bolus should be prepared before CT scanning and included in images

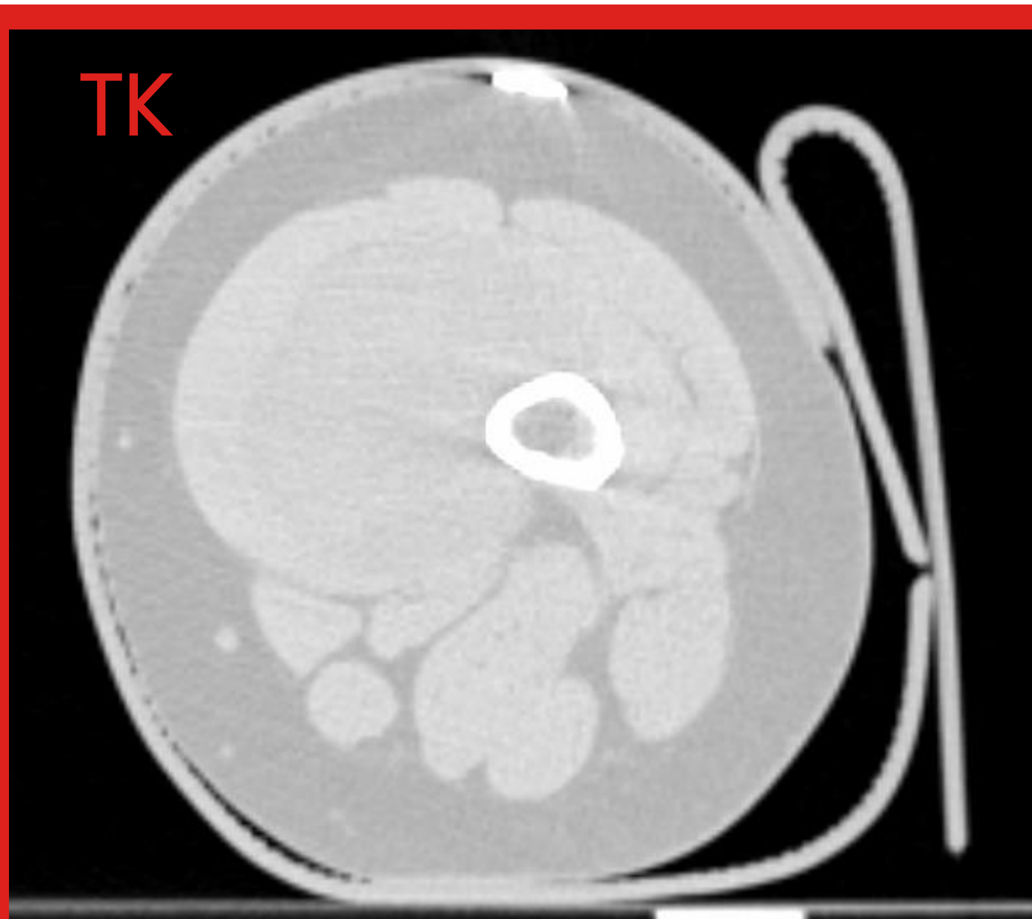


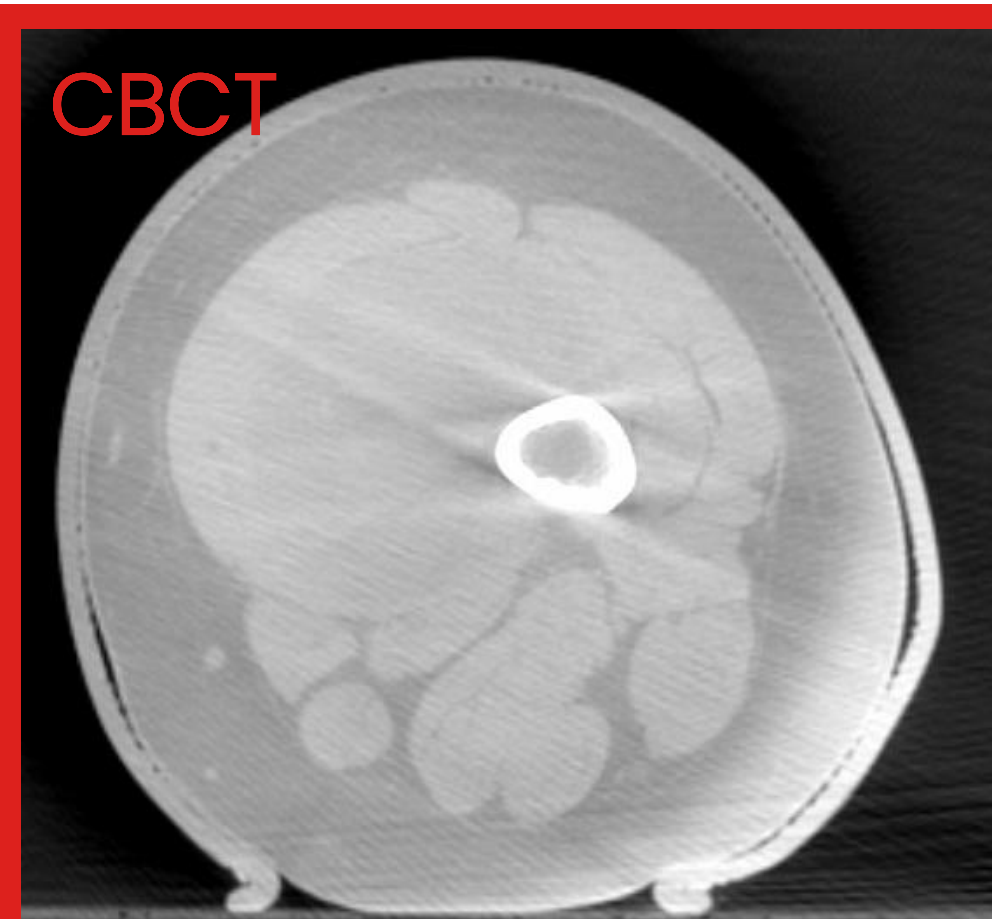


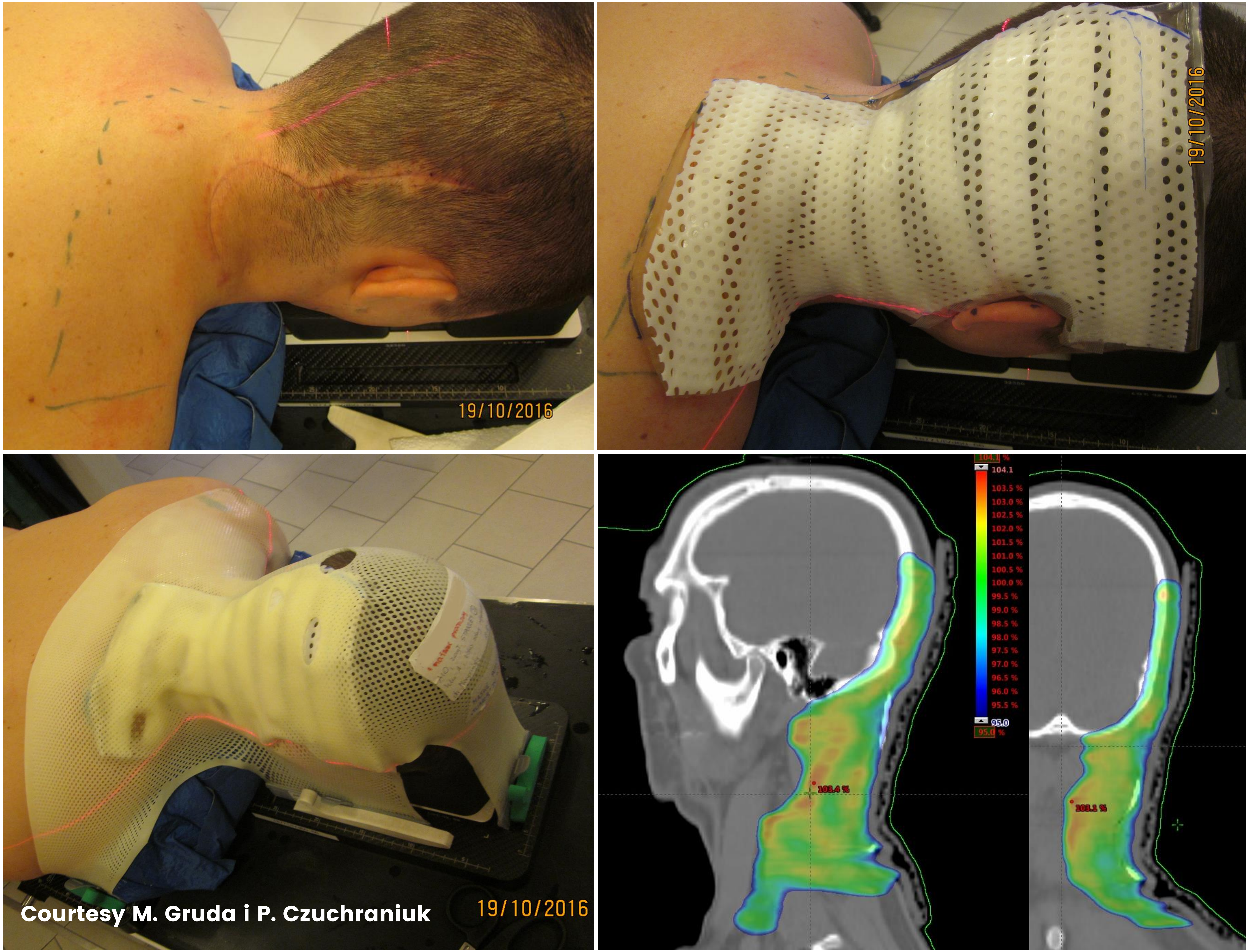



# Bolus placement during Treatment (CBCT)

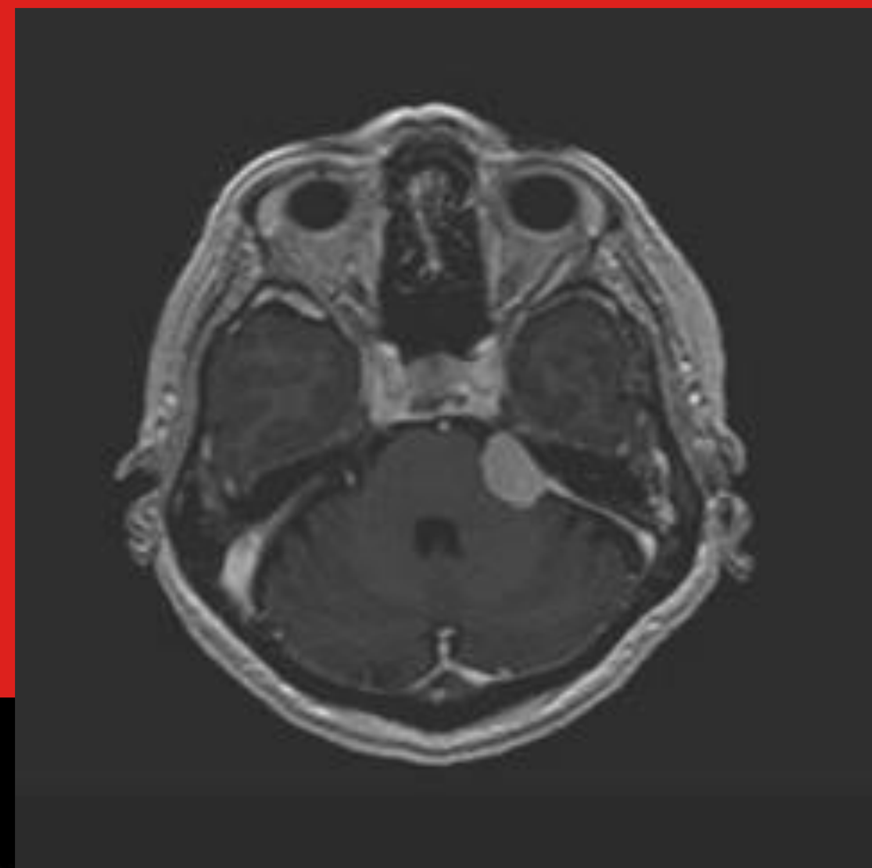


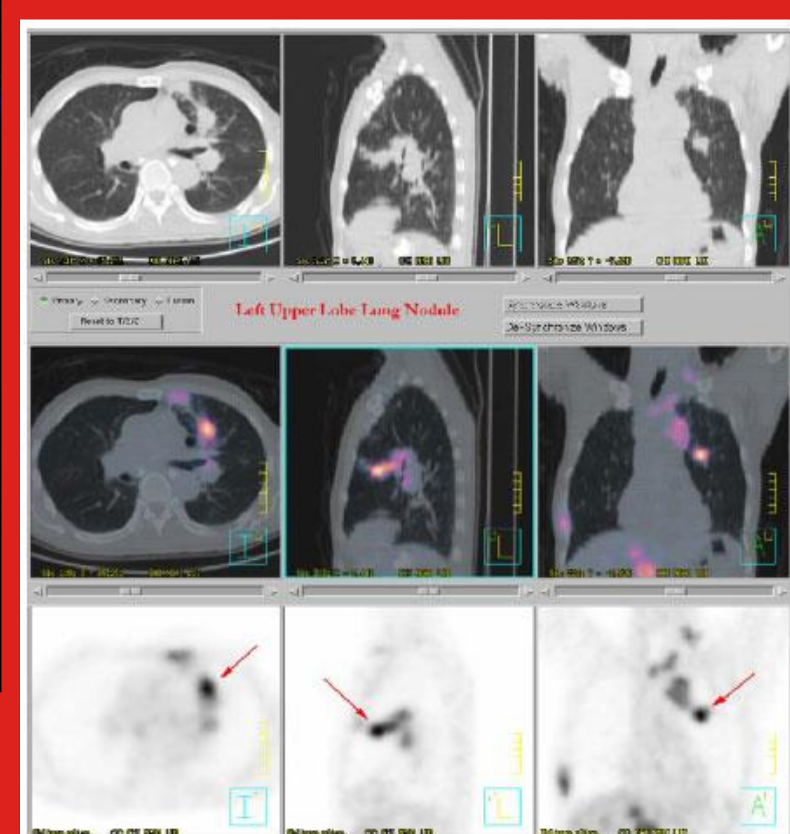


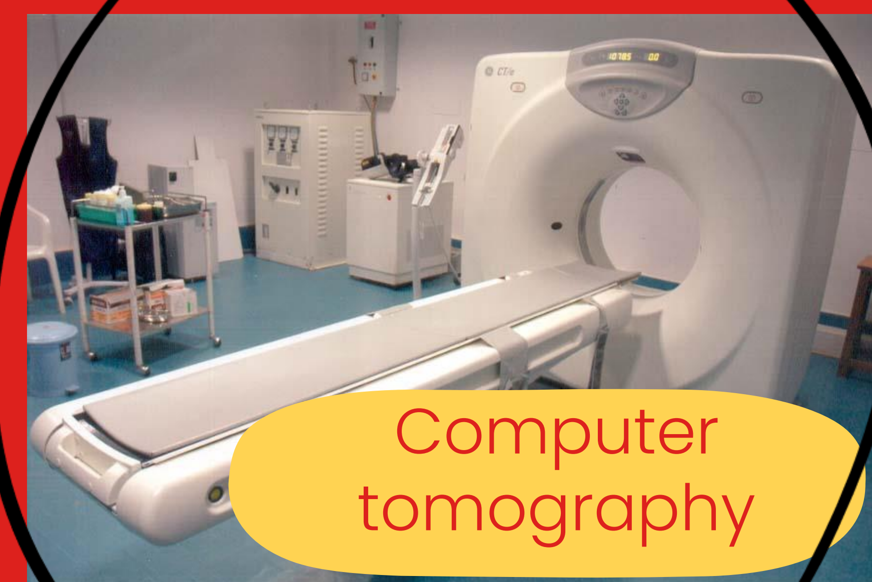

19/10/2016
19/10/2016
19/10/2016
Courtesy M. Gruda i P. Czuchraniuk
104.1
103.5 %
103.0 %
102.5 %
102.0 %
101.5 %
101.0 %
100.5 %
100.0 %
99.5 %
99.0 %
98.5 %
98.0 %
97.5 %
97.0 %
96.5 %
96.0 %
95.5 %
95.0
103.4 %
103.1 %

# Additional Imaging should be included to improve delineation accuracy.

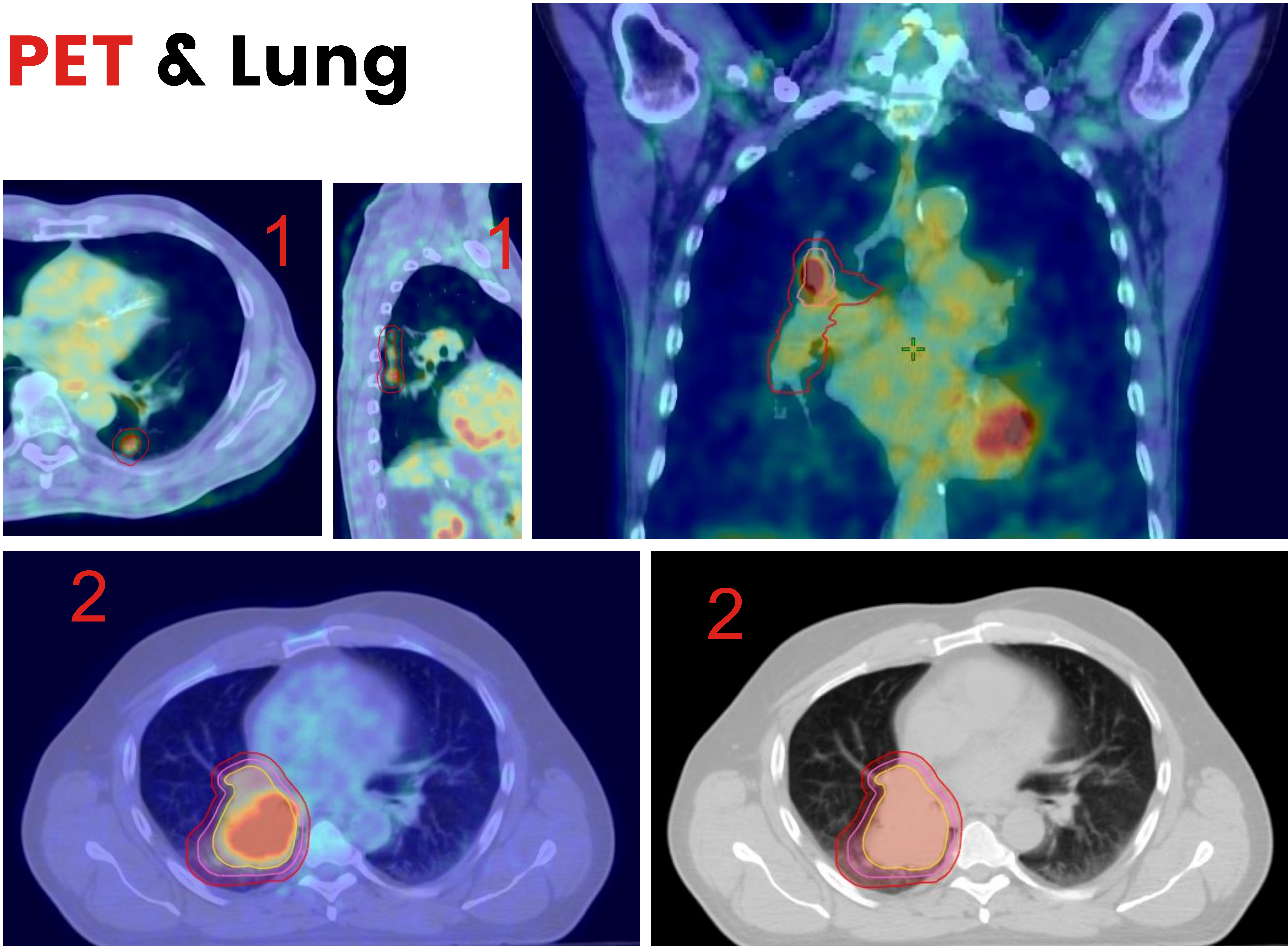




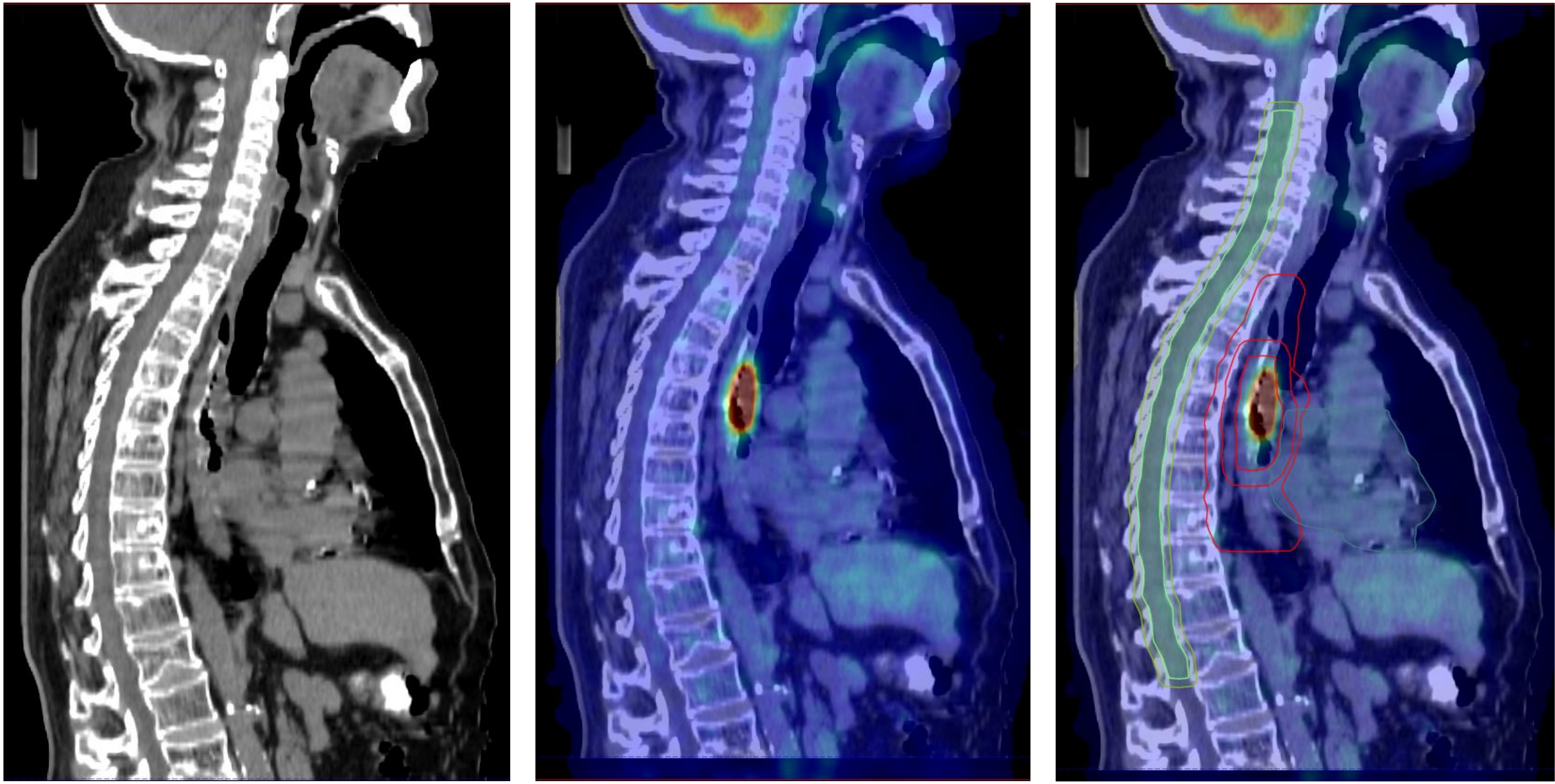





# PET & Lung



# PET-CT & esophagus



# PET

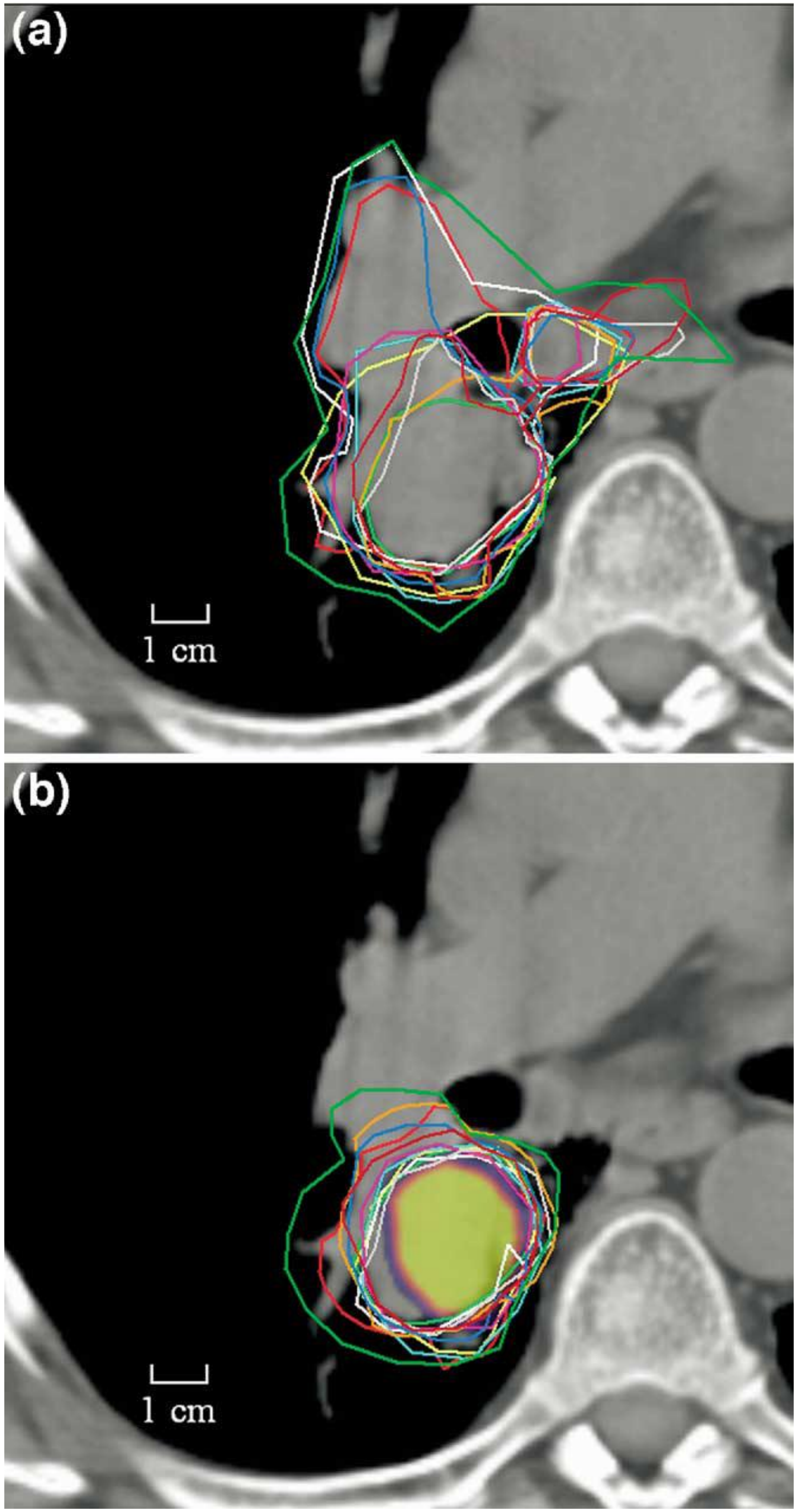
11 oncologists contoured 22 patients with an interval of 1 year

1 time only CT

2nd time CT + PET

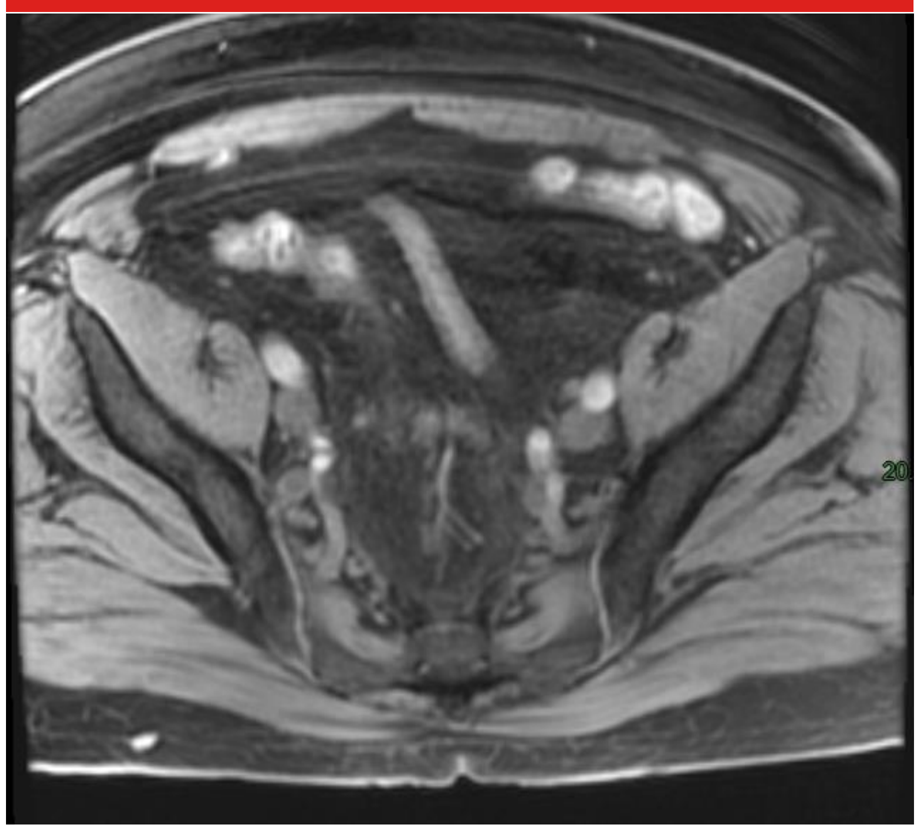
- 3D observer variation 1.0 cm (SD, CT) vs 0.4 cm (SD, CT–FDG-PET).
- The largest differences were the area of atelectasis (SD 1.9 cm vs to 0.5 cm).
- Smaller differences in interpretation (number of discrepancies 45% vs 18%)
- Average contouring time 12 vs. 16 min, $p < 0.001$
- Average number of corrections 25 vs. 39, $p < 0.001$)

Roel J H M Steenbakkers et al. IROBP 64(2), 2006

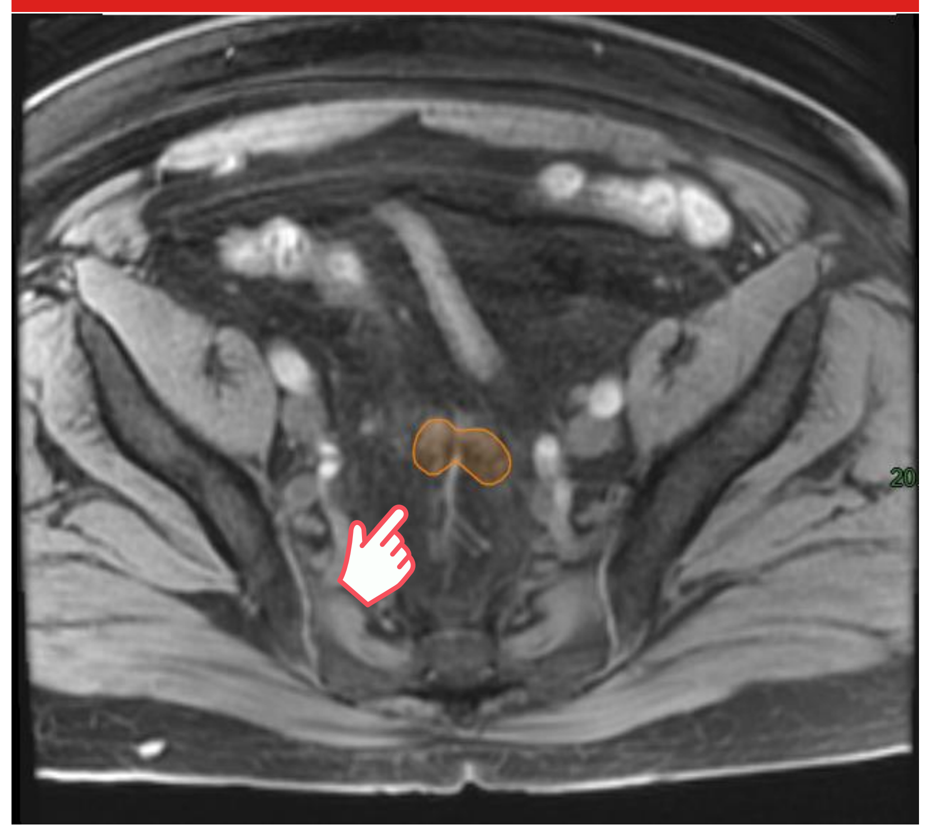


# CT vs. Nuclear Magnetic Resonance

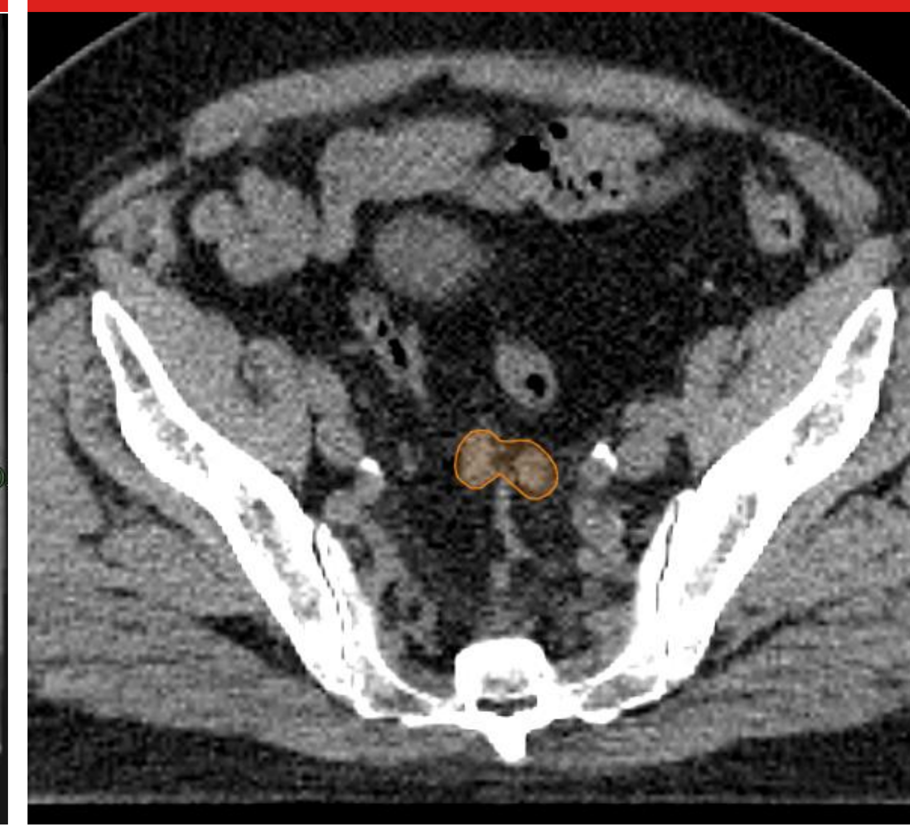
NMR



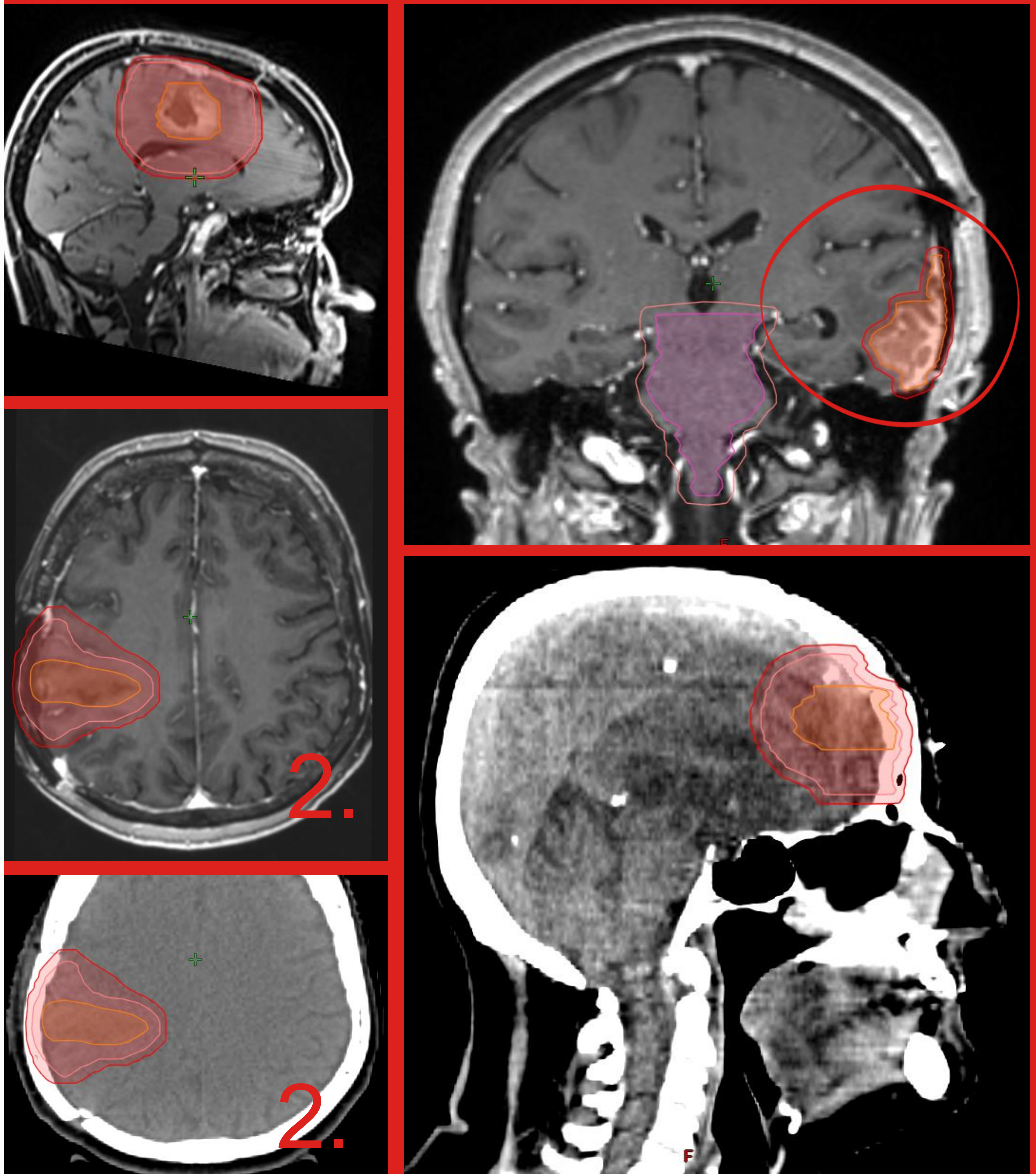

NMR GTV - N

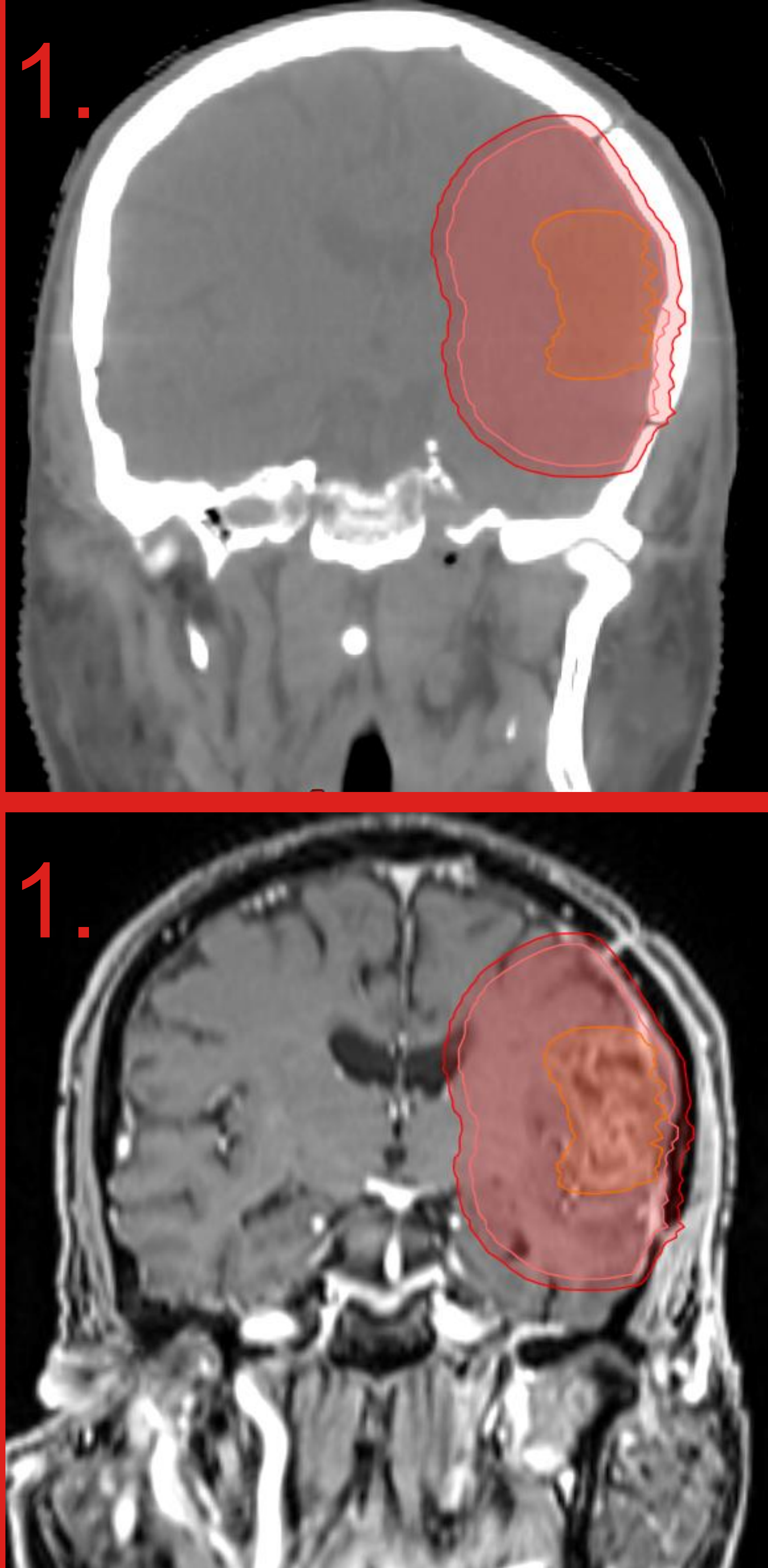



NMR GTV - N

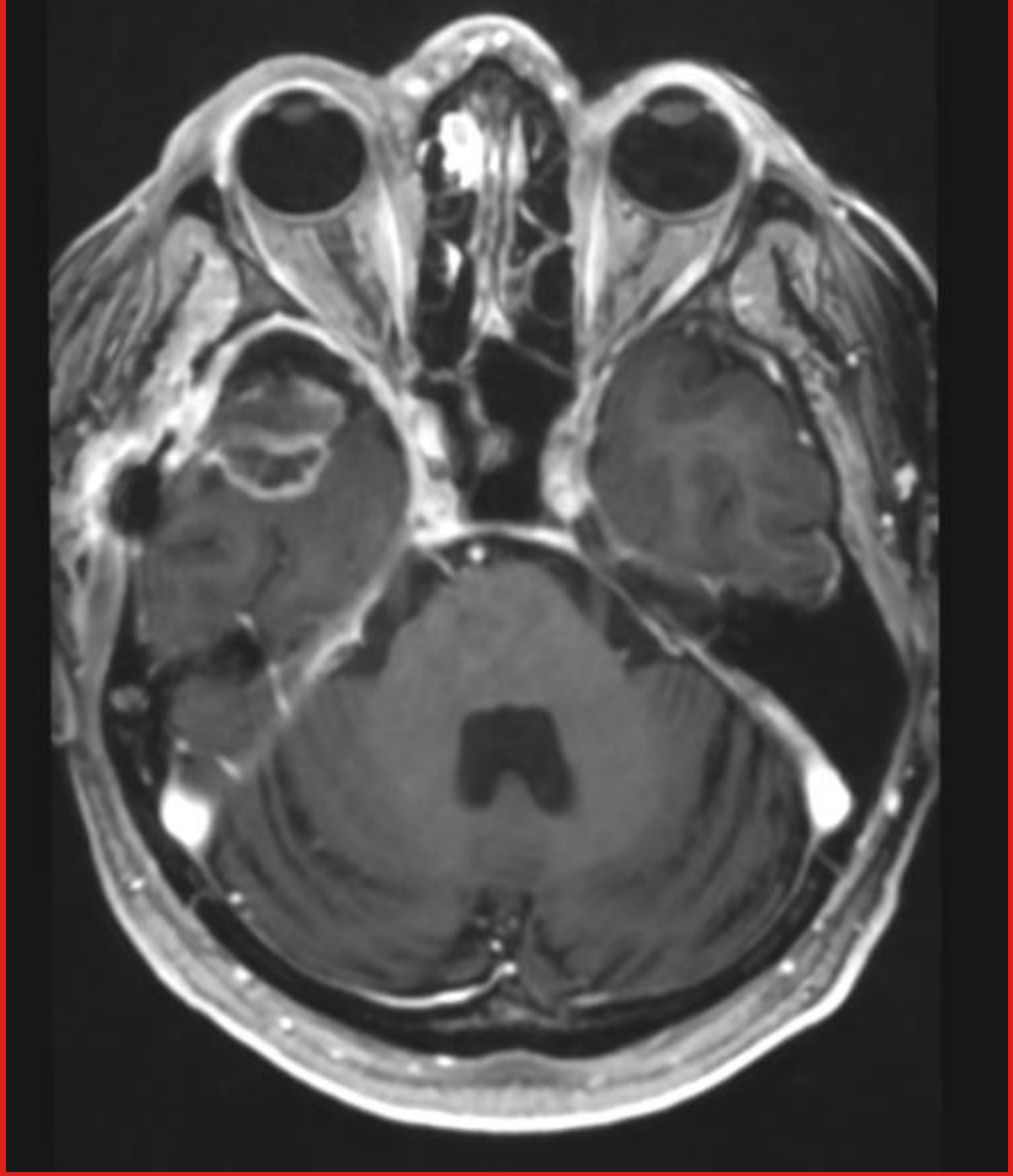


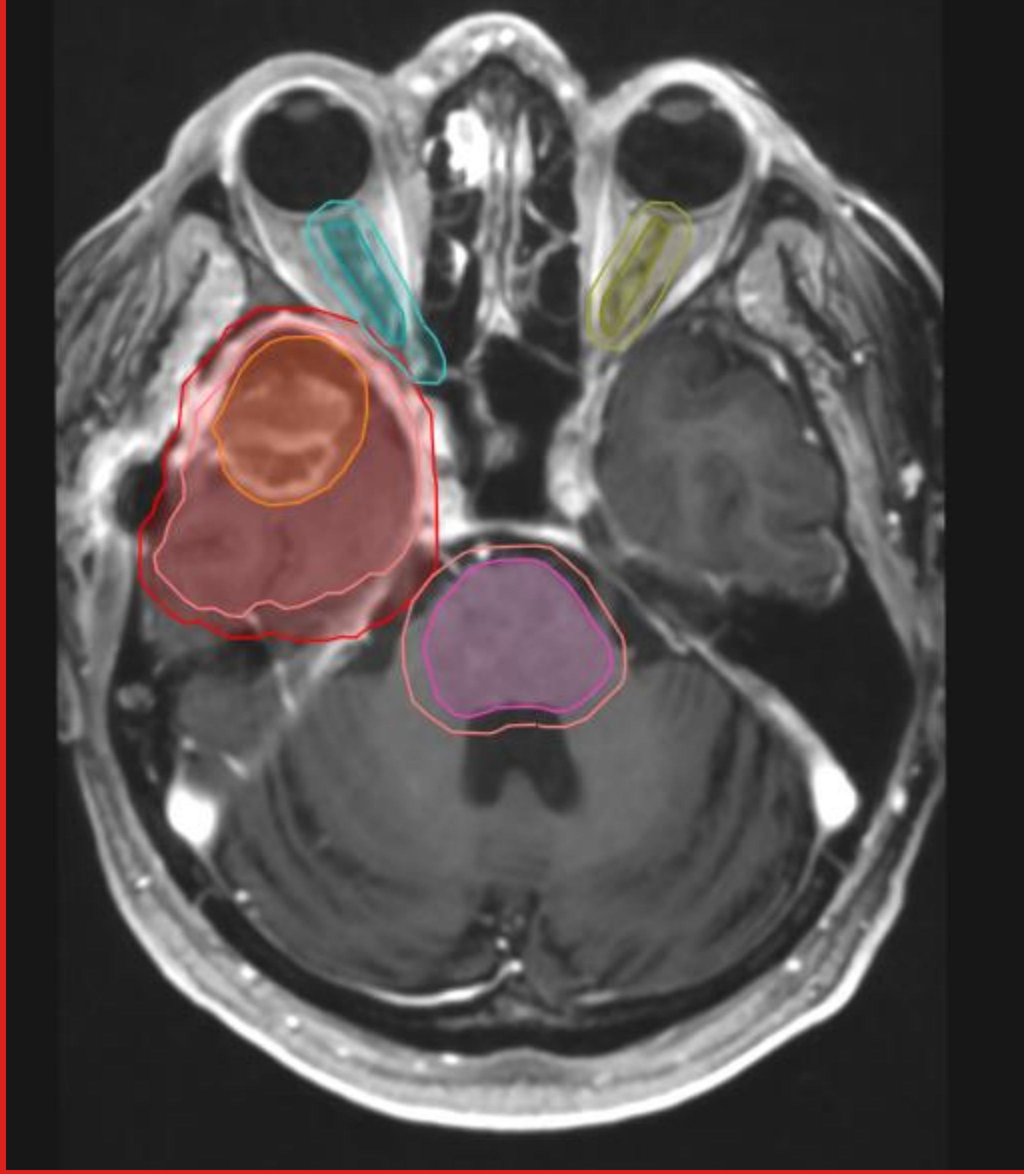
# GTV/CTV/PTV - brain

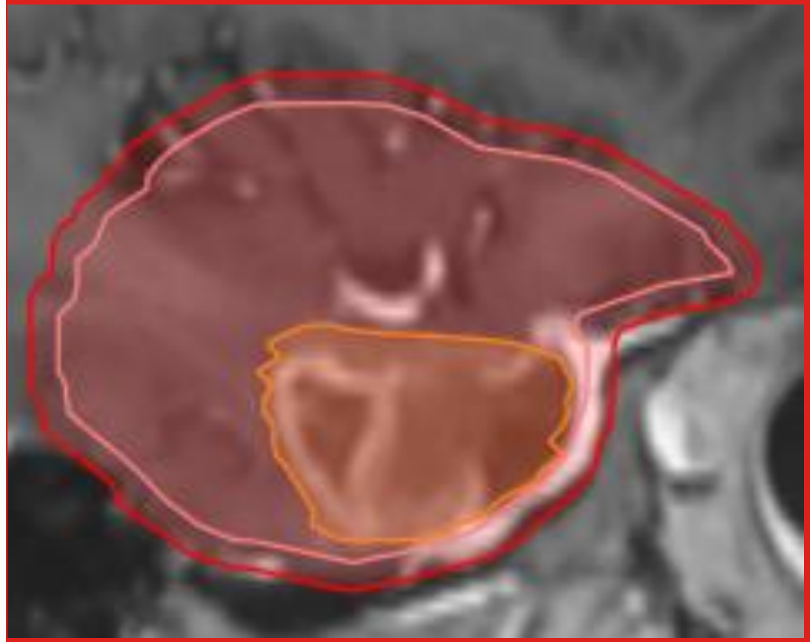


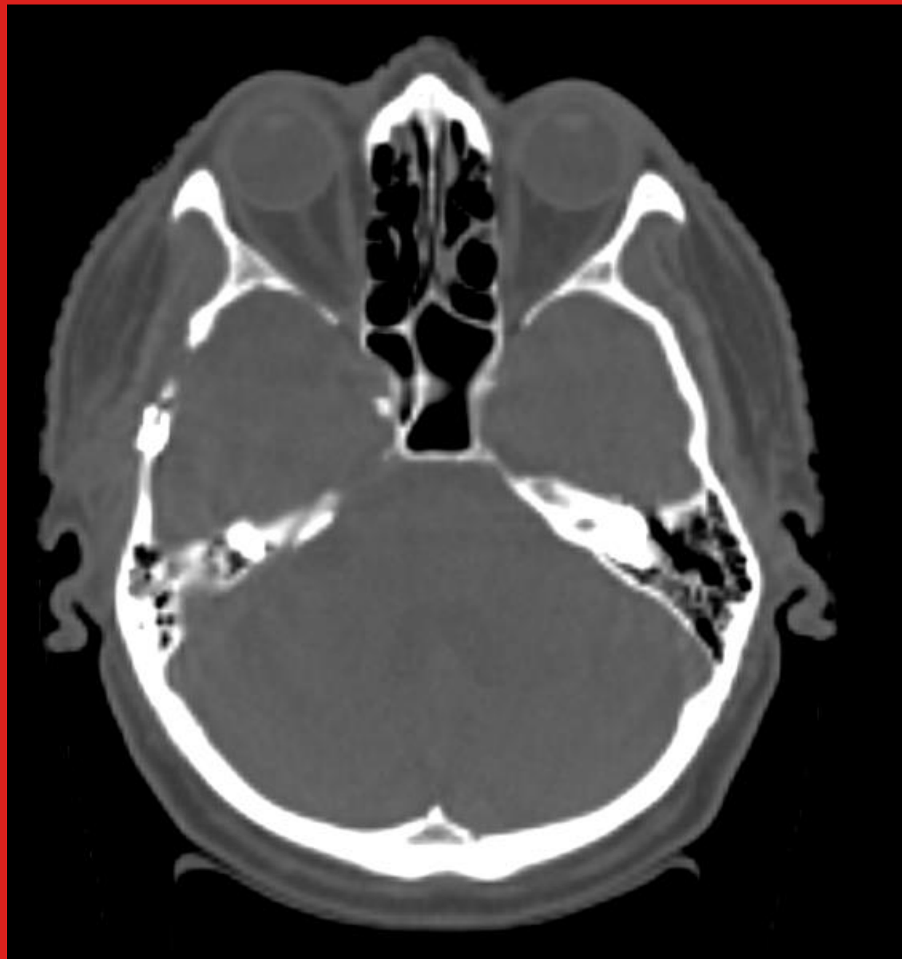



## brain









# QA - NMR

## Geometric distortion in clinical MRI systems Part I: evaluation using a 3D phantom

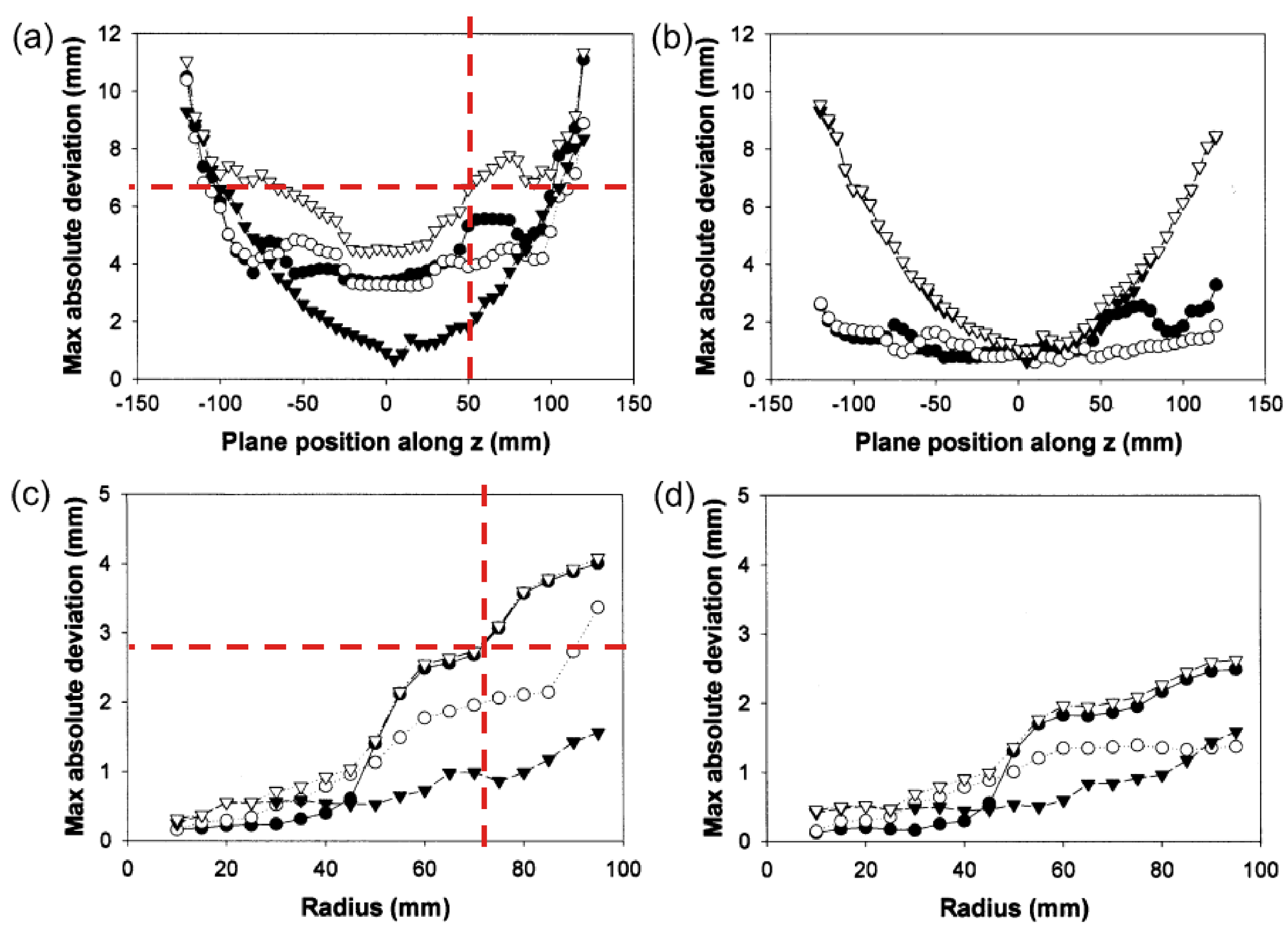
Deming Wang[a,*], Wendy Strugnell[b], Gary Cowin[a], David M. Doddrell[a], Richard Slaughter[b]

[a]Centre for Magnetic Resonance, The University of Queensland, St. Lucia, QLD 4072, Australia
[b]Cardiovascular MRI Research Centre, The Prince Charles Hospital, Chermside, QLD 4032, Australia

Received 28 July 2004; accepted 1 August 2004

Without correction | With correction | $\nabla$ $dr_{max}$

Siemens Sonata 1.5T



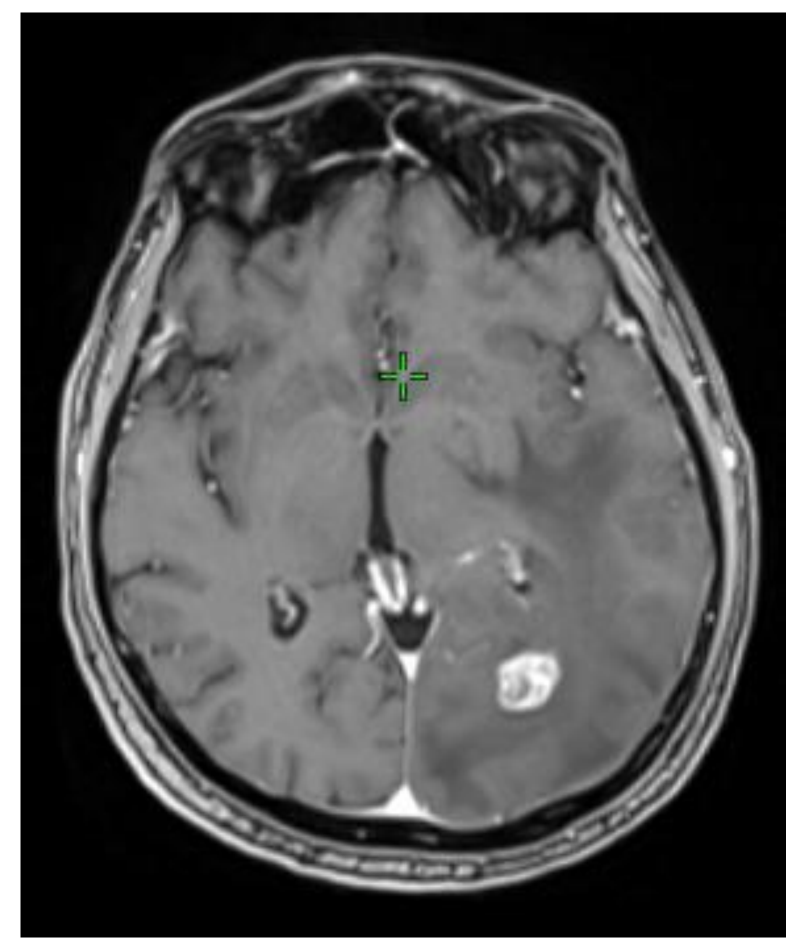
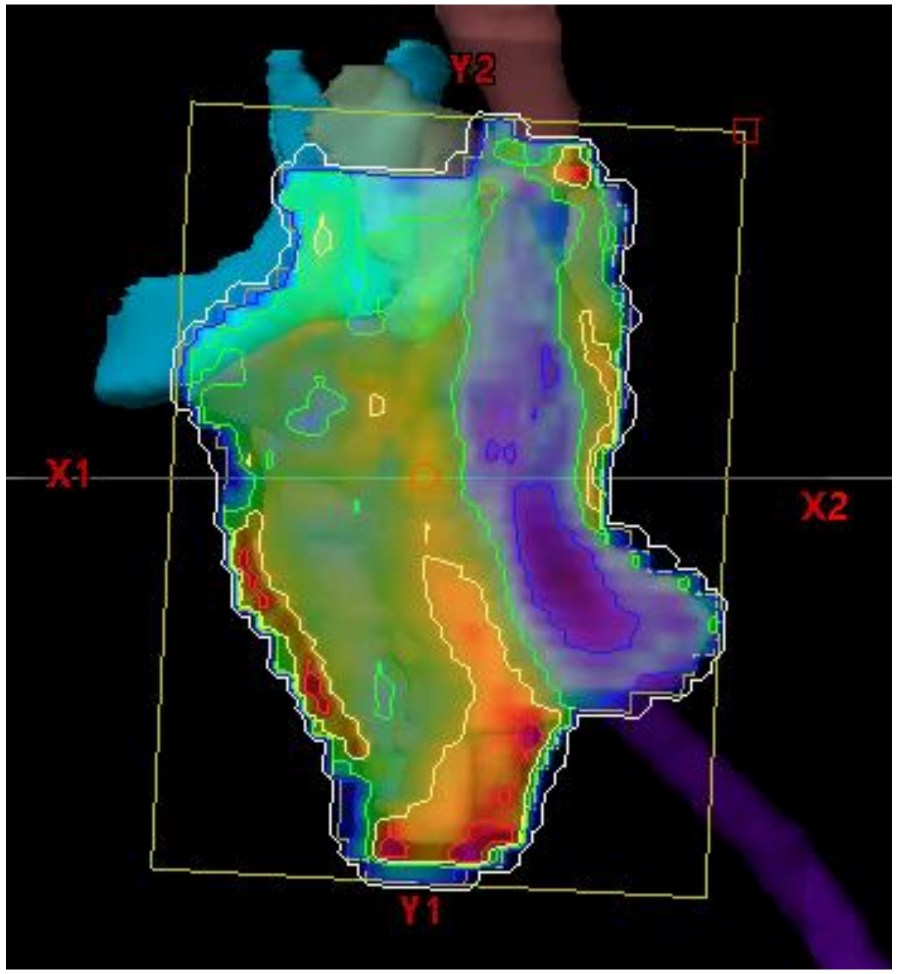
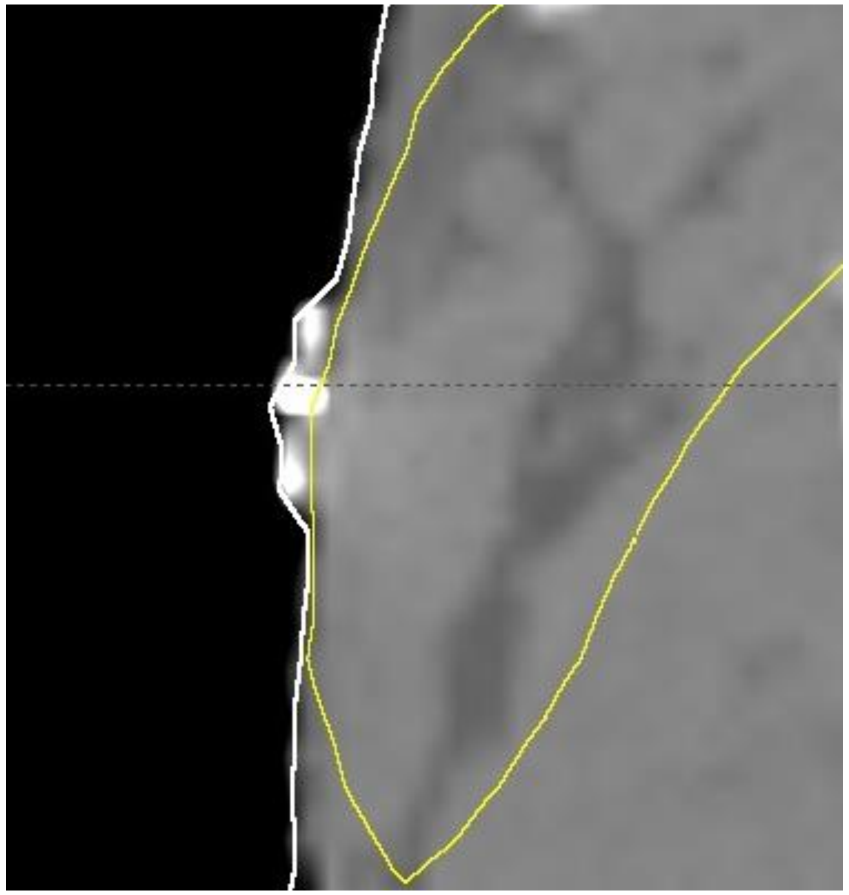



Fig. 2. The maximum absolute deviations (●, $|dx|_{max}$; ○, $|dy|_{max}$; ▼, $|dz|_{max}$; ▽, $dr_{max}$) of the geometric distortion measured in a Siemens Sonata 1.5-T MRI system: (a) in axial planes on images acquired with no vendor's correction; (b) with vendor's correction applied in the *xy* plane; (c) on surface of spheres of different radius (no correction); (d) on surface of spheres of different radius (with correction).

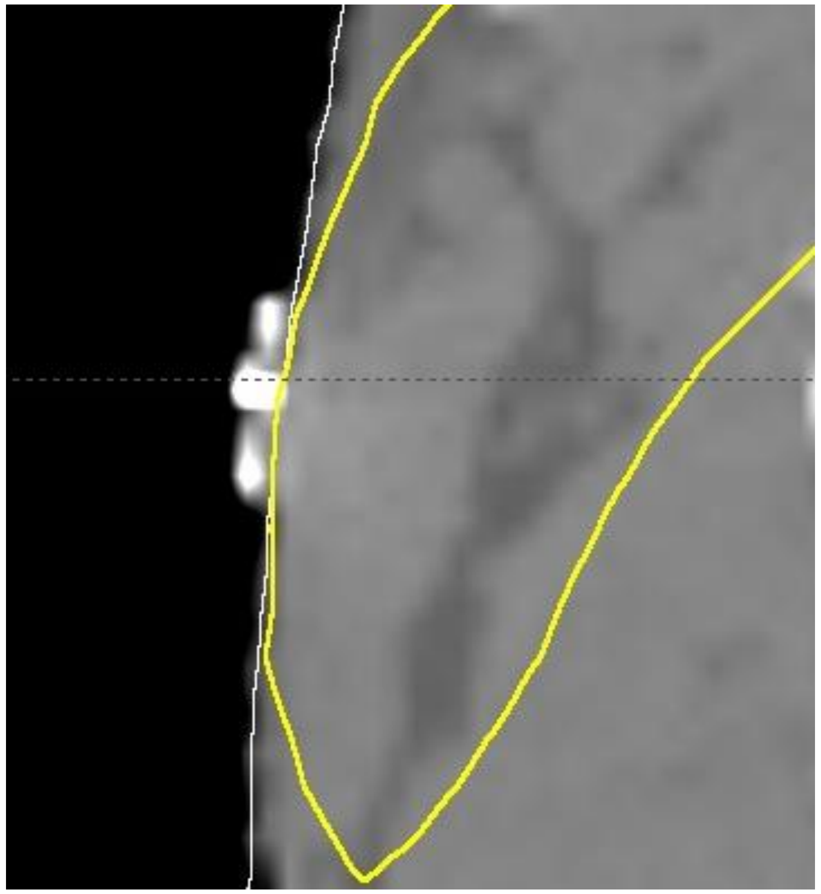
# Contouring

- Body outline should not contain markers
- PTV should not be close to the skin surface (for optimization purposes cut off, e.g. 3mm)

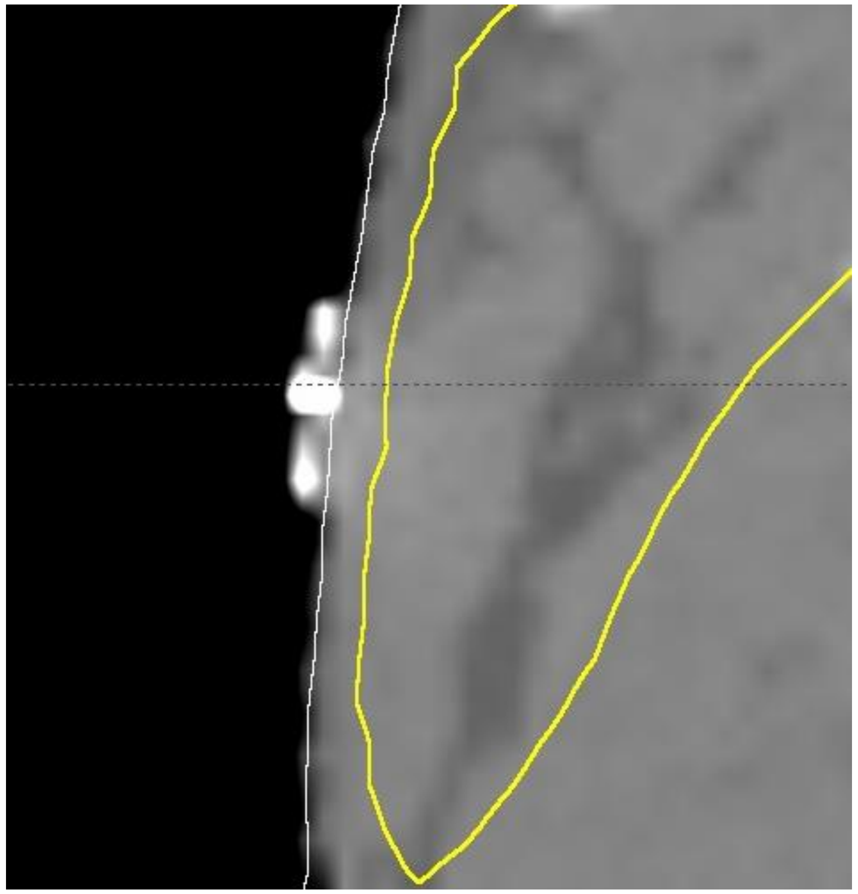


X

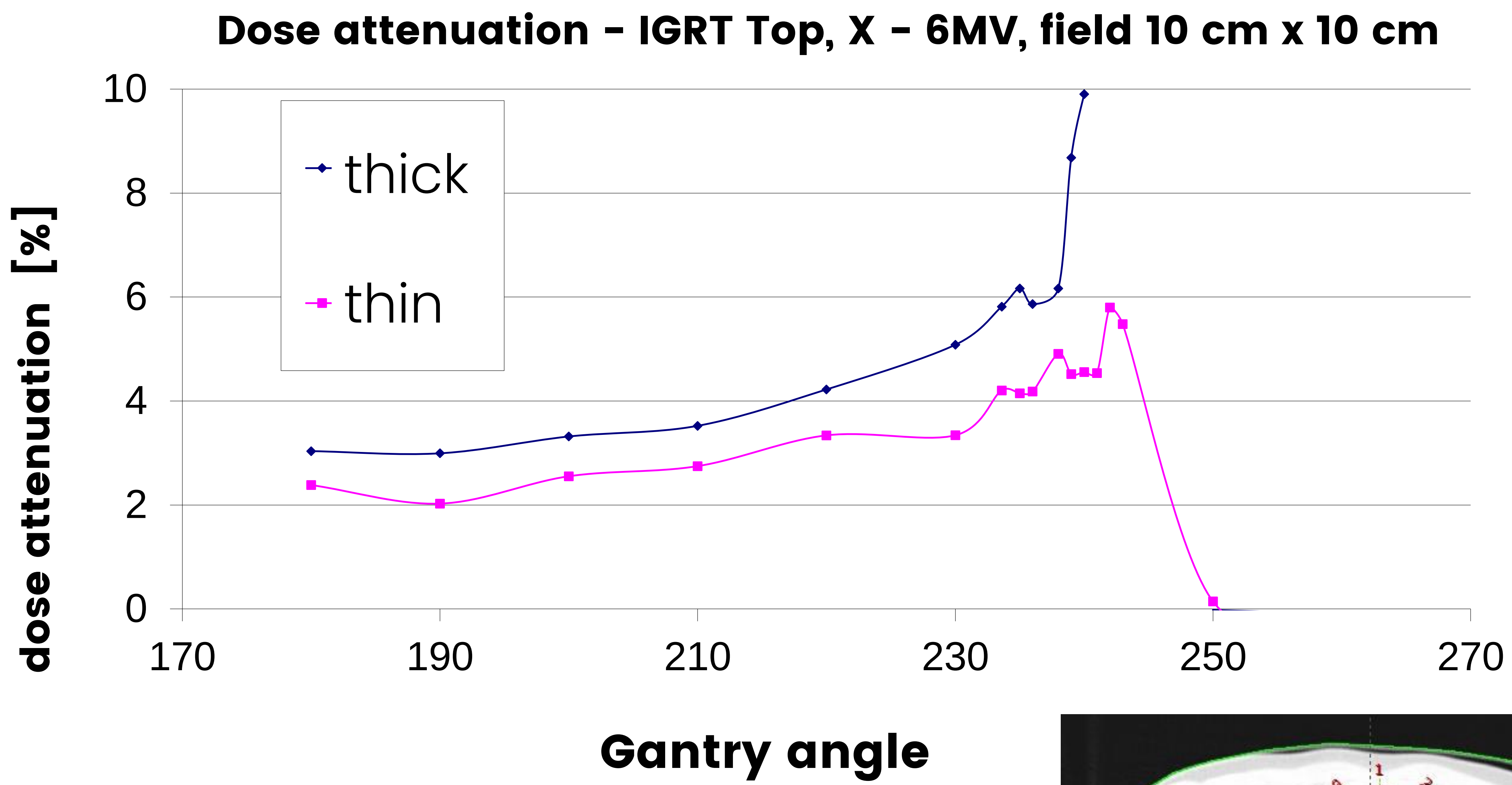


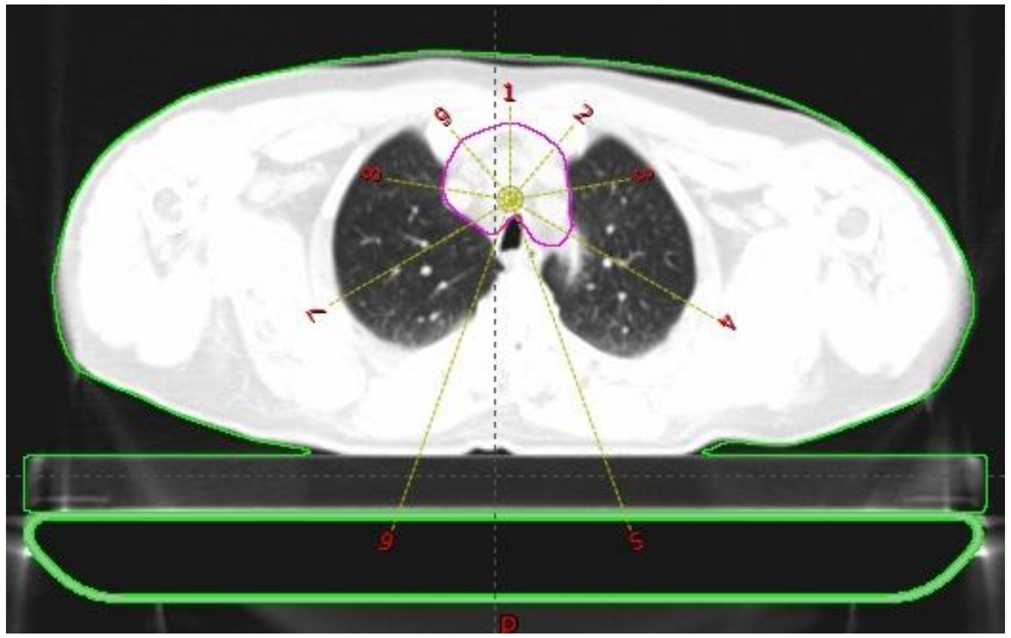
X



✓

# Accessories and treatment table



Dosimetric effects caused by couch tops and immobilization devices: Report of AAPM Task Group 176

Arthur J. Olch[a)]
*Radiation Oncology Department, University of Southern California and Children's Hospital Los Angeles, Los Angeles, California 90027*

**Courtesy Dosimetry team – Agnieszka Walewska**

## Plan preparation

1. Defining the beam geometry
   number of fields, beam angles, collimator angles, technique IMRT vs. VMAT

2. Defining score function, the cost function

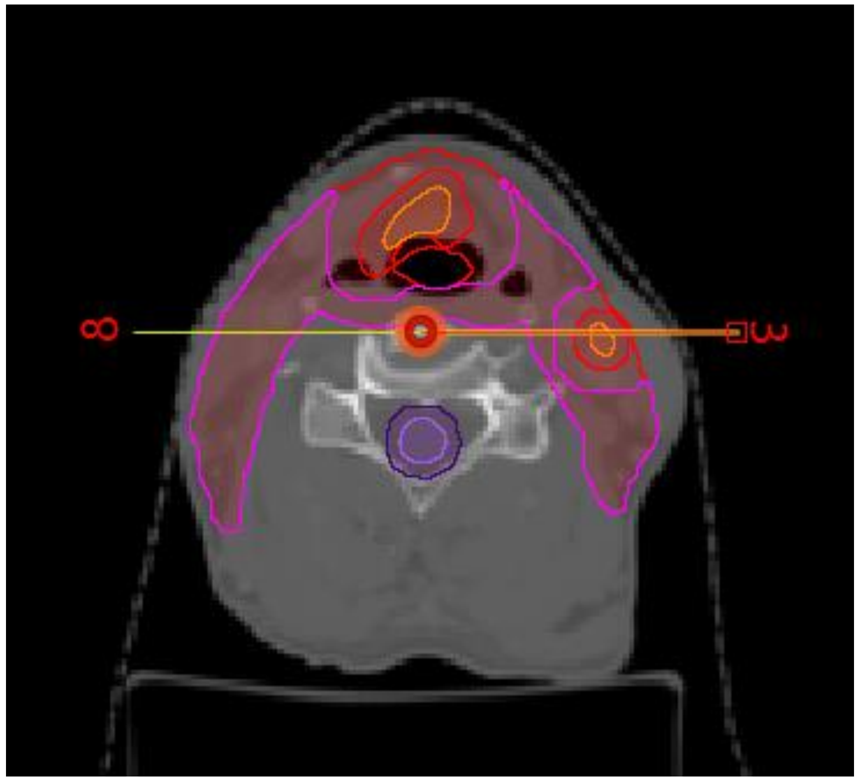
3. Defining objectives and constraints for PTV and OAR's

4. Computer optimization – finding the optimal solution (dose distribution)

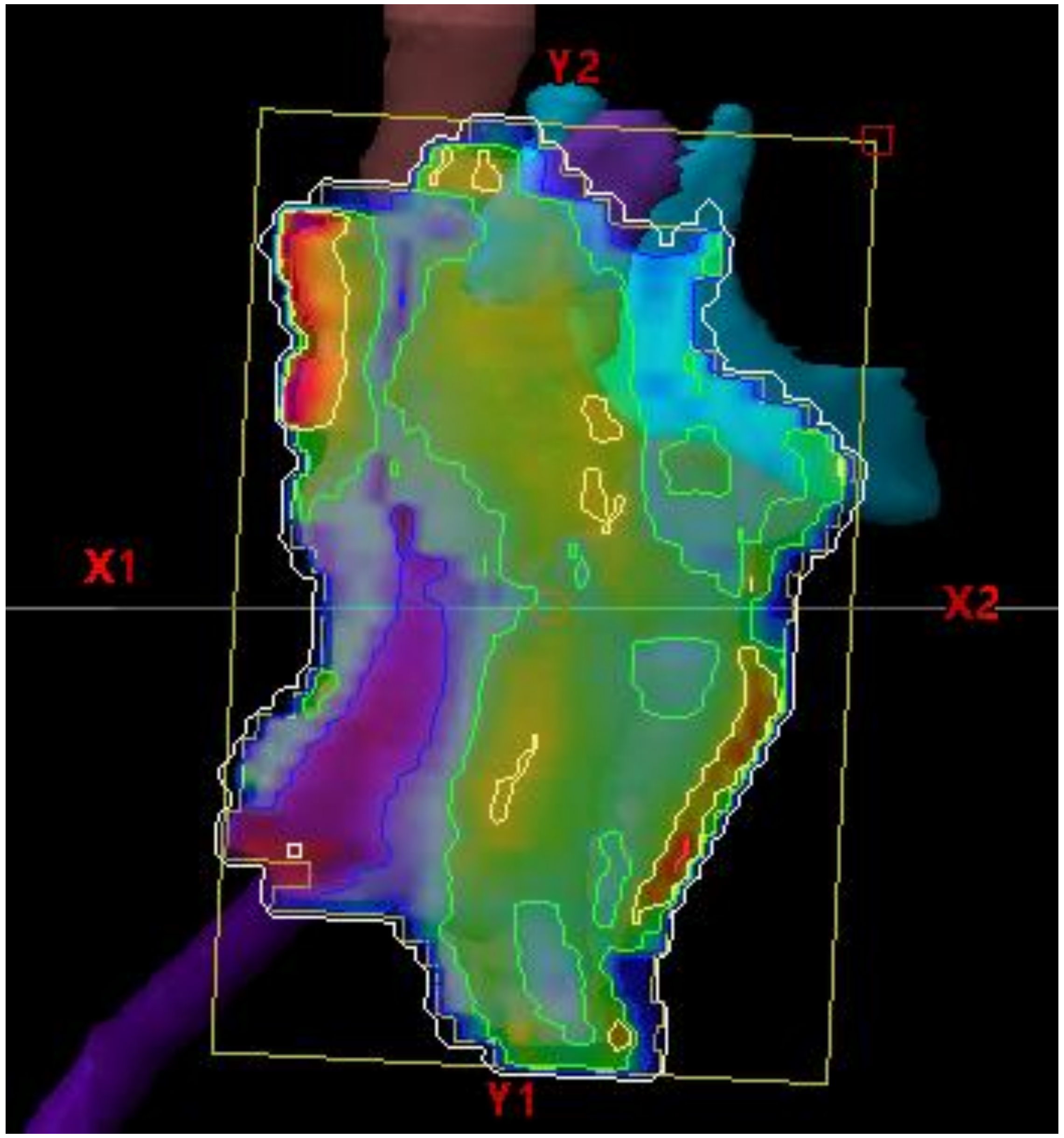
# Beam geometry

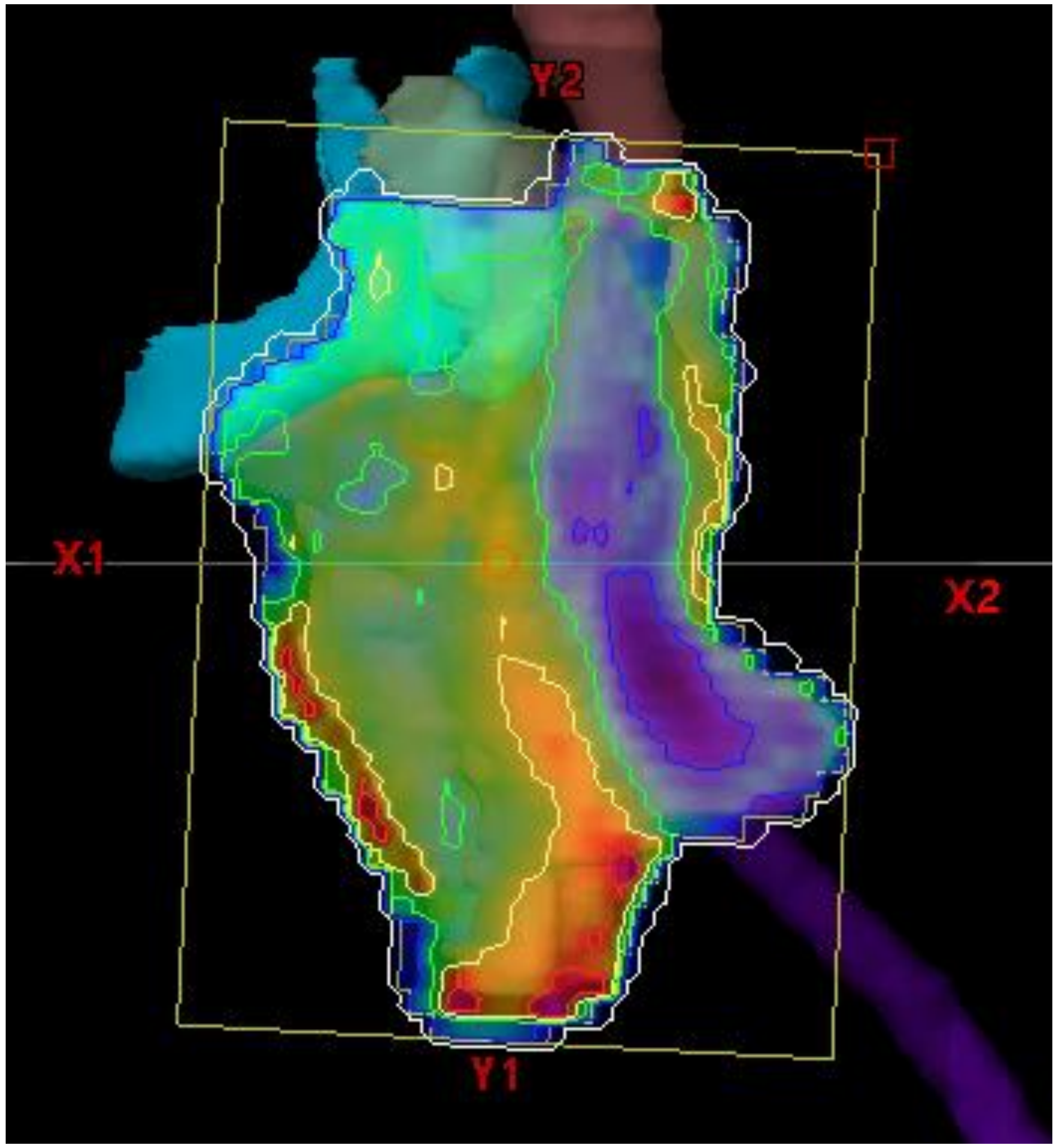
- Opposite beams should be avoided

# Opposite beams



- same task, same result
- expand possibilities




not clinical fluences

# Beam geometry

- Opposite beams should be avoided
- Odd number of beams (?)
- Non-coplanar beams provide less benefit compared to 3D-CRT (?)

# Non-coplanar beams

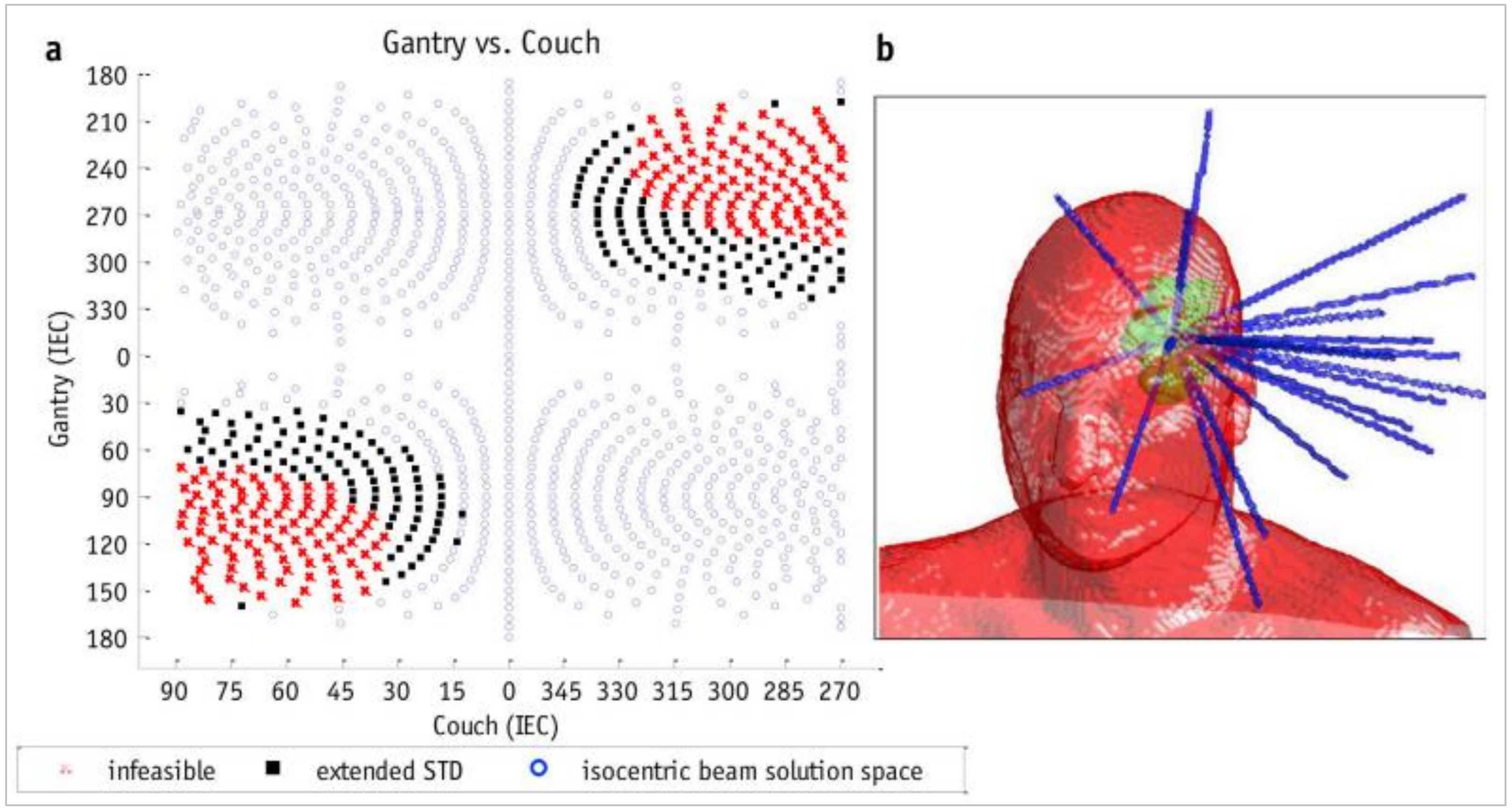
- Reduces the beam overlap and „smear" the dose
- More common in:
  - intracranial stereotactic radiotherapy
  - SRS (single-fraction radiosurgery)
  - SBRT (stereotactic body radiotherapy)
  - APBI (accelerated partial breast irradiation)
- For C-arm linear accelerators (linacs) it is achieved by rotating a treatment couch to a different position
- For C-arm linacs -> time-consuming (increased delivery time) -> for IMRT/VMAT less useful in practice
- O-ring (?)

**However**, modern linacs allow automated rotations (fully automated delivery) -> the view is being reconsidered.

Smyth G at el, . Recent developments in non-coplanar radiotherapy. Br J Radiol 2019

# Non-coplanar beams

- Automated optimization of beam orientation for non-coplanar beams
- Collision




non-collisional search space for non-coplanar beam orientation

Smyth G at el, Br J Radiol 2019; 92

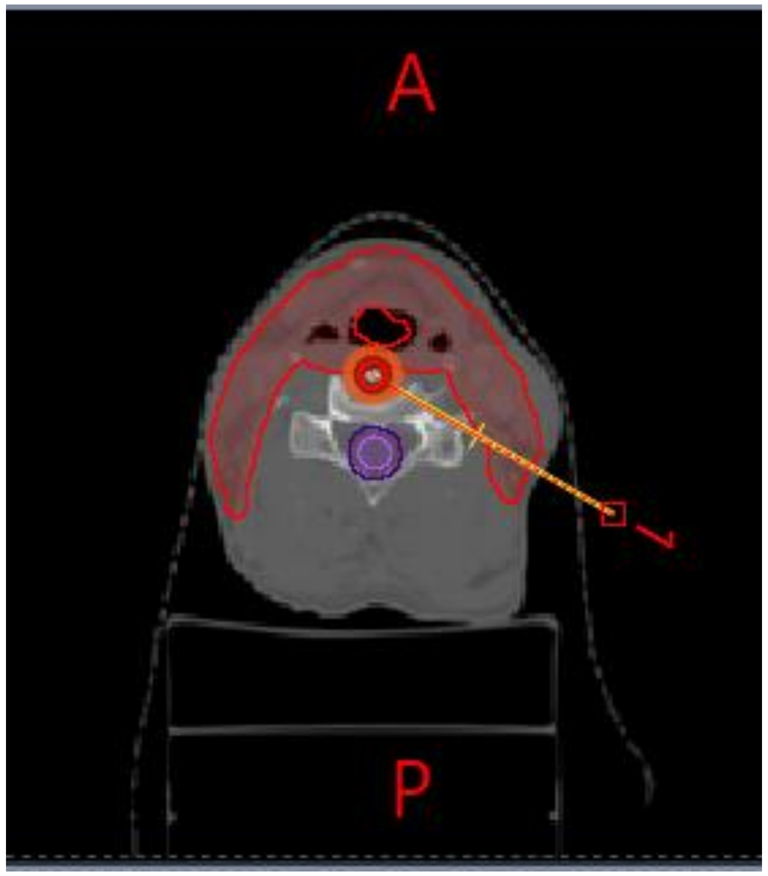
- Intrafraction motion
- Cyberknife, gammaknife – already non-coplanar beams

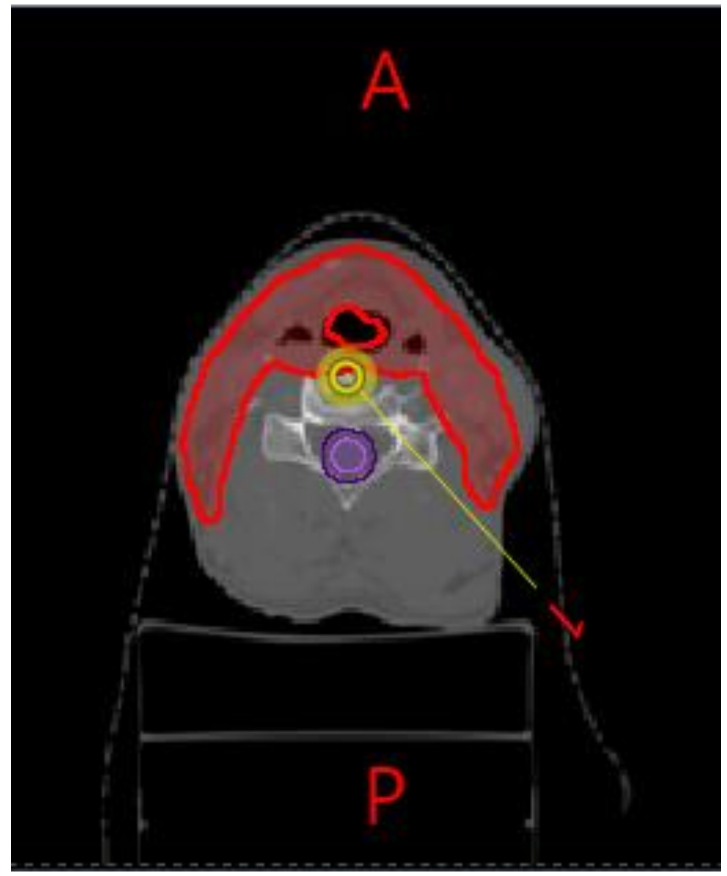
## Beam geometry

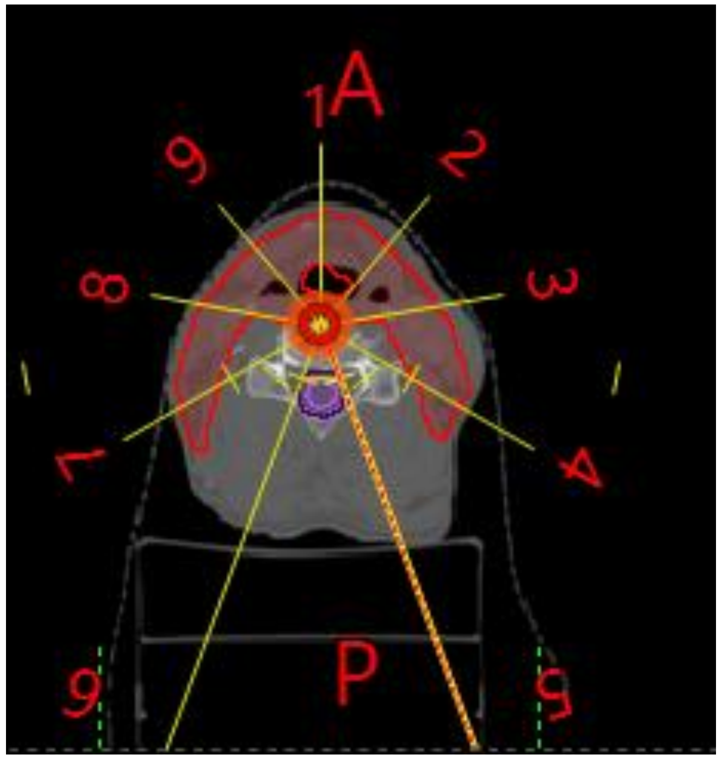
- Opposite beams should be avoided
- Odd number of beams (?)
- Non-coplanar beams provide less benefit compared to 3D-CRT (?)
- For the number of beams above 7, optimization of the head angles does not significantly improve the results compared to equally spaced beams

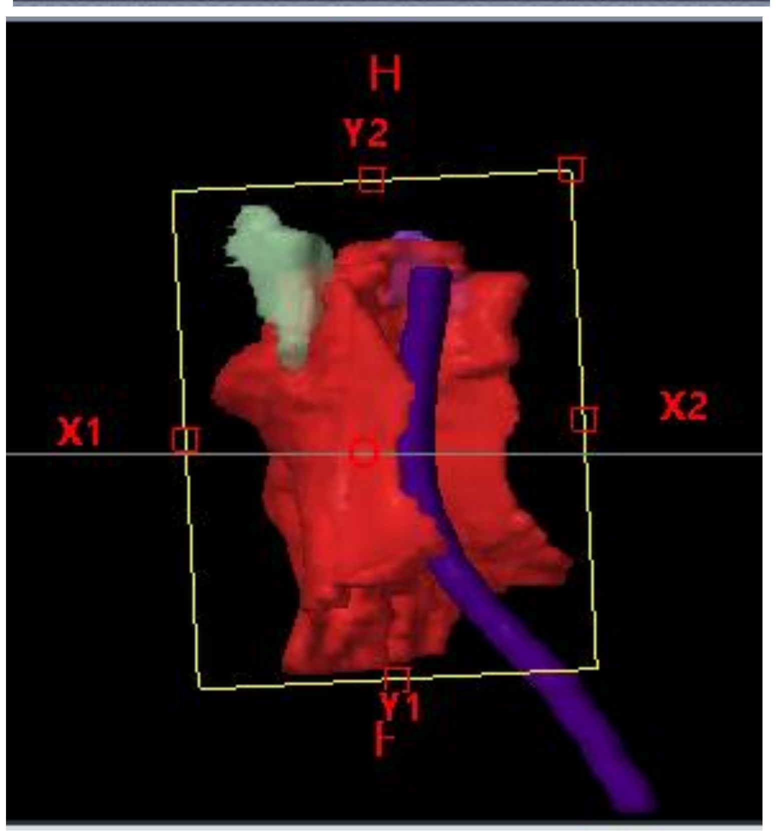
# Beam angle

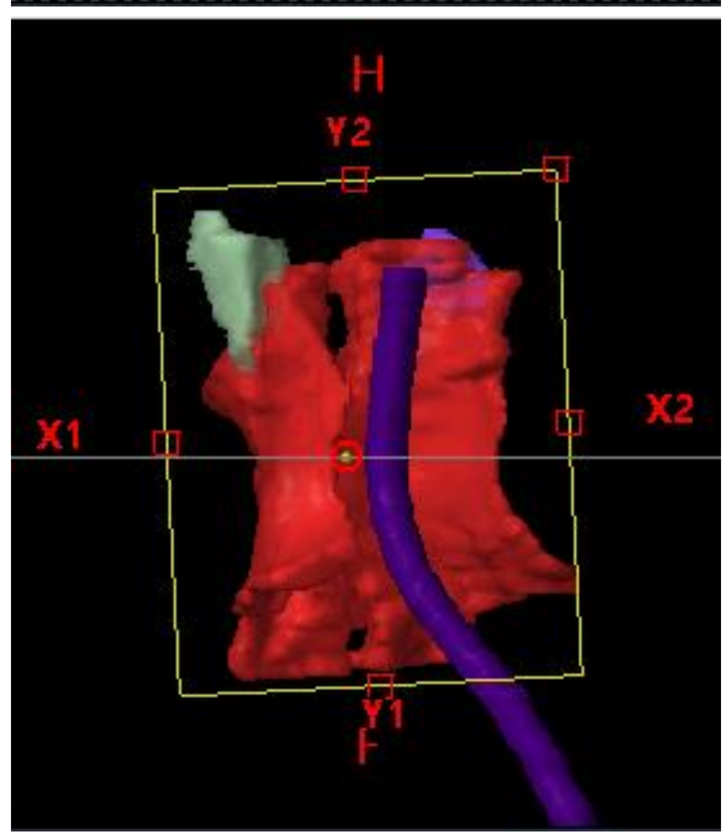
- 120°
- 138°
- 9 beams

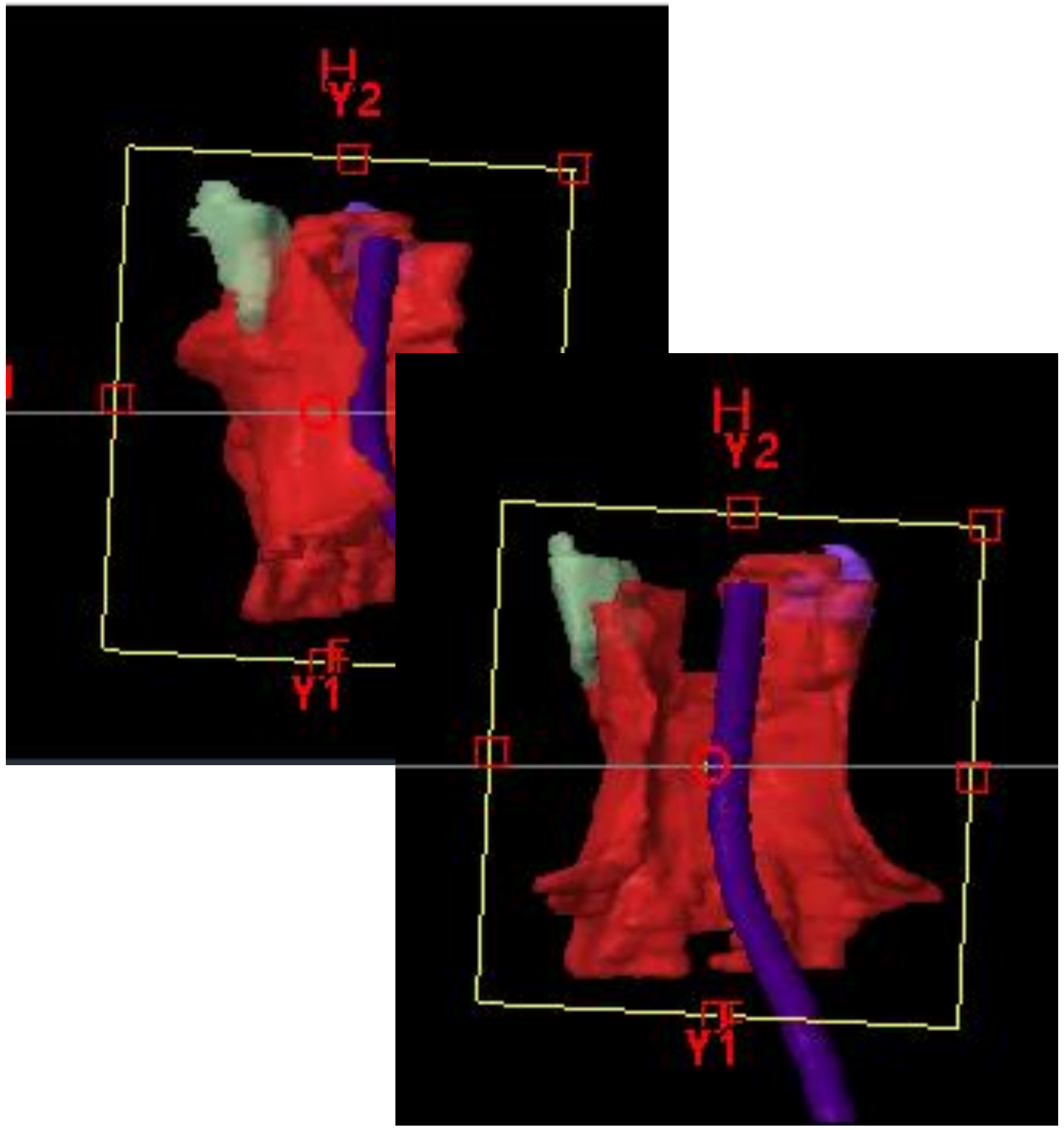


# Beam geometry

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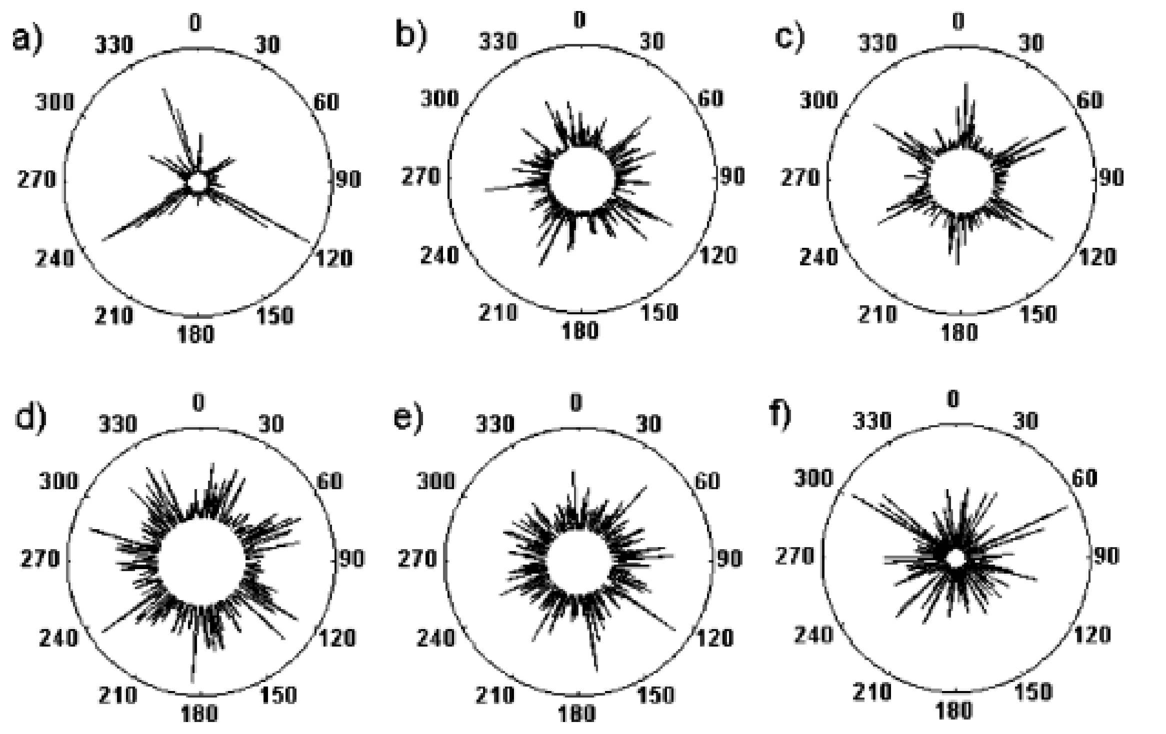
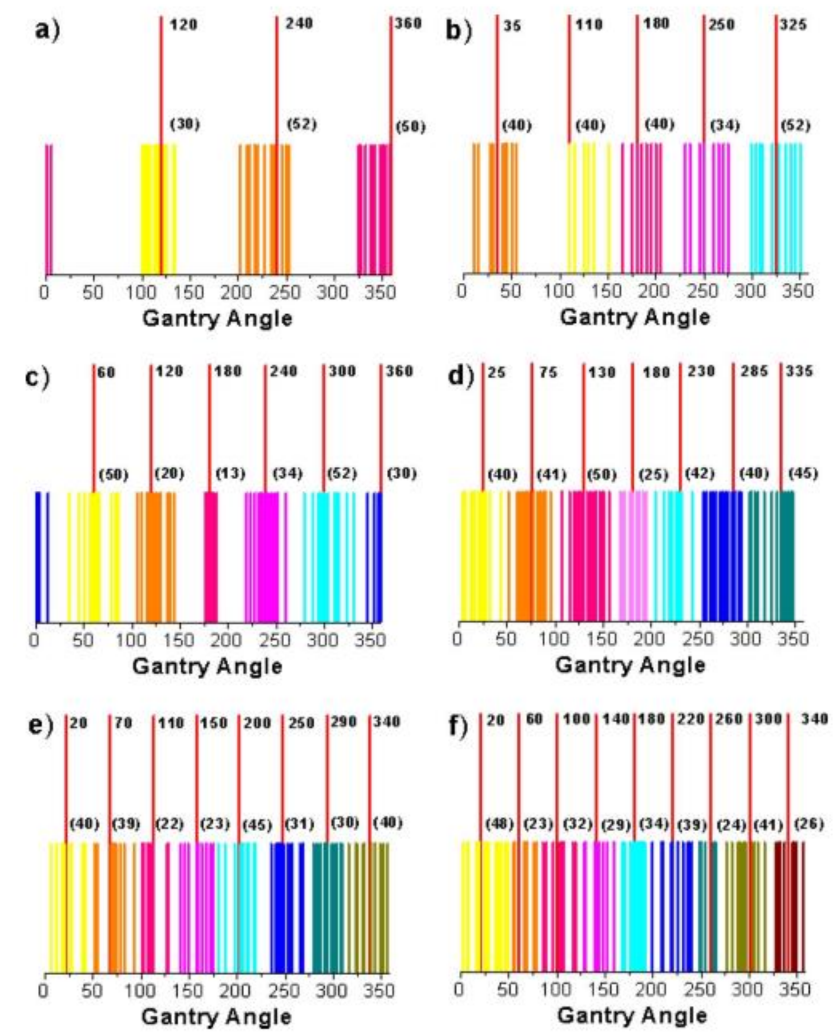
- Odd number of beams (?)
- Non-coplanar beams provide less benefit compared to 3D-CRT (?)
- For the number of beams above 7, optimization of the head angles does not significantly improve the results compared to equally spaced beams
  - *the beam angles as additional parameters to optimize - long execution time*
  - *select the beam angles first (a simplified objective function based on some prior knowledge)*
  - *class solution*

# Class solution - prostate

15 prostate cases -> Pareto front -> most frequent beam configuration -> the optimal one

One patient



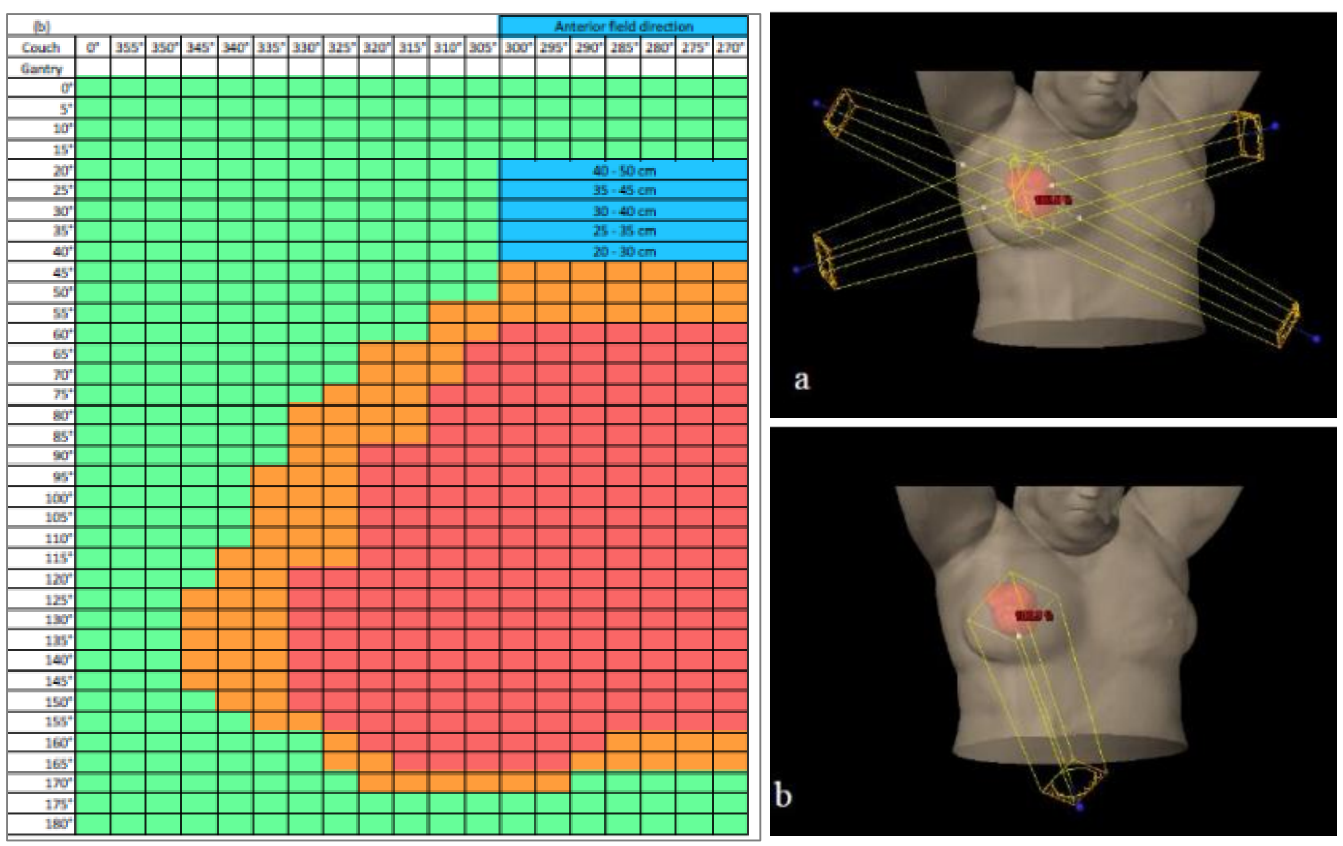




Results: **3 beams** (0°, 120°, 240°), **5 beams** (35°, 110°, 180°, 250°, 325°), **6 beams** (0°, 60°, 120°, 180°, 240°, 300°), **7 beams** (25°, 75°, 130°, 180°, 230°, 285°, 335°), **8 beams** (20°, 70°, 110°, 150°, 200°, 250°, 290°, 340°), **9 beams** (20°, 60°, 100°, 140°, 180°, 220°, 260°, 300°, 340°)

E. Schreibmann and L. Xing: Medical Physics, 31(10), 2004

# Class solution - APBI

- IMRT - better sparing of contralateral breast and lung than VMAT
- Non-coplanar beams - clearance - patient collisions
- 40 patients (17 right-sided, 23 left-sided)
- 6 MV five-field non-coplanar beam




Sarah Quirk, Practical Radiation Oncology, 8(3), 2018

# Beam geometry

- Opposite beams should be avoided
- Odd number of beams (?)
- Non-coplanar beams provide less benefit compared to 3D-CRT (?)
- For the number of beams above 7, optimization of the head angles does not significantly improve the results compared to equally spaced beams
  - *the beam angles as additional parameters to optimize - long execution time*
  - *select the beam angles first (a simplified objective function based on some prior knowledge)*
  - *class solution*
- Collimator angle ≠ 0° (e.g. ± 3 °)

# Beam Energy

Radiotherapy and Oncology 82 (2007) 55–62
www.thegreenjournal.com

*Lung cancer IMRT*

An analysis of 6-MV versus 18-MV photon energy plans for intensity-modulated radiation therapy (IMRT) of lung cancer

Elisabeth Weiss[a,b,*], Jeffrey V. Siebers[a], Paul J. Keall[a]

[a]*Department of Radiation Oncology, Virginia Commonwealth University, Richmond, VA, USA,* [b]*Department of Radiotherapy, University of Göttingen, Göttingen, Germany*

Radiotherapy and Oncology 82 (2007) 63–69
www.thegreenjournal.com

*Lung cancer IMRT*

Comparison of 6 MV and 18 MV photons for IMRT treatment of lung cancer

Indira Madani[a,*], Barbara Vanderstraeten[a,b], Samuel Bral[a], Marc Coghe[a], Werner De Gersem[a], Carlos De Wagter[a], Hubert Thierens[b], Wilfried De Neve[a]

[a]*Department of Radiotherapy, Ghent University Hospital, Belgium,* [b]*Department of Medical Physics, Ghent University, Belgium*

## Beam Energy

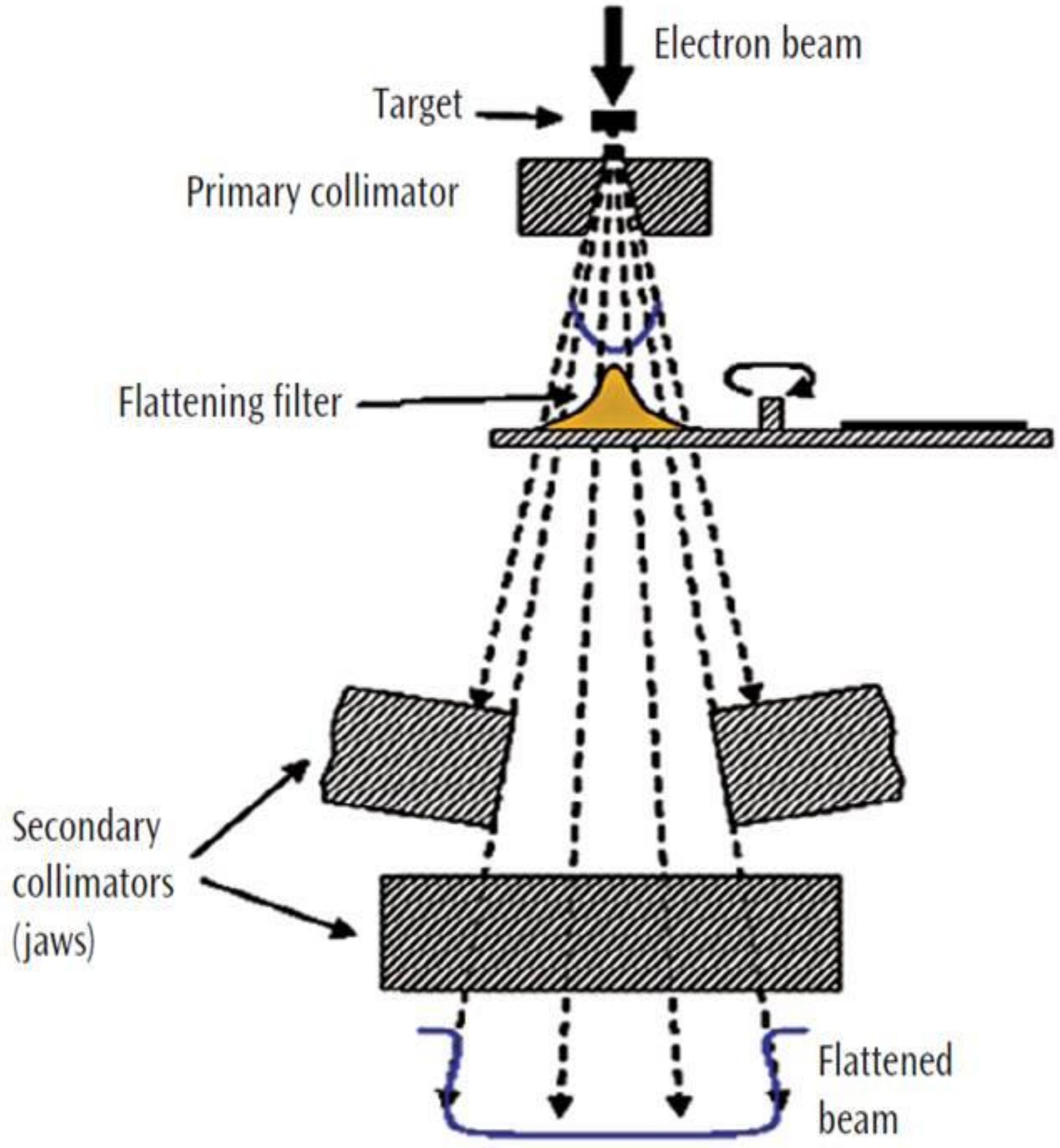
1. Generally lower energies (below 10 MV) are recommended – not all TPS algorithms model properly electron transport

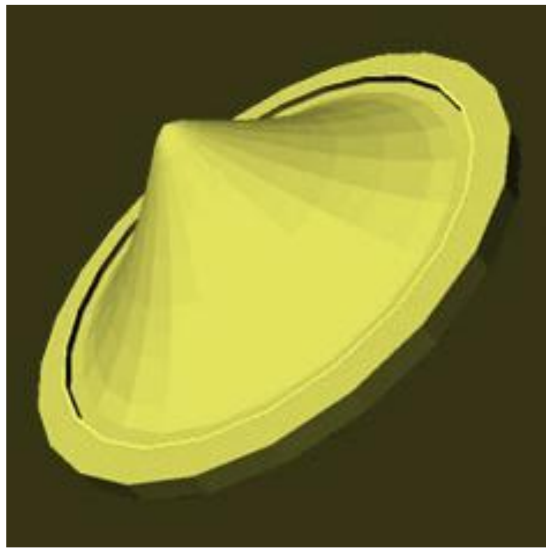
2. Difference between energies of negligible importance compared to a definition of CTV, taking into account tumour mobility, and determining the total dose

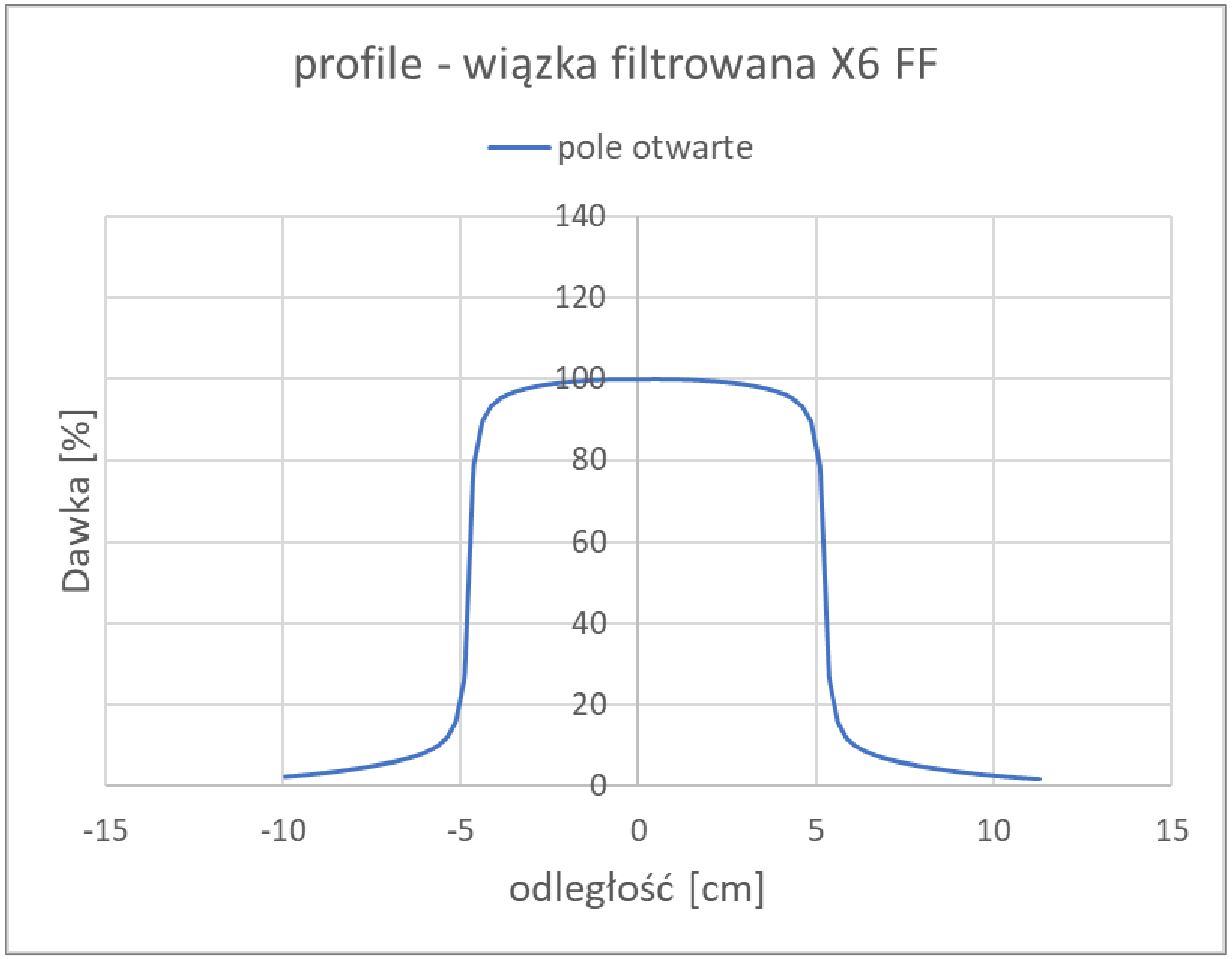
3. More important than energy is using the correct algorithm – especially in lung

4. For tumours surrounded by lung tissue - low energy is recommended (rebuild-up effect)

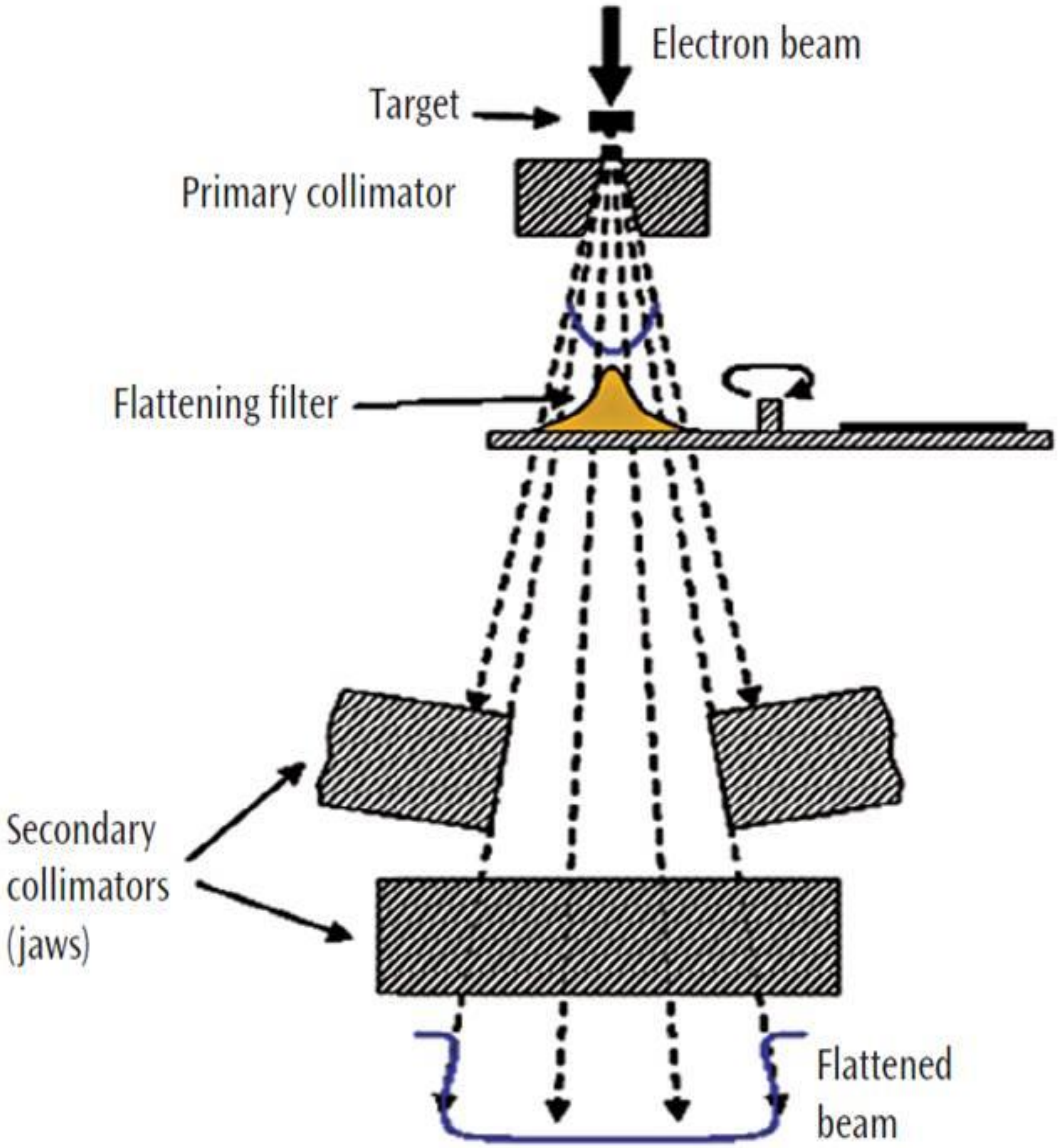
# Flattening Filter Accelerators

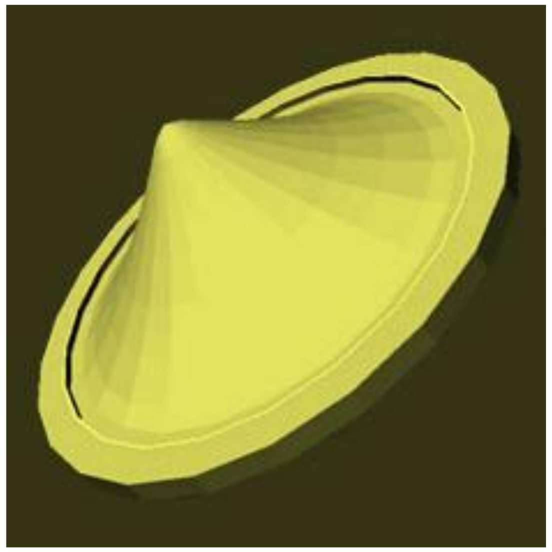


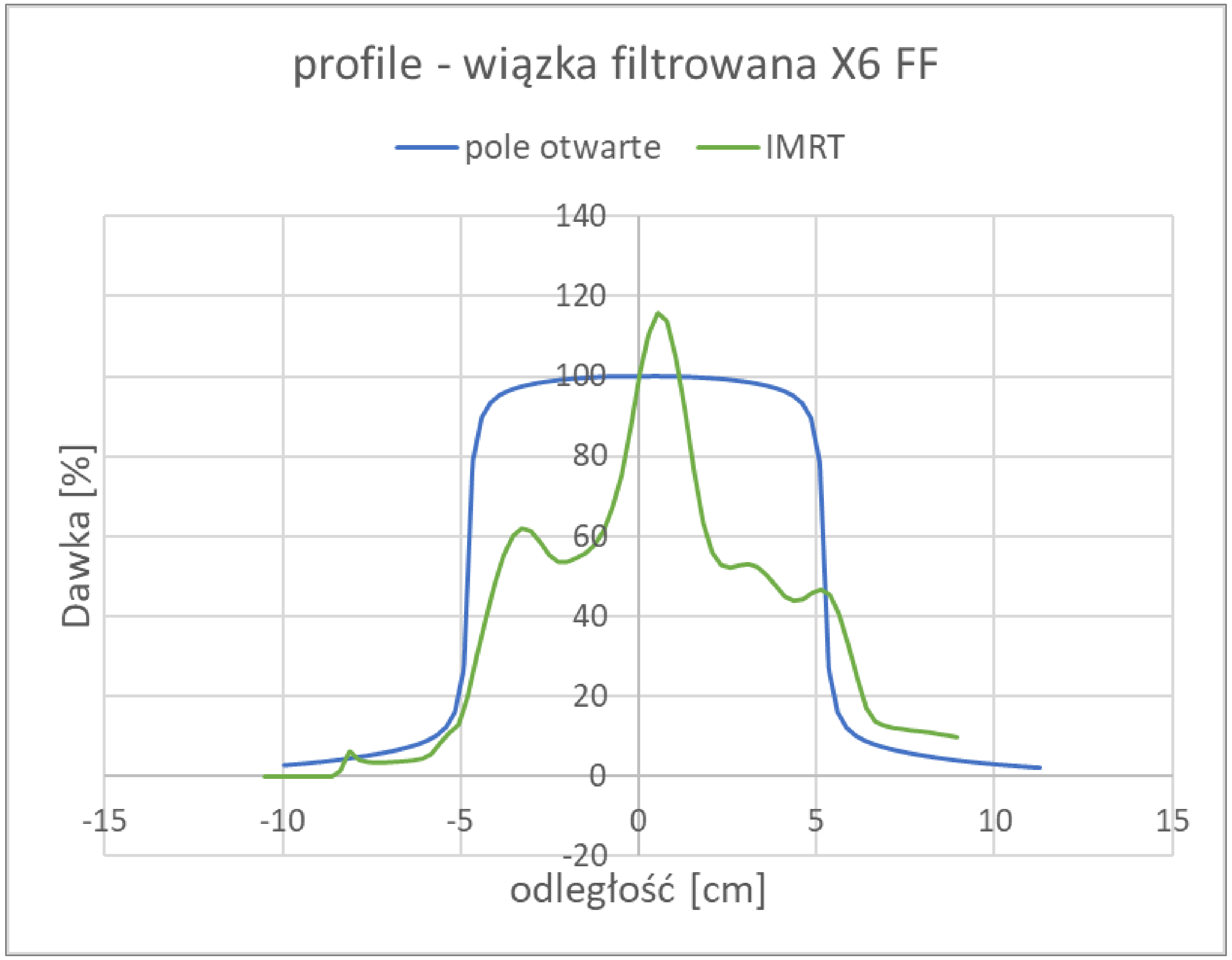





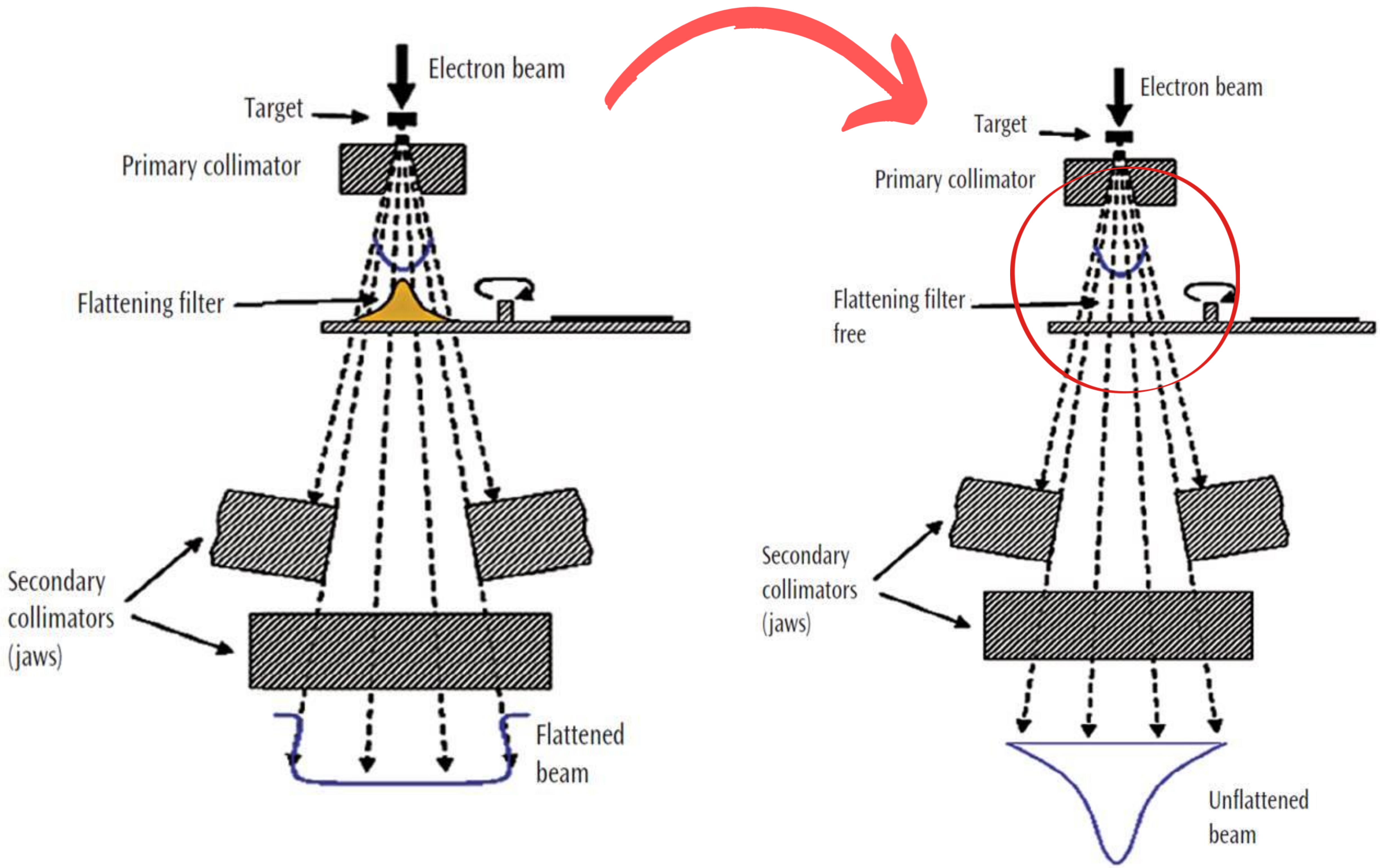
# Flattening Filter Accelerators





# Flattening Filter Free Accelerators



# Profiles FF vs FFF

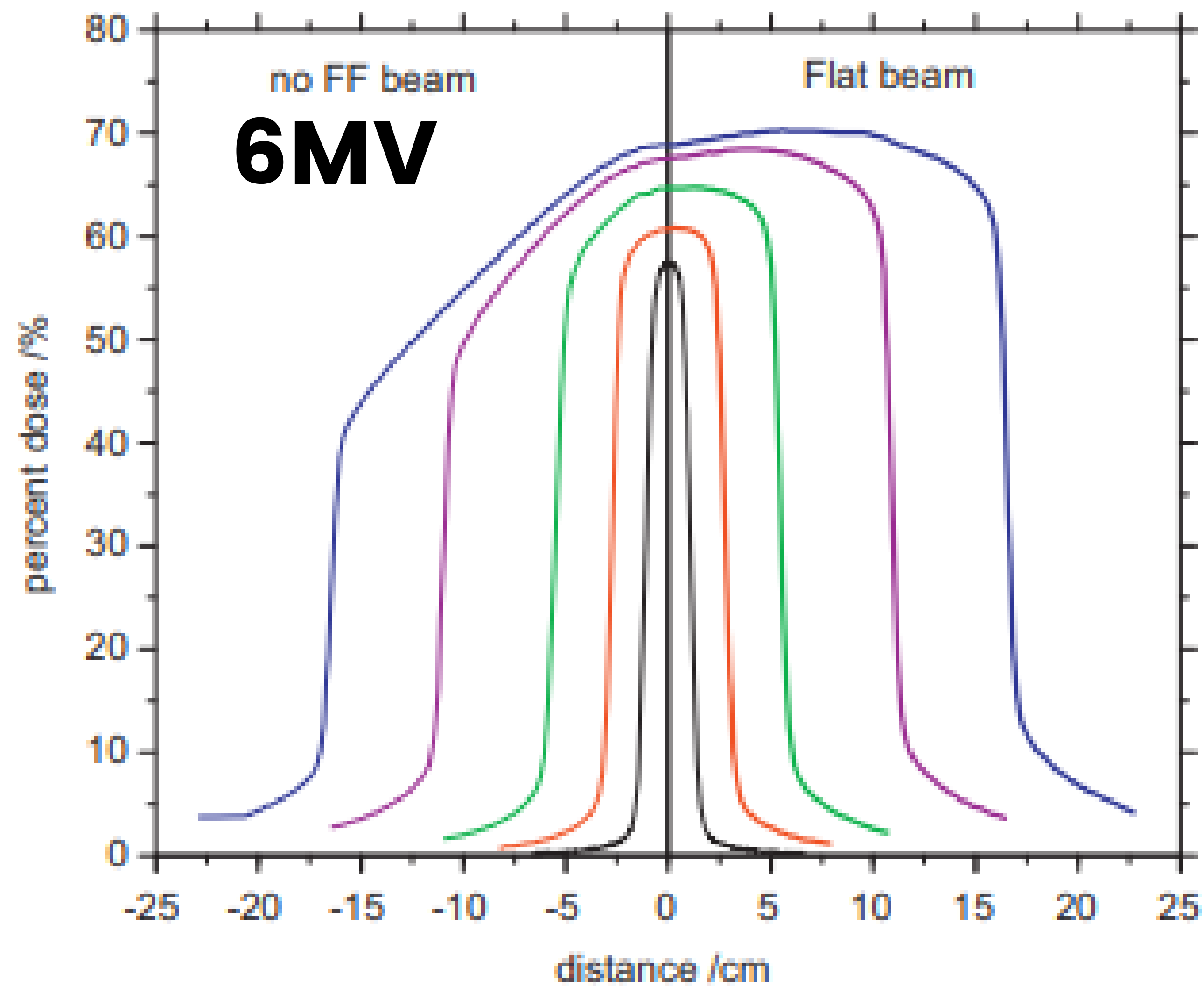



**Fig. 7.** Half profiles for the 2 × 2, 5 × 5, 10 × 10, 20 × 20 and 30 × 30 cm² fields of the 6 MV photon beams. The unflattened profiles are shown with thin lines (left) and the flat ones with thick lines (right).

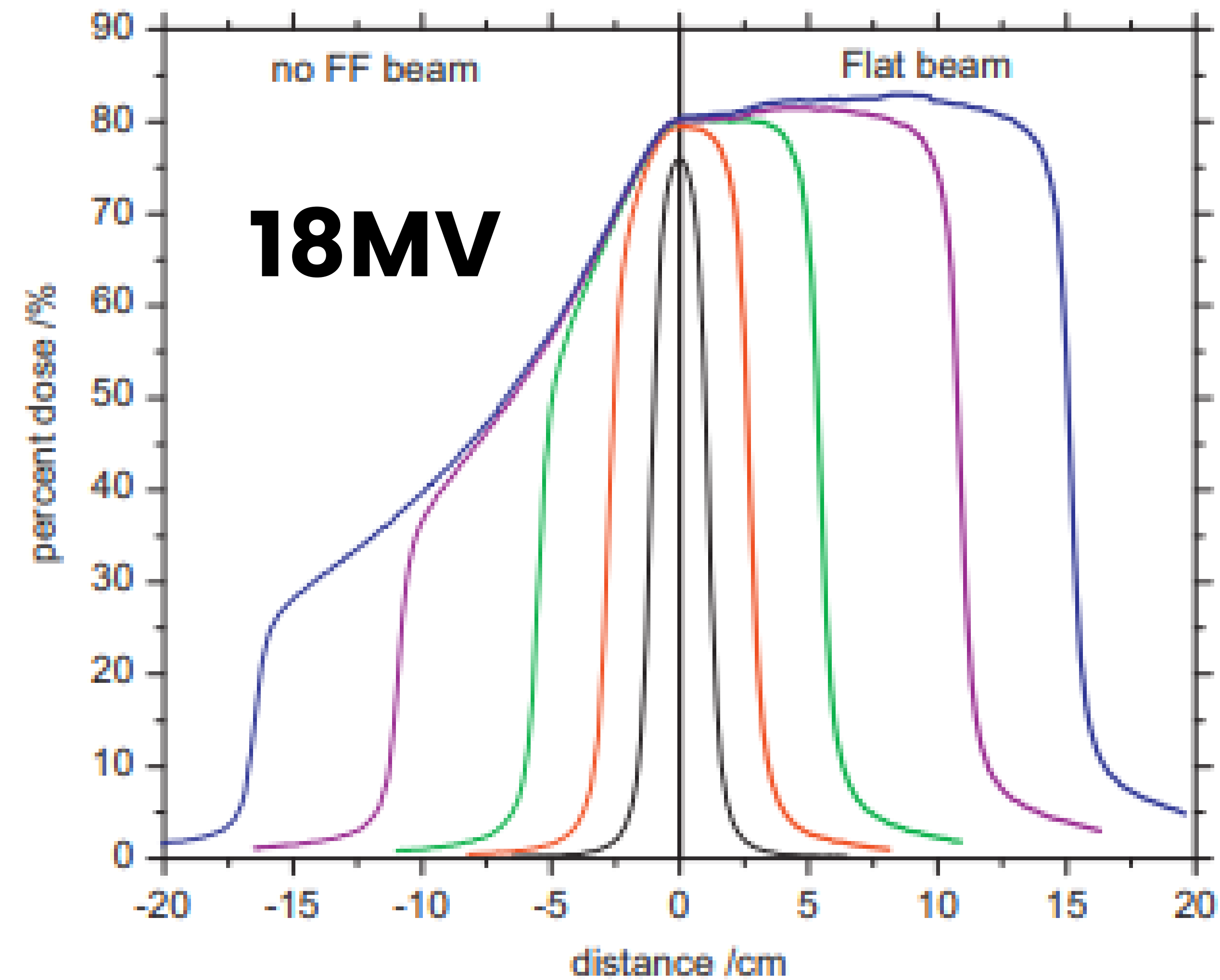
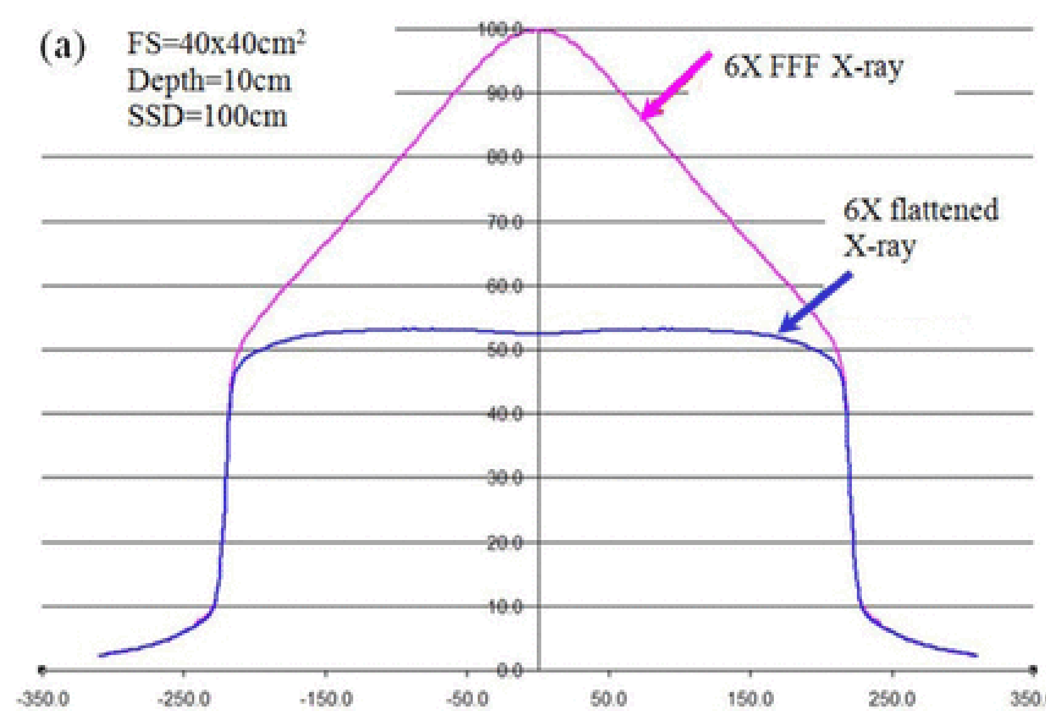



**Fig. 8.** Half profiles for the 2 × 2, 5 × 5, 10 × 10, 20 × 20 and 30 × 30 cm² fields of the 18 MV photon beams. The unflattened profiles are shown with thin lines (left) and the flat ones with thick lines (right).




Higher dose rate

TrueBeam

6 MV - 1400 MU/min

10 MV - 2400 MU/min

S. Stathakis et al. Applied Radiation and Isotopes 67 (2009)

# FF vs FFF – energy

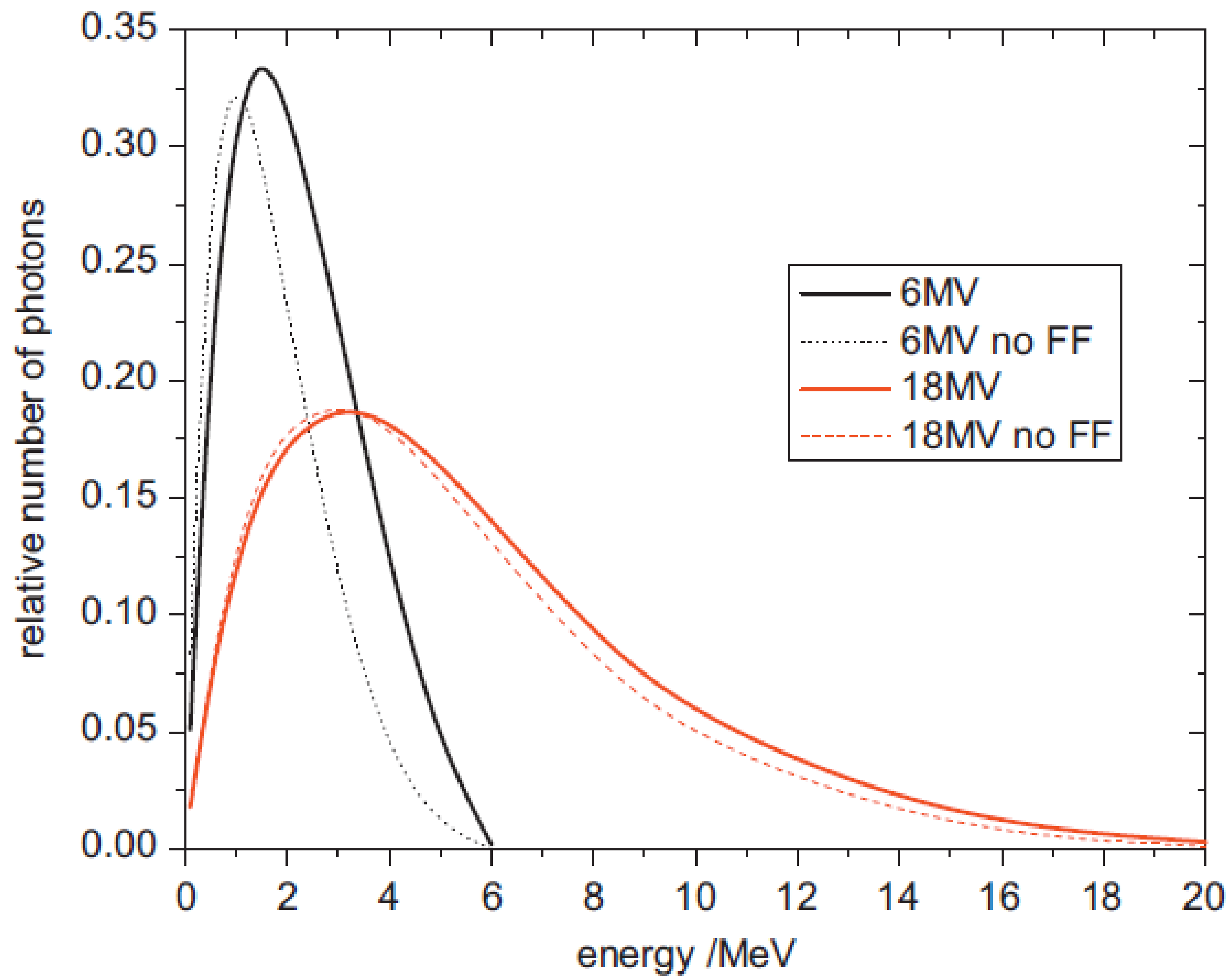



**Fig. 1.** Relative energy fluence for the 6 and 18 MV photon beams. Comparison between flatened and flattening filter-free beams.

*Stathakis, Applied Radiation and Isotopes 67 (2009)*

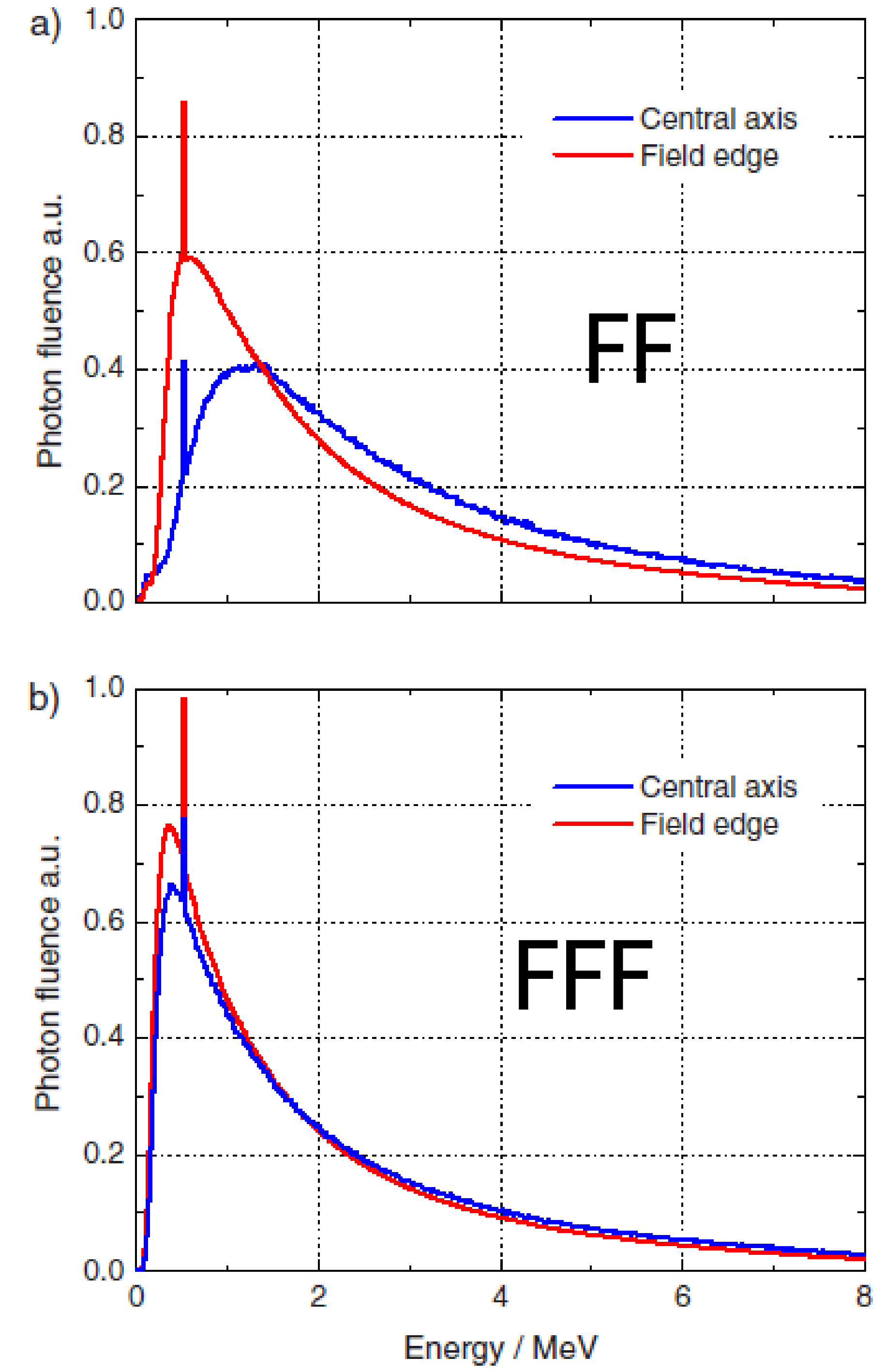
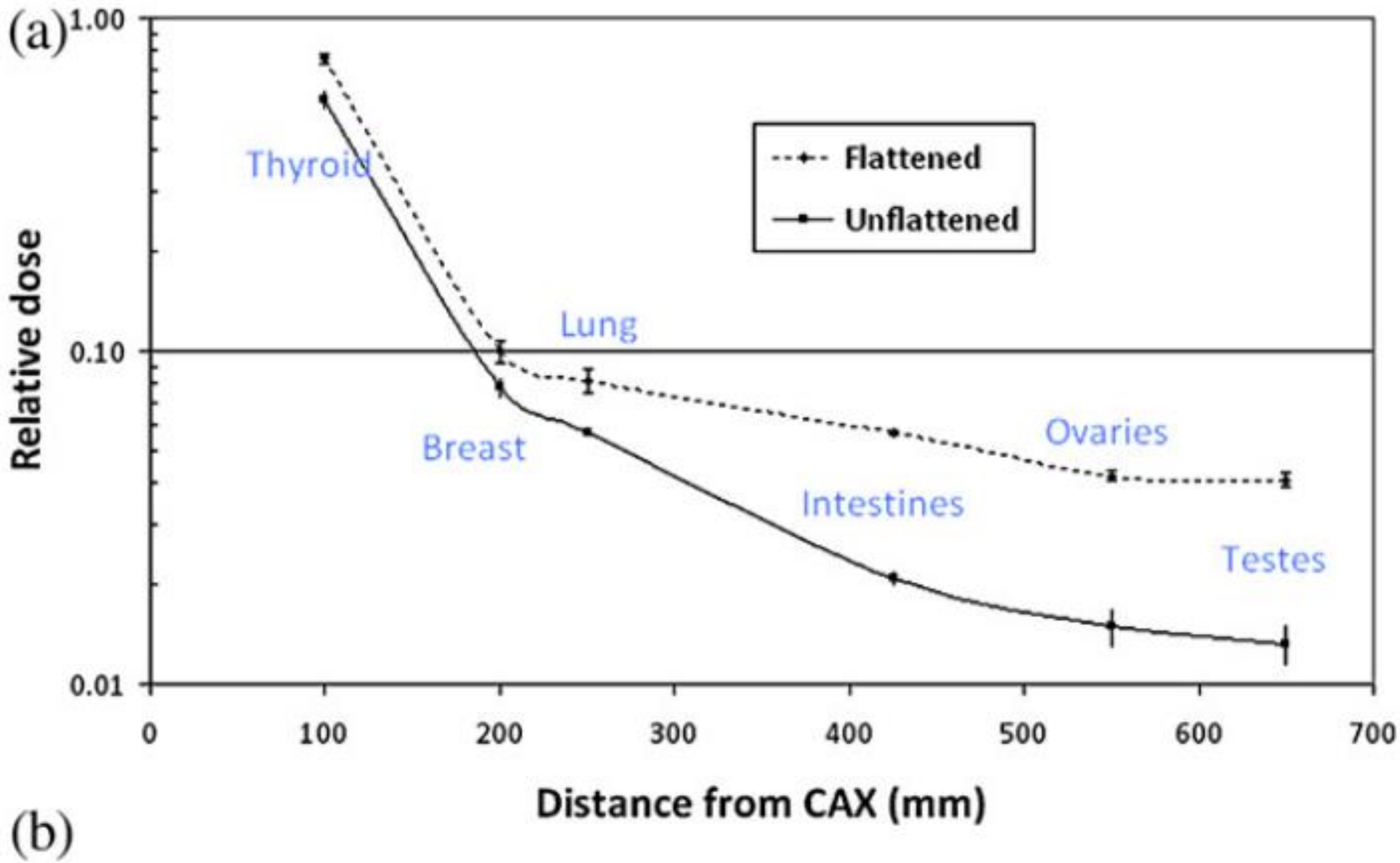



FIG. 1. Comparison of Monte Carlo simulated x-ray spectra on the central beam axis and the field edge of (a) flattened and (b) unflattened 10 MV beams provided by an Elekta linac. More details concerning the MC simulations w.r.t. this figure can be found in Dalaryd *et al.* (Ref. 32).

*Dietmar, Med. Phys. 38(3), March 2011*

- FF – beam hardening
- Fluence increases in a field's central part
- 6MV FFF -> 4-5MV FF, 18MV FFF -> 15MV FF

# Dose outside the field




Pediatric IMRT with unflattened photon beams d J. CASHMORE et al. IJROBP 80(4), 2011

# FFF and 3D – CRT conventional fractionation

- Increased dose rate
- Lower peripheral dose (less scatter on the head)
- Field-in-field technique

- no differences in PTV coverage
- V5Gy and V10 Gy significant differences in favor of FFF
- more MUs and more fields

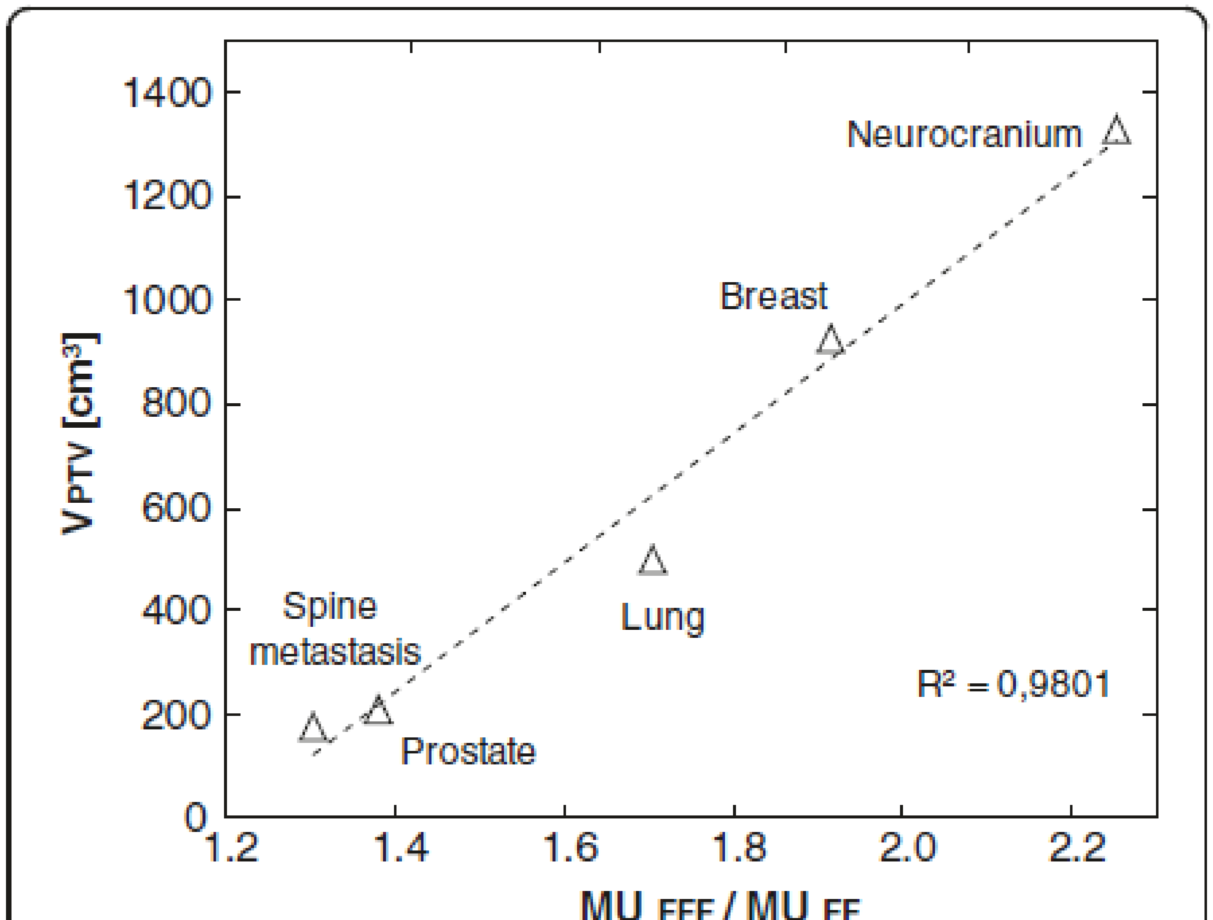



**Figure 4 Diagram correlating PTV volume with MU coefficient.** Correlation between the mean values for the location-specific coefficient $MU_{FFF}/MU_{FF}$ and the volume of PTV. The quality of the linear fit is given with $R^2$.

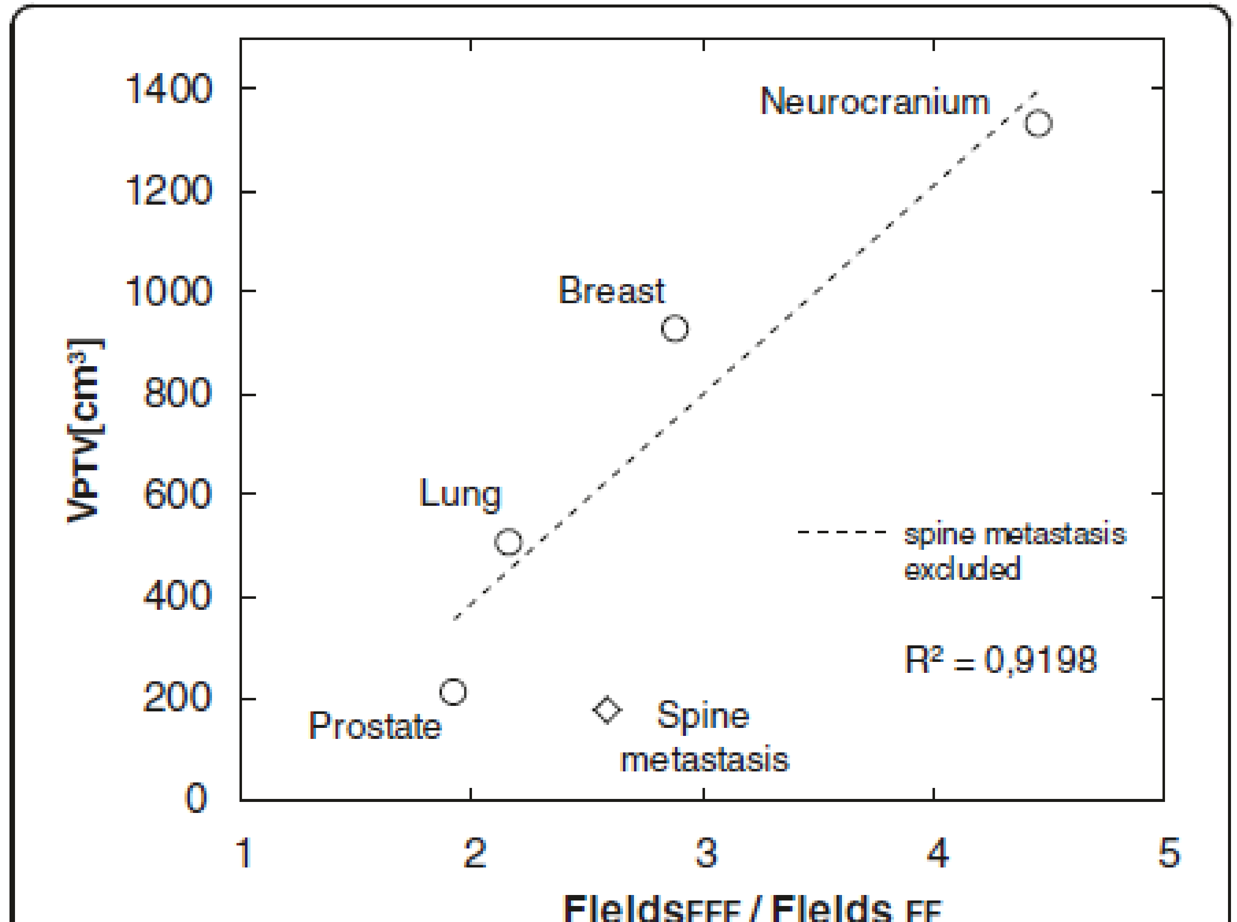
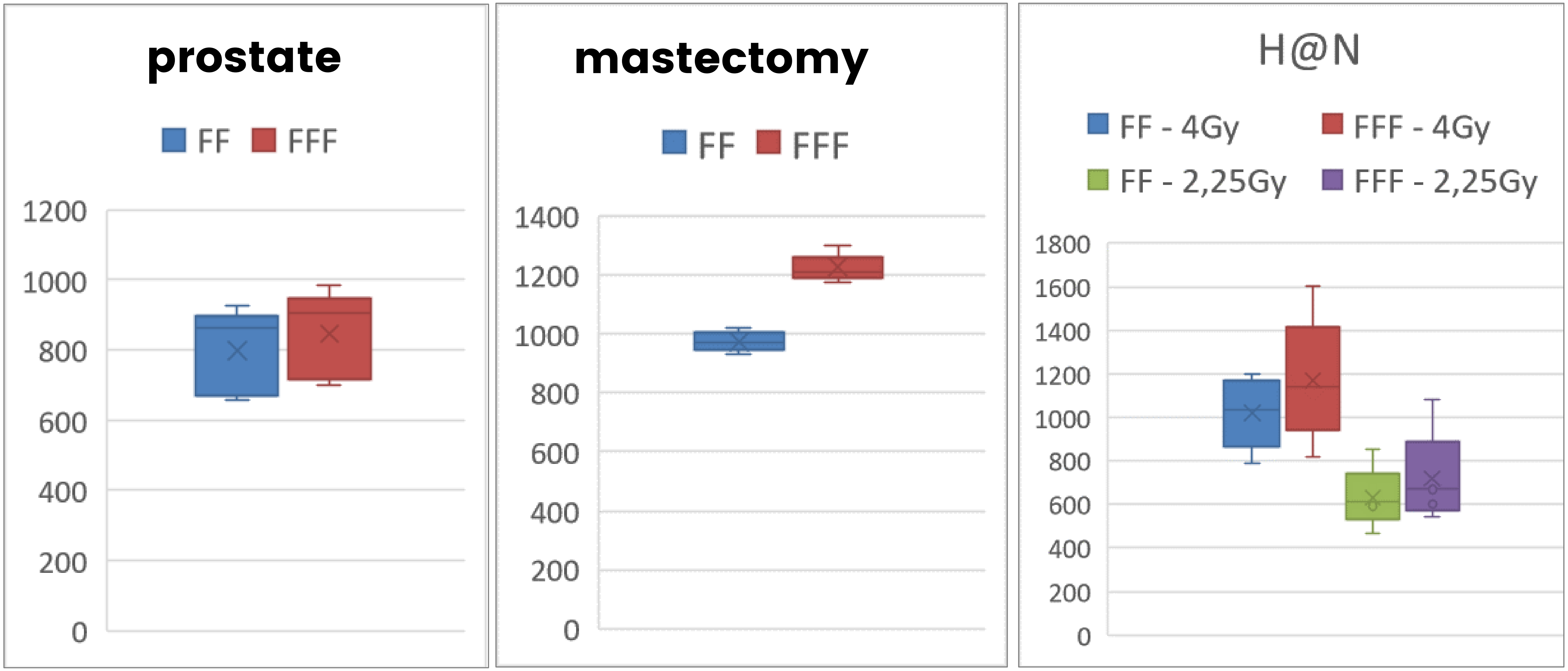



**Figure 3 Diagram correlating PTV volume with the field number coefficient.** Correlation between the mean values for the location-specific coefficient $Fields_{FFF}/Fields_{FF}$ and the volume of PTV. The quality of the linear fit is given with $R^2$. The data point spine metastasis was excluded due to inconsistent FF planning with virtual wedge fields.

*Kretschmer et al. Radiation Oncology 2013, 8:133 The imapct of flattening-filter-free beam technologu on 3D conformal RT*

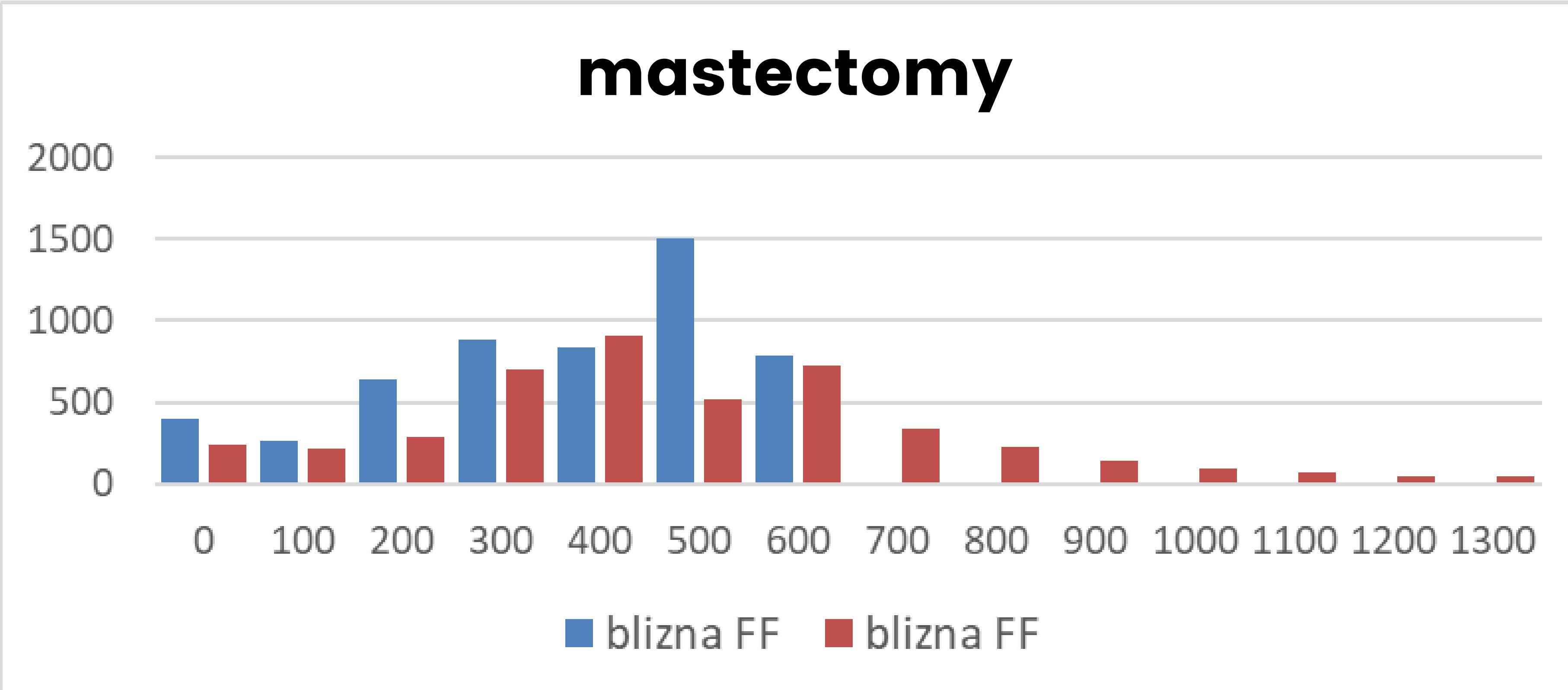
# MU numer

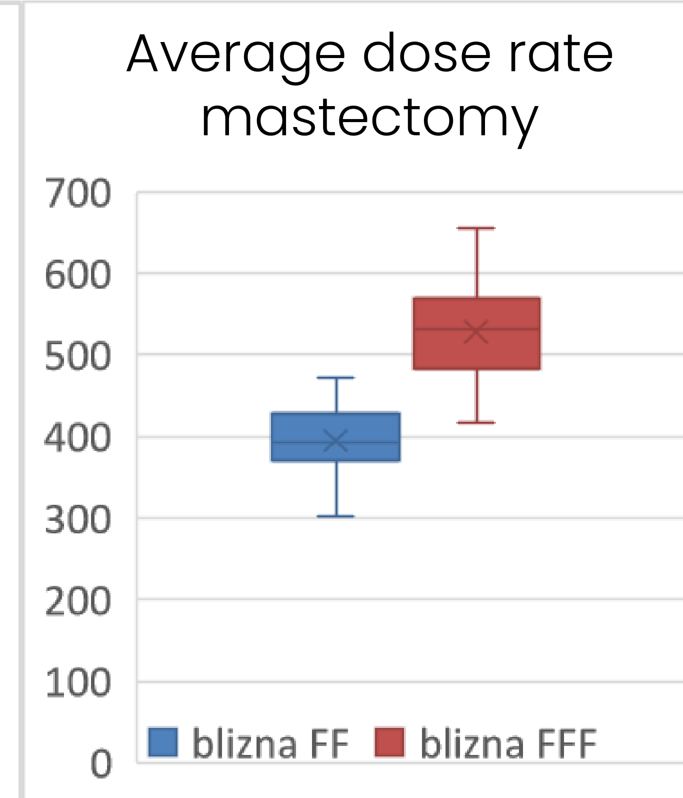


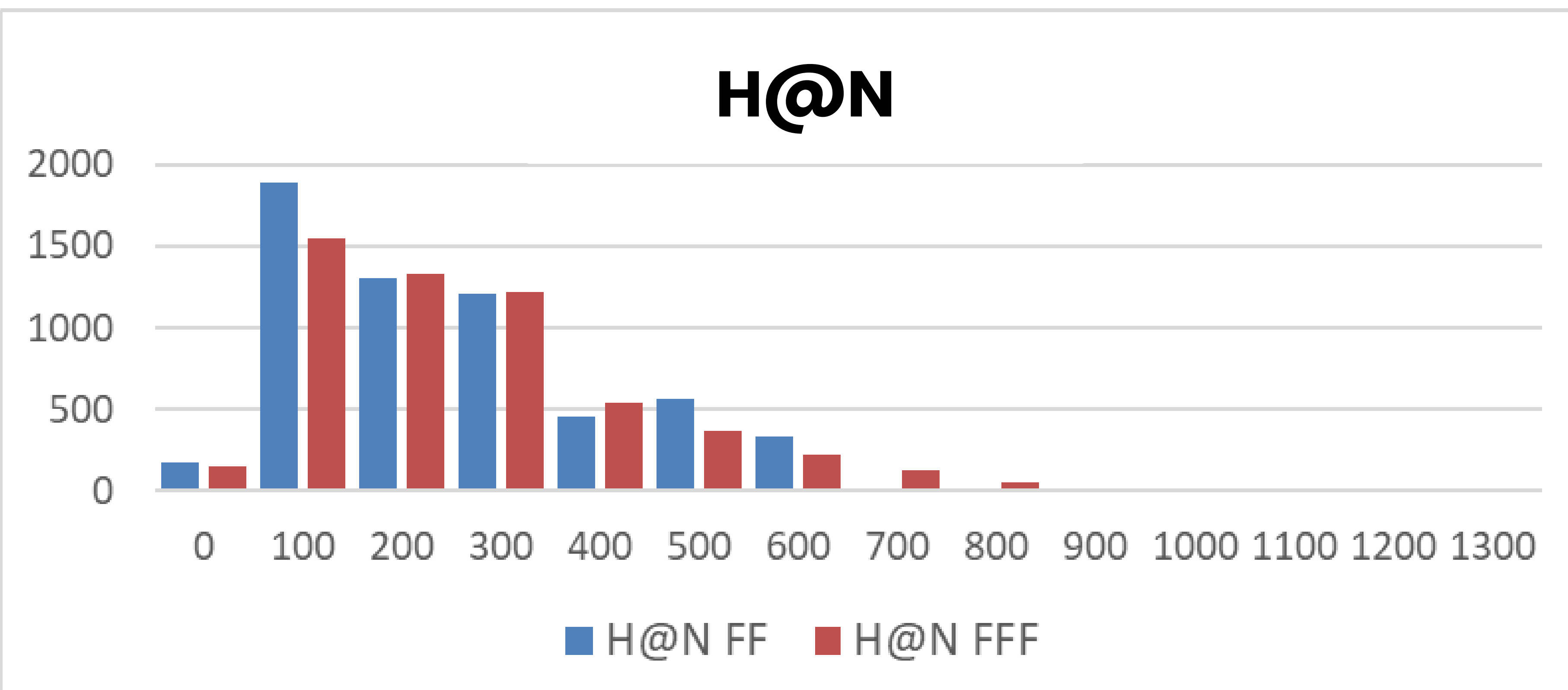

prostate – 5 patients ($d_{fr}$ = 2,6Gy)
mastectomy – 10 patients ($d_{fr}$ = 2,25Gy)
H@N – 2 x 5 patients ($d_{fr}$ = 4Gy, $d_{fr}$ = 2.25)

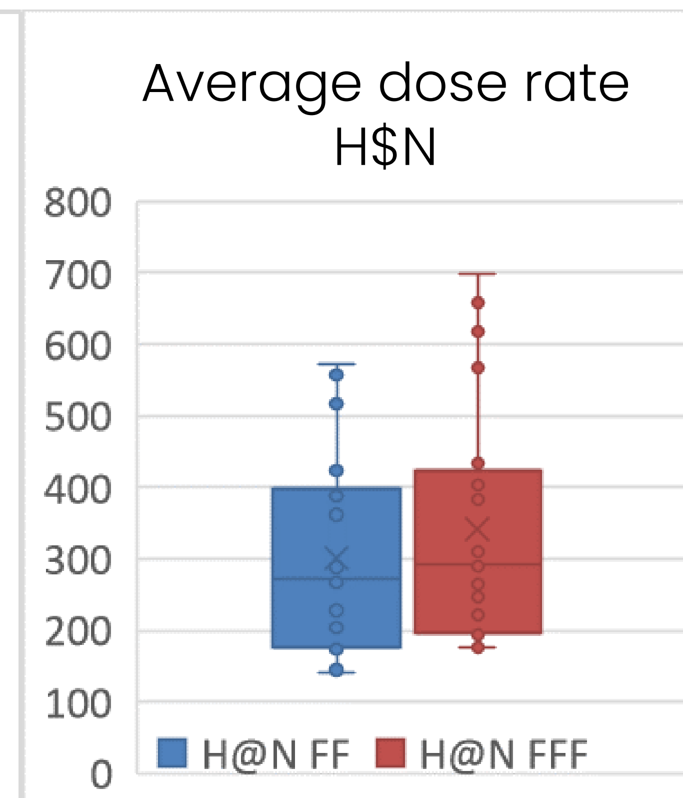
Higher MU number for FFF

# Dose rate

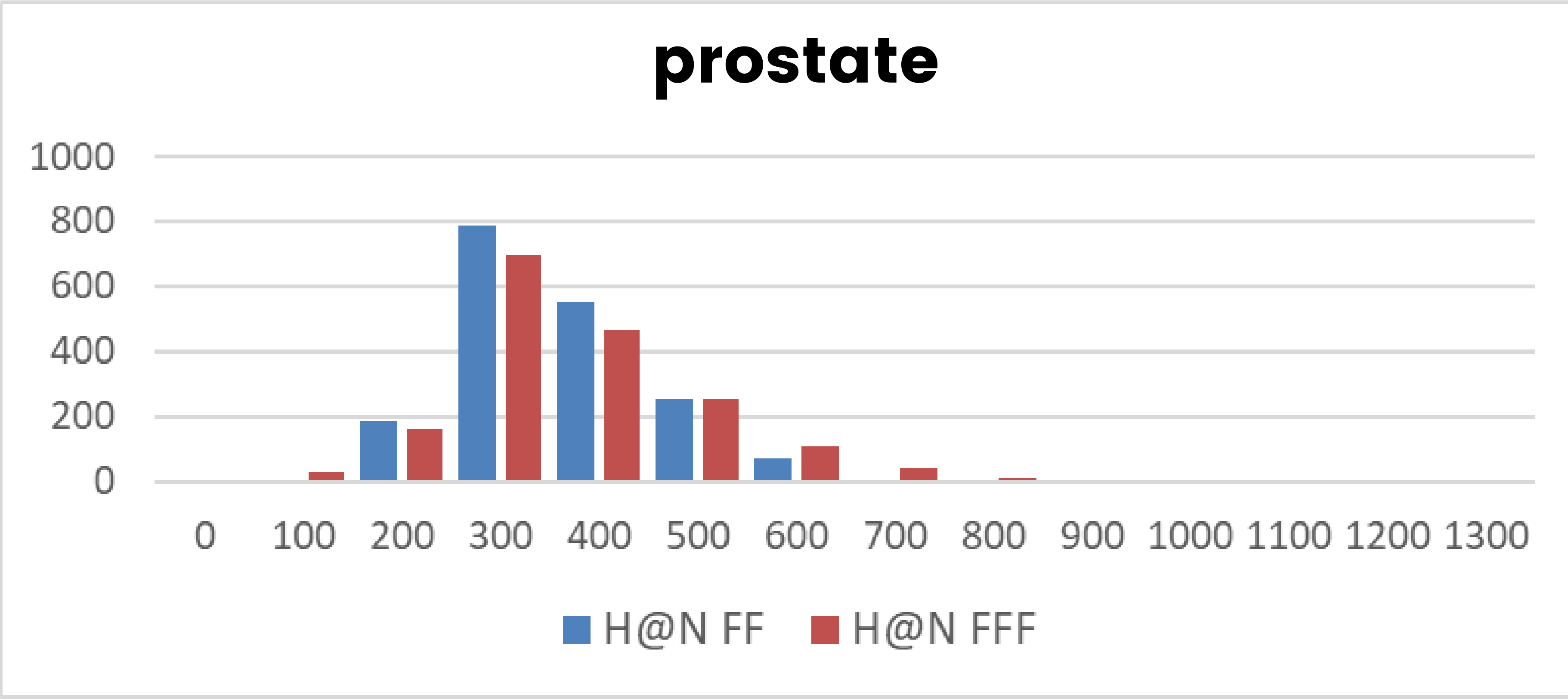

mastectomy
2000
1500
1000
500
0
0 100 200 300 400 500 600 700 800 900 1000 1100 1200 1300
blizna FF
blizna FF

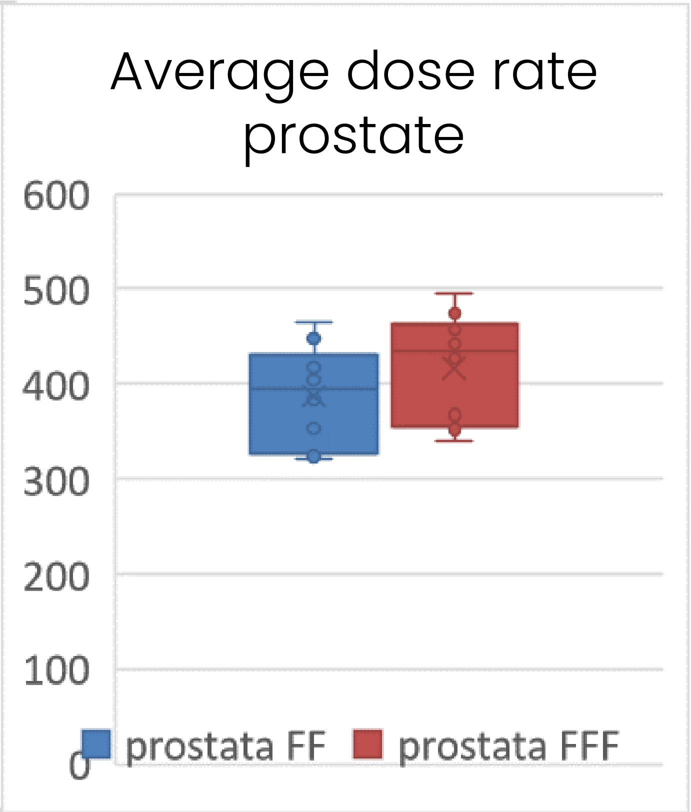

Average dose rate mastectomy
700
600
500
400
300
200
100
0
blizna FF
blizna FFF


H@N
2000
1500
1000
500
0
0 100 200 300 400 500 600 700 800 900 1000 1100 1200 1300
H@N FF
H@N FFF


Average dose rate H$N
800
700
600
500
400
300
200
100
0
H@N FF
H@N FFF

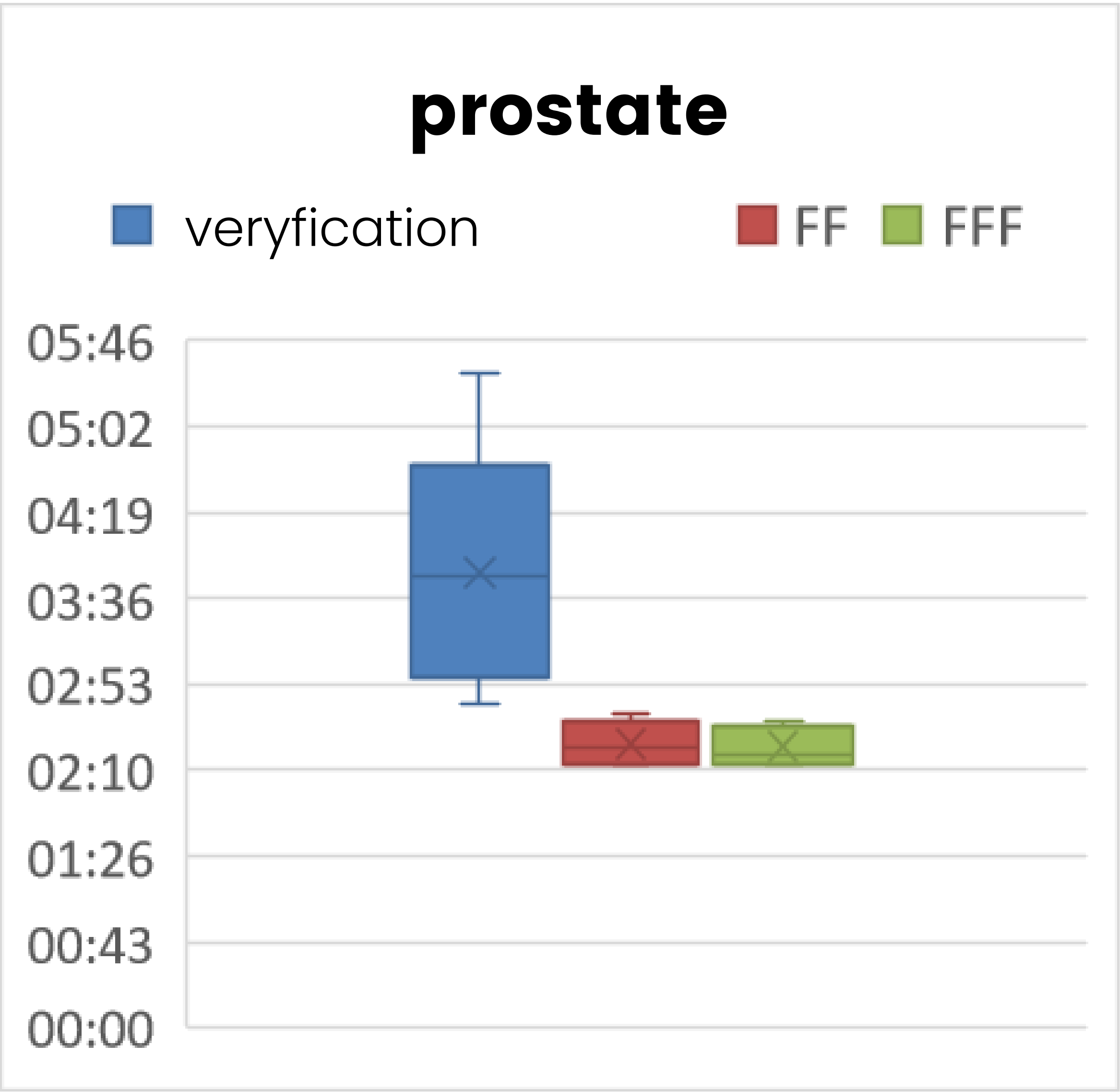
# Dose rate



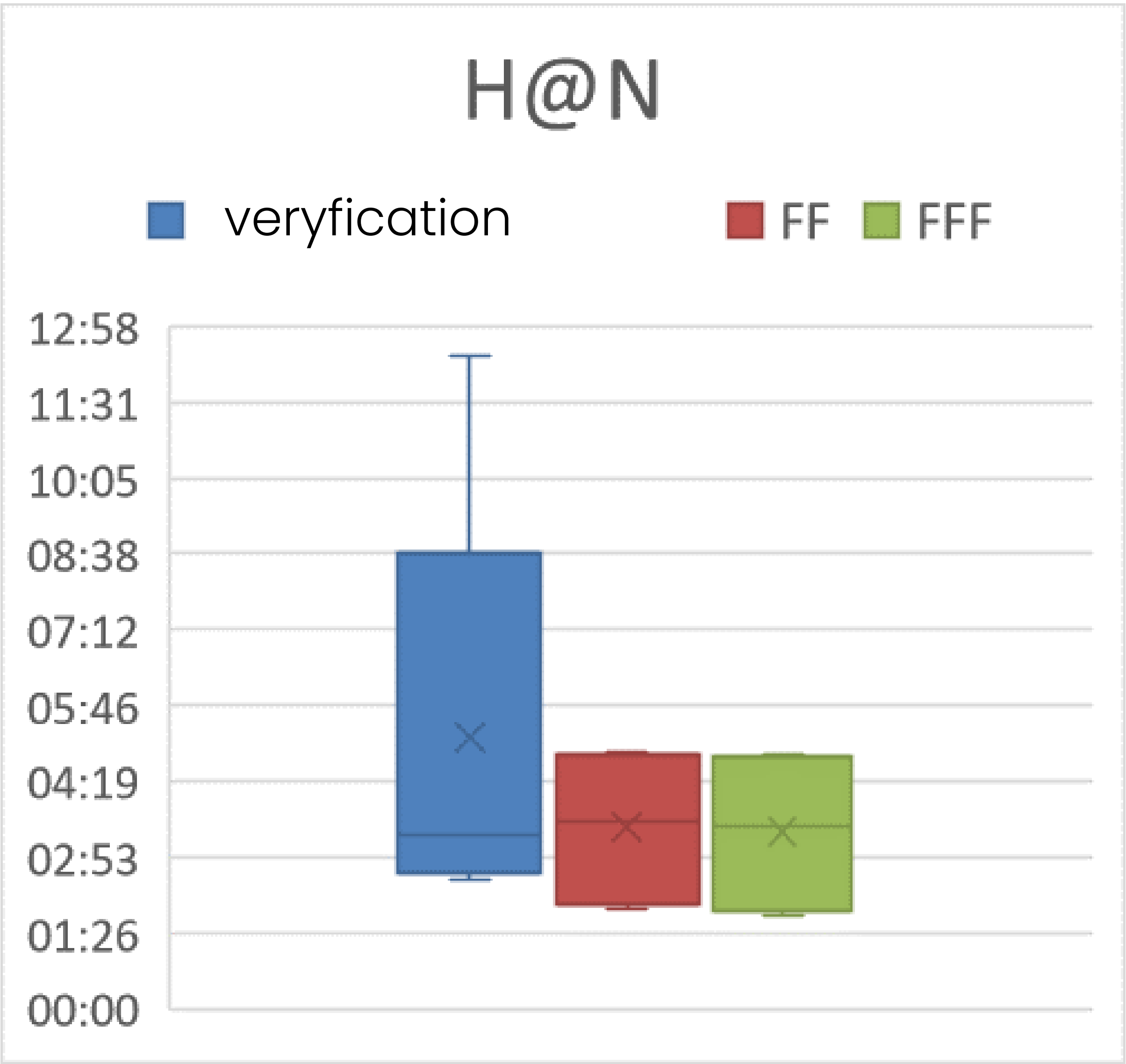




No significant differences in dose rate
Smaller number of arcs to consider

# Patient position weryfication

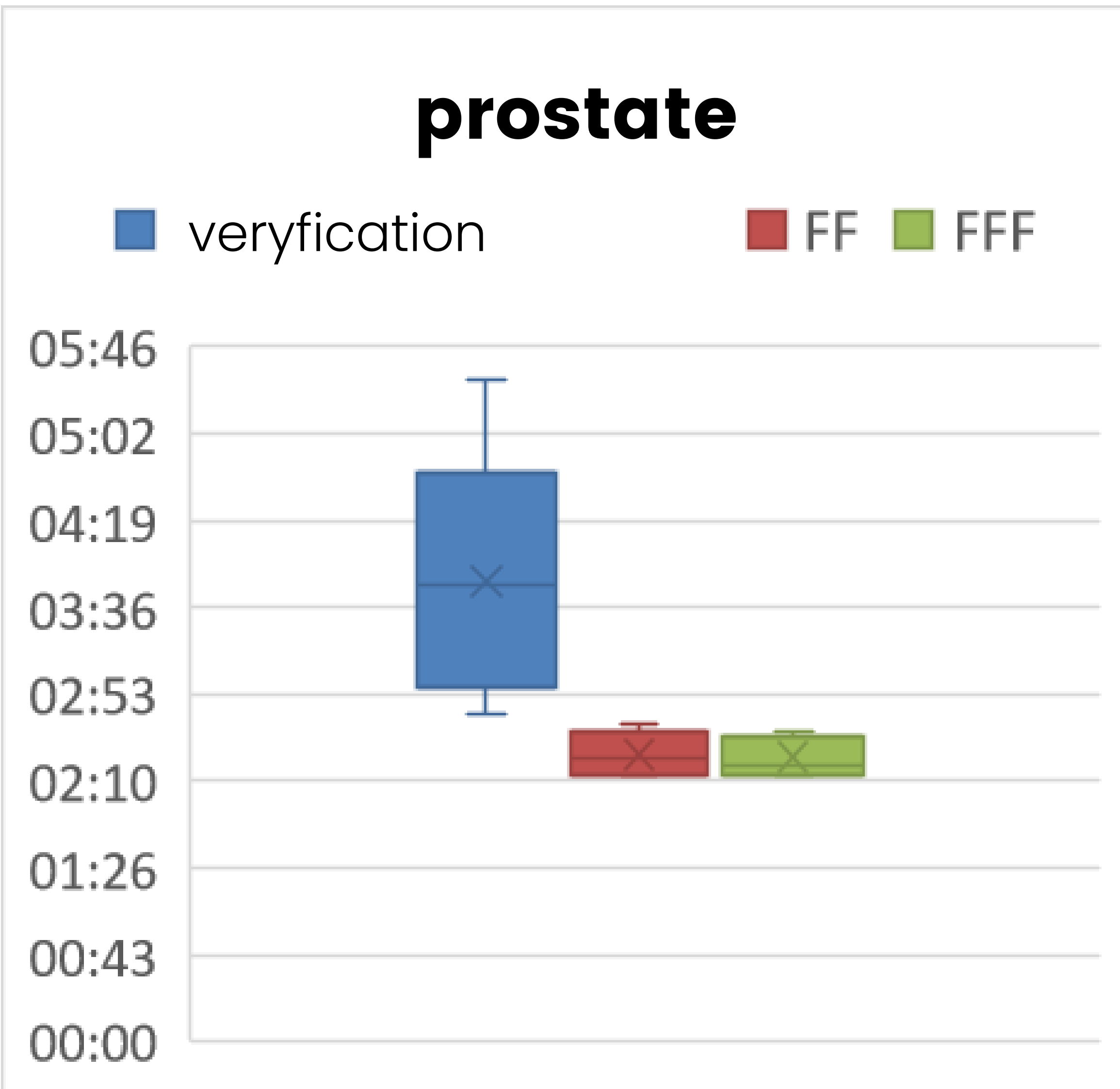



for 5 patients (5fr, 6fr)
2.6 Gy -> 70,20 Gy

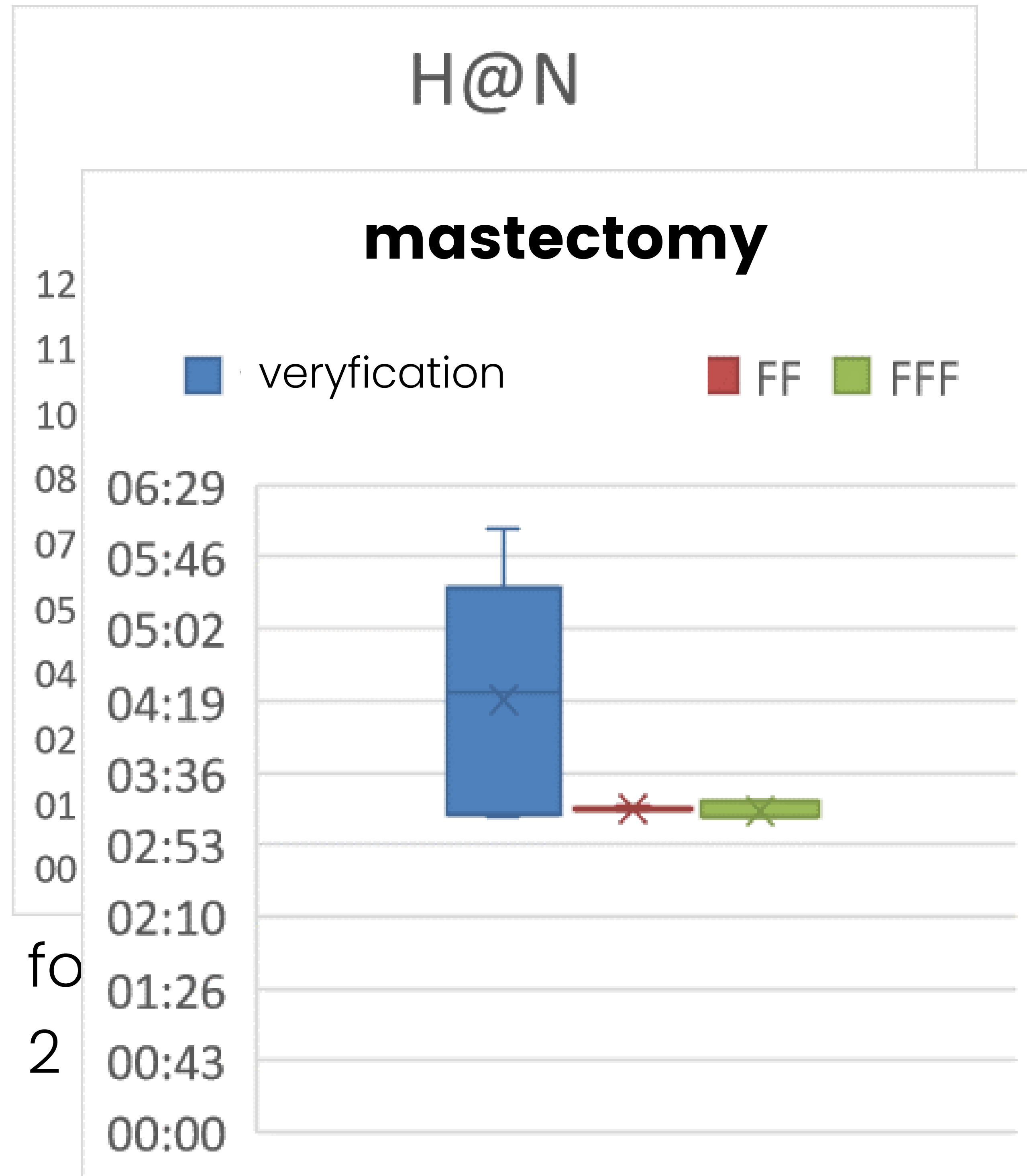



for 5 patients (3fr, 4fr)
2 etapy -> 5 x 4Gy

# Patient position weryfication




for 5 patients (5fr, 6fr)
2.6 Gy -> 70,20 Gy




fo
2

for 5 patients (5fr, 10fr)

## Plan preparation

1. Defining the beam geometry
   number of fields, beam angles, collimator angles, technique IMRT vs. VMAT

2. Defining score function, the cost function

3. Defining objectives and constraints for PTV and OAR's

4. Computer optimization – finding the optimal solution (dose distribution)

# Optimization - objective function

The objective function (score function, cost function)

Quantitative definition of clinically meaningful goals and constraints.

The formula in most systems is predefined.

**Physical Functions** – the balance between OAR's (healthy tissues) sparing and PTV coverage - steered with weighting factors/penalty factors/importance factors. The most widely employed method -> physical objectives reflect clinical practice and outcome.

**Biological functions** – EUD, TCP/NTCP.

# Objective function – example of physical function

The method of least squares

$$\min[F(\vec{\omega})] = \frac{I_{PTV}}{T_{PTV}} \sum_{i \in T_{PTV}} C_{PTV}^{-}(d_i - d_{PTV}^{-})^2 + \frac{I_{PRV}}{T_{PRV}} \sum_{i \in T_{PRV}} C_{PRV}^{-}(d_i - d_{PRV}^{+})^2$$

$$C_{PTV}^{-} = \begin{cases} 1 & if\ d_i < d_{PTV}^{-} \\ 0 & otherwise \end{cases}$$

$$C_{PRV}^{+} = \begin{cases} 1 & if\ d_i > d_{PRV}^{+} \\ 0 & otherwise \end{cases}$$

- $I_{PTV}$, $I_{PRV}$ – importance (weight)
- $T_{PTV}$, $T_{PRV}$ – number of points within the structure
- $d_{PTV}^{-}, d_{PTV}^{+}$ – minimum and maximum **dose constraints** for PTV and OAR, respectively.

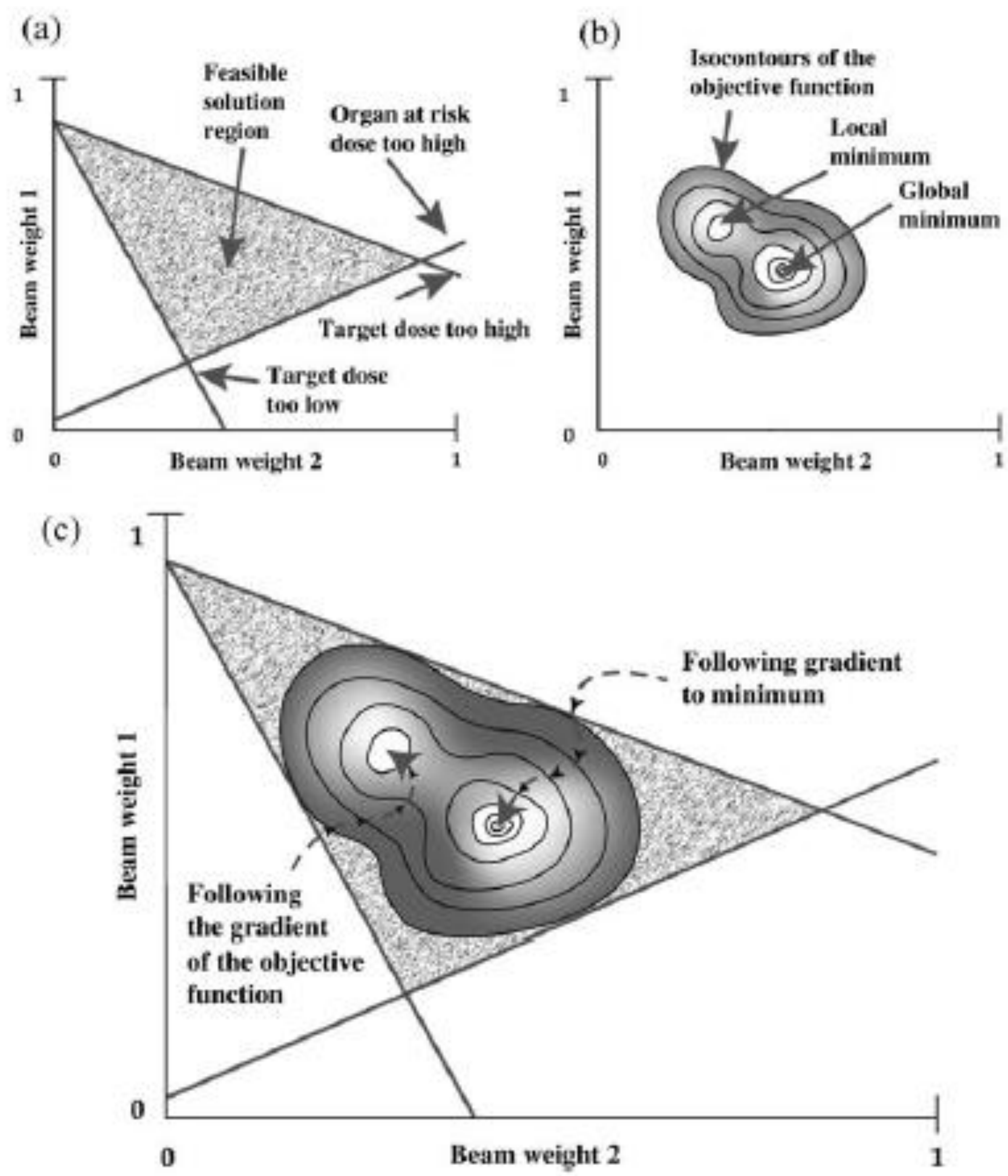
# Optimization constraints

## Hard constrains

Define the boundaries of the permissible solution set.

They can't be violated (negative intensity, unfeasible file size)

Solutions based only on hard constraints do not provide an optimal solution.

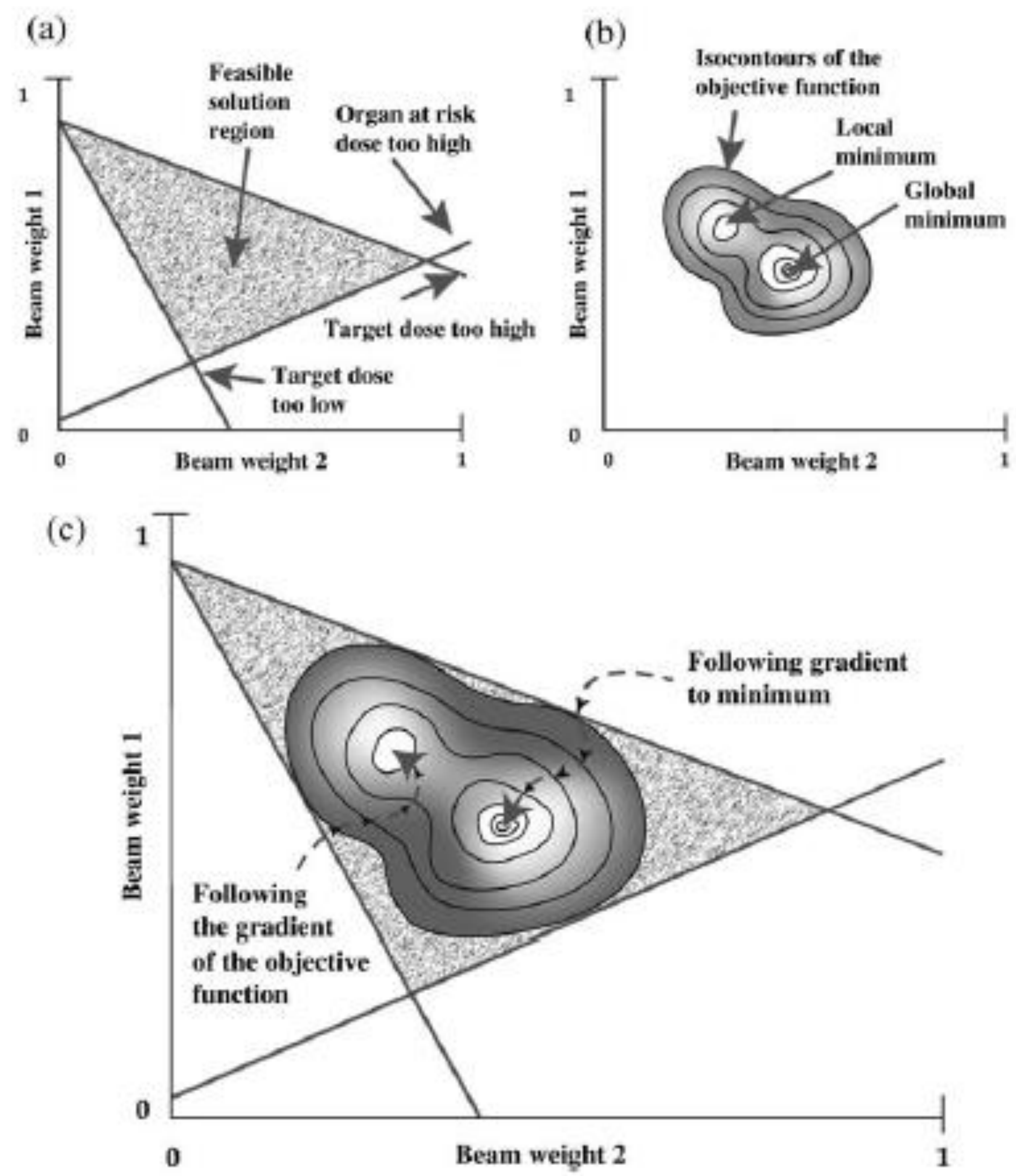


# Optimization constraints

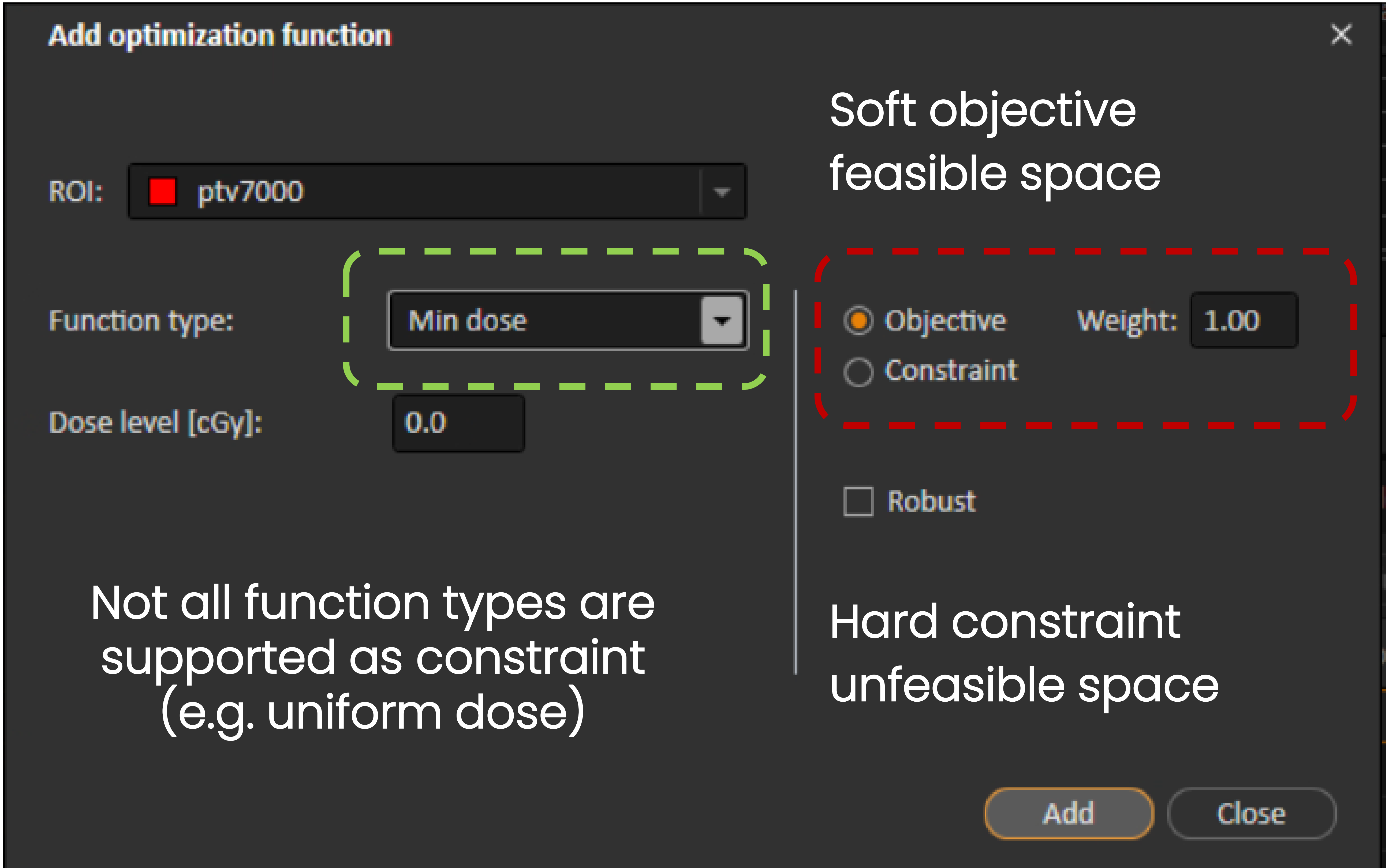
## Soft objectives

They define the "global" minimum or "best" solution for a given objective function.

Depending on the starting point, other solutions can be obtained (not always the global minimum).



# RayStation Treatment Planning System




Soft objective feasible space

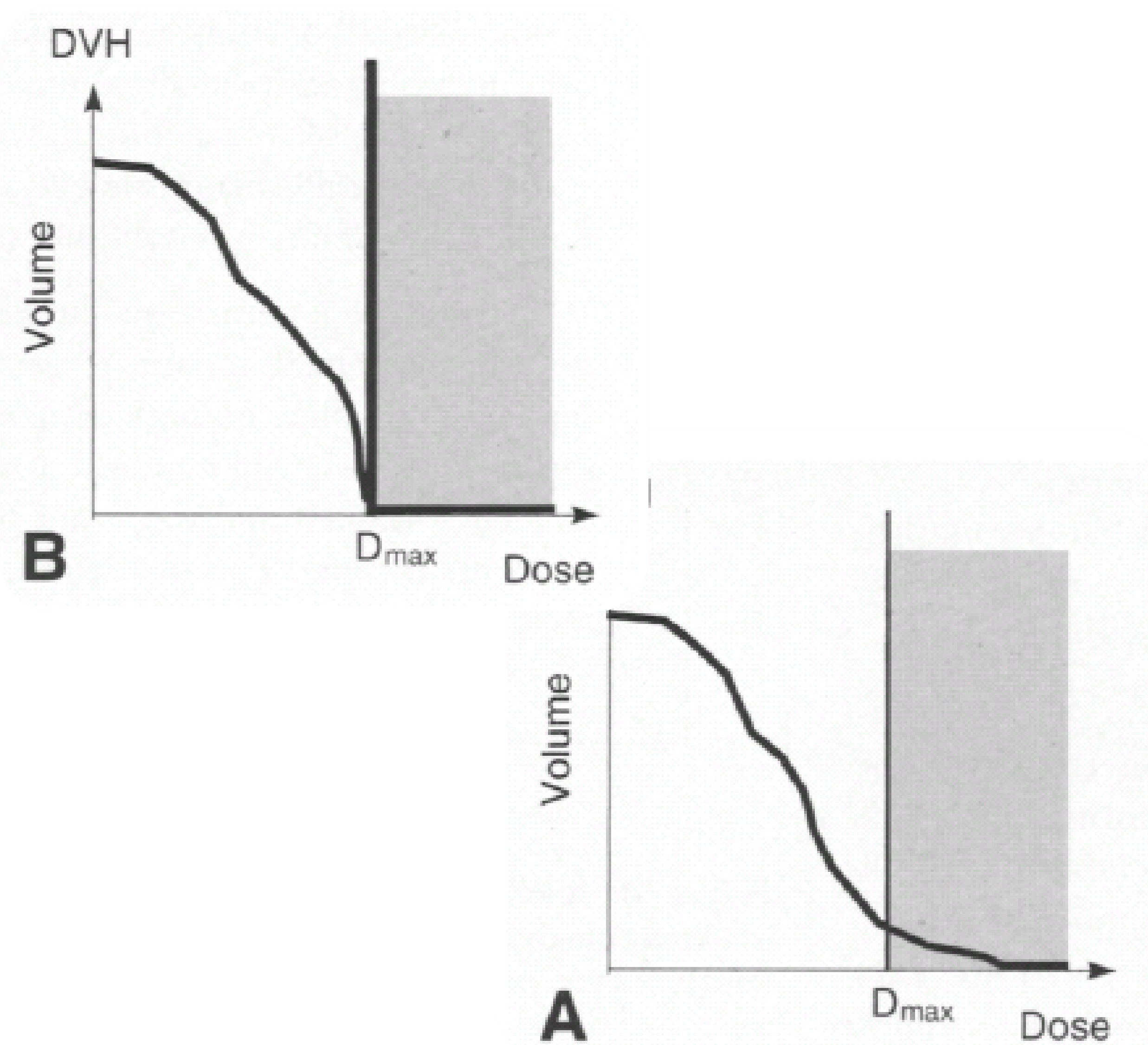
Hard constraint unfeasible space

Not all function types are supported as constraint (e.g. uniform dose)

# Dose constraints – OARs

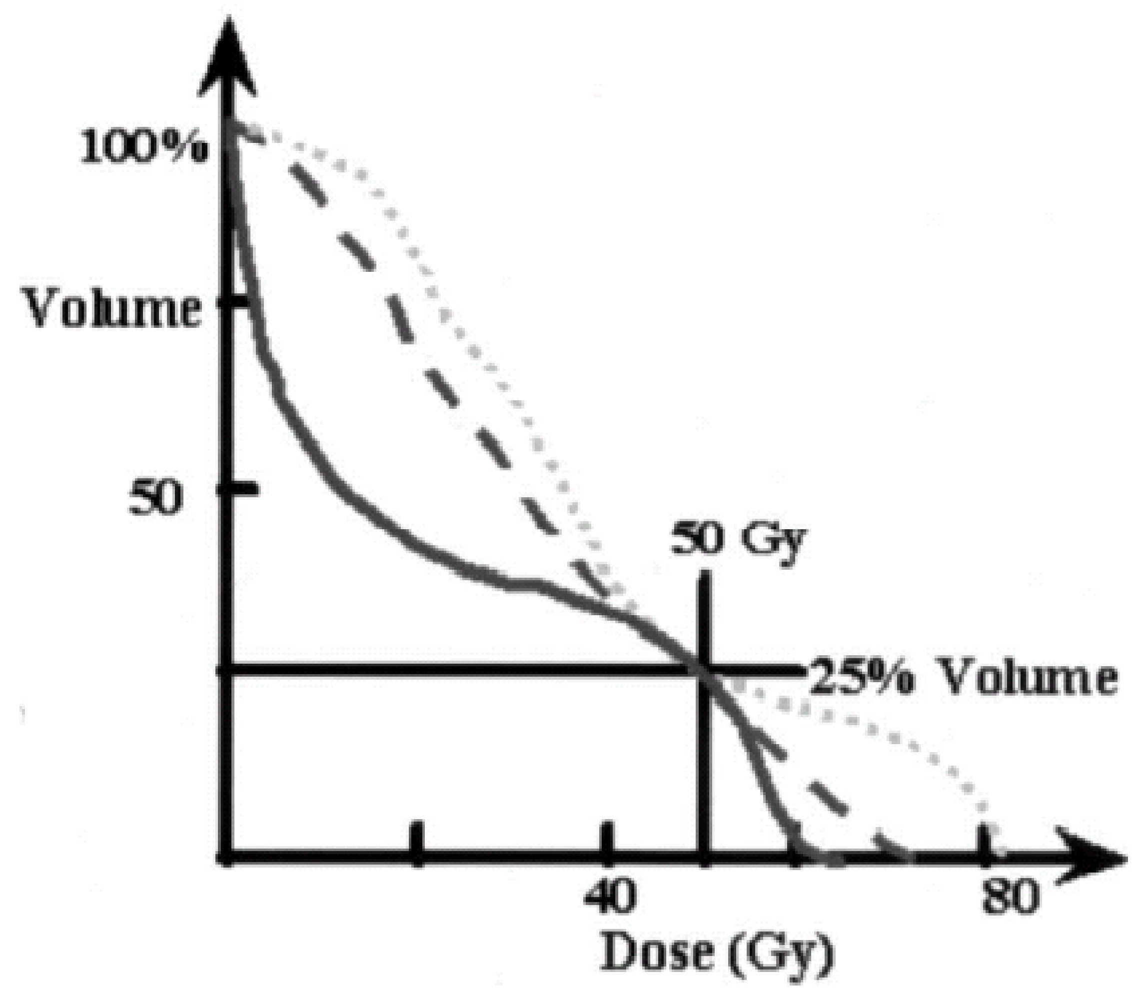
## Serial OARs

Maximum dose constraints
In most cases, the penalty is proportional to the square of the dose exceeding the tolerance level

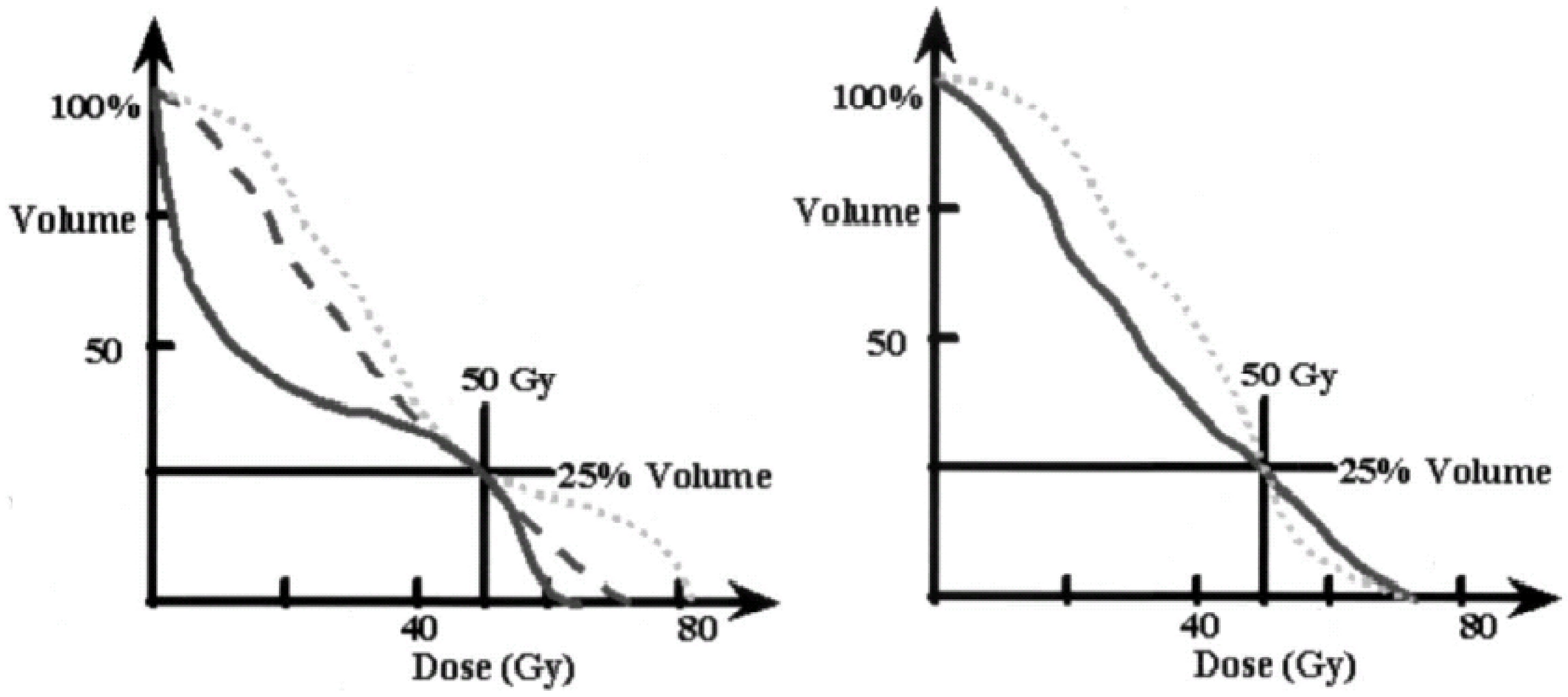



## Paralele OARs

Dose – Volume constraints



# Dose–Volume constraints – OARs



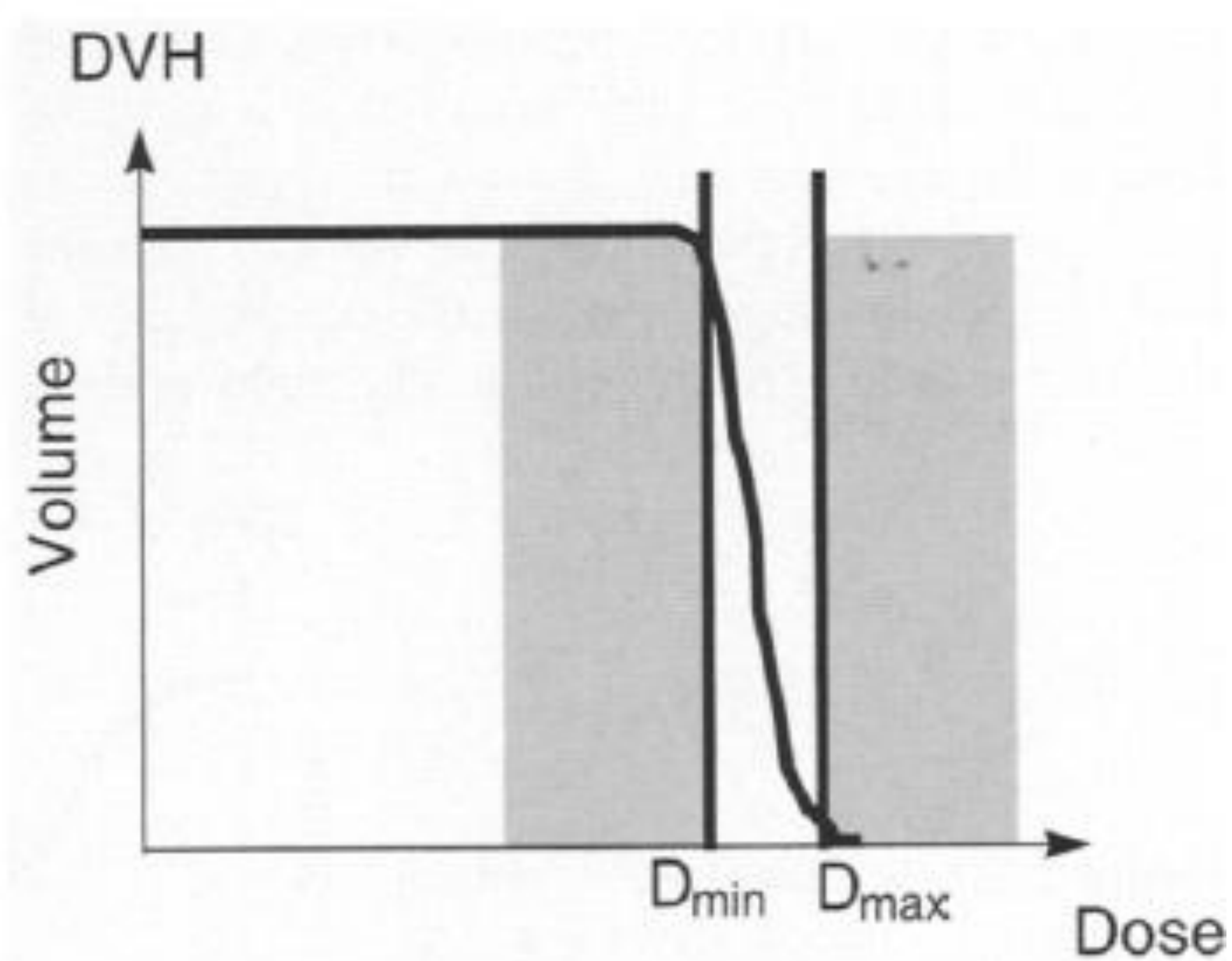

- What more a given clinical endpoint may be caused by a variety of dose distributions or DVH

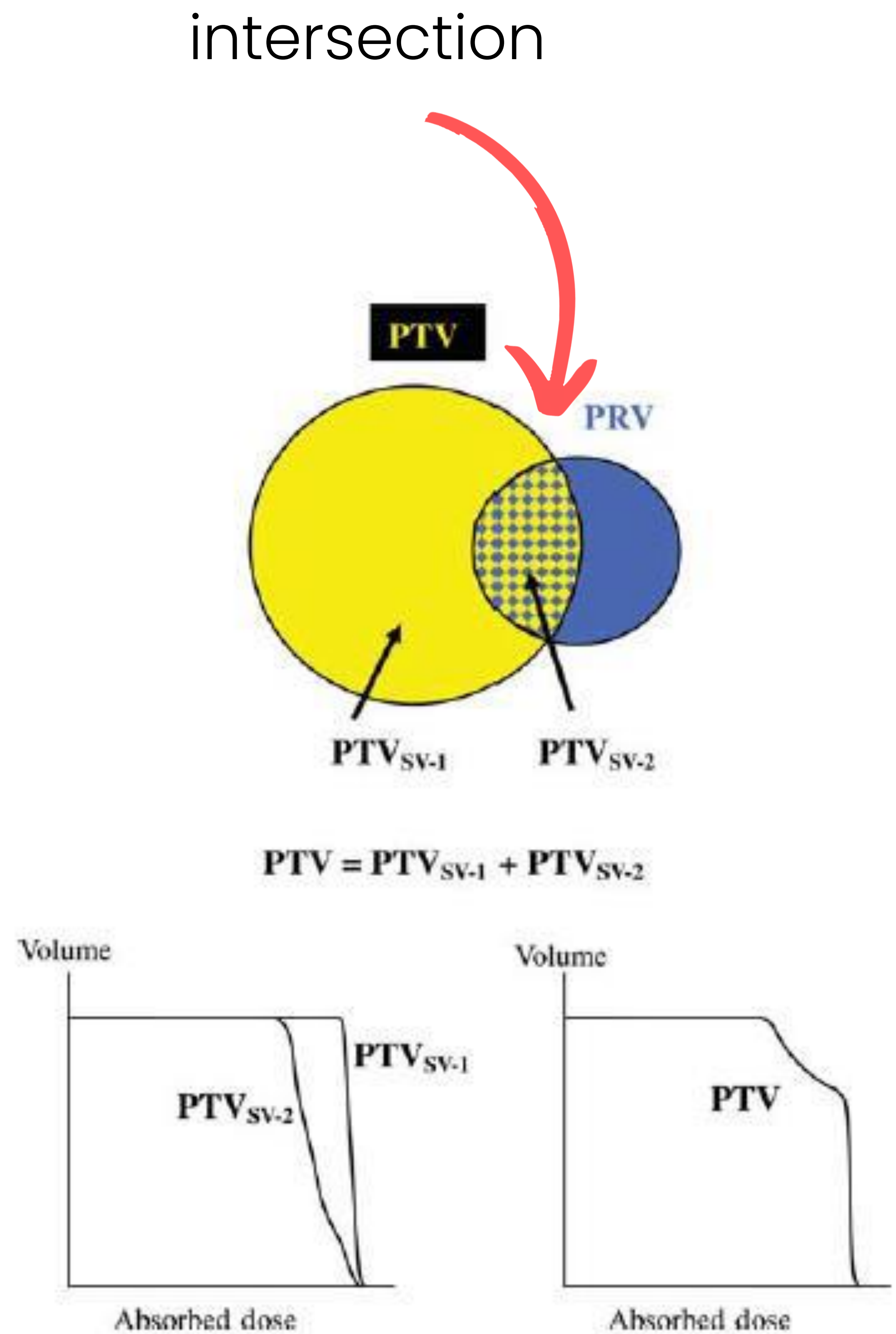
# Dose constraints – PTVs

- ~~$D_{min}$ > 95%, $D_{max}$ < 107%~~

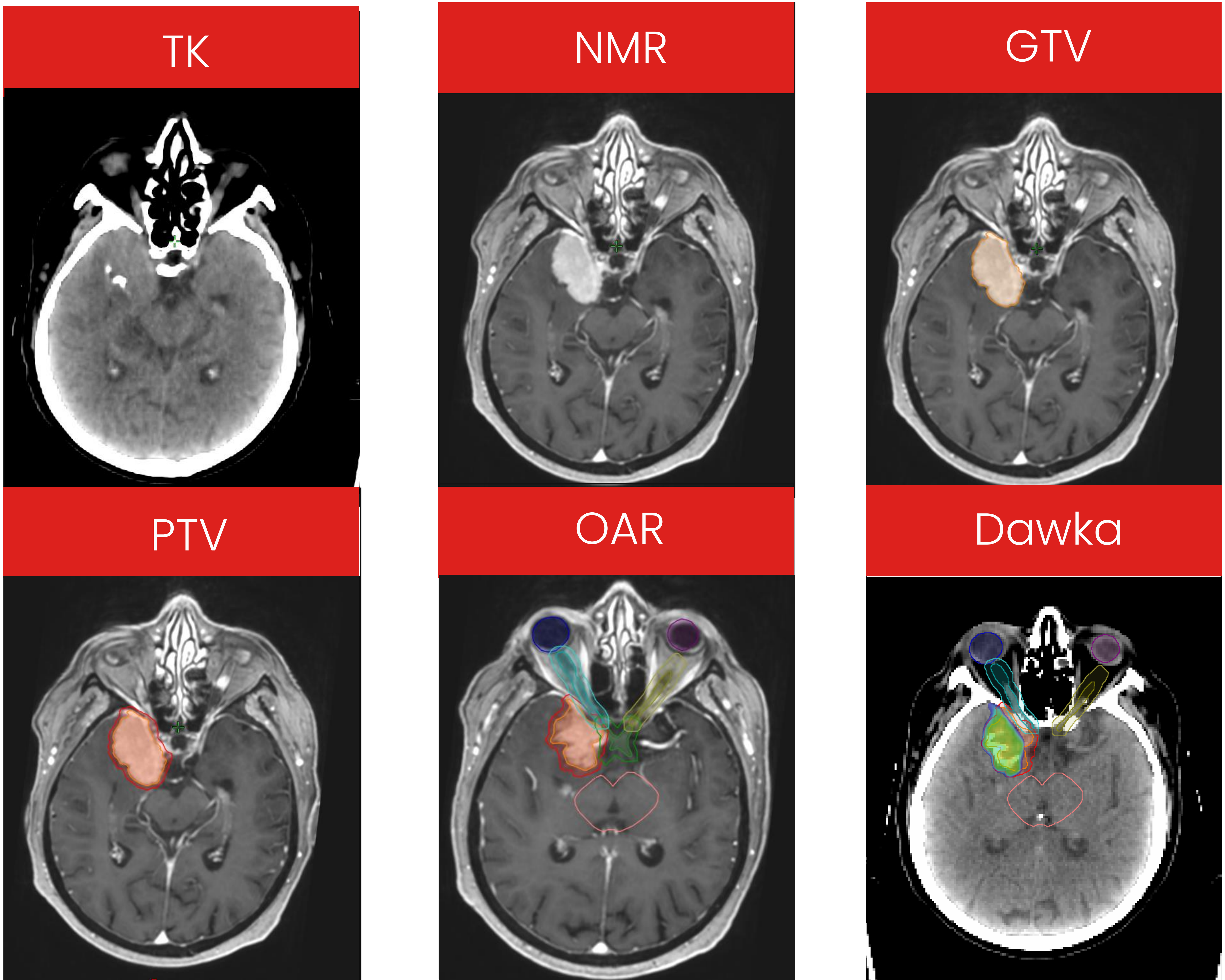
- **D98% > 95%, D2% < 107%**

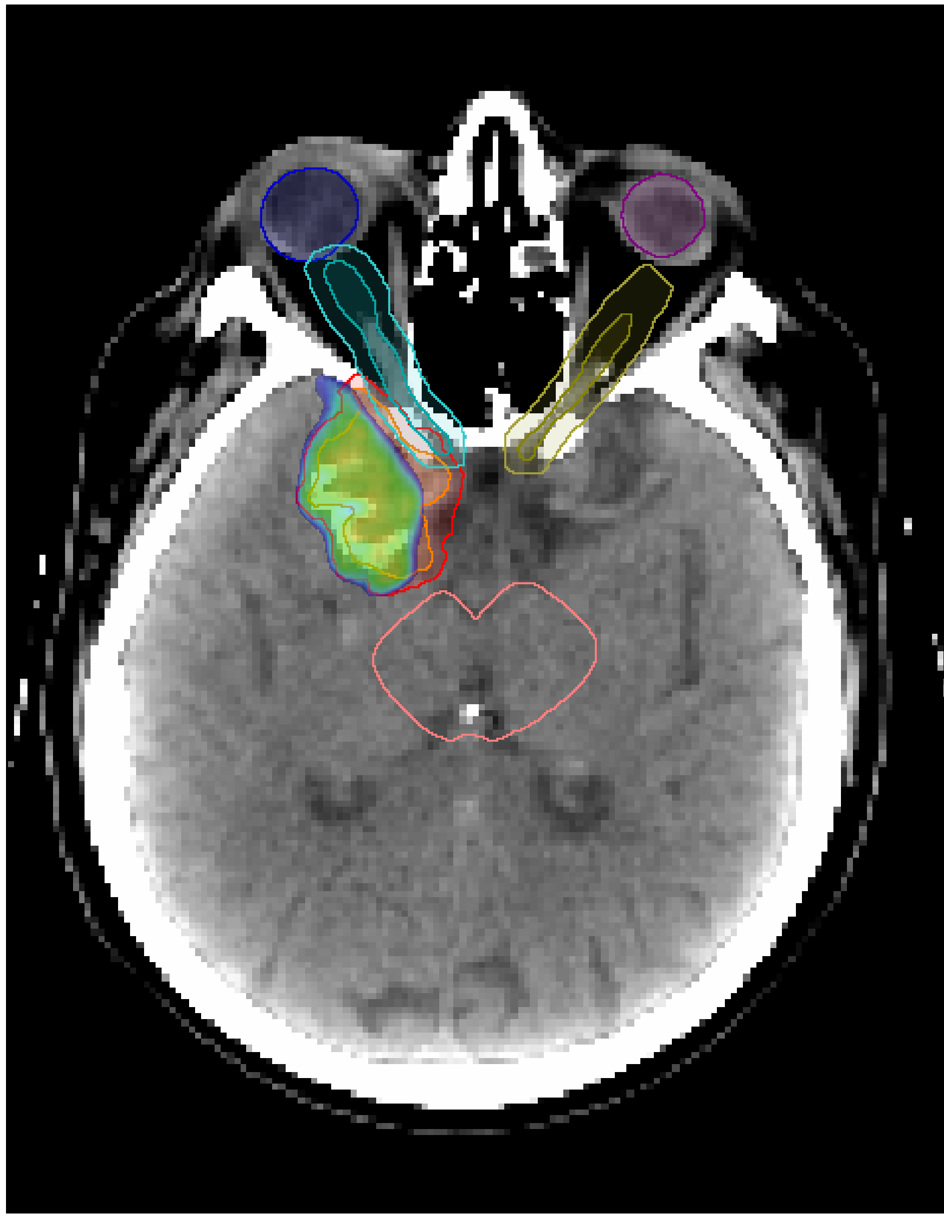
- D95%> 95%

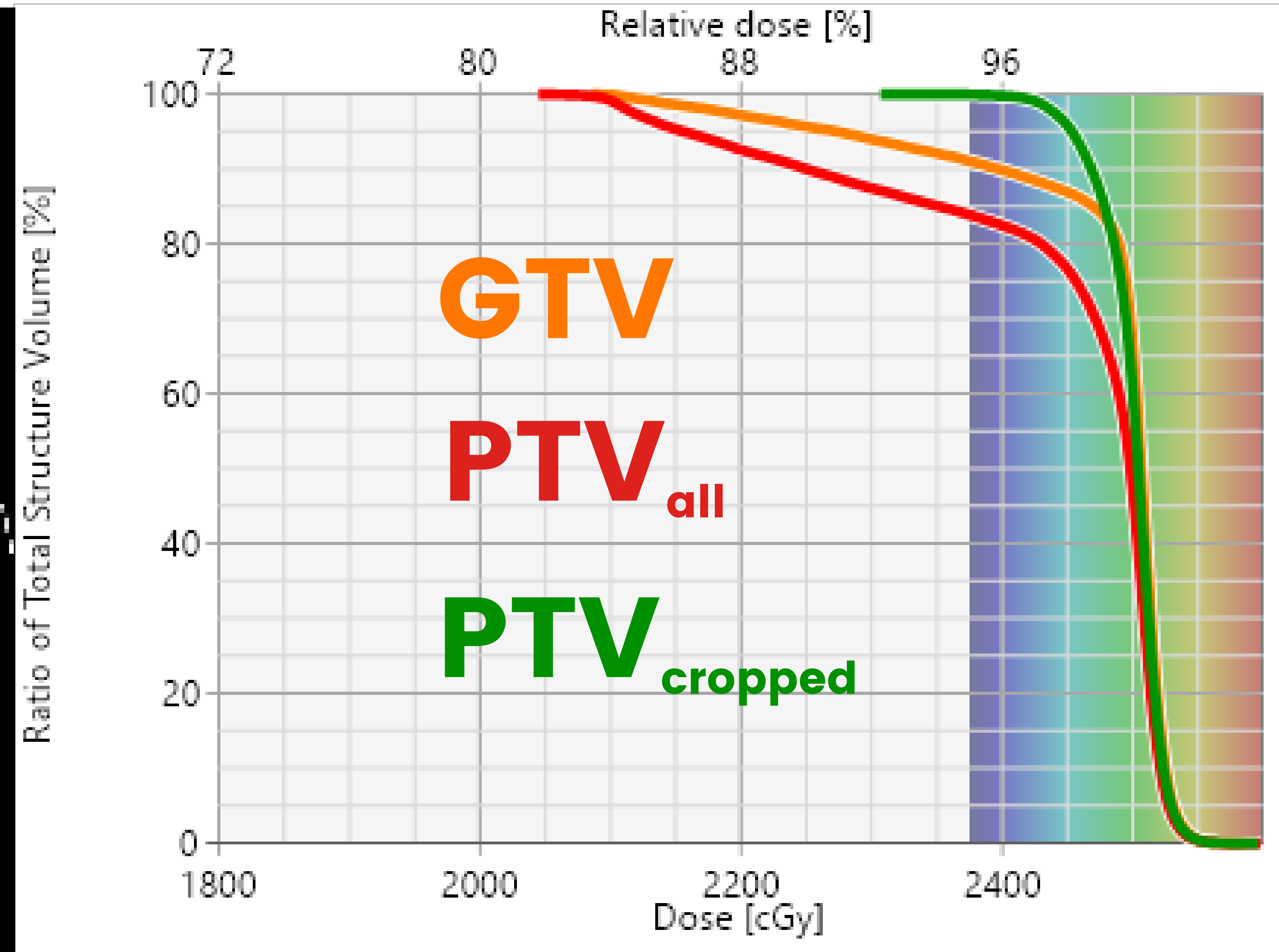


# OAR i PTV overlapping





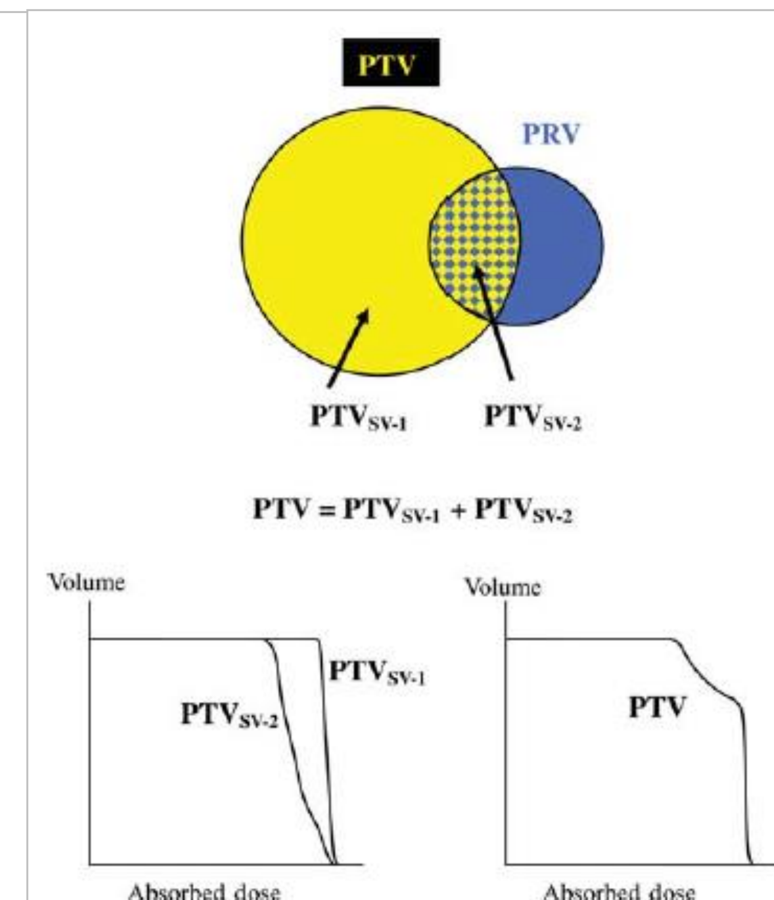



## Contradictory expectation

## PTV & OAR'S

If it is not possible to deliver a therapeutic dose to the entire PTV, under dosage areas should be reported.

**Important in analyzing future potential failures.**



# Objective function – EUD-based formalism

EUD, the dose given uniformly, which results in the same cell killing as the actual nonuniform dose distribution

$$EUD = \left(\frac{1}{N}\sum_i D_i^a\right)^{\frac{1}{a}}$$

$$F = \prod_j f_j \quad \Longleftarrow \quad f_{OAR} = \frac{1}{1+\left(\frac{EUD}{EUD_0}\right)^n}; \quad f_{PTV} = \frac{1}{1+\left(\frac{EUD_0}{EUD}\right)^n}$$

- Better results for OAR -> objective determined based on the whole organ, not a partial volume
- Wider search space -> search for plans with different DVH but the same EUD

# Weighting factors

- A combination of objectives is combined in the form of a single objective function.
- The weighting factors are often incorporated into the optimization process
- The influence of these factors on the final solution is not known until the end of optimization (exception – when you can change them during optimization)
- A good understanding of how the weighting factor works and training on how to use them is required!

**FIAT 500**

**Alfa Romeo TONALE**


Even if you have a driving license, you have to learn how to drive each vehicle.

## IMRT planning process - steps

Plan geometry ✓

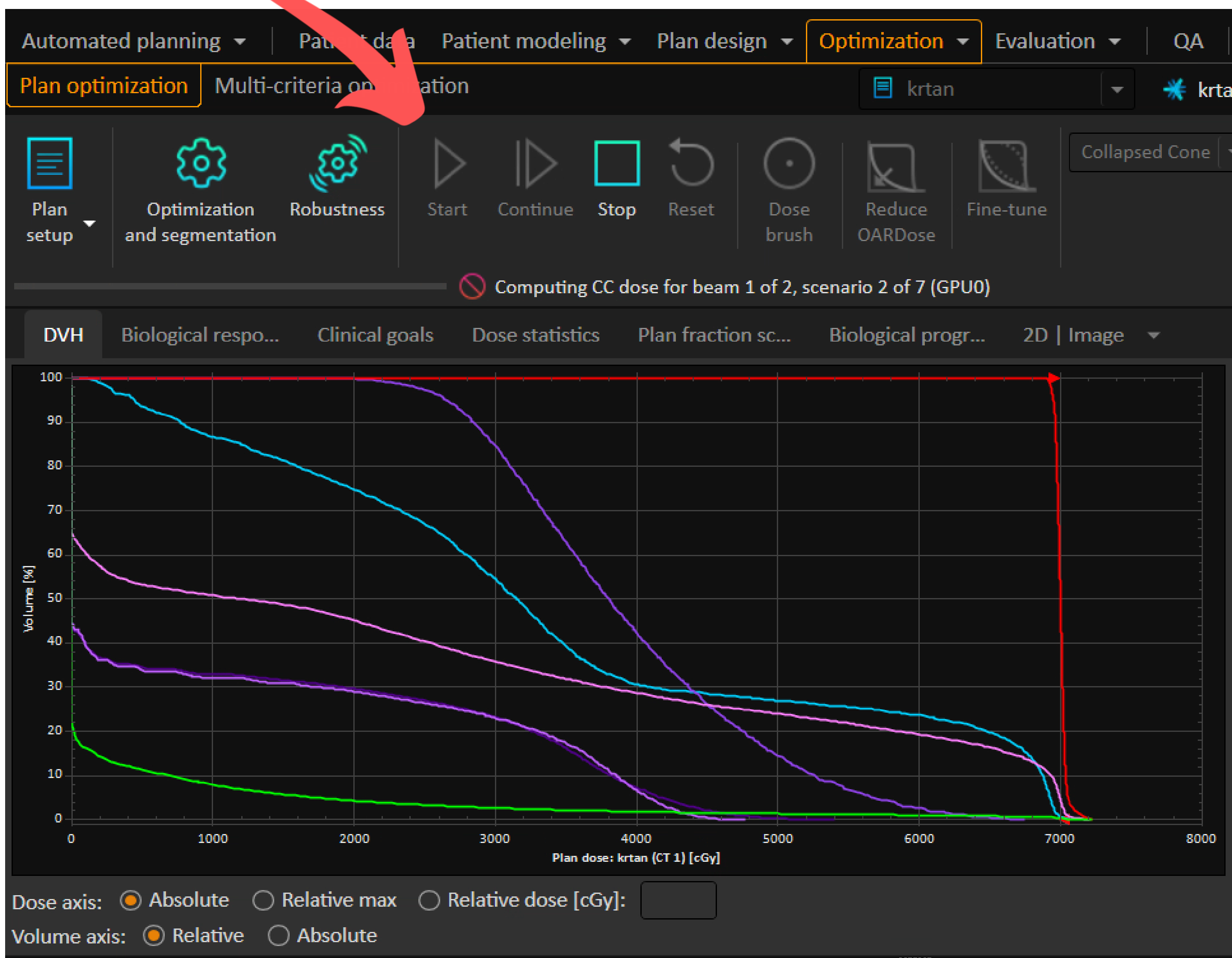
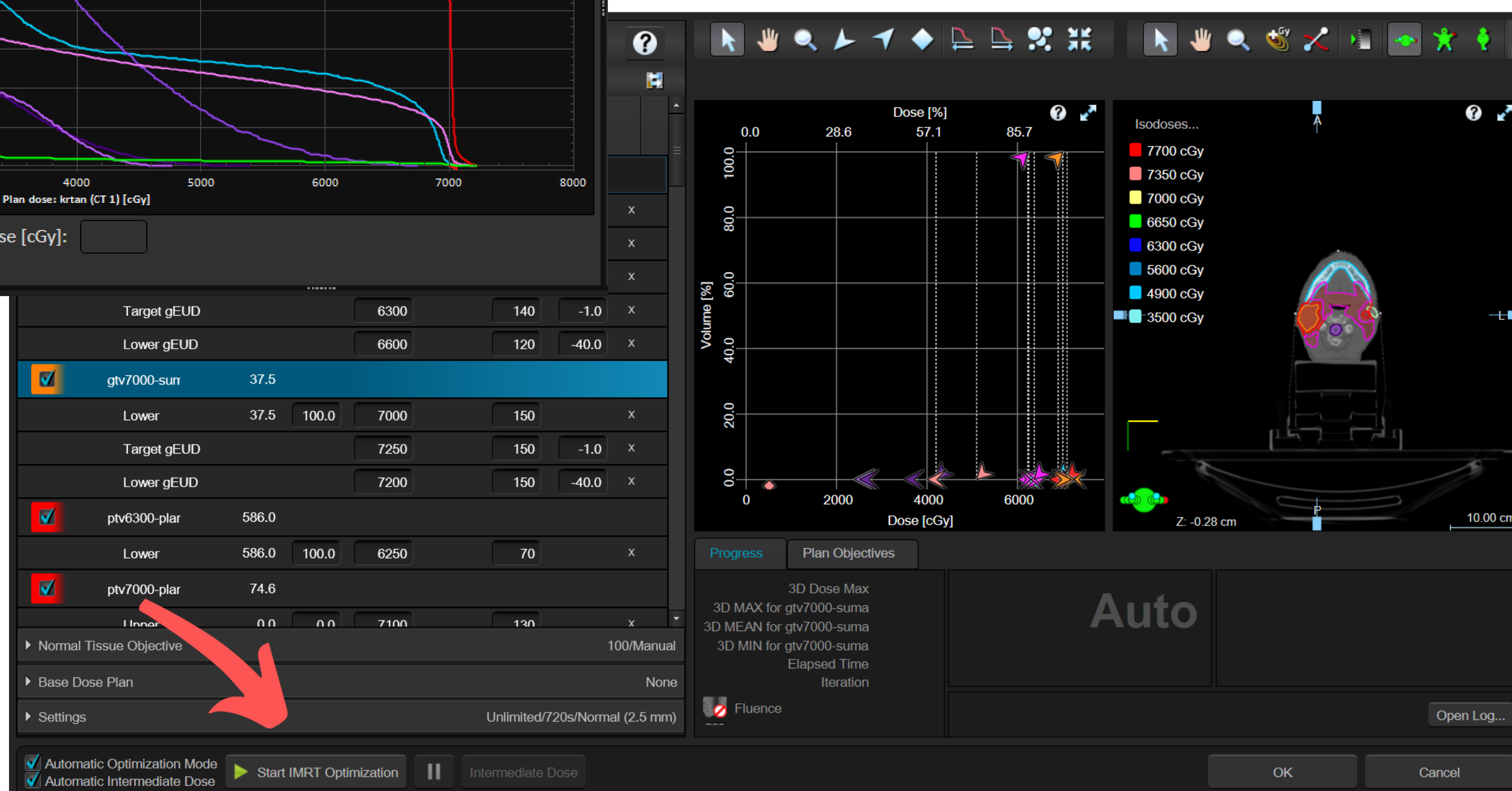
Cost function (a mathematical measure of meeting expectations) ✓

Dose constraints (Dmax, Dmin, dose-volume, Dmean, biological measures) ✓

Importance/penalty (the relative importance of individual constraints) ✓

**What next?**

# Push the button!!!


# Take a break!!!

*Józef Chełmoński – Storks (1900) - National Museum in Warsaw*

# Optimization

1. For all possible **beamlets, initialization** of intensities - zero or the same value for all those not passing through the PTV.

2. **Dose** and objective function **calculation**.

3. Iterative **objective function maximization**
   Methods: *stochastic, deterministic*

4. Finding **corresponding fluence maps**

5. **Calculation of the final dose** distribution

# Iteration method

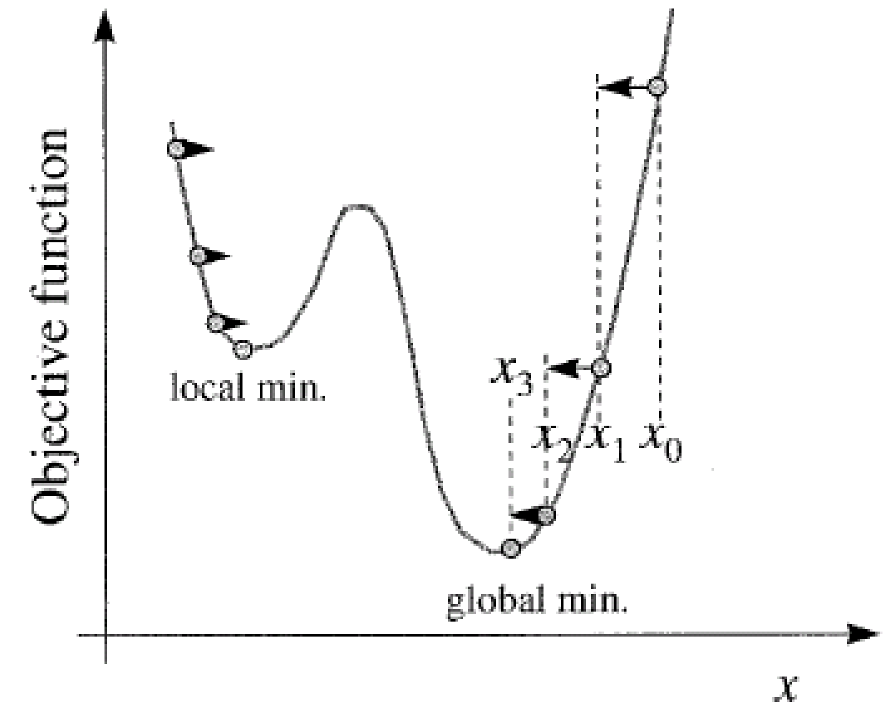
Optimization technique depends on objective function.

**Deterministic methods** – the rules for making changes to beam profiles in each iteration are determined (i.e.: there is no element of randomness)

**Stochastic Methods** – changes made based on a random search for a new position in each iteration step

# Deterministic methods

1. quick methods
2. they may fall into local minima
3. only changes related to a decrease in the cost function are accepted.



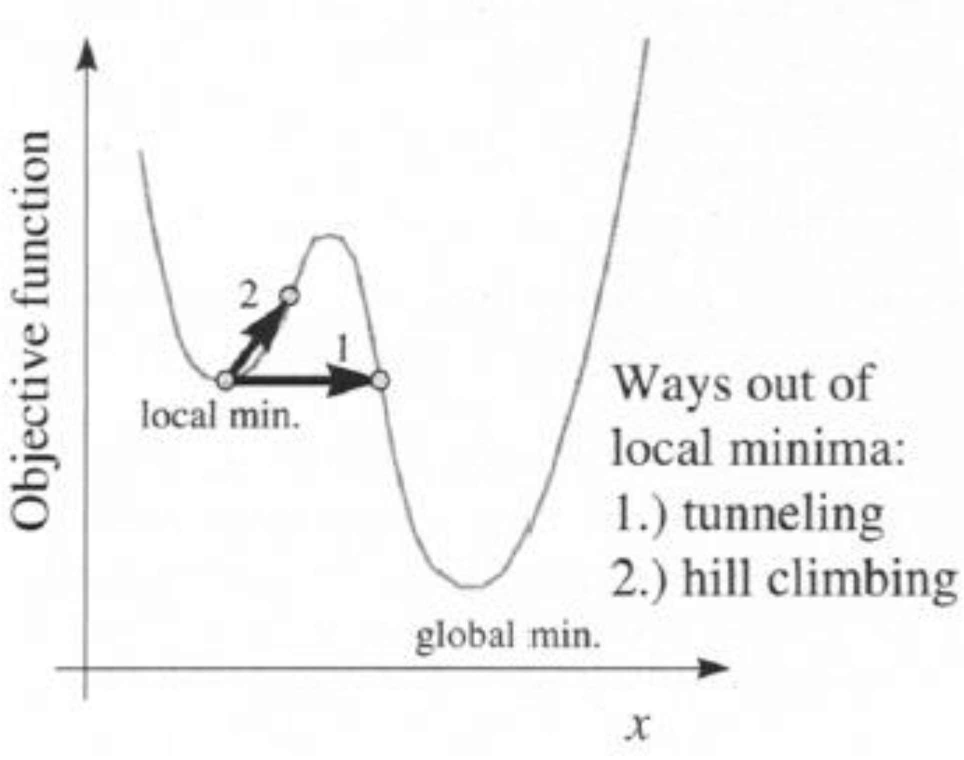

but:

- local minima do not necessarily have to occur. For simple objective functions they do not exist
- Choosing a good starting point helps you avoid getting "stuck„
- Local minima may be close to the global minimum

Gradient technique
Linear programming (limited to linear objective functions, not accurately describing tumor response and OAR's irradiation)
linear least squares (least squares method, gradient descent)

# Stochastic method

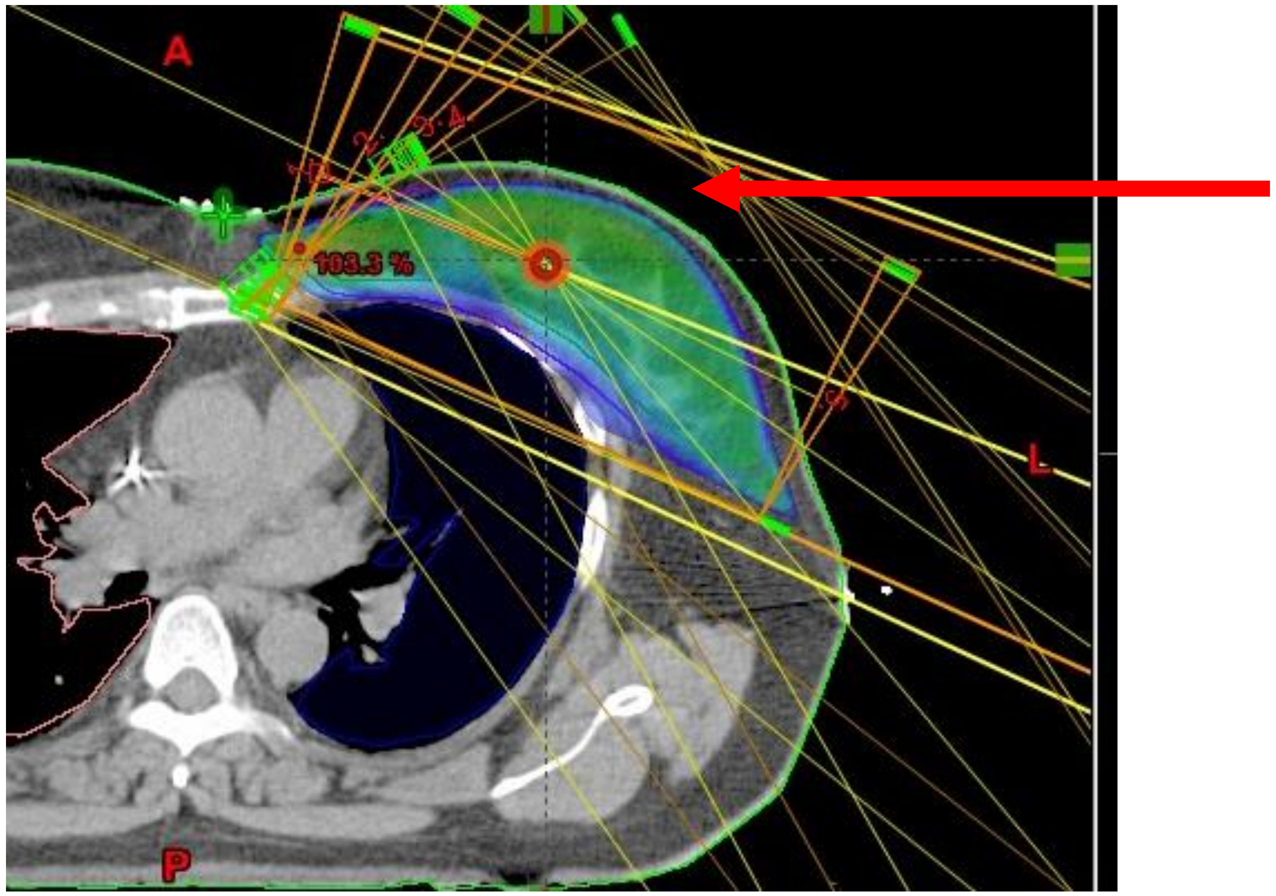
1. Slower
2. They enable finding the global minimum
3. Changes associated with an increase in the cost function are accepted with a certain probability




**Simulated Annealing**, fast simulated annealing, genetic algorithm

# TREATMENT PLANNING
# WHAT ELSE IS WORTH REMEMBERING?

# Expand your beam




1. Skin flas tool (Eclispe)
2. Artificial bolus
3. Tools included in your TPS

Where:

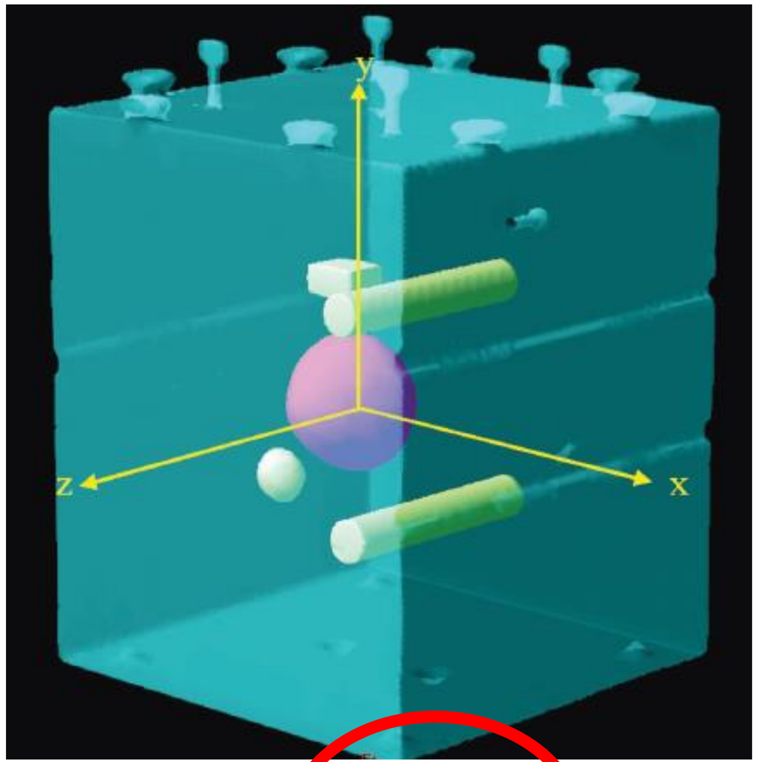
- Breast – breathing, size changing
- Sarkomas – gtv size change

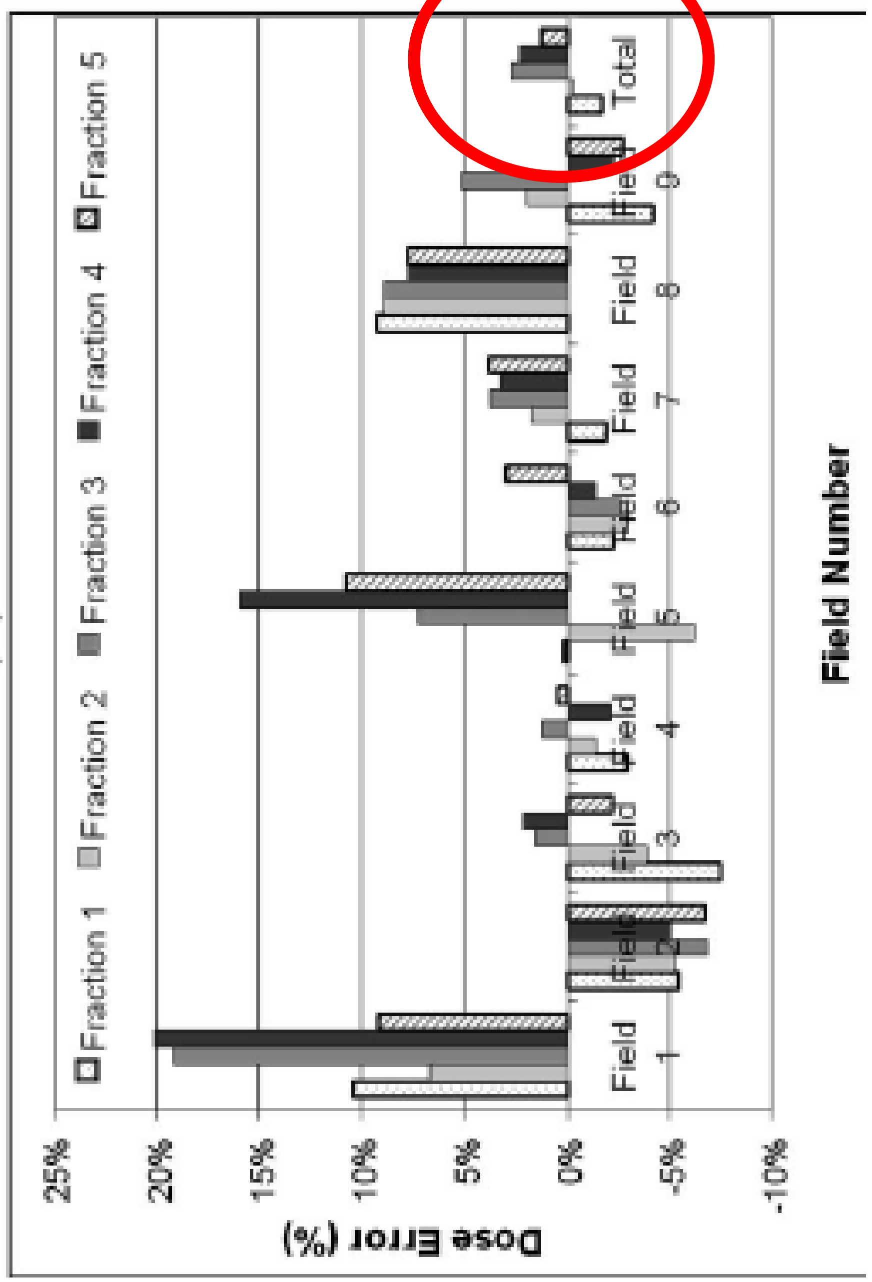
# IMRT – interplay effect

- DMLC-IMRT with a different number of fields
- Measurements with a chamber (0.125) and films in moving phantom
- Measurements for a different number of fractions with and without respiratory mobility
- TCP, EUD

- For one field -11.7% to 47.8%
- For a sum of fields -1.7% to 3.5%
- $D_{min}$ -18.8%, $D_{max}$ +19.7, but due to randomness it was averaged

3D dose distributions, DVHs, TCPs, and EUDs for stationary and moving cases showed good agreement after two or more fractions, suggesting that tumors affected by respiration motion may be treated using IMRT without significant dosimetric and biological consequences.






Duan et. al. Med. Phys. 33 (5) May 2006

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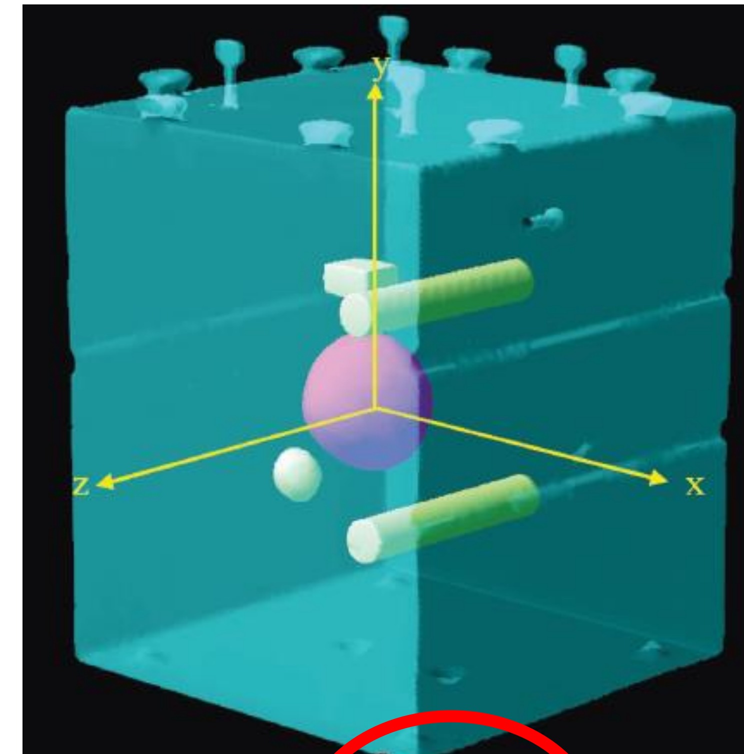
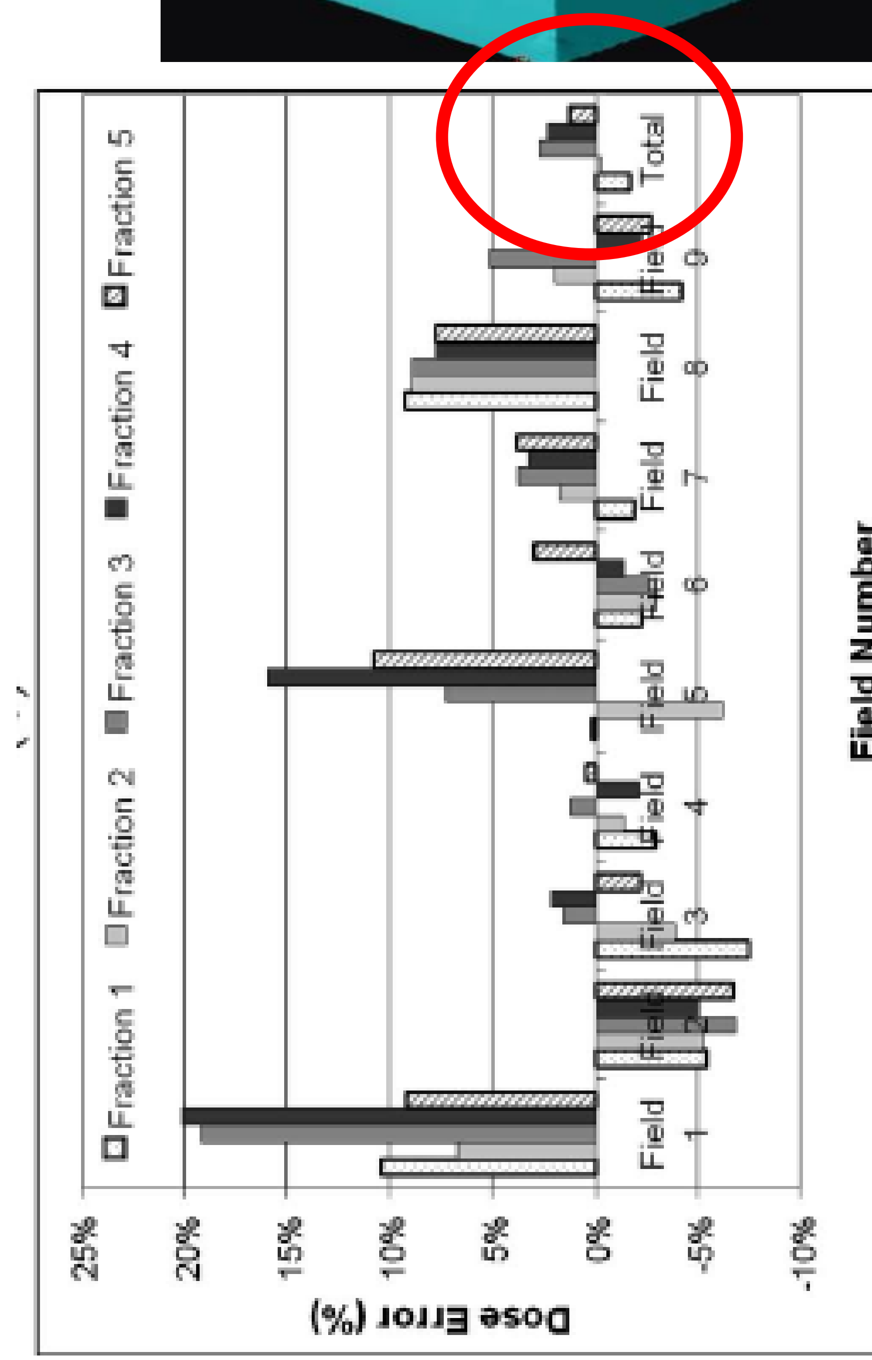
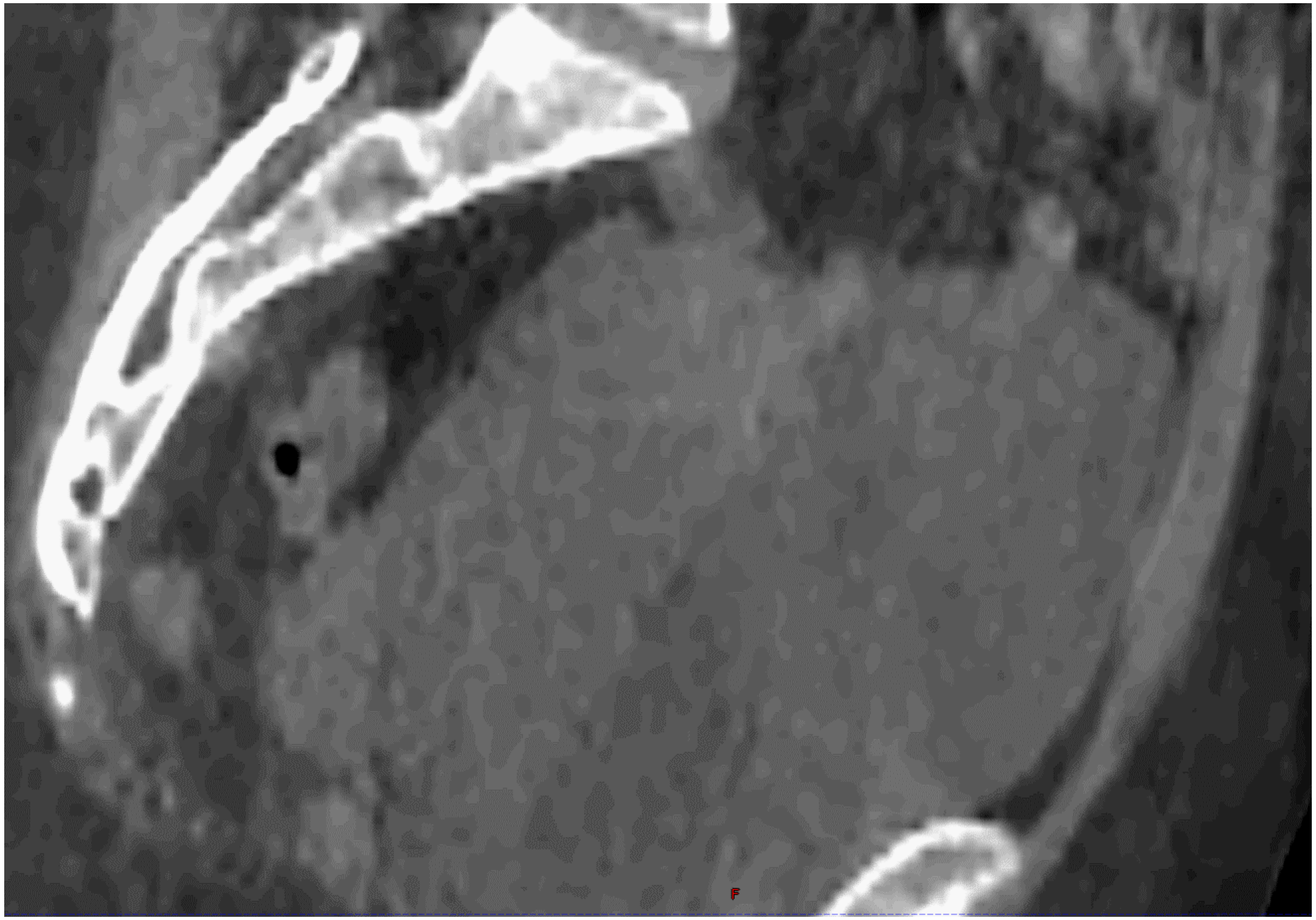
good habit
avoid small segments
e.g: aperture shape controller

Duan et. al. Med. Phys. 33 (5) May 2006



# Movement it not only lungs!!!



# IMRT - pros and cons

- More conformal dose distribution for concave targets, for PTV close to OAR's, the possibility of dose increase (increasing local control/cure)
- reducing complications (e.g. xerostomia, diarrhoea)
- SIB (distributions with different dose levels)

**but**

- Machine QA - more advanced QA of machines required
- TPS QA - new procedures for verifying the system. More detailed acceptance of the TPS dose calculations is required.
- Patient-specific QA - additional dosimetric verification required
- More precise contouring of both PTV and OAR required
- More precise patient position verification

# literature

- Journal of the ICRU Vol 10 No 1 (2010) Report 83: Prescribing, Recording, and Reporting Photon-Beam Intensity-Modulated Radiation Therapy (IMRT)
- AAPM reports
- Intensity Modulated Radiation Therapy: A Clinical Perspective Arno J. Mundt, John C. Roeske (2005)