Integration of Magnetic Resonance Imaging and Proton Therapy

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OncoRay National Center for Radiation Research in Oncology Dresden

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OncoRay structure

Centre for Innovation Competence supported by:

- Technical University Dresden
- Helmholtz-Center Dresden-Rossendorf
- University Hospital Carl Gustav Carus

OncoRay is situated on the Medical Campus and belongs to the Medical Faculty / University Hospital

Department of Radiotherapy

Director

Prof. Dr. med. Mechthild Krause

Personell

- 8 ROs (*Strahlentherapeuten*)
- 10 MPEs (*Medizinphysikexperte*)
- 16 MTAs (*medizin-technische Assistent*)
- 25 KS (*Krankenschwester*)

Radiation treatments per year

- 2400 oncological indications
- 500 benign indications

Expertise focus

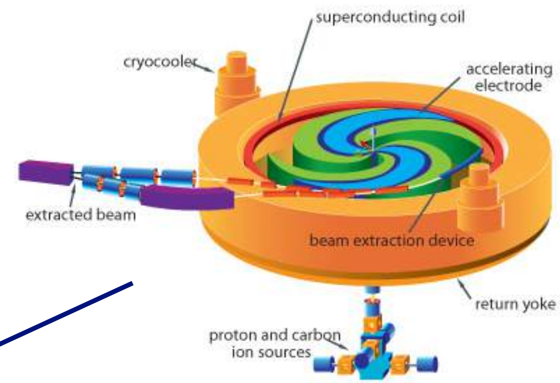
- rectal cancer
- head and neck cancer
- lung cancer

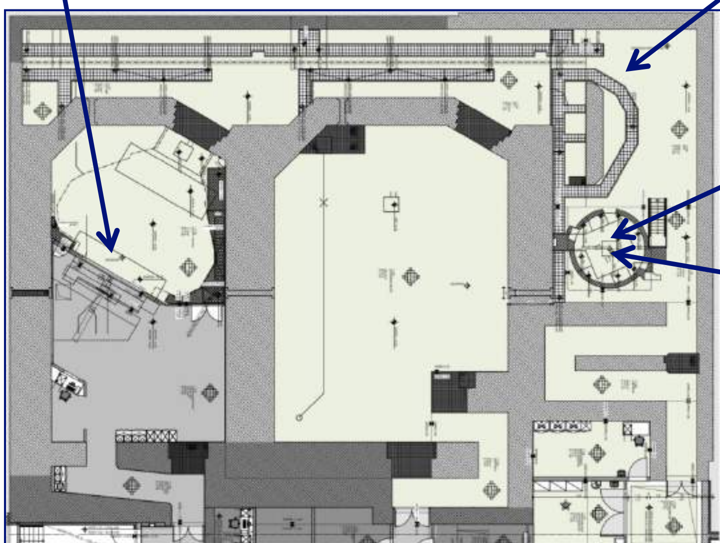
Photon therapy using high-energy X-rays

Particle therapy using high-energy protons

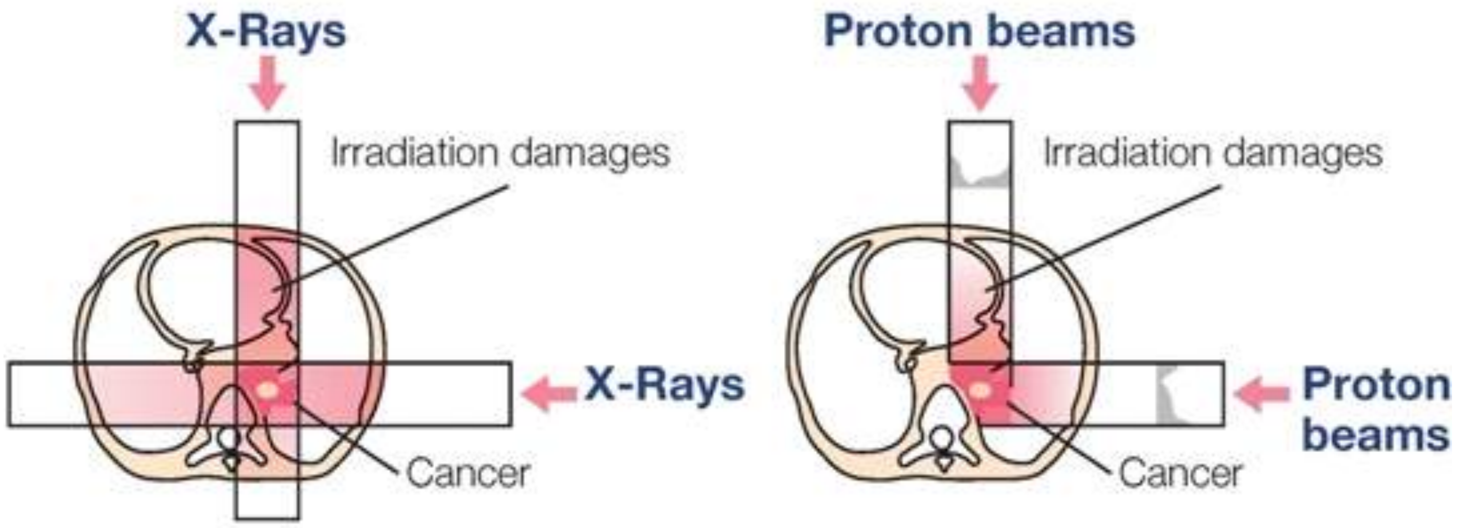
University Proton Therapy Dresden

Universitäts Protonen Therapie Dresden





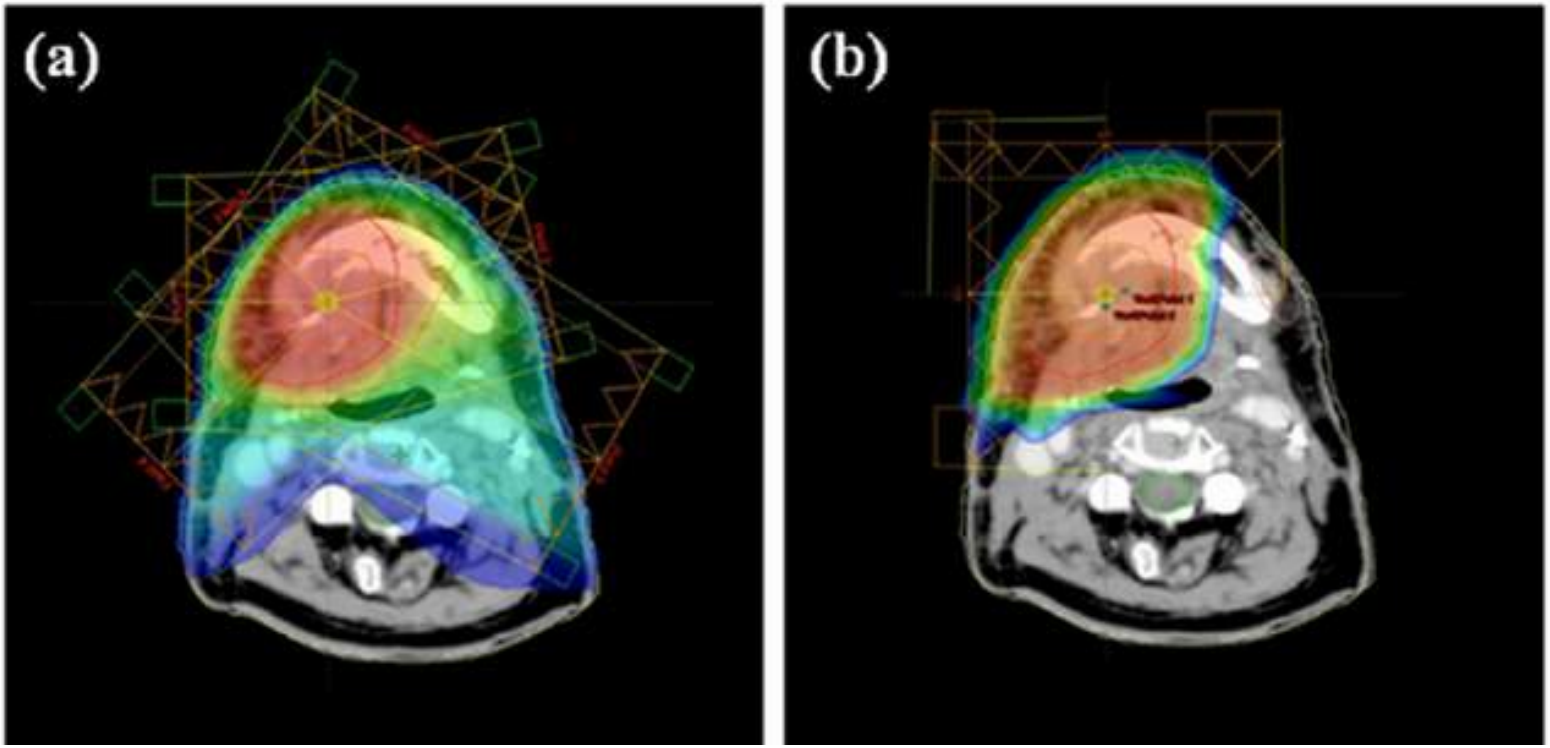
X-ray therapy *vs*. Proton therapy



- Proton beam <u>stops</u> inside the patient → **improved** normal tissue sparing
- Density changes in beam path → **range** (penetration depth) **uncertainty**

Treatment planning: dose distributions

- **Head and neck tumour**



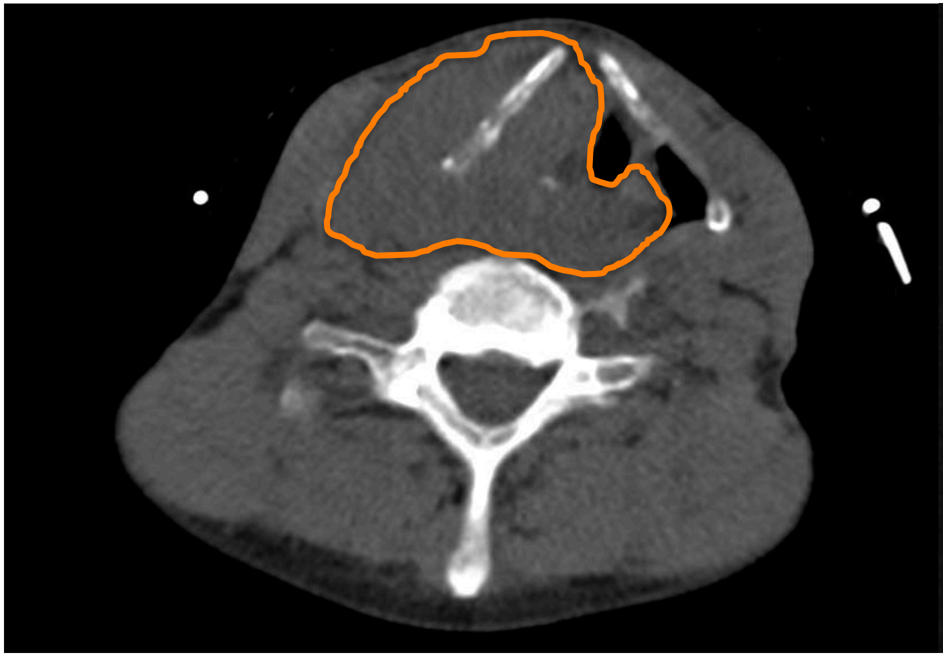
X-ray therapy

Proton therapy

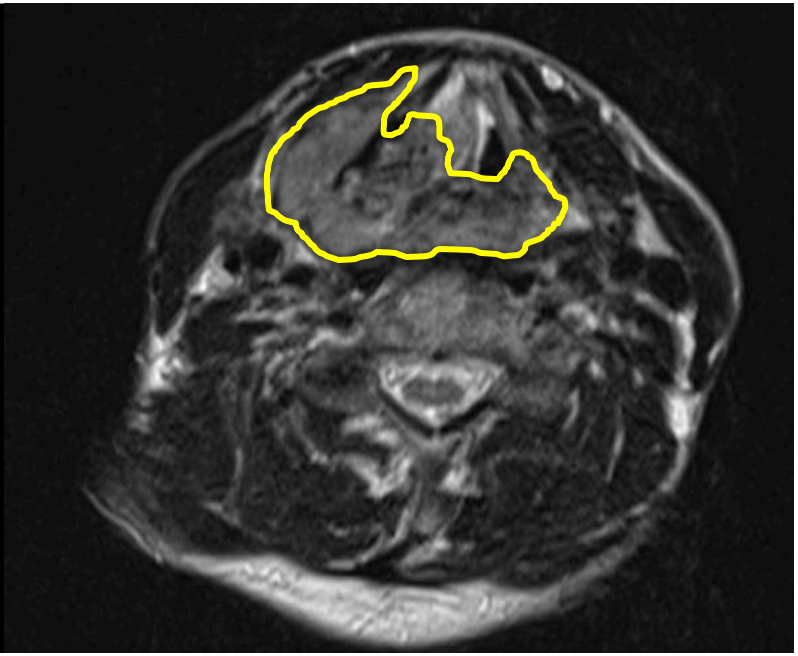
Image-guided target volume definition

- **Head and neck tumour**

T4N1M0 laryngeal carcinoma



Computed Tomography (CT)

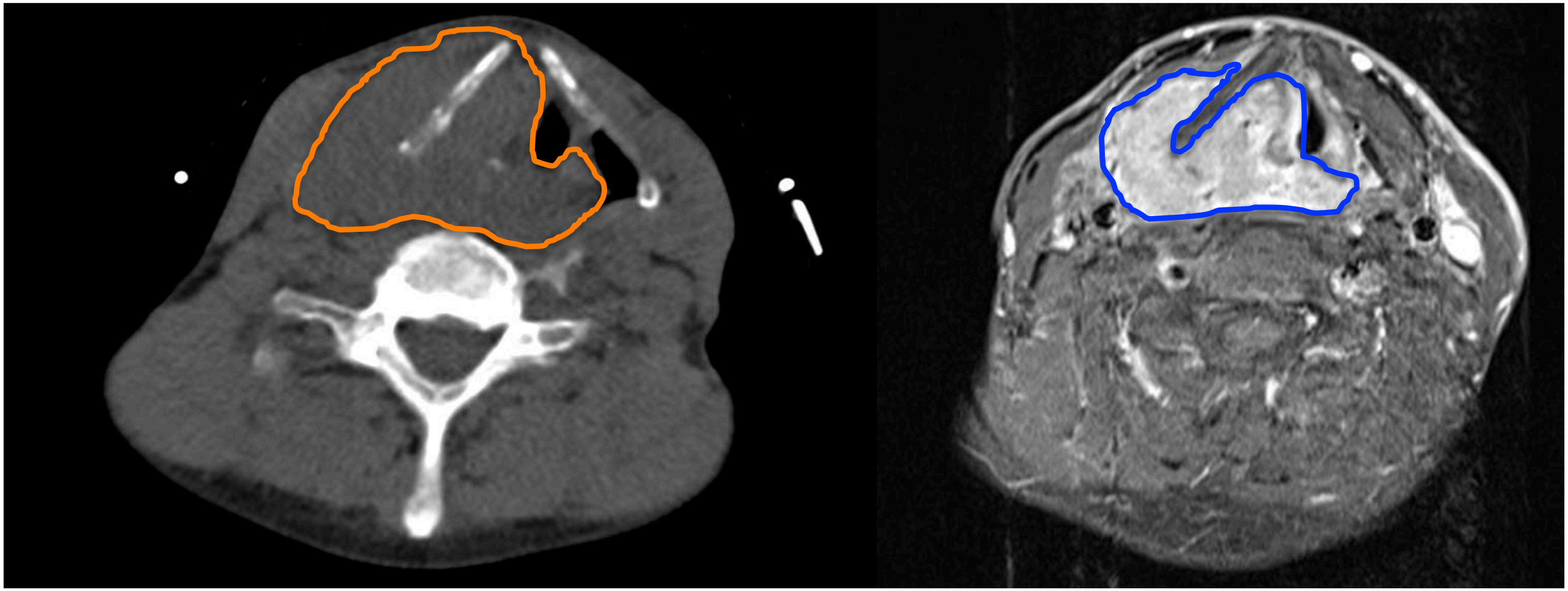


Magnetic Resonance Imaging (MRI)

Image-guided target volume definition

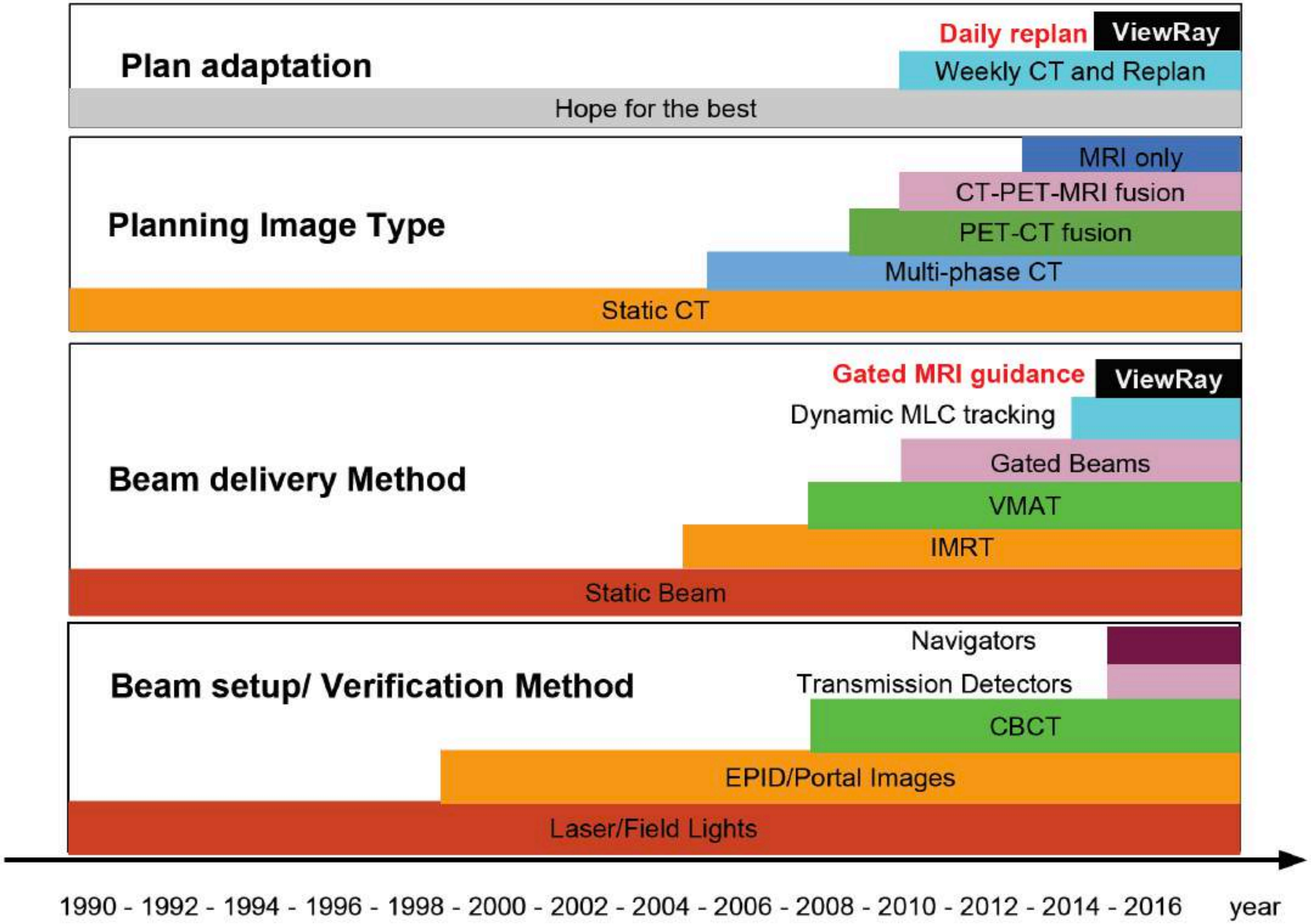
- **Head and neck tumour**

T4N1M0 laryngeal carcinoma



Computed Tomography (CT) Magnetic Resonance Imaging (MRI)

Technical developments in radiotherapy



Courtesy: dr. Brad Oborn (Univ. Wollongong)

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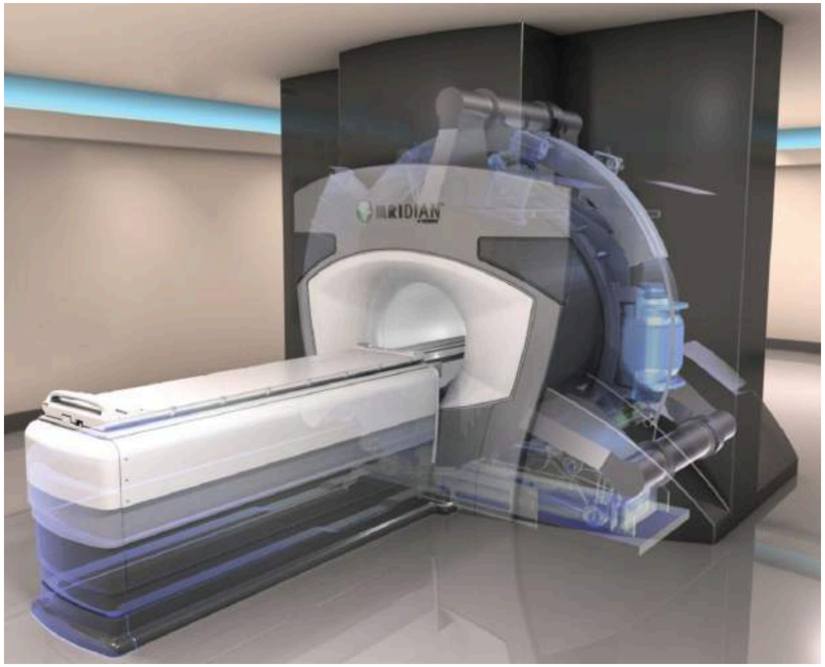
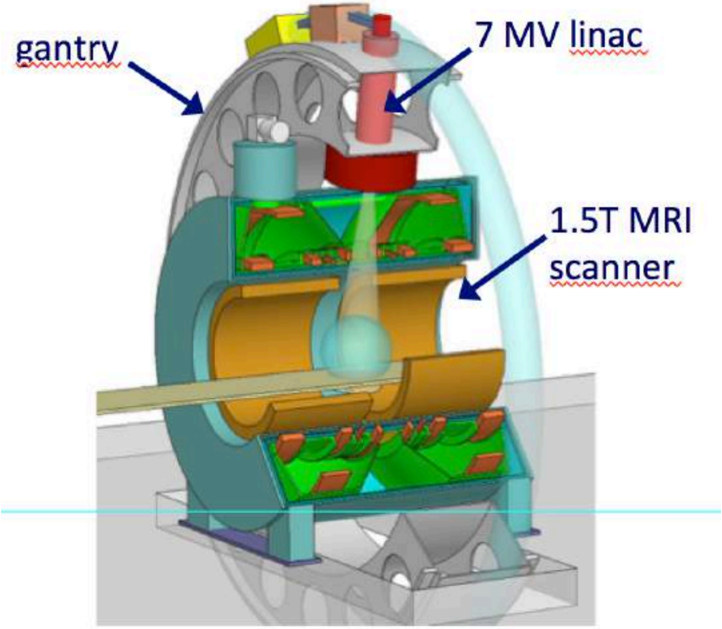
Real-time MRI-guided radiotherapy

MRIdian (ViewRay Inc.)



- gantry with ^{60}Co source heads
- split bore 0.35 T magnet

Integrated MRI-Linac



Elekta/Philips (Utrecht group)

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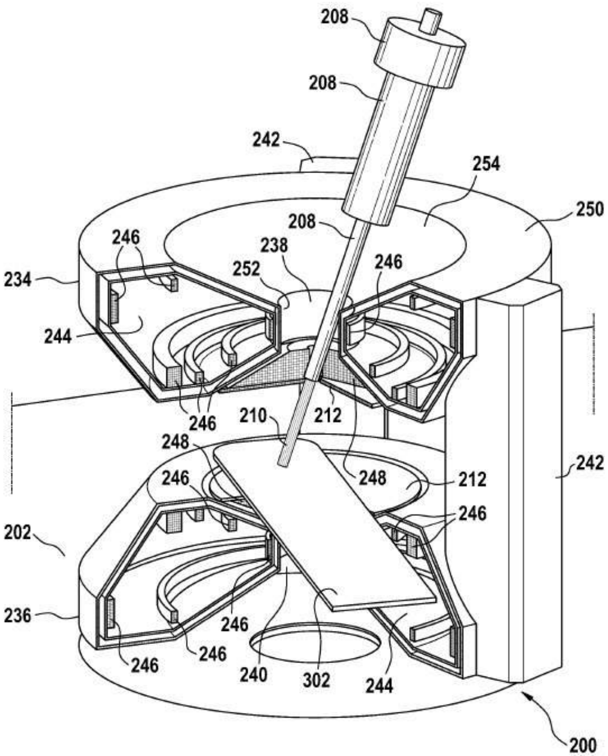
Integration of MRI and proton therapy

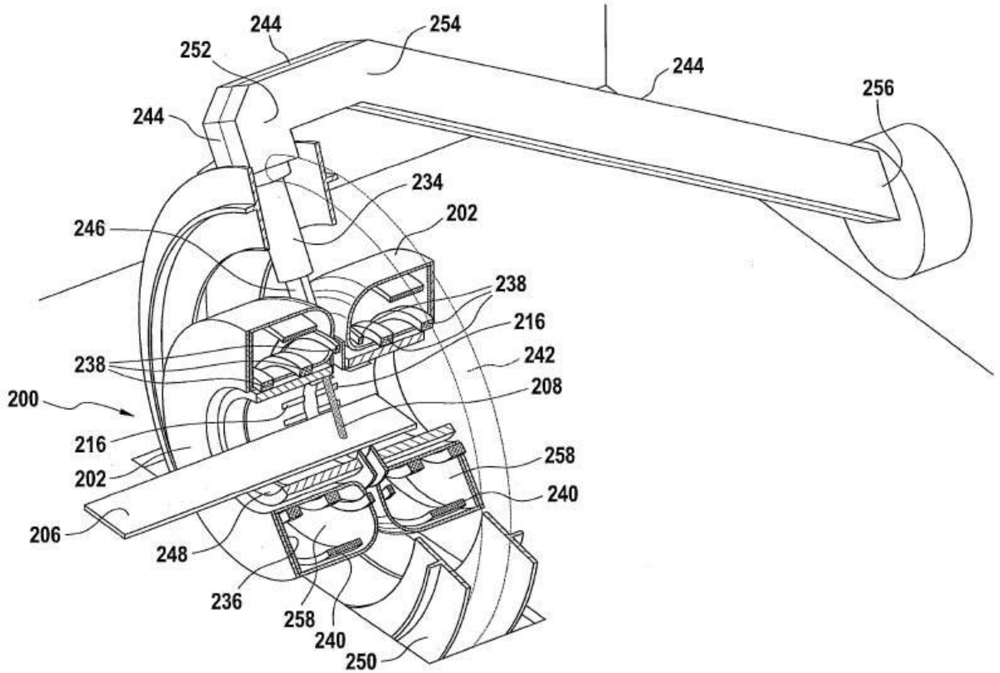
Vision: *treat what you see, track what you treat*

- MRI scanner at beam isocenter

patent EP2376195



Open MRI scanner



Split-bore MRI scanner

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Rationale of MRiPT

1. Image-guidance in proton therapy lags behind IGXT

- 2D X-ray imaging (throughout available)
- in-room CT (only available in some centers)
- on-board CBCT (recently released product)

X-ray based systems:

- limited intra-fractional imaging capabilities
- limited soft-tissue contrast

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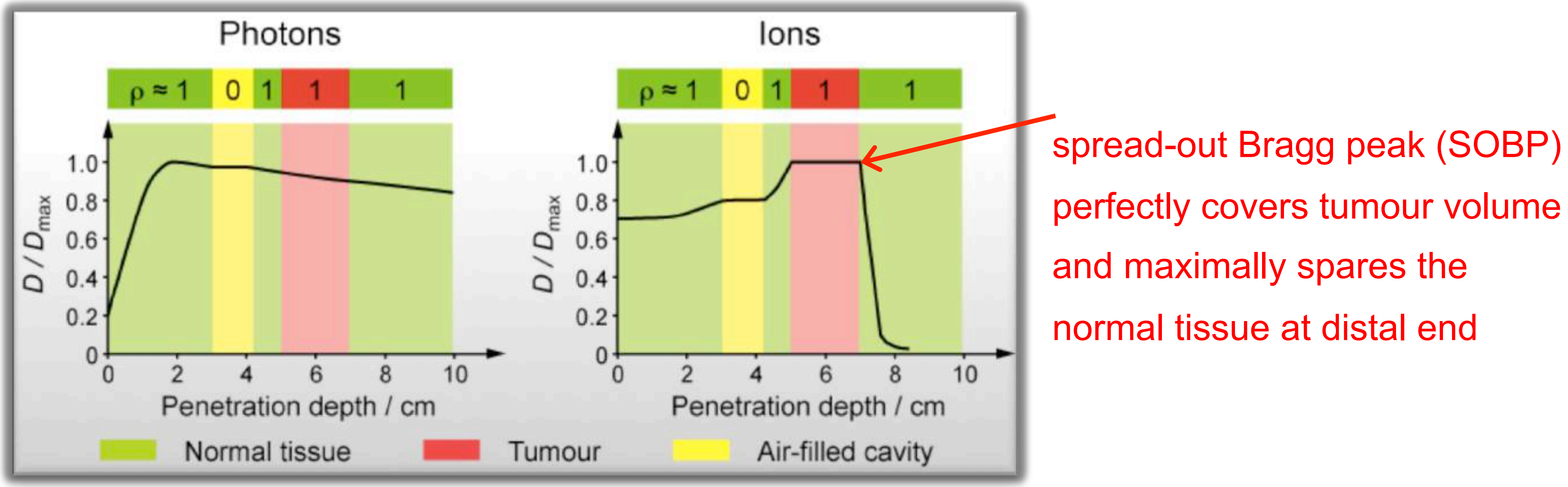
Rationale of MRiPT

2. Protons are **more sensitive to anatomical variations** than photons
   - **material composition** in beam path determines **Bragg peak location**



- What happens if the air-filled cavity (ρ ≈ 0) is replaced by normal tissue (ρ ≈ 1)?

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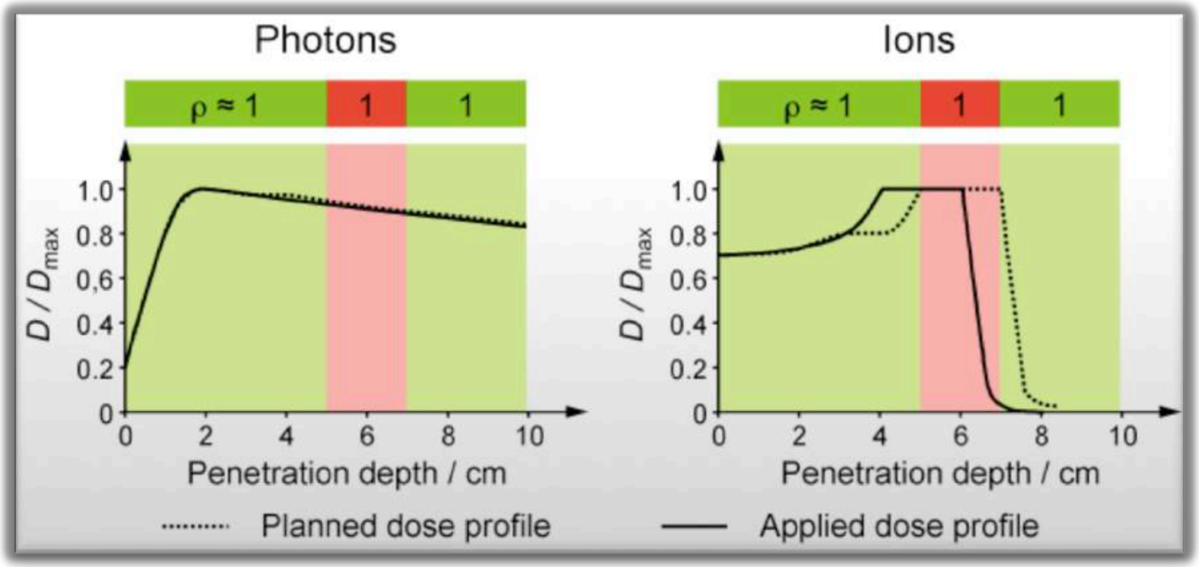
Rationale of MRiPT

2. Protons are **more sensitive to anatomical variations** than photons
   - **material composition** in beam path determines **Bragg peak location**



- **Protons**: SOBP will shift stream upwards
- **Photons**: hardly any dosimetric effect between planned and applied dose

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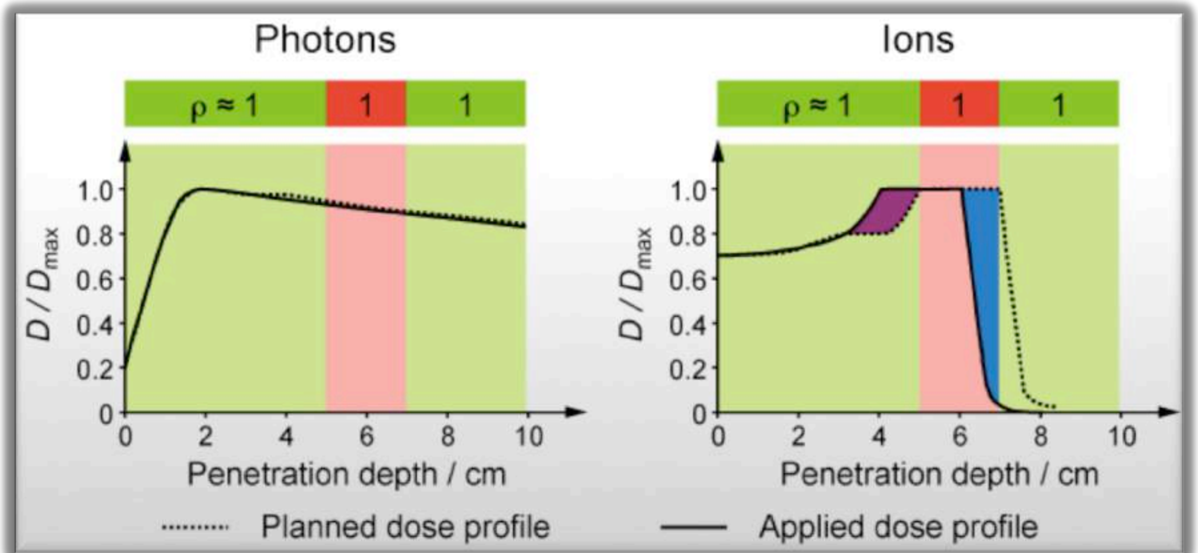
Rationale of MRiPT

2. Protons are **more sensitive to anatomical variations** than photons
   - **material composition** in beam path determines **Bragg peak location**



changed ion range will cause

- **overdose** in normal tissue
- **underdose** in tumour

- Because of these uncertainties, relatively **large margins** are still needed

Currently the dosimetric benefit of proton therapy is **not** fully exploited !!

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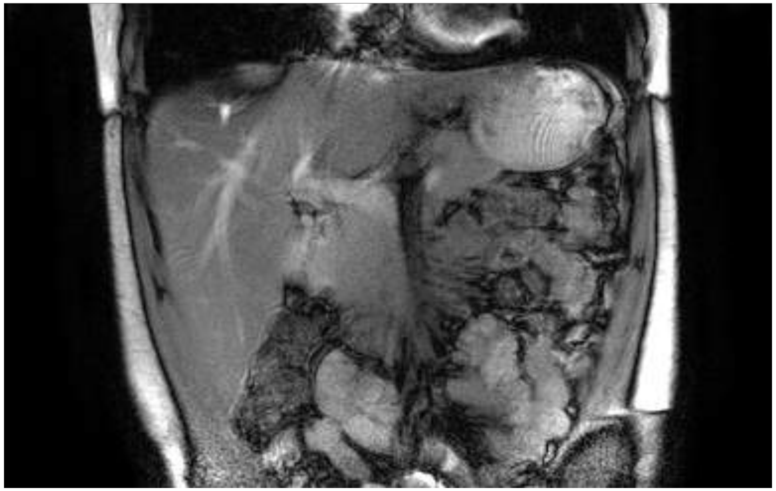
Motivation for MRI

- **MRI** offers
  - ✓ Fast real-time imaging
  - ✓ Superior soft tissue contrast
  - ✓ Freedom from radiation dose

- **Challenge**

Integration of MRI and PT for on-line image-guidance faces the challenge of their **mutual interaction**



2D-cine MRI scan showing intrafractional motion in the abdomen

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Technical challenges in MRiPT

Vision: integrate MR scanner at beam isocenter

R
I
S
K

- MR image degradation from gantry
- MR image degradation during irradiation
- Radiation harness of magnet
- Beam interaction with RF antenna
- Planning on MR images
- Range detection
- Dosimetry in the presence of a magnetic field
- **Beam deflection due to magnetic field of MR scanner**

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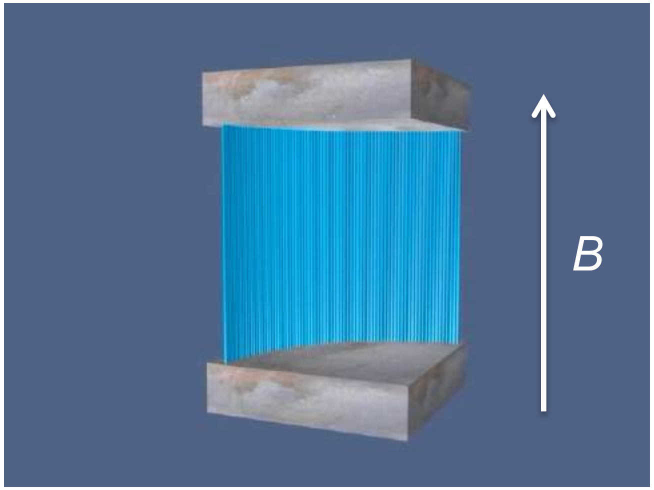
Challenges in MRiPT

Why are moving protons deflected in a magnetic field?

- protons are **charged particles**
- charged particles experience the **Lorentz force** in a magnetic field



Deflected beams have to be taken into account for **treatment planning**!

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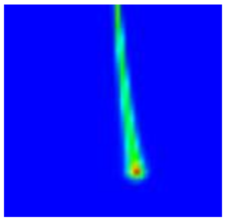
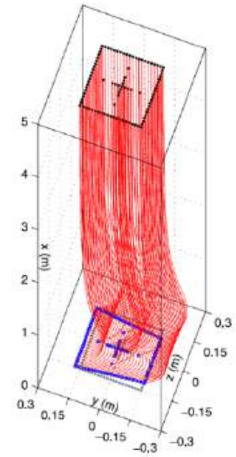
Developments in MRiPT

Beam deflection simulation studies:



- **2008 Dose to phantom in uniform $B_{\perp}$ field**
  Raaymakers *et al*., Phys. Med. Biol. 53(20), 2008
  Wolf & Bortfeld, Phys. Med. Biol. 57(17), 2012
- **2014 Dose to patient in uniform $B_{\perp}$ field**
  Moteabbed *et al*., Med. Phys. 41(11), 2014
  Hartman *et al*., Phys. Med. Biol. 60(11), 2015
- **2015 Dosimetric effects of MRI fringe field**
  Oborn *et al*., Med. Phys. 42(5), 2015



Experimental proof-of-principle:

- **2016 First „*in magnet*“ film dosimetry in slab phantom**
  Hoffmann *et al*., OncoRay (Dresden)

Goals

1. Develop fast and accurate **beam trajectory** prediction model
   - use to design experimental setup with real magnet and phantom
   - facilitate non-uniform *B* fields and inhomogeneous media
   - compare with existing analytical and numerical methods

2. Develop a Monte Carlo model for **full dose** simulations
   - quantify magnetic field induced **dose distortions**
   - estimate **demagnetization** and **radioactivation** effects

3. Realize measurement setup for ***„in magnet"*** **experiments**
   - show **dosimetric proof-of-principle** with proton pencil beams

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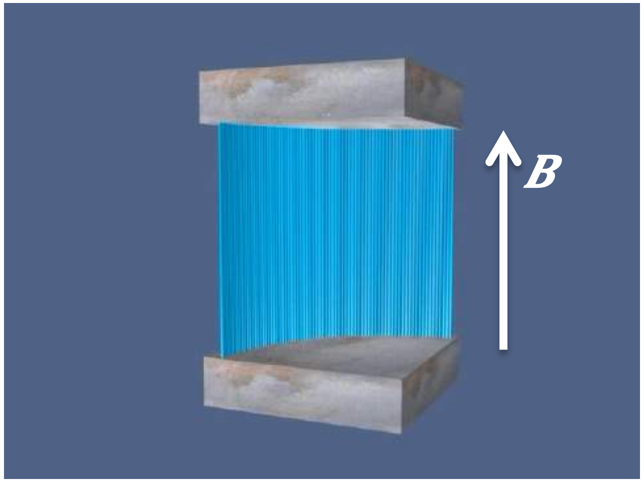
Fast beam trajectory prediction method

1. Moving proton is **deflected in magnetic field** through Lorentz force

$$\vec{F} = m\frac{\mathrm{d}\vec{v}}{\mathrm{d}t} = q(\vec{v} \times \vec{B})$$



2. Protons **loose energy** while interacting with matter
   - protons have a **finite range** (Bragg-Kleeman rule): $R_0 = \alpha E_0^p$

     least-square fit based on ICRU 49:

     $p = 1.75$, $\alpha = 2.43\times10^{-3}$ cm/MeV^p for $E_0 \leq 250$ MeV (Bortfeld, 1997)

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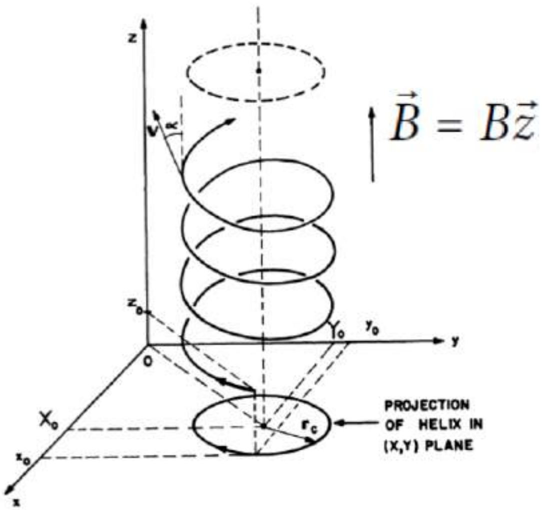
Fast beam trajectory prediction method

In vacuo: orbit of charged particle is **spiral trajectory** parallel to B field



Gyroradius can be expressed in terms of **kinetic energy**

$$r = \frac{\sin\theta}{qBc}\left[\left(E_k + m_o c^2\right)^2 - \left(m_o c^2\right)^2\right]^{\frac{1}{2}}$$

m_o rest mass of proton

c speed of light

θ angle between velocity vector $\vec{v}$ and magnetic field vector $\vec{B}$

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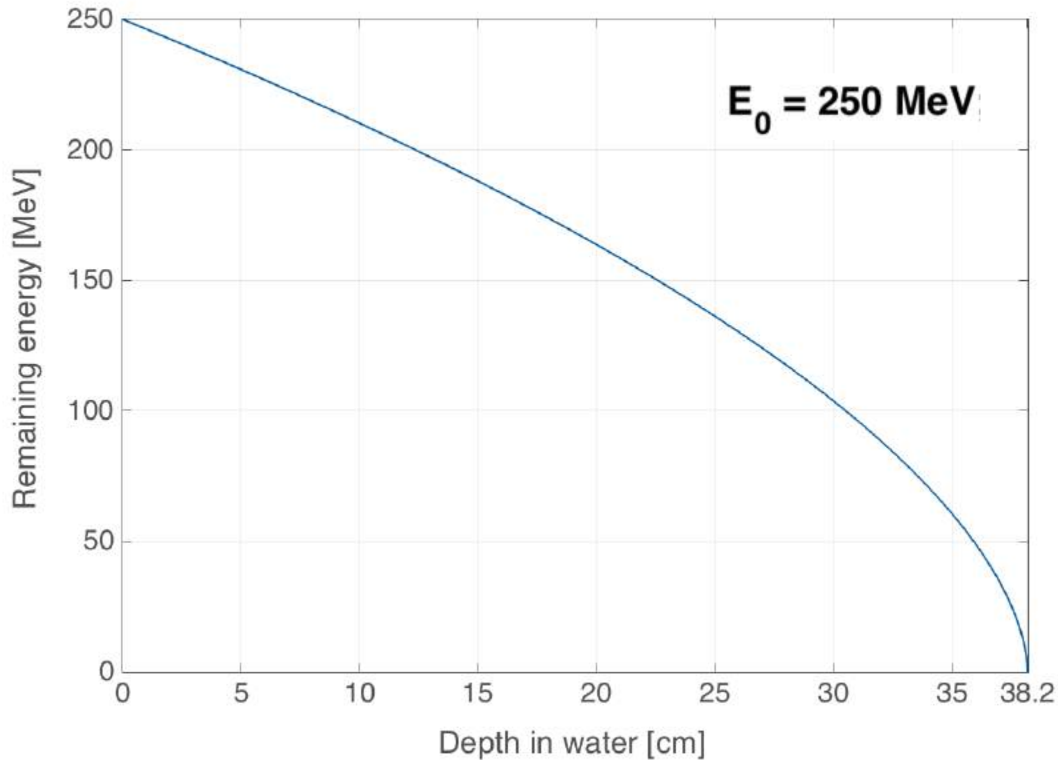
Fast beam trajectory prediction method

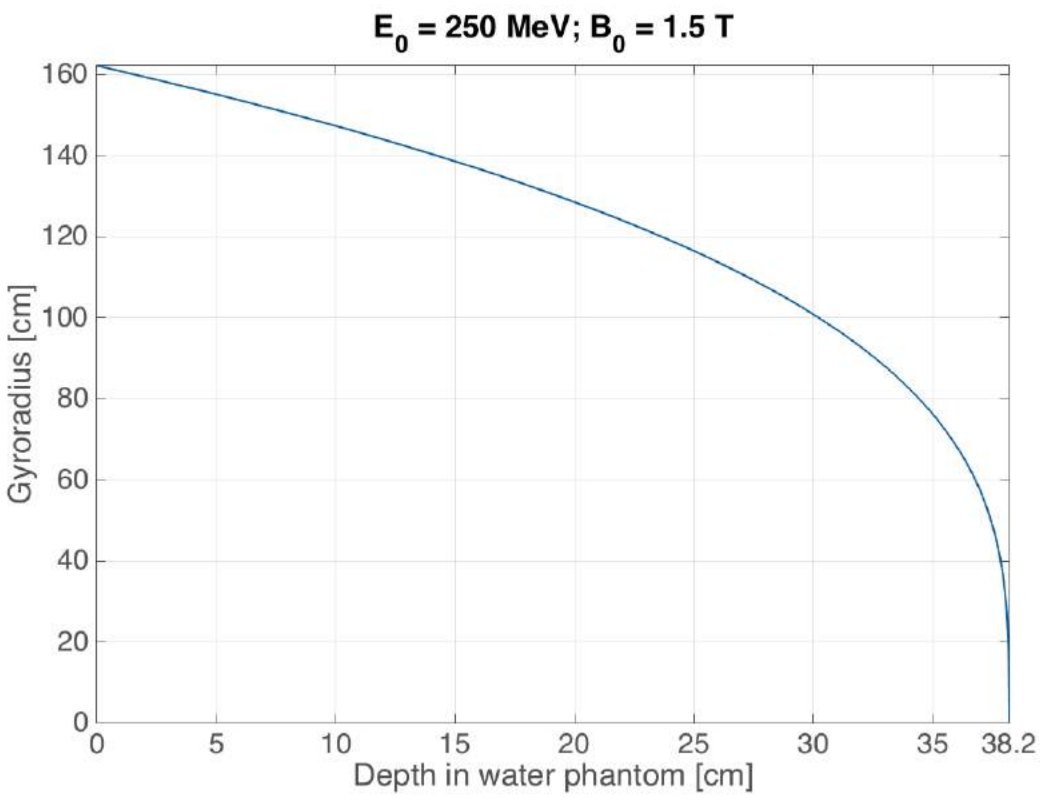
CSDA: **remaining energy** of proton as function of depth *s* in water:

$$E(s) = \alpha^{-1/p} (R_0 - s)^{1/p}$$



Fast beam trajectory prediction method

Subsitute **remaining energy** formula into the **gyroradius** formula to obtain the **gyroradius as function of depth**



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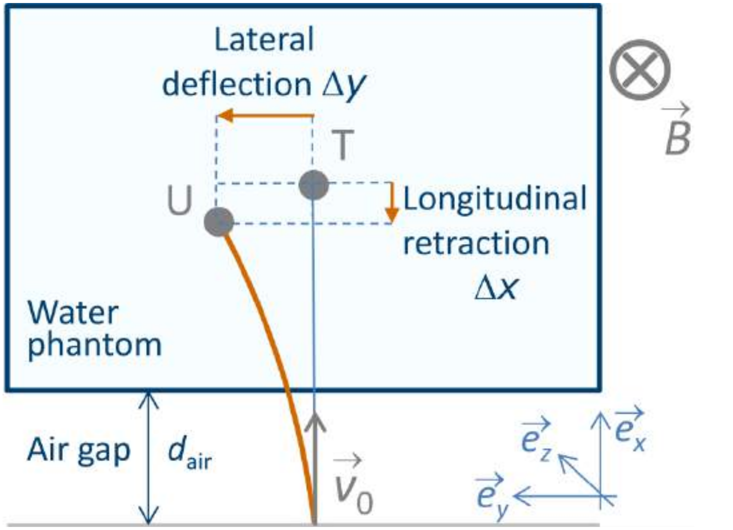
Fast beam trajectory prediction method

- **Iterative reconstruction** of proton beam trajectory in water
- **Discretization** in steps of constant **energy**
- **Radius of gyration** depends on energy:

$$r_i = \frac{mv_i}{eB_0} = \frac{\sqrt{2mE_i(1 + \frac{E_i}{2mc^2})}}{eB_0}$$

e: proton charge, m: rest mass, v: velocity,
B_0: magn. flux density, E: energy, c: lightspeed



T = intended Bragg peak spot
U = actual Bragg peak spot

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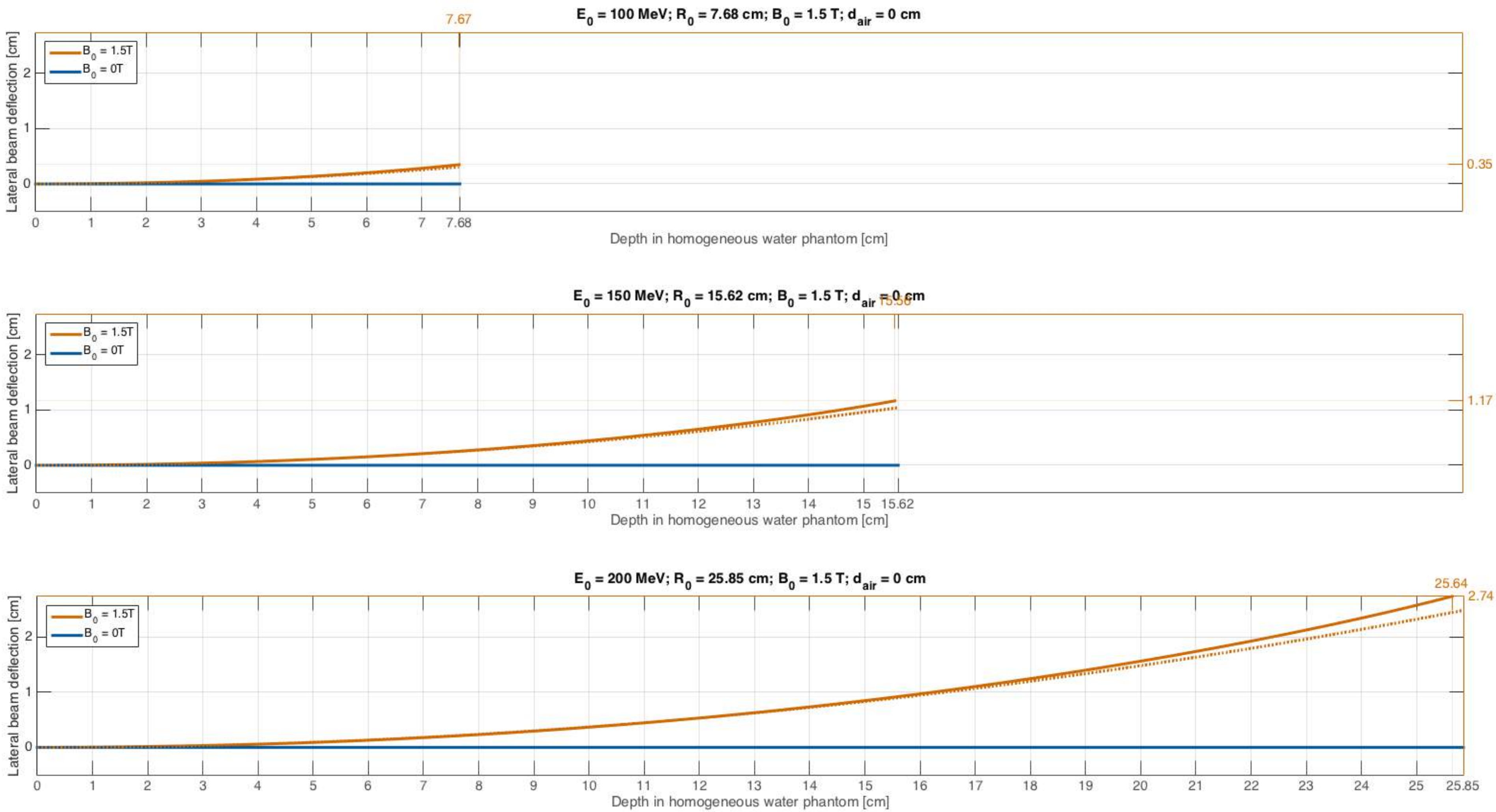
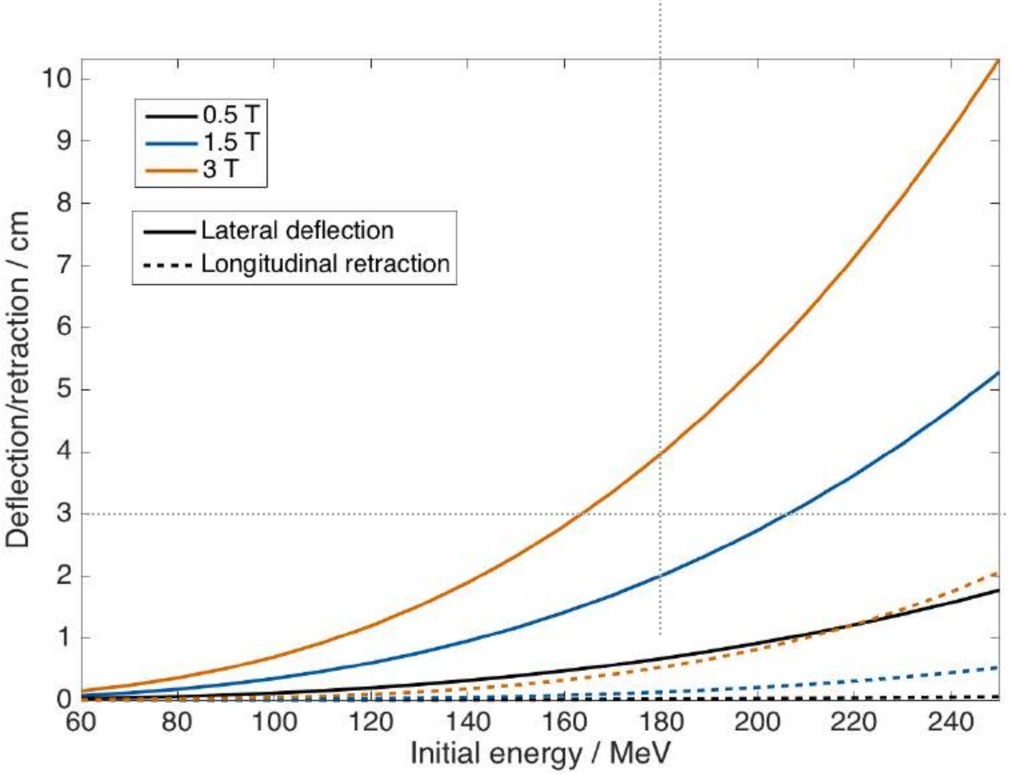
Fast beam trajectory prediction method



new model no energy loss

Beam trajectory prediction: *in water*



- **Lateral deflection** depends on B and E_0 (proportional to 3rd power)
- Lateral deflection dominates over **longitudinal retraction**
- **Relativistic corrections** are small, but non-negligible at higher E_0

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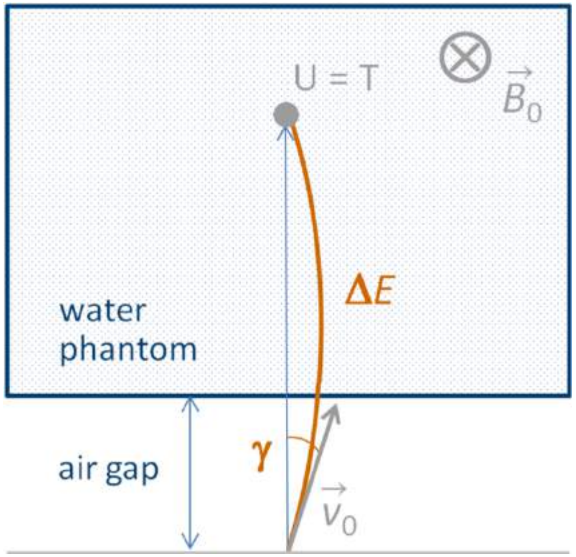
Can the deflection be corrected for?

- Altered **entrance angle** γ corrects for lateral deflection
- Altered **initial energy** ΔE corrects for longitudinal retraction
- Numerical **optimization** minimizes distance to intended Bragg peak position



B_0/T	E_0/MeV	ΔE/MeV	γ/°
0.5	60	0.054	3.55
	150	0.068	3.25
	250	0.123	3.97
1.5	60	0.500	10.67
	150	0.618	9.77
	250	1.122	11.98
3	60	2.034	21.46
	150	2.525	19.71
	250	4.730	24.37

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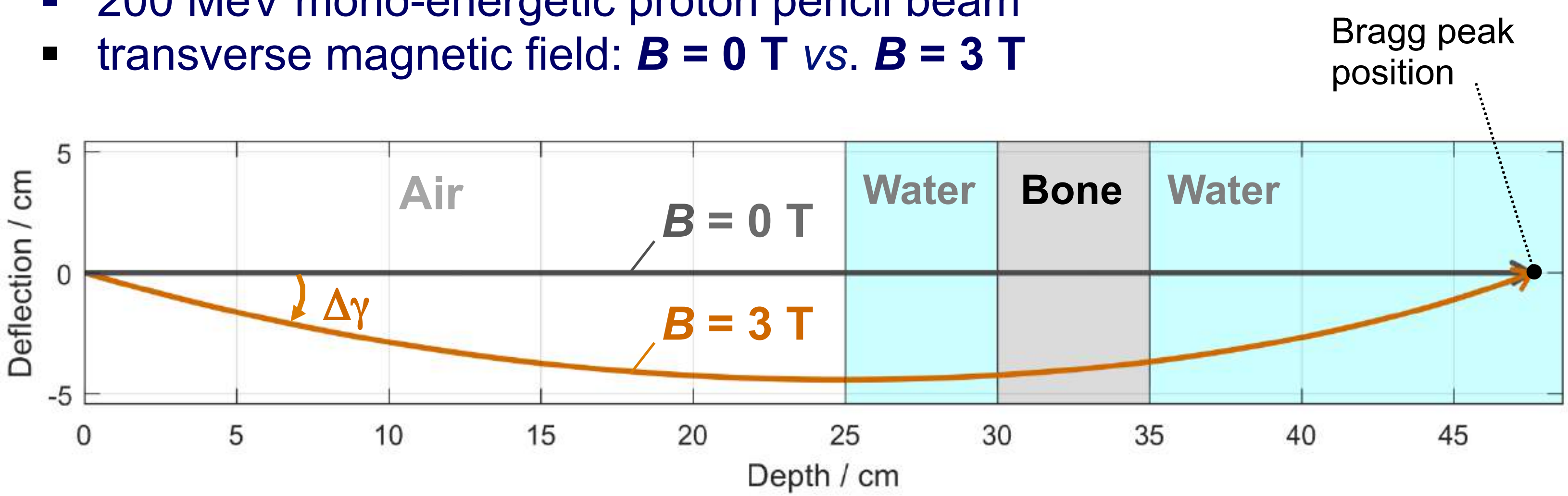
How about heterogeneous media?

Fast beam trajectory prediction in heterogeneous media

- 200 MeV mono-energetic proton pencil beam
- transverse magnetic field: ***B* = 0 T** *vs.* ***B* = 3 T**



Optimized beam correction parameters

- **angle** adjustment: $\Delta\gamma$ = 20.1 deg
- **energy** adjustment (due to increased pathlength): ΔE_0 = +3.23 MeV

Fast beam trajectory prediction method

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Phys. Med. Biol. 62 (2017) 1548–1564 doi:10.1088/1361-6560/62/4/1548

Prediction and compensation of magnetic beam deflection in MR-integrated proton therapy: a method optimized regarding accuracy, versatility and speed

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Beam trajectory prediction

Limitations of our current model:

1. range straggling effect due to proton scattering and nuclear reactions has been neglected
2. energy dispersion has not been included
3. no **realistic magnetic field** considered so far
4. no **experimental validation** performed so far

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Experimental validation at UPTD

- isochronous cyclotron (IBA)
- beam energy: 70−230 MeV
- passive scattering + active scanning
- first patient treated: 2014
- 15×20 m^2 experimental room
- static beam line
- intelligent beam switching system

Experimental room

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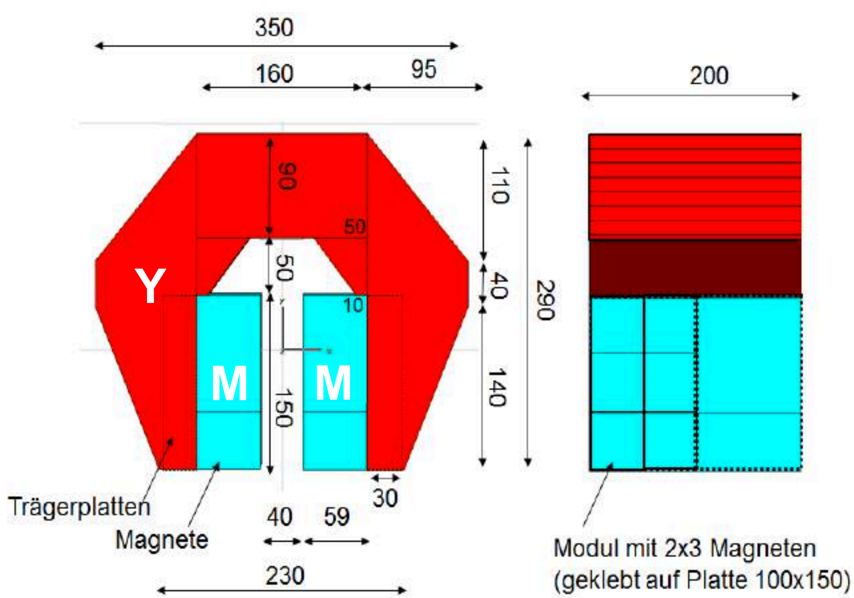
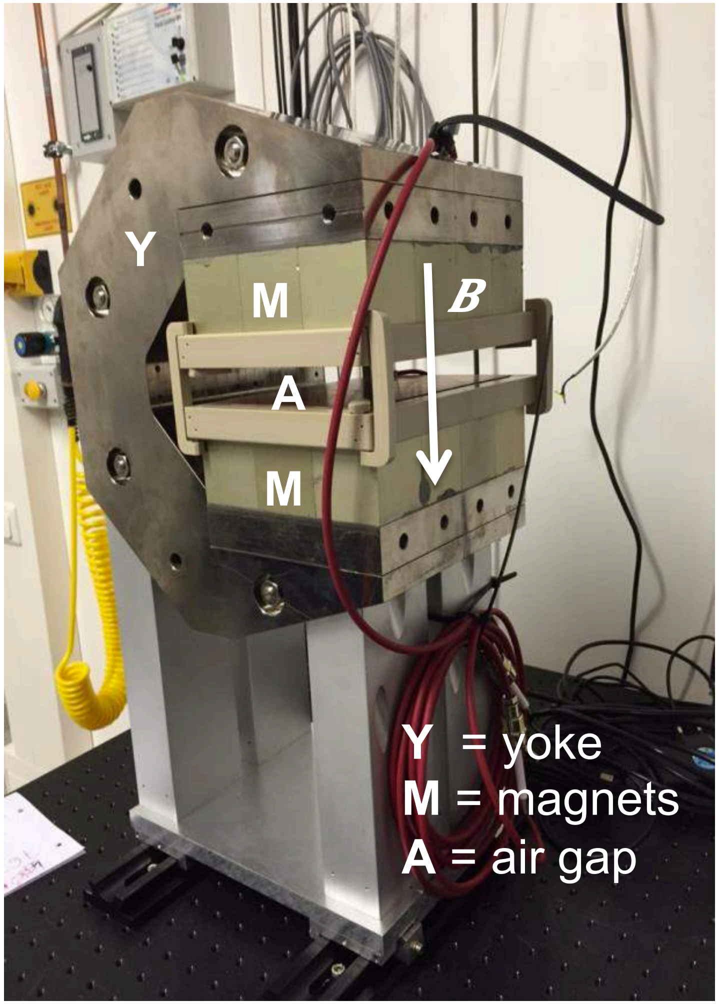
Experimental validation

Permanent $Nd_2Fe_{14}B$ dipole magnet

- 0.95 T
- 15 × 20 cm^2 field >0.5 T
- transverse magnetic field
- yoke: steel grade 1008
- magnets: NdFeB grade 764 TP





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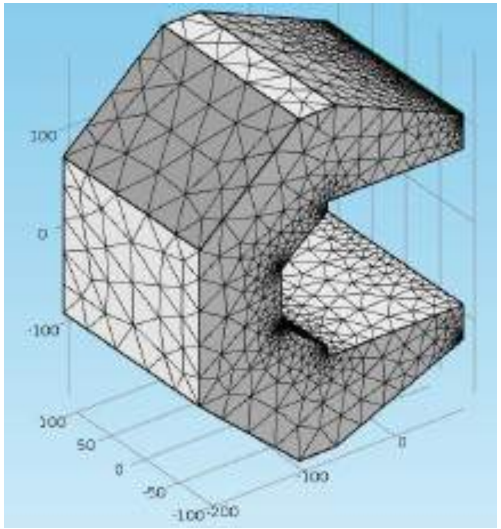
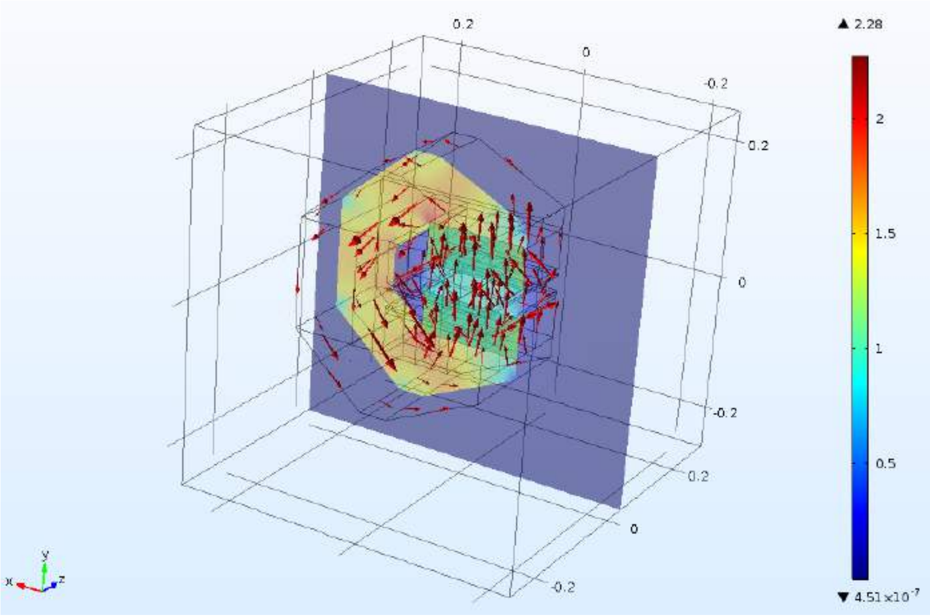
Magnetic field modelling

FEM simulations (COMSOL Multiphysics®)

- define **geometry** and **material properties** (μ_r, HB curves for iron, …)
- solve **stationary Maxwell equation** on a mesh under boundary conditions

$\vec{\nabla} \times \vec{H} = 0$ where $\vec{B} = \mu_r \mu_0 \vec{H} + \vec{B}_r$ and $\vec{B} = \vec{\nabla} \times \vec{A}$ $\qquad B_r = 1.37$ T ($Nd_2Fe_{14}B$)





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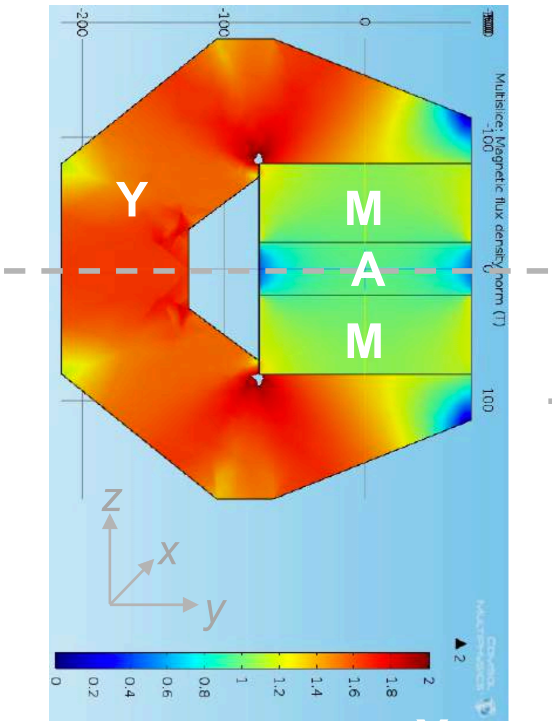
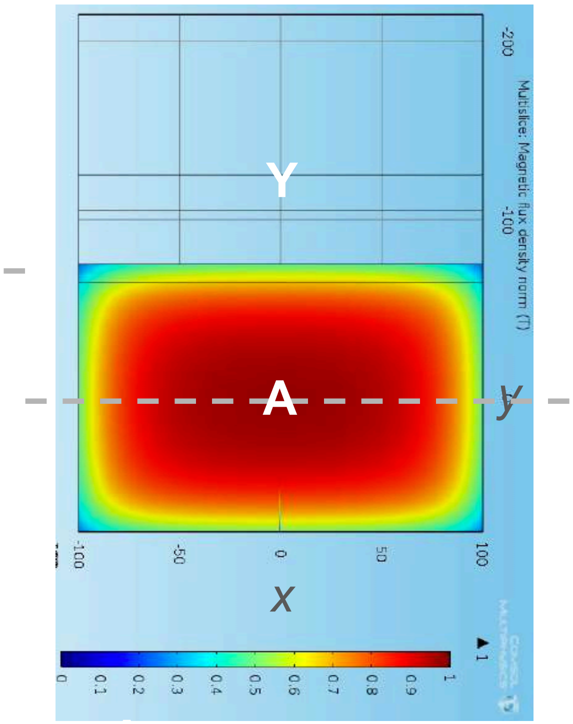
Magnetic field modelling

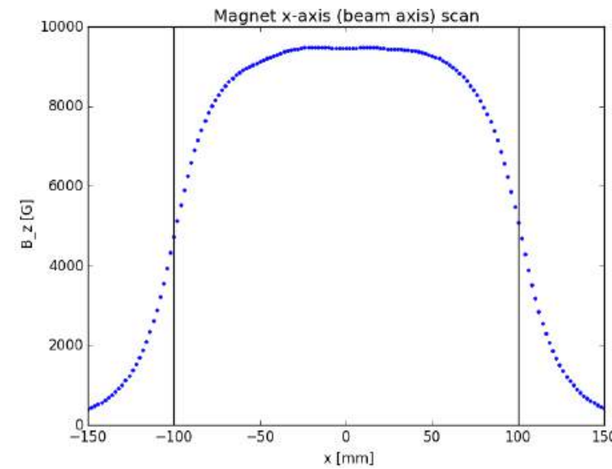
Result: 3D map of magnetic flux density



sagittal view



transversal view
through center of air gap



B_z profile along central x-axis

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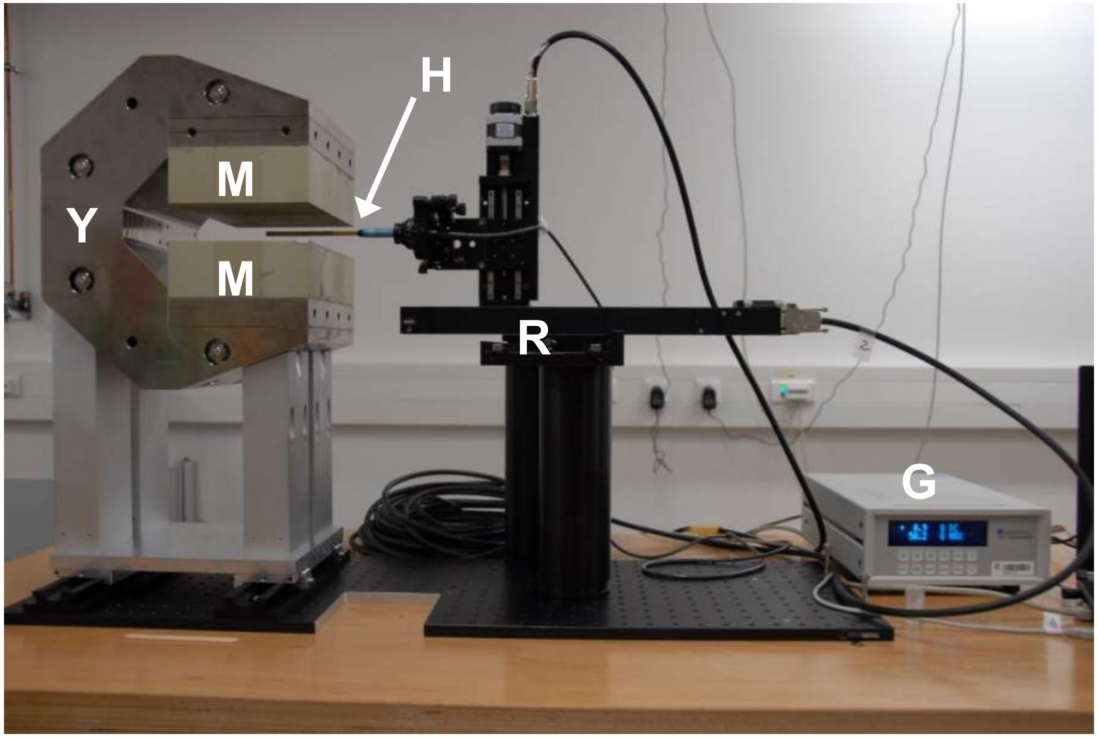
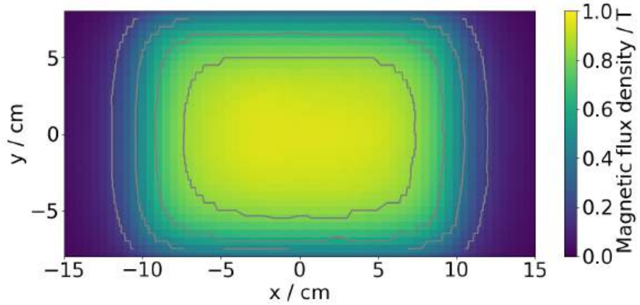
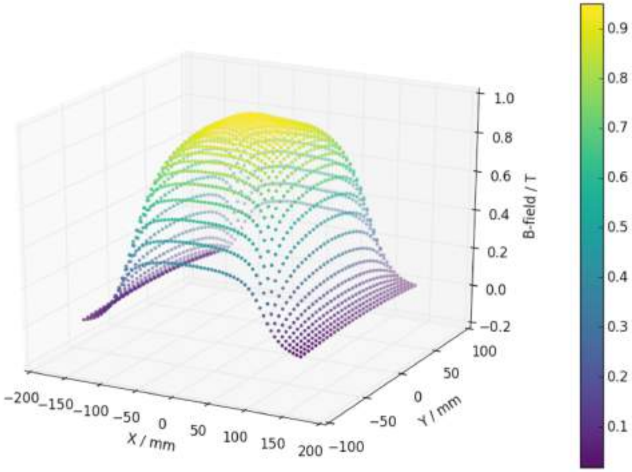
Magnetic field measurements

Automated magnetometry







Y = yoke
M = magnets

H = Hall probe
G = Gauss meter
R = 3D robotic positioner

B_z component

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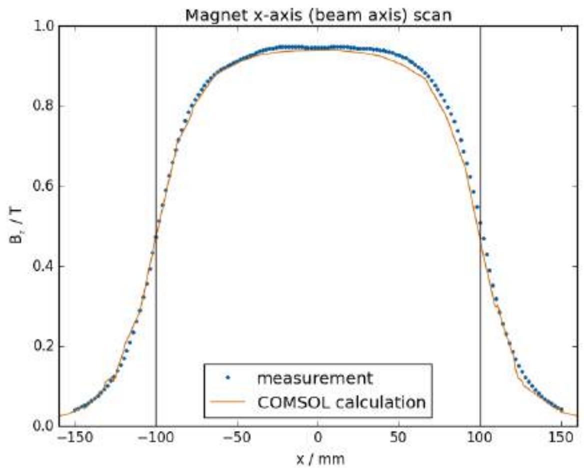
Magnetic modelling and measurements

Comparison of FEM simulations and magnetometry

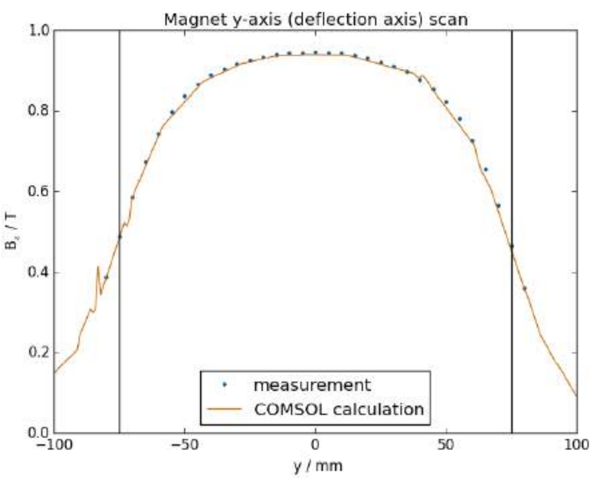
B_z on central x axis

- Max difference:
  - 40 mT (4%) in high gradient region
  - 23 mT (2.4%) in plateau region



B_z on central y axis

- Max difference:
  - 19 mT (2%) in high gradient region
  - 2 mT (0.2%) in plateau region



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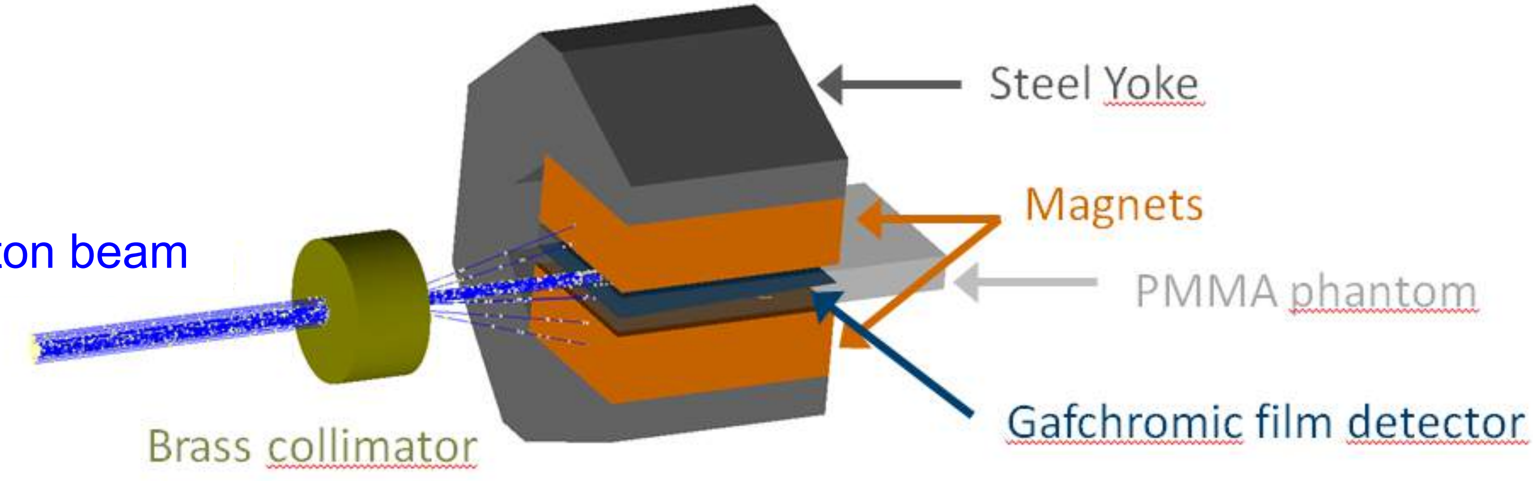
Monte Carlo simulation (Geant4)



- **Beam** : E_0 = 70−180 MeV, σ_r = 4 mm, d = 170 cm
- **Collimator** : ∅ = 5, 10 mm, r_{out} = 9 cm, l = 6.6 cm, d = 20 cm, brass
- **Phantom** : 30 × 15 × 3 cm^3
- **Film** : 20 cm × 15 cm × 28 µm, tilted by 1°
  Gafchromic® EBT3 material = polyester + LiPAD
- **Magnets** : magnetic field extension: 50 × 50 × 50 cm^3

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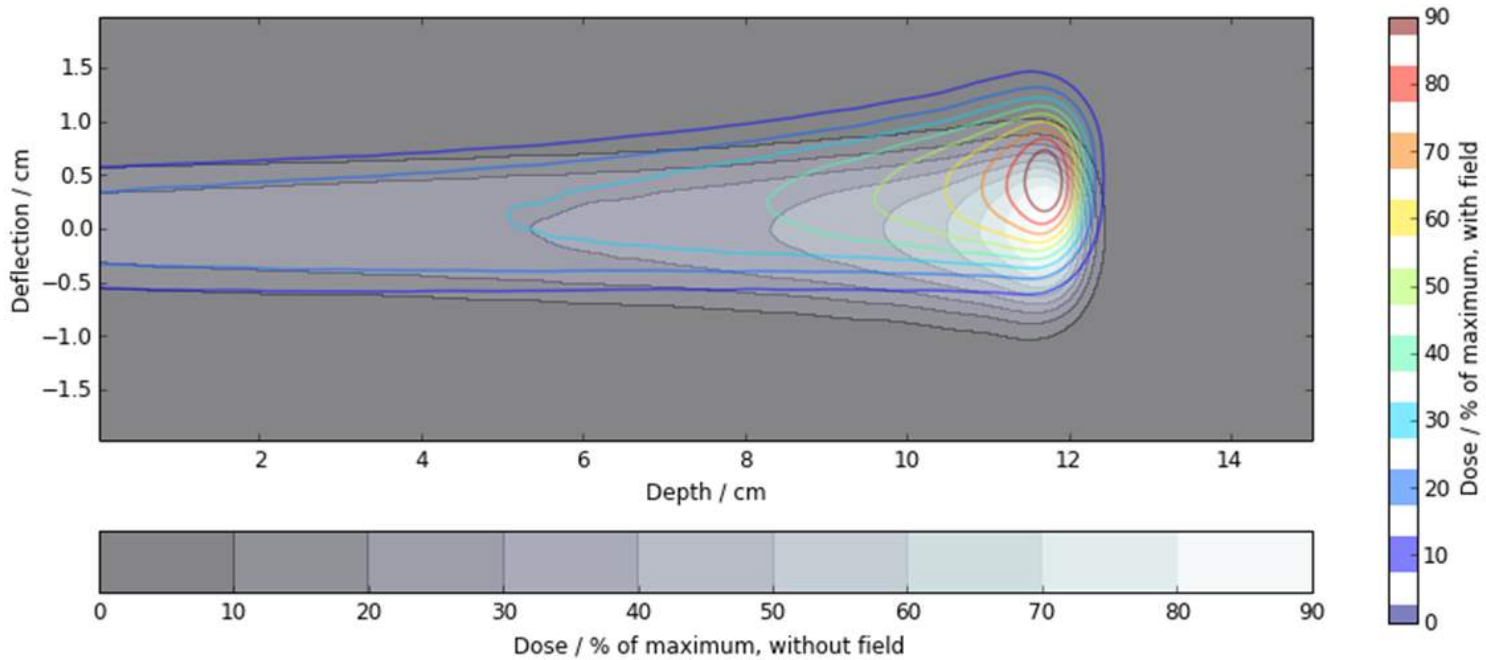
Results of MC simulation

- poly-energetic 140 MeV proton pencil beam (σ_E = 1 MeV)
- ∅10 mm collimated beam
- with and without magnetic field



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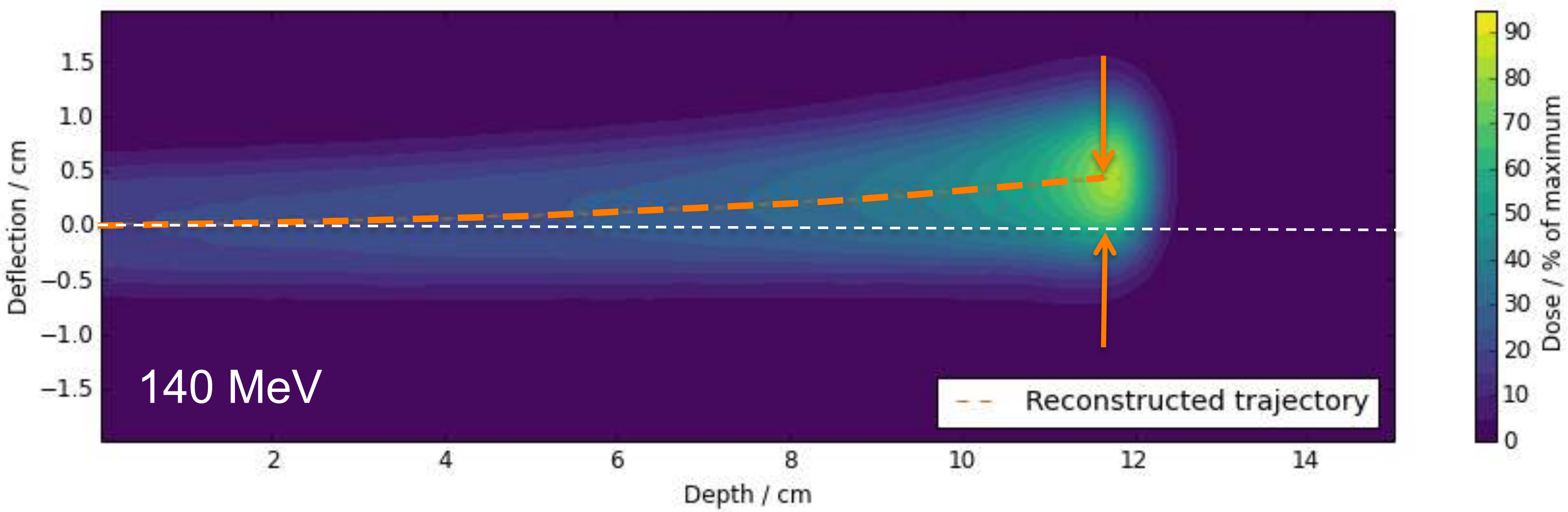
Results of MC simulations

- Reconstruction of central **beam path** by Gaussian fit of lateral profile



- Lateral beam deflection at Bragg peak: 5 mm
- **Conclusion**: these effects should be measurable with EBT3 film dosimetry

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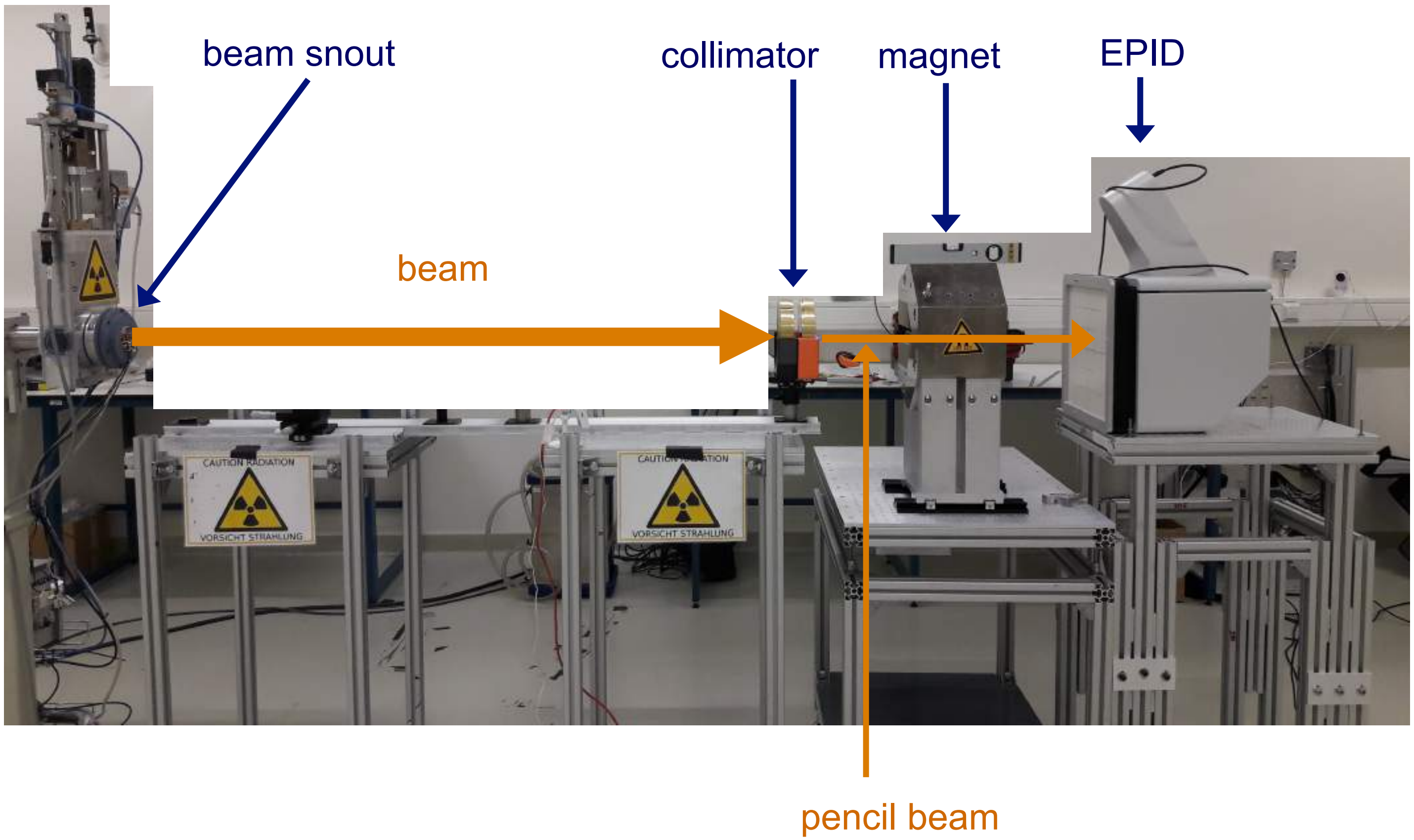
Transmission experiment

Measurement setup

- Purpose: measure **in-plane** and **out-of-plane** beam deflection



collimators
∅5 mm, ∅10 mm

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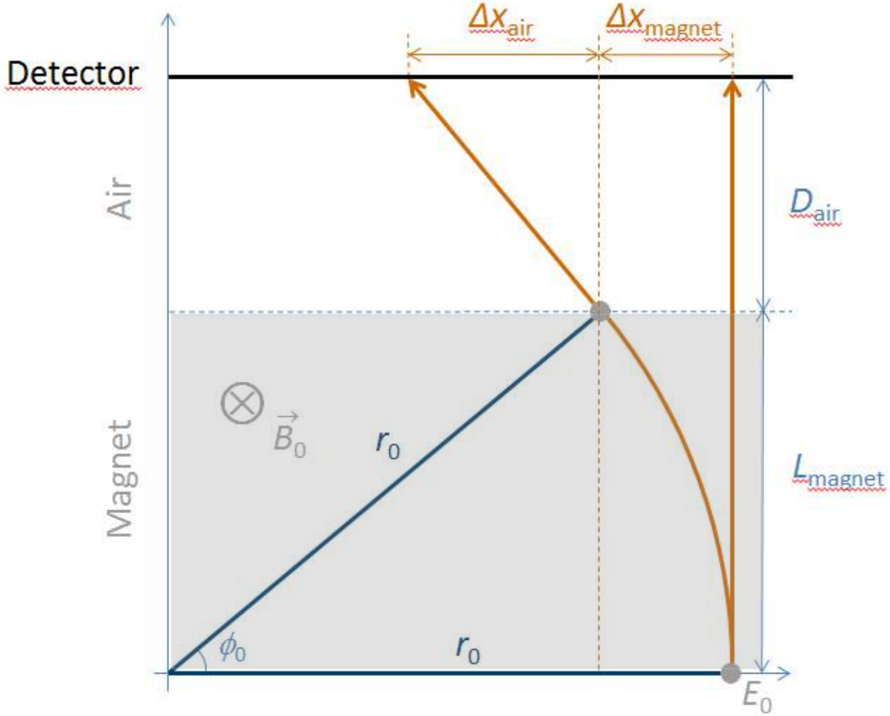
Transmission experiment

In-plane beam deflection



$$\Delta x = \Delta x_{magnet} + \Delta x_{air}$$
$$= r_o - \frac{r_o^2 - L_{magnet}^2 + D_{air} \cdot L_{magnet}}{\sqrt{r_o^2 - L_{magnet}^2}}$$
$$r_o \approx 14.4 \frac{\sqrt{E_o}}{B}$$

	Lateral beam deflection / mm		
E_0 / MeV	predicted	measured	difference
	D_{air} = 24 cm		
70	55.2	56.0	0.8
90	48.6	49.0	0.4
110	43.8	44.0	0.2
120	42.0	42.0	0.0
140	38.8	38.5	-0.3
160	36.3	36.0	-0.3
180	34.2	34.0	-0.2
200	32.4	32.0	-0.4
210	31.6	31.0	-0.6
225	30.5	30.0	-0.5
			0.4 RMSE

Out-of-plane beam deflection

- measured: <0.5 mm
- main component of *B* field is perpendicular to beam direction

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Measurement setup

Proton beam

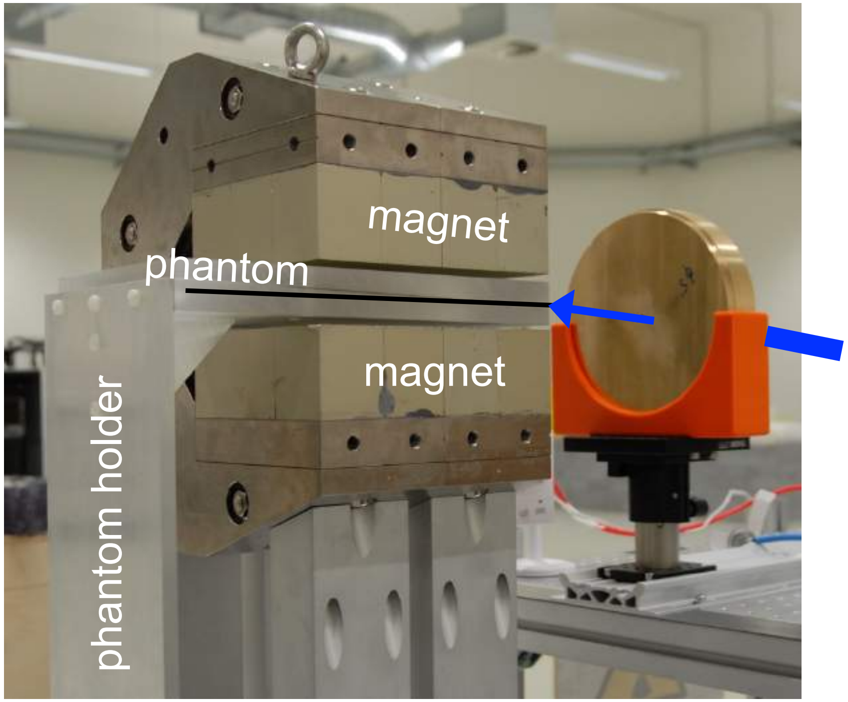
- Brass collimators with circular voids (∅10 mm)
- Pencil beams (blue arrow in figure)
- Energy: 80, 100, …, 180 MeV

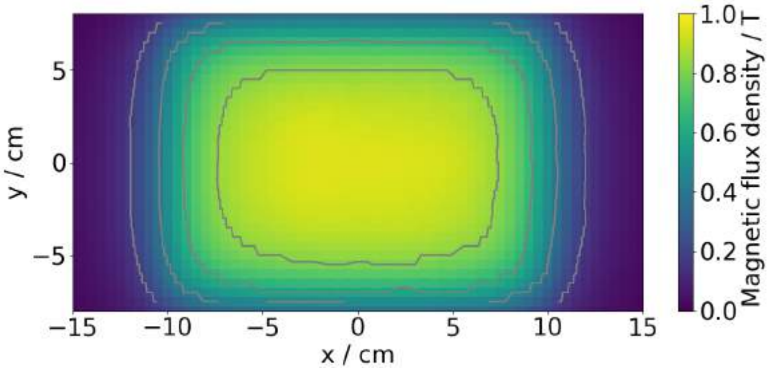
Tissue equivalent phantom

- 2 horizontal PMMA slabs
- placed between magnet poles
- 2D dose measurement with Gafchromic EBT3 film placed in central plane (1° inclination)

Magnetic field

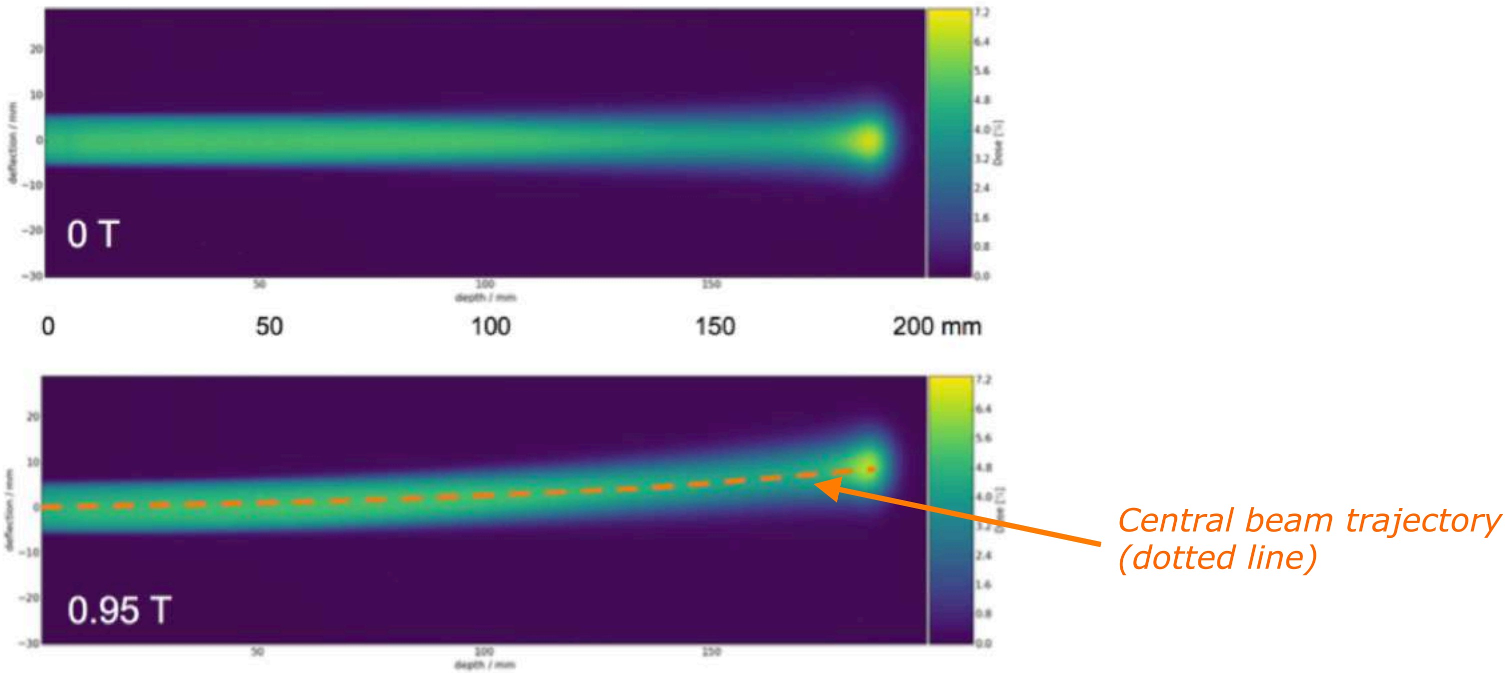
- C-shaped 0.95 T permanent $Nd_2Fe_{14}B$ dipole magnet (20 × 15 cm^2)
- 3D Hall probe magnetometry used to map out the main and fringe field





Irradiation experiment

- All irradiations were conducted **with** and **without magnetic field**
- Data without magnetic field serve as intrinsic reference
- **Depth-dose curves** reconstructed by radial integration of dose distributions
- Central **beam trajectory** estimated from fitting lateral profile with Gaussian



Planar dose distributions of 180 MeV proton pencil beam in PMMA with and without magnetic field

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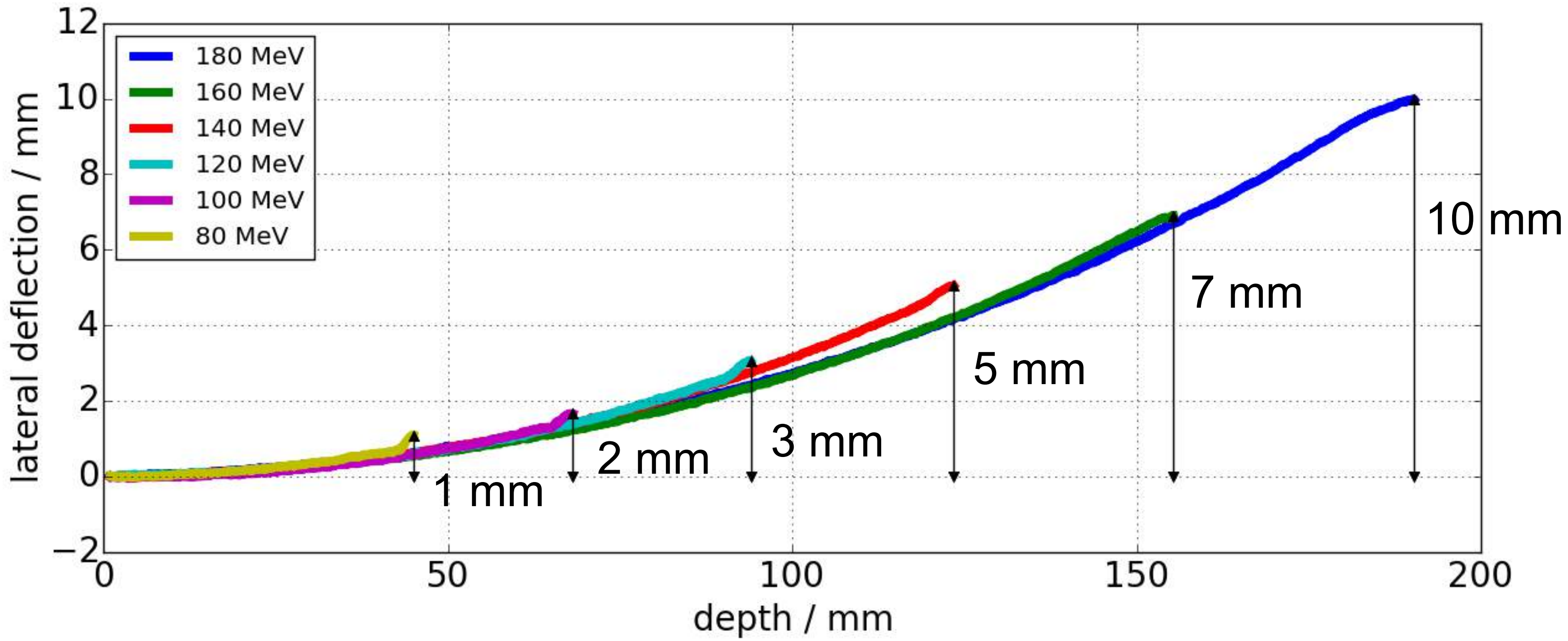
Film dosimetry results

Measured proton pencil beam trajectories in magnetic field



- Lateral beam deflection ranges from 1−10 mm for energies of 80−180 MeV

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Conclusions

1. **FEM model** accurately predicts the 3D magnetic field of our magnet measurement setup

2. **Monte Carlo model**
   - *B* field induced dose distortions are significant and measurable
   - deflected beam trajectory with high accuracy and precision
   - beam deflection can be compensated for during treatment planning

3. **„*In magnet* “ experiment**
   - a „***nortolcyc***“ was created with our 0.95 T magnet
   - **first dosimetric proof-of-principle** with proton pencil beams

4. Detailed **comparison of simulations and measurements** is work in progress

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Acknowledgements

OncoRay (Medical Radiation Physics)

Sonja Schellhammer

Sebastian Gantz

Armin Lühr

Wolfgang Enghardt

HZDR (Institute of Radiation Physics)

Michael Bussmann

Karl Zeil

University of Wollongong (Centre for Medical Radiation Physics), Australia

Bradley Oborn