Physics and Dosimetry of Hadron Therapy

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Ionizing Radiation of Interest for External Radiation Therapy

- Photons (γ, X ray)
  - Charge: 0
  - Indirect Ionization

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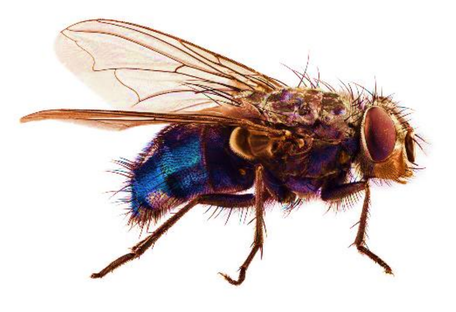
Ionizing Radiation of Interest for External Radiation Therapy

- Photons (γ, X ray)
  - Charge: 0
  - Indirect Ionization

- Electrons
  - Charge: -1
  - Direct Ionization
  - Mass: 0.512 MeV



Ionizing Radiation of Interest for External Radiation Therapy

- **Photons (γ, X ray)**
  - Charge: 0
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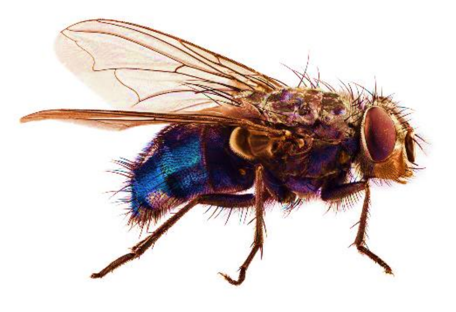
- **Electrons**
  - Charge: -1
  - Direct Ionization
  - Mass: 0.512 MeV



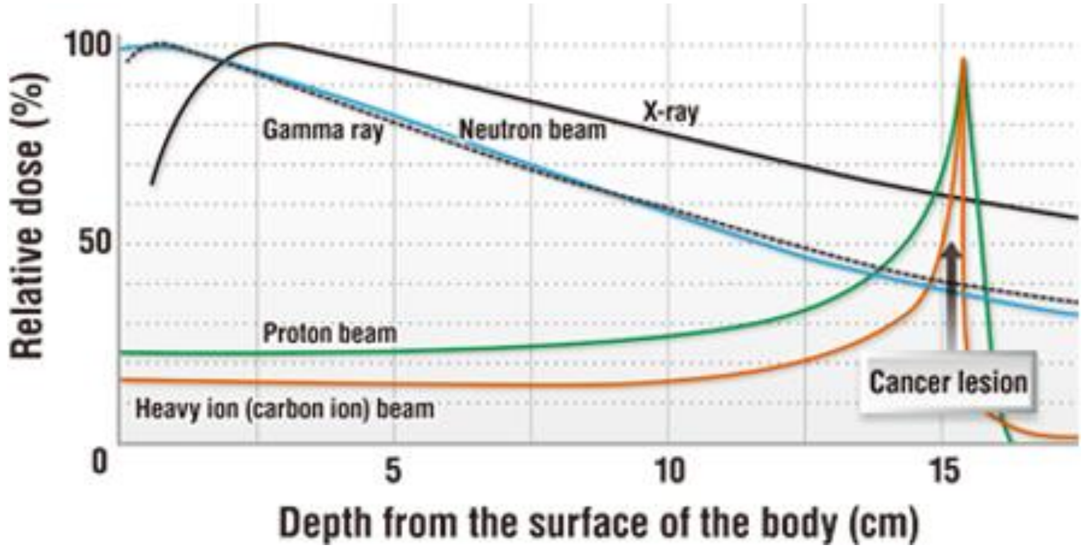
- **Pions$^-$ (π^-)**
  - Charge: -1
  - Unstable
  - The pion is absorbed by the nucleus, and the π^- rest mass of 140 MeV appears in the form of kinetic energy of nuclear fragments, except for about 40 MeV, which is used in overcoming the binding energy of the nucleus.
  - Mass: 138 MeV (273 x m_e)

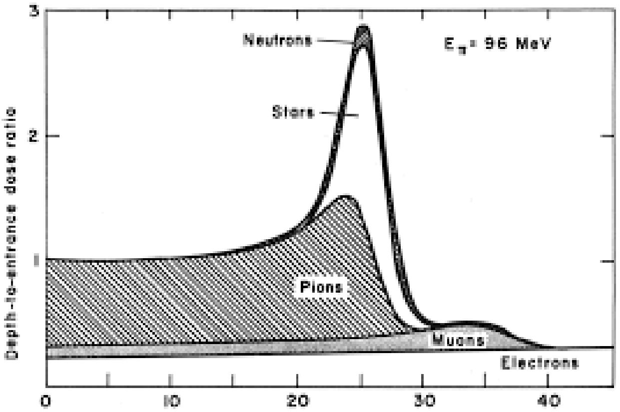
Particles of Interest for External Radiation Therapy

- Protons
  - Charge: +1
  - Direct Ionization
  - Mass ~ 938 MeV (2,000 x m_e)

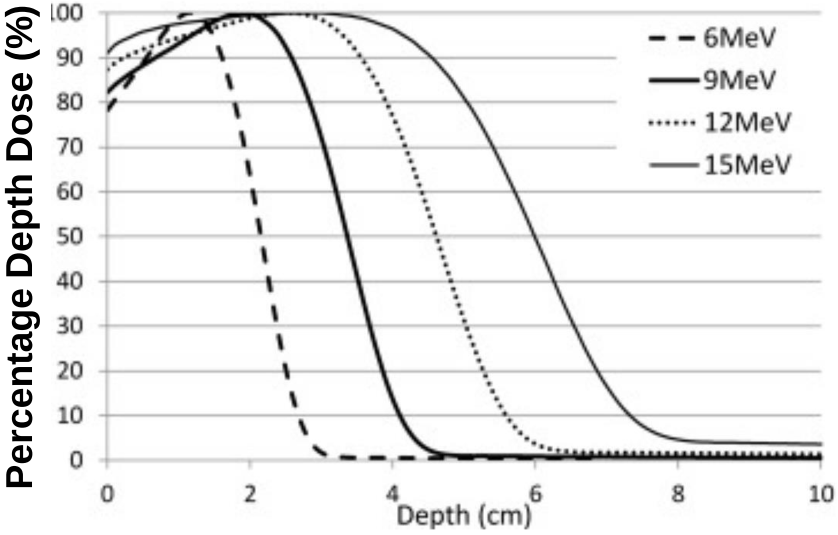
- Carbon ions
  - Charge: +6
  - Direct Ionization
  - Mass ~ 12 x m_p

Different Deth-Dose Distributions





$$\begin{aligned}\pi^- + {}^{16}O &\rightarrow {}^{12}C + p + 3n,\\ &\rightarrow 3\alpha + p + 3n,\\ &\rightarrow {}^{14}N + 2n,\\ &\rightarrow {}^{10}B + \alpha + 2n.\end{aligned}$$



Ionizing Radiation of Interest for External Radiation Therapy

- Neutrons
  - Charge: 0
  - Indirect Ionization
  - Mass ~ 938 MeV (2,000 x m_e)

- Protons
  - Charge: +1
  - Direct Ionization
  - Mass ~ 938 MeV (2,000 x m_e)

HADRONS

Subatomic particles made of two or more quarks held together by the strong interaction

- Carbon ior
  - Charge: +6
  - Direct Ionization
  - Mass ~ 12 x m_p

Particles of Interest for External Radiation Therapy

- Protons
  - Charge: +1
  - Direct Ionization
  - Mass ~ 938 MeV (2,000 x m_e)

- Carbon ions
  - Charge: +6
  - Direct Ionization
  - Mass ~ 12 x m_p

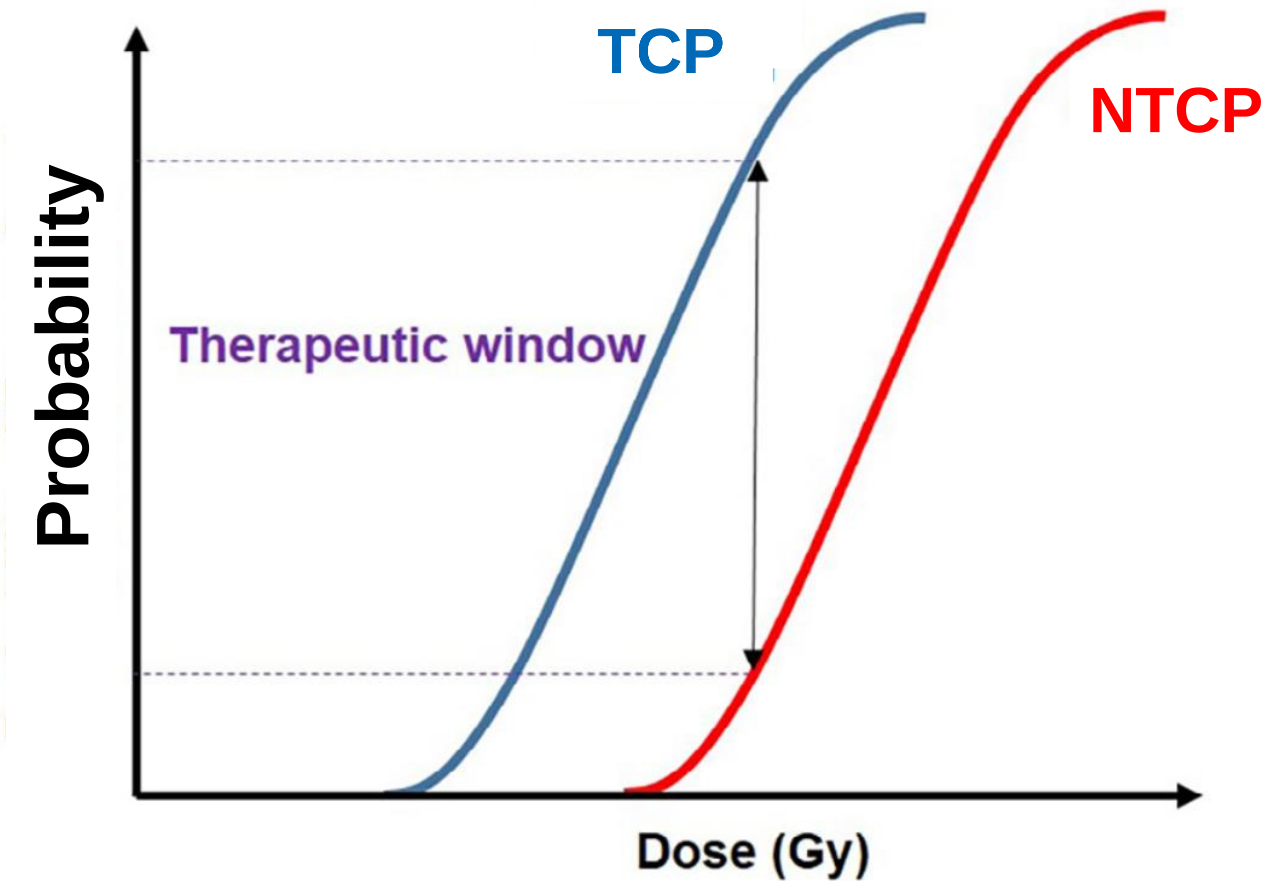
Dosimetry

Dosimetry is the determination of absorbed dose in matter or tissue resulting from exposure to ionizing

$$D = \frac{dE}{dm} \left[\frac{\mathrm{J}}{\mathrm{kg}} = \mathrm{Gy}\right]$$

Why do we care about the dosimetry?

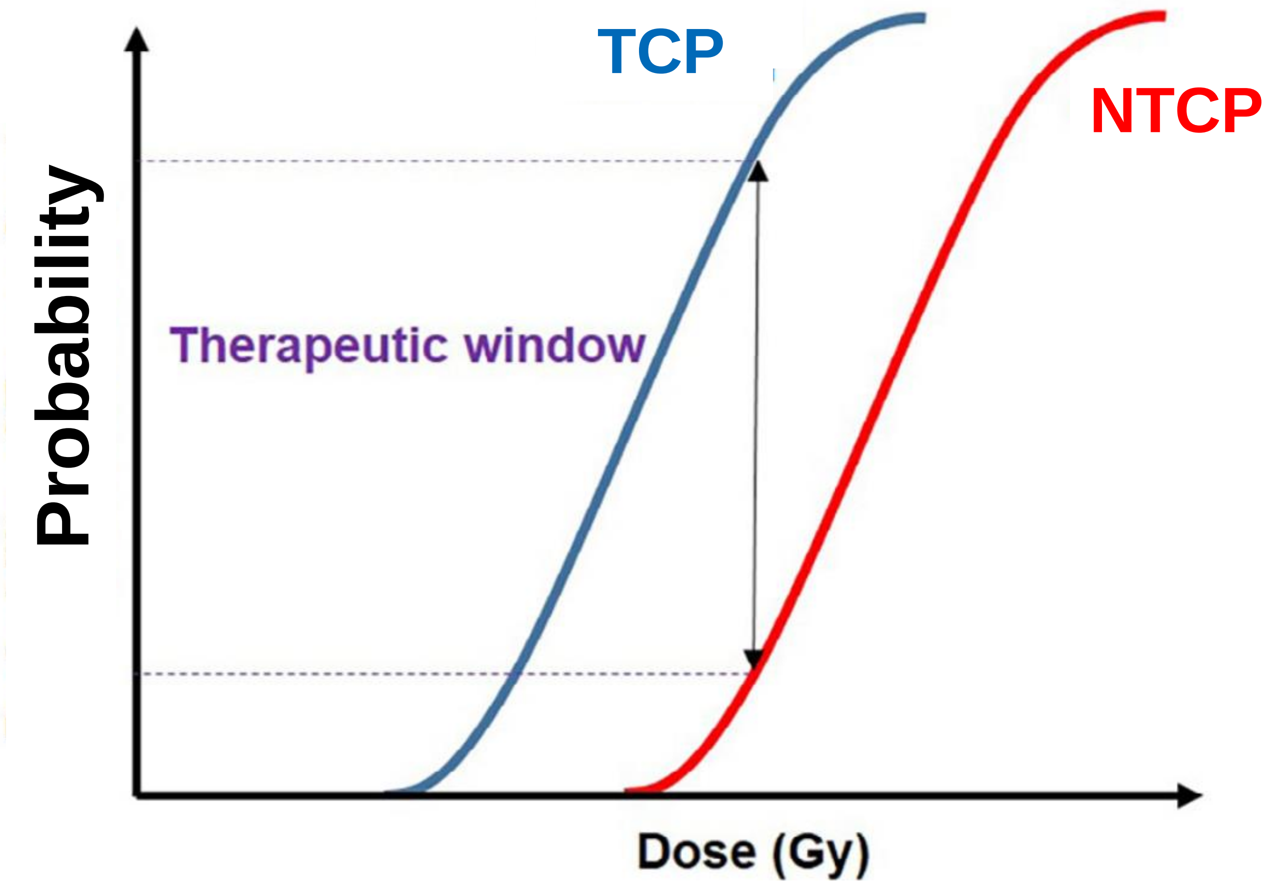
The Need for Dosimetry



TCP = Tumor Control Probability

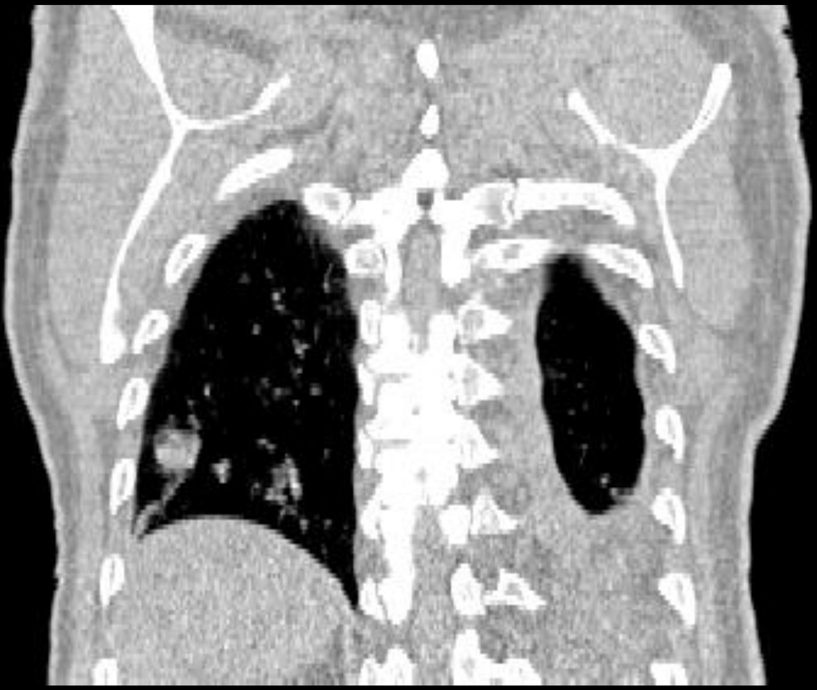
NTCP = Normal Tissue Complication Probability

The Need for Dosimetry



Due to the steep slopes of TCP and NTCP, a 5% dose error can lead to a TCP change of 10-20% and even more for NTCP!!

The movements and changes in the body makes it especially challenging!

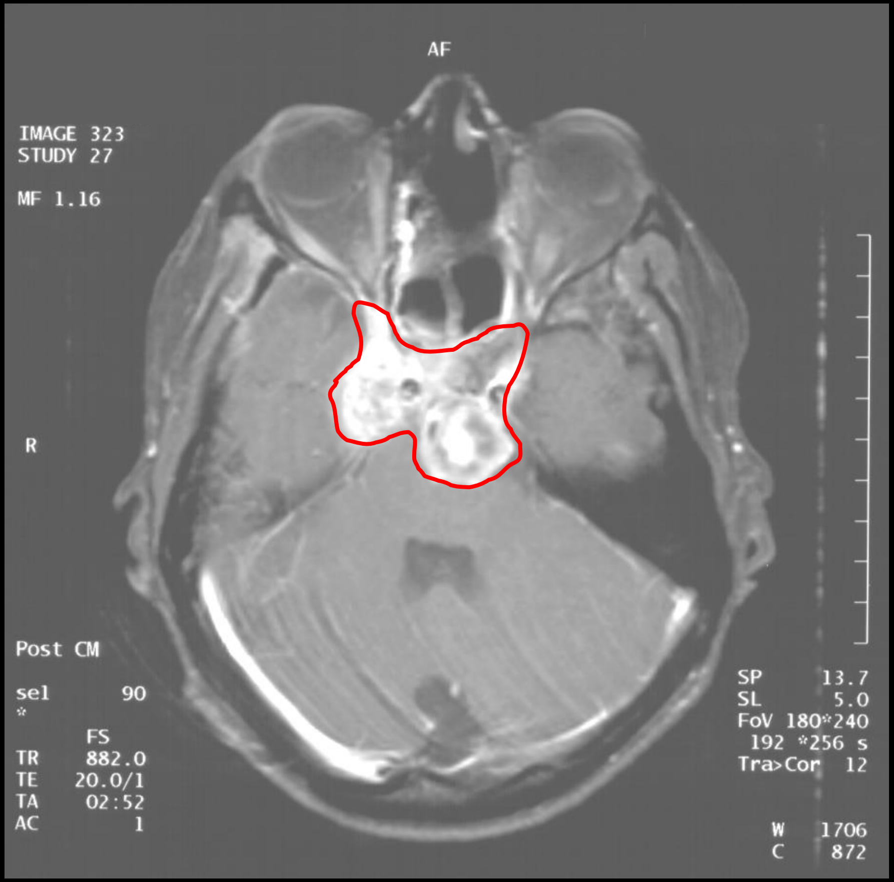


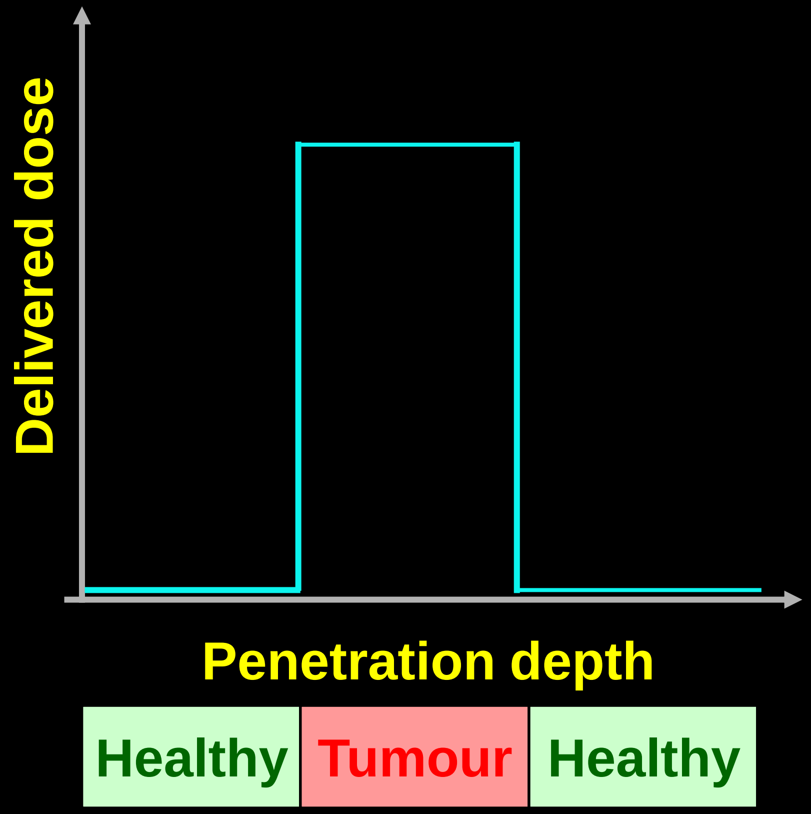
Before we will go into the dosimetry and its challenges in hadron therapy, let’s look at the basic physics of hadron therapy

Physical Advantages of Ion Beams

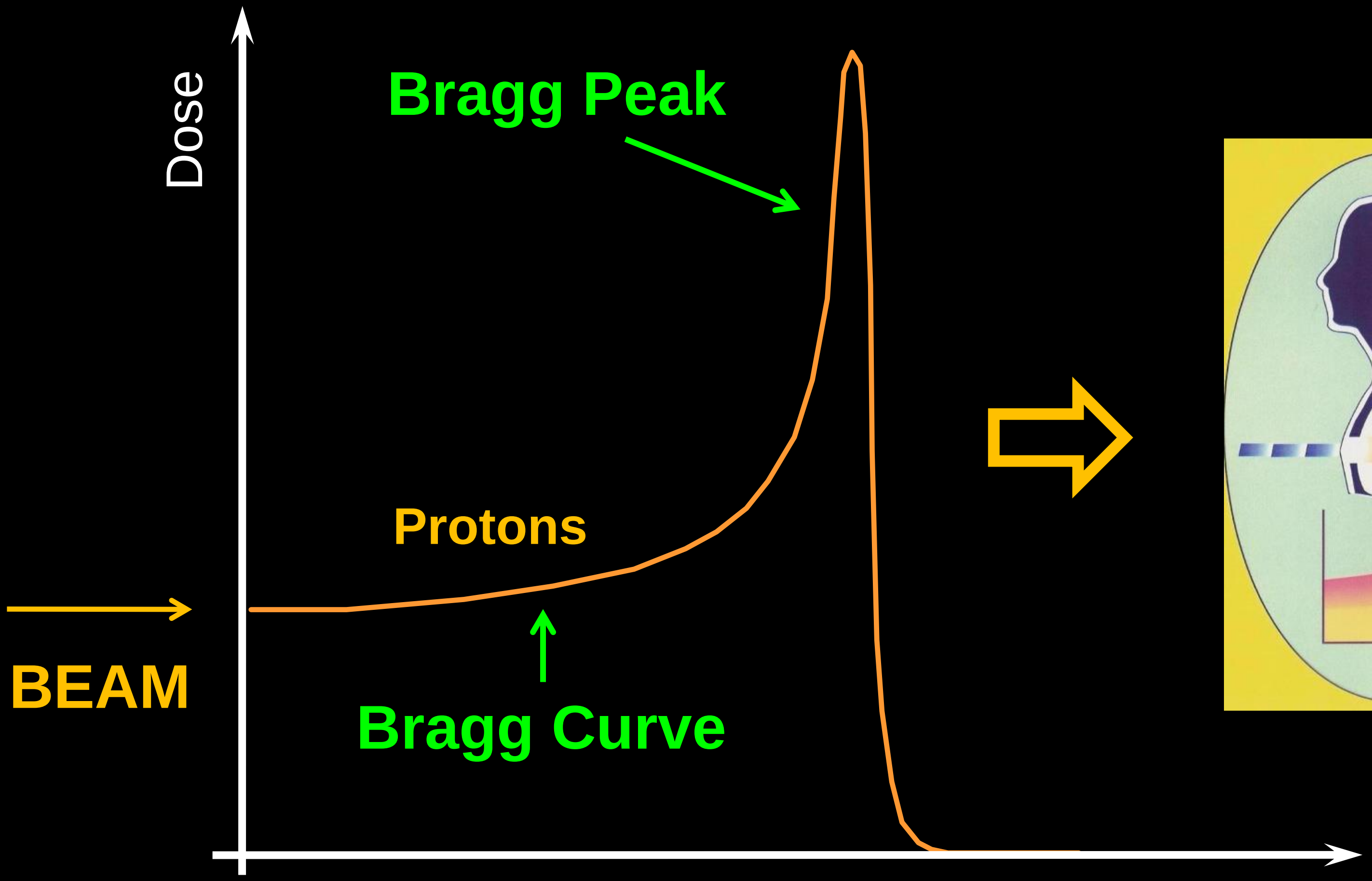
The request of the radiooncologist

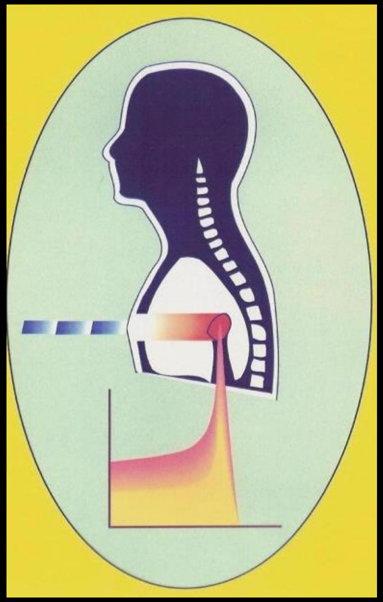
- destroy the tumour tissue
- spare the healthy tissue





Why use Proton Beams for Radiotherapy ?

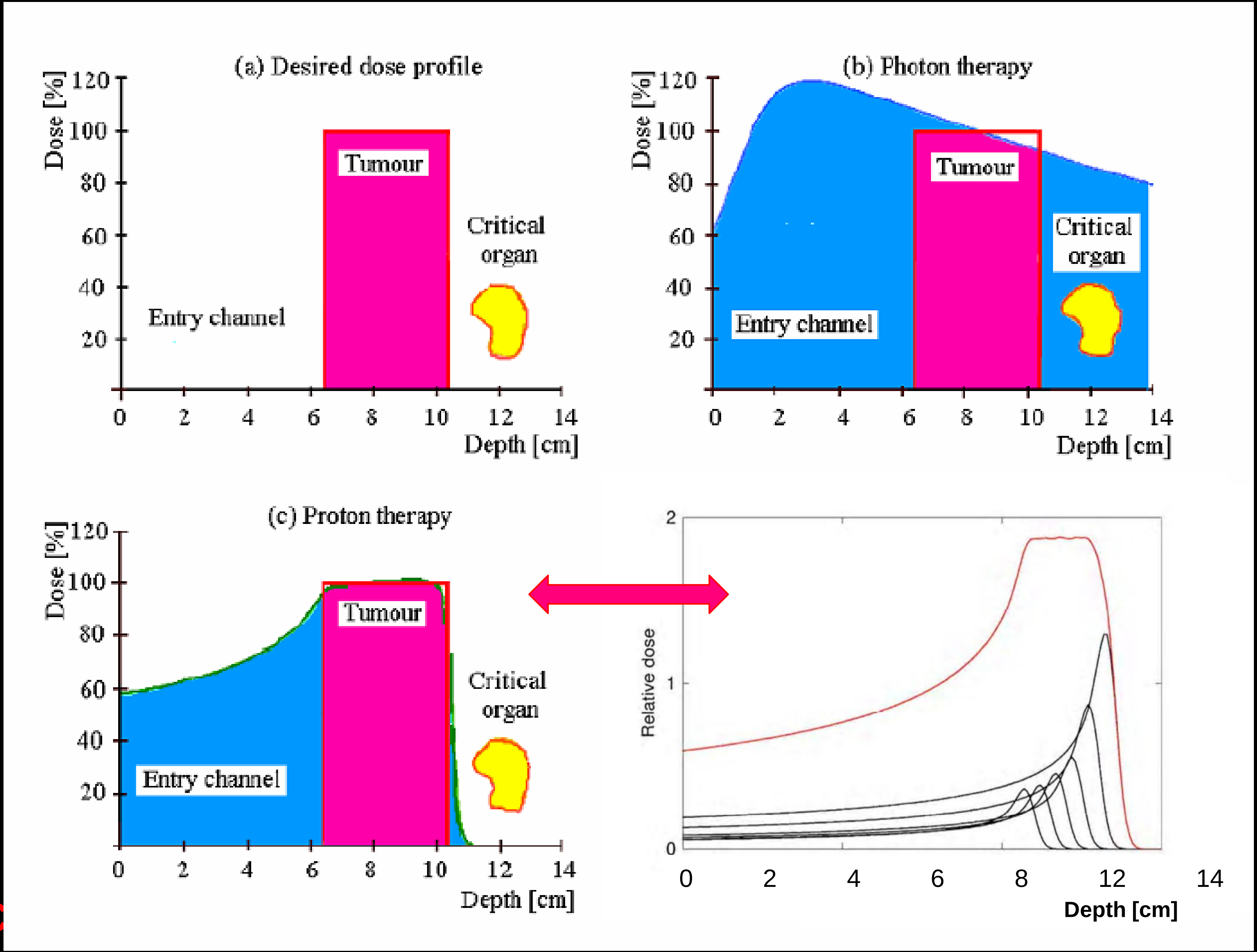




Good dose localization

(depth-dose distributions)

Physical Advantages of Ion Beams



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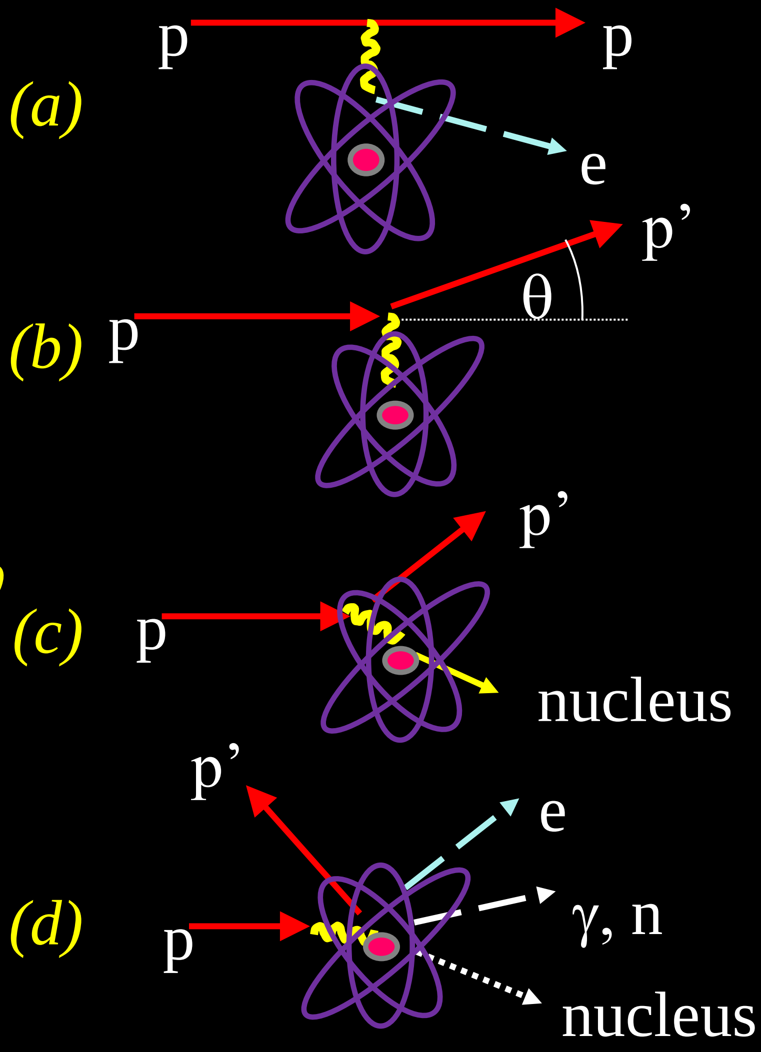
Proton Interactions with Matter

- *With electrons* mediated by Coulomb force (*a*)
  - *Excitation*
  - *Ionization*
- *With nucleus* mediated by Coulomb & nuclear forces (*b-d*)
  - *Multiple Coulomb scattering (b), small θ*
  - *Elastic nuclear collision (c), large θ*
  - *Inelastic nuclear interaction (d)*

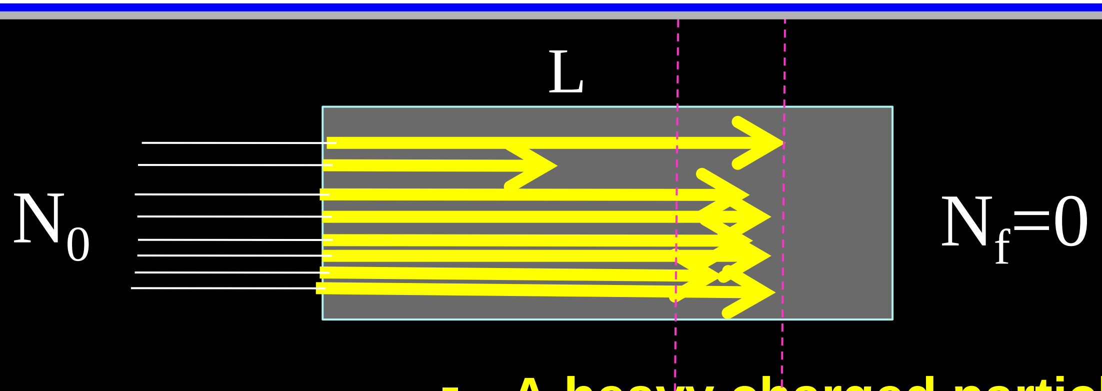
Mean electron energy E_{mean} very low (m_p>> m_e)

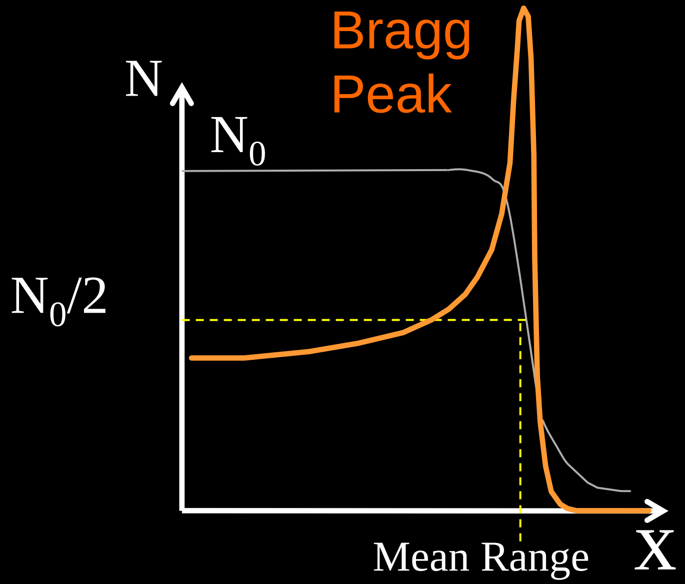
E_{mean} independent of proton kinetic energy

Interaction probability higher for slower protons



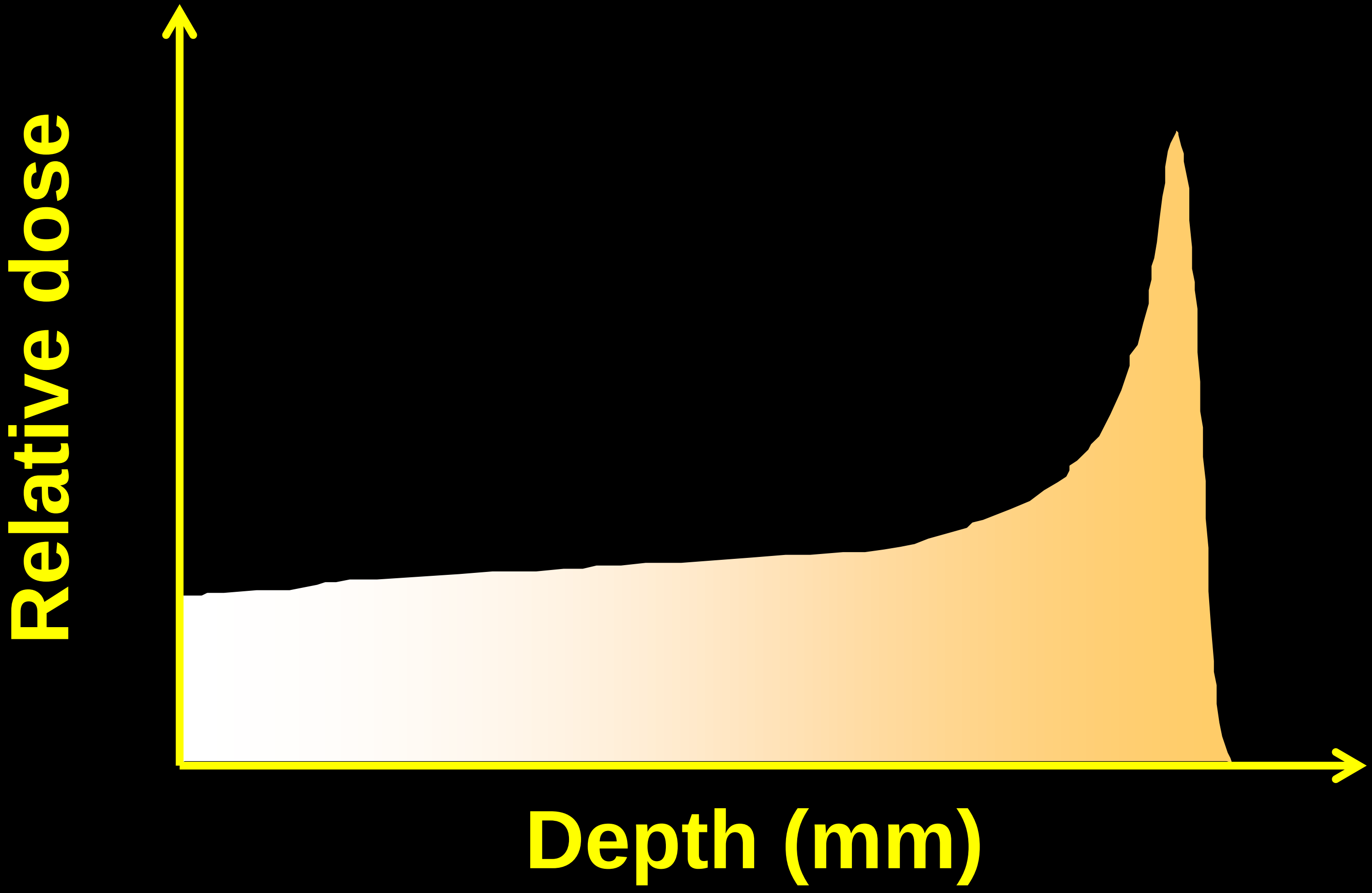
Particle Beams – Nearly No Attenuation





- A heavy charged particle endures multiple interactions through matter, but “stays” in the beam, because it is deflected only slightly.
- It loses only a small fraction of its energy in each interaction (except in “rare” nuclear interactions) until it stops, i.e., continuous slowing down.
- It deposits most energy near the end!!

How do we calc. the 1D depth-dose distribution (“Bragg Curve”)



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Linear Stopping Power (S)

$$S = -\left[(dE/dx)_{el} + (dE/dx)_{nuc} + (dE/dx)_{rad}\right]$$

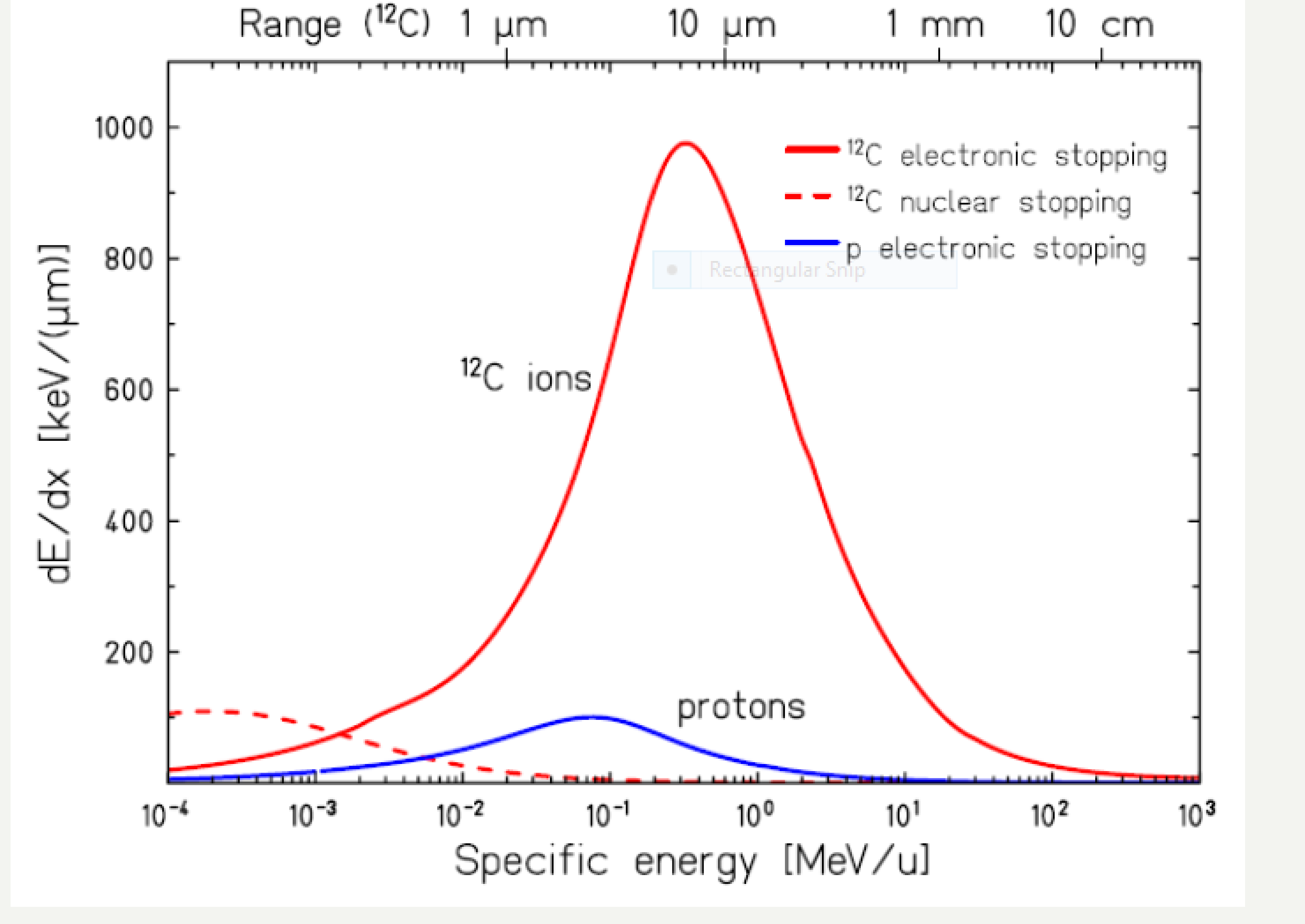
- **S is the loss of energy (*E*) of a charged particle per unit path length (*x*)**
- **S has the unit MeV/cm (or more common keV/μm)**

- ***$(dE/dx)_{el}$: electronic or collision stopping power***

- ***$(dE/dx)_{nuc}$: nuclear stopping due to elastic Coulomb scattering***

- ***$(dE/dx)_{rad}$: radiative stopping power due to the emission of bremsstrahlung in the electric fields of the particles in the material traversed***
  - ***only important for ions at extremely high energies***

Linear Stopping Power (S)

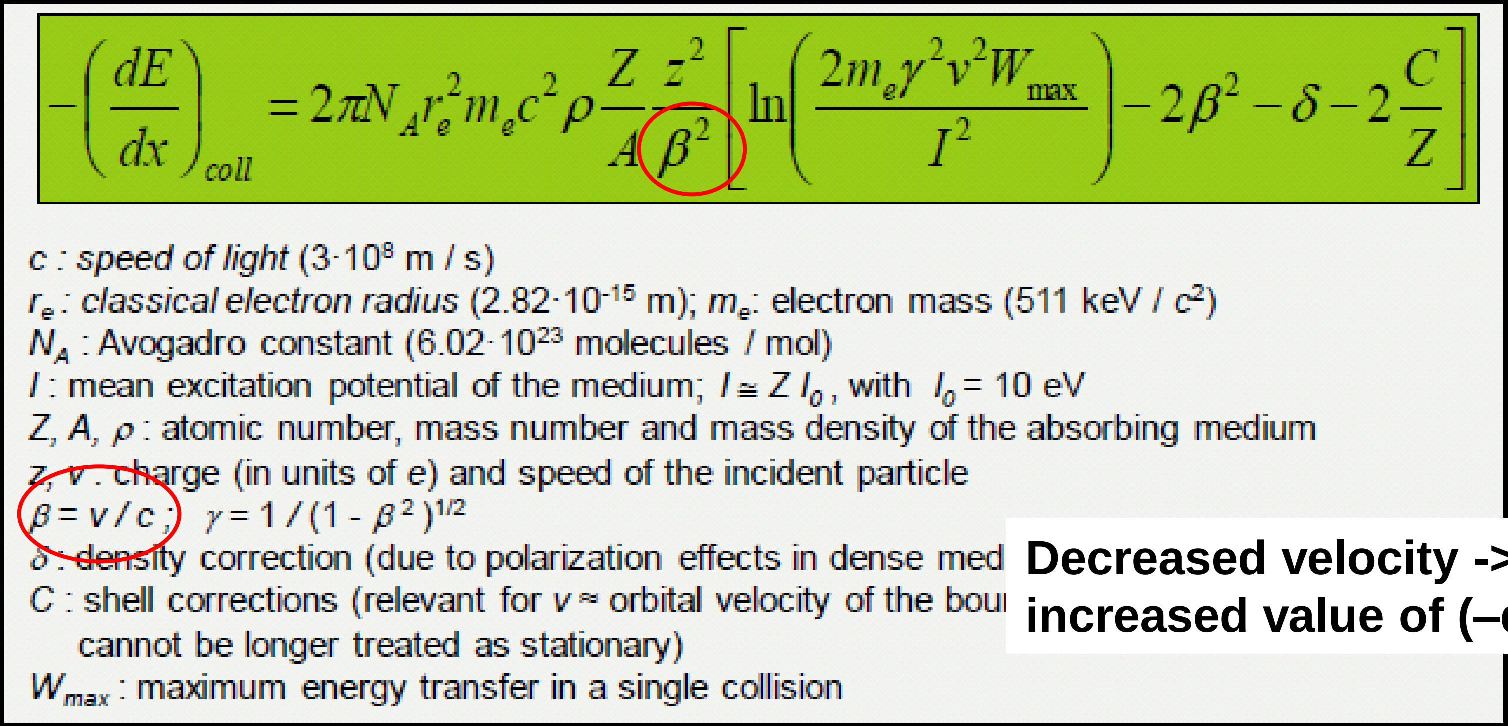


Linear Stopping Power (S)

- Within the range of therapeutically relevant energies for protons and carbon ions, the process of energy loss is dominated by *electronic collisions* and can be described by the:

Bethe-Bloch Formula (E > ~1 MeV/nucleon)

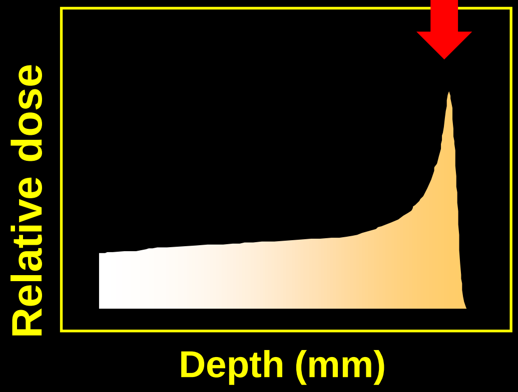
Bethe-Bloch Formula for Stopping Power



$$-\left(\frac{dE}{dx}\right)_{coll} = 2\pi N_A r_e^2 m_e c^2 \rho \frac{Z}{A}\frac{z^2}{\beta^2}\left[\ln\left(\frac{2m_e\gamma^2 v^2 W_{max}}{I^2}\right) - 2\beta^2 - \delta - 2\frac{C}{Z}\right]$$

c : *speed of light* ($3 \cdot 10^8$ m / s)
r_e : *classical electron radius* ($2.82 \cdot 10^{-15}$ m); m_e: electron mass (511 keV / c^2)
N_A : Avogadro constant ($6.02 \cdot 10^{23}$ molecules / mol)
I : mean excitation potential of the medium; $I \cong Z I_0$, with $I_0 = 10$ eV
Z, A, ρ : atomic number, mass number and mass density of the absorbing medium
z, v : charge (in units of e) and speed of the incident particle
$\beta = v / c$; $\gamma = 1 / (1 - \beta^2)^{1/2}$
δ : density correction (due to polarization effects in dense med
C : shell corrections (relevant for $v \approx$ orbital velocity of the bou
cannot be longer treated as stationary)
W_{max} : maximum energy transfer in a single collision

Decreased velocity -> increased value of (–dE/dx)



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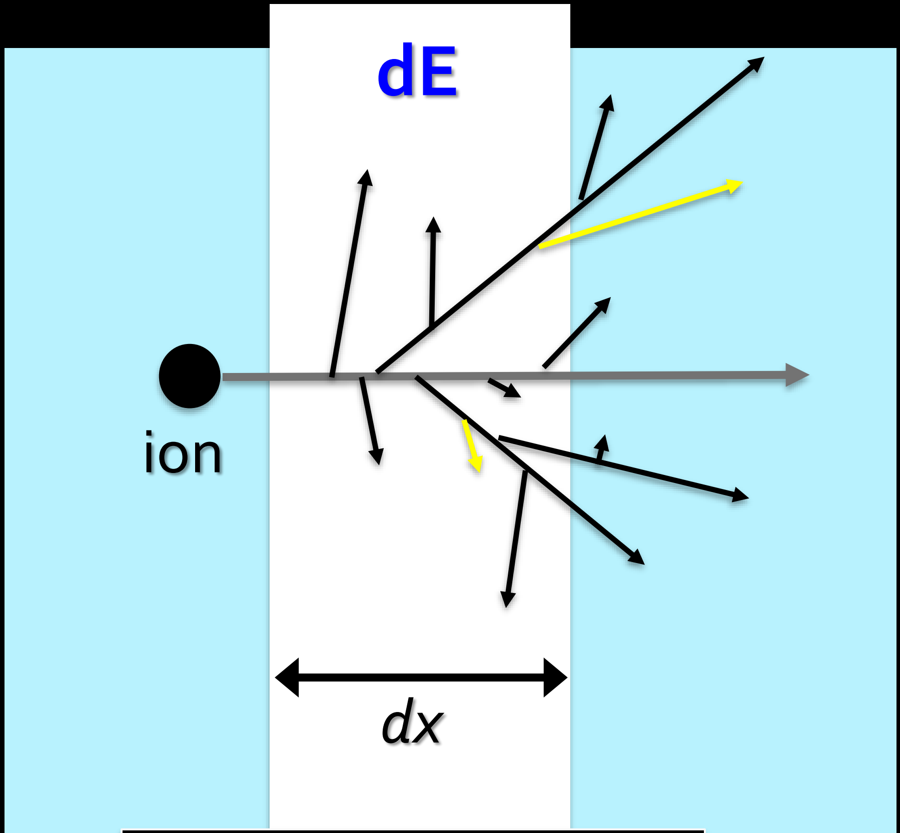
Linear Energy Transfer (LET)

Macroscopic: $\text{LET}_\infty = (dE/dx)_\infty$



LET_∞ is the unrestricted linear energy transfer

LET_∞ is the amount of energy deposited per unit length of a material as a charged particle traverses the material

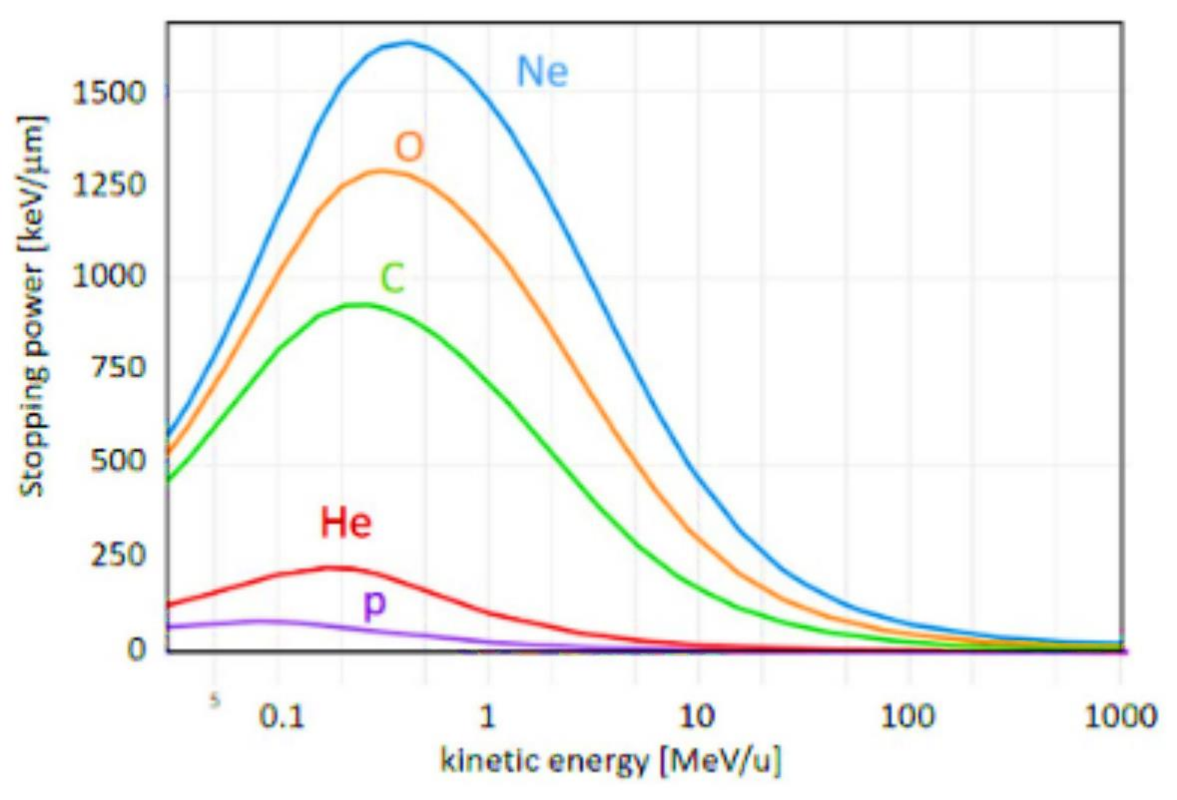
LET_∞ has the unit MeV/cm (or more common keV/µm)

Independent of radial dose distribution

Linear Stopping Power (S)

$S = -LET_{el,\infty} = -(dE/dx)_{el,\infty}$

From now on labelled $LET_{\infty} = (dE/dx)_{\infty}$



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Mass Stopping Power

S/ρ = mass stopping power [MeV,cm^2/g]

ρ = density

$$S/\rho = \left(\frac{dE}{\rho dx}\right) \quad [\mathrm{MeV,cm^2/g}]$$

The dose from charged particled in some medium (e.g. water):

$$D_w = \int_0^{E0} \phi_E \left(\frac{dE}{\rho dx}\right)_w dE \quad [\mathrm{Gy}]$$

ϕ_E = the particle fluence

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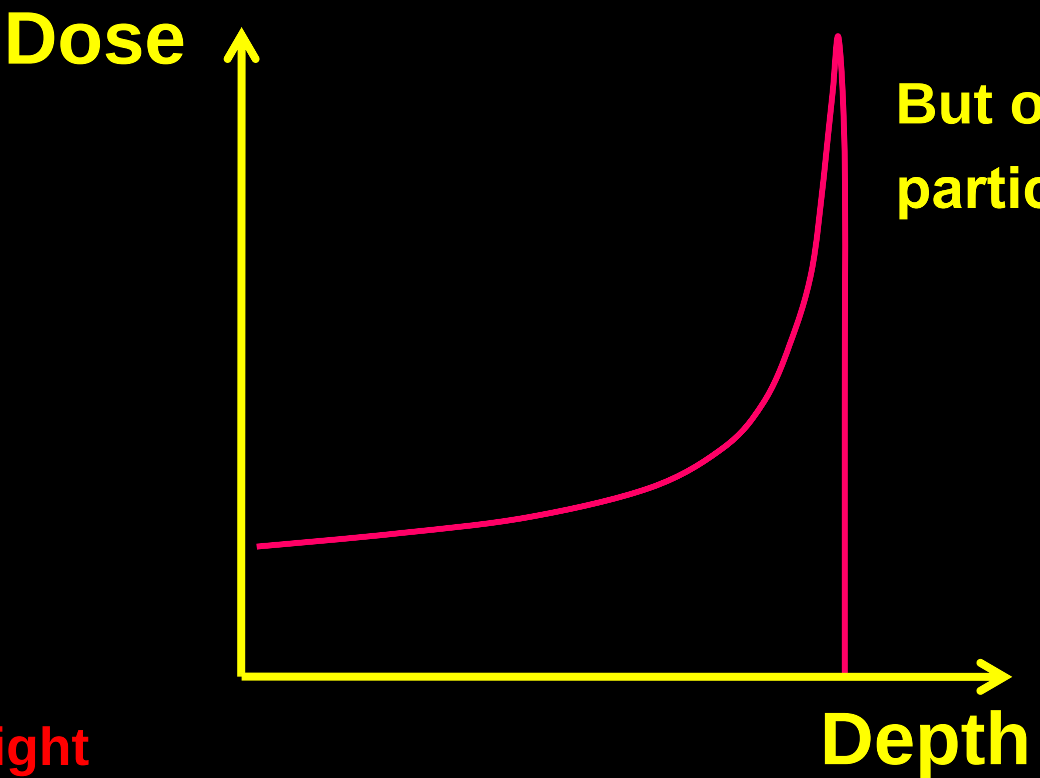
1D Depth Dose Distribution - Bragg Curve

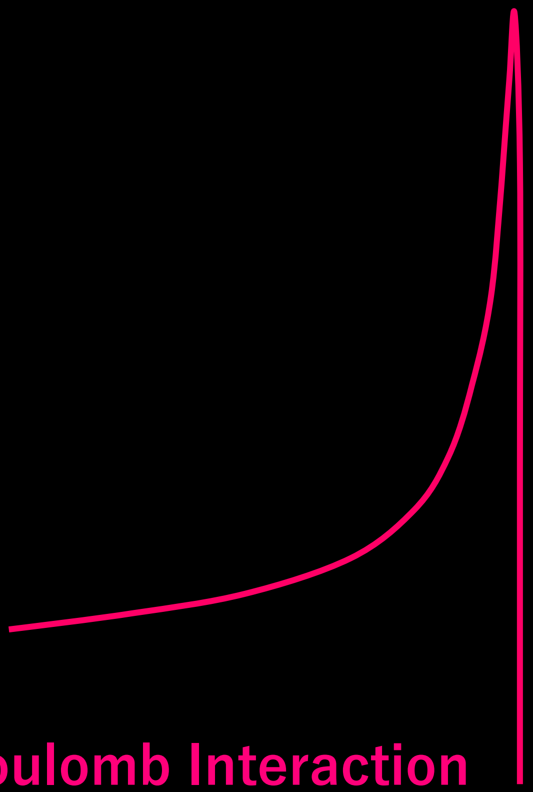
Now we can calc. depth-dose distributions:

$$D_w = \int_0^{E0} \phi_E \left(\frac{dE}{\rho dx}\right)_w dE \quad [\text{Gy}]$$



But only for primary particles and “ideal case”!

1D Depth Dose Distribution - Bragg Curve



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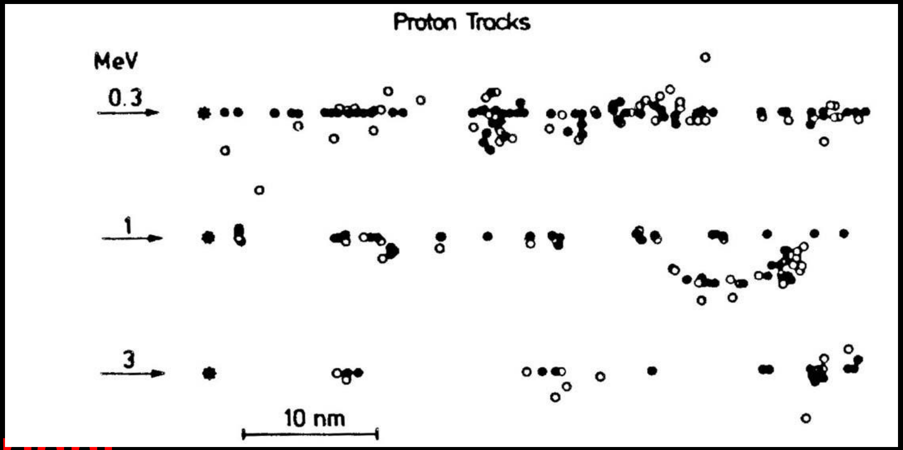
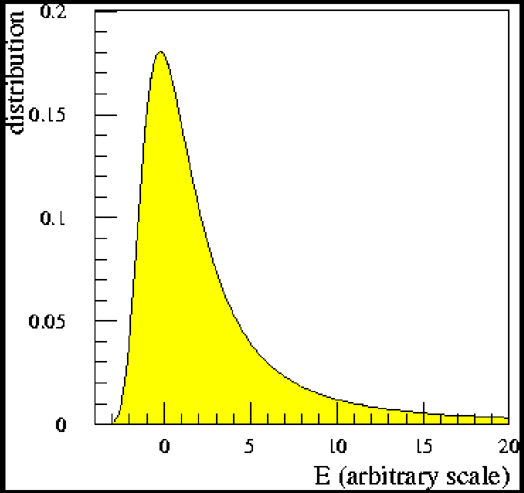
Energy Straggling

- So far we used the Continuously Slowing Down Approximation (CSDA)
- The Bethe-Bloch formula gives *the mean energy* lost per unit path length
- In reality, ions lose their energy in individual collisions with electrons
  - Actual energy loss will scatter around the mean value
  - Energy loss distribution is not Gaussian around mean
- The stochastic behavior of energy deposition in matter (*energy straggling*) is described by stochastic distributions such as the Landau-Vavilov distribution

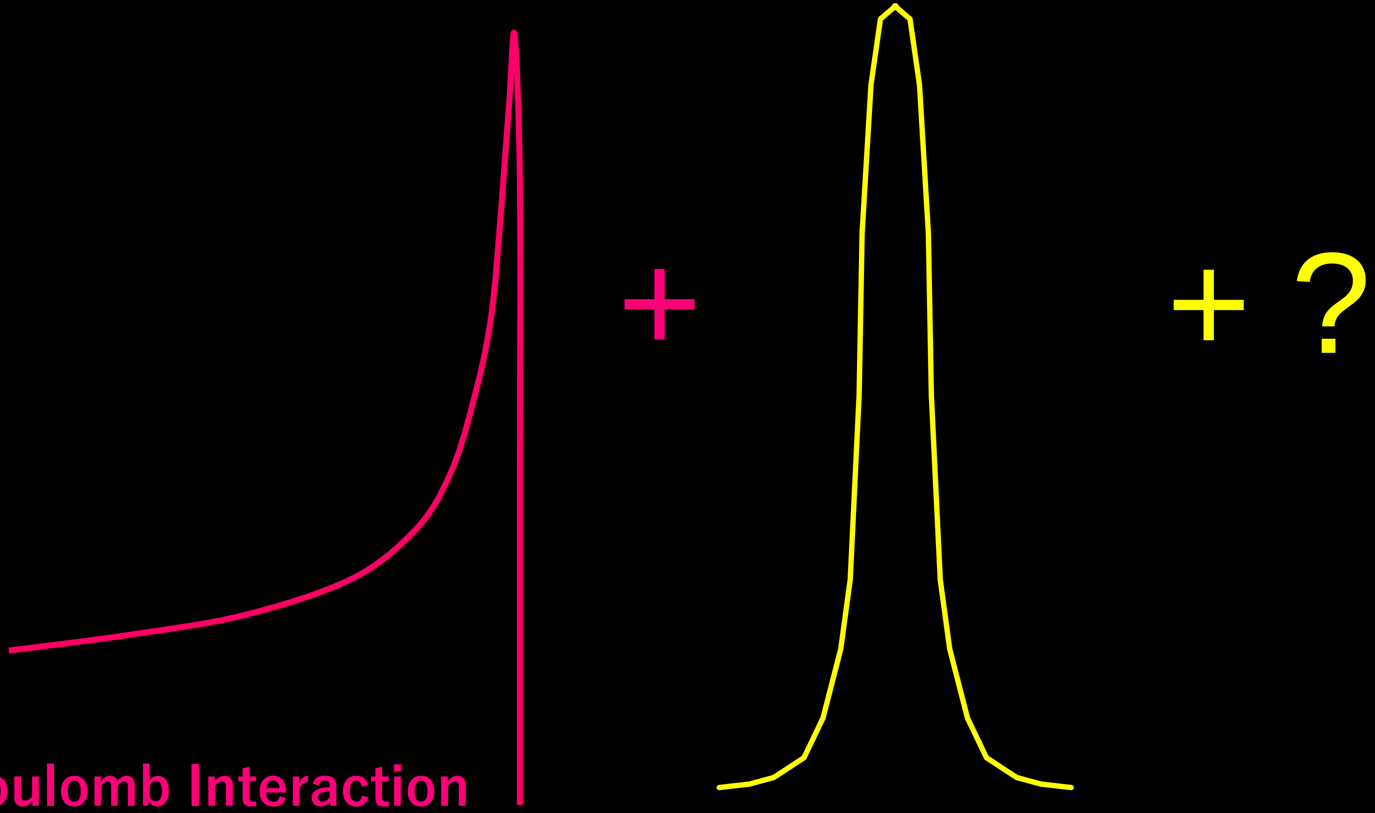




Landau's theory J. Phys (USSR) 8, 201 (1944)

Range Straggling

Energy straggling → range straggling



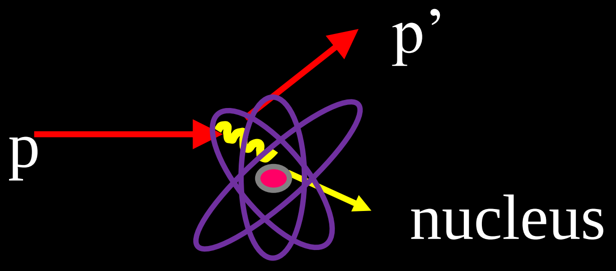
Theoretical
w/o Straggling

Range Straggling
Distribution

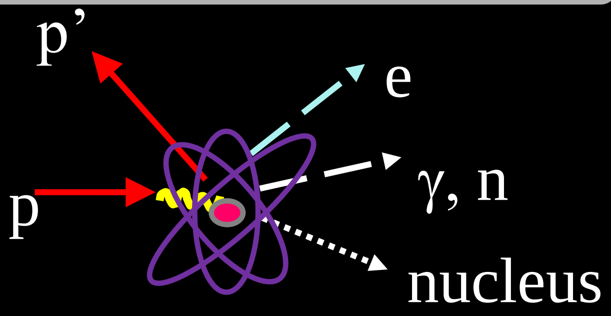
Coulomb Interaction

Range Straggling

Nuclear Interactions of Protons



Elastic collision (large q)

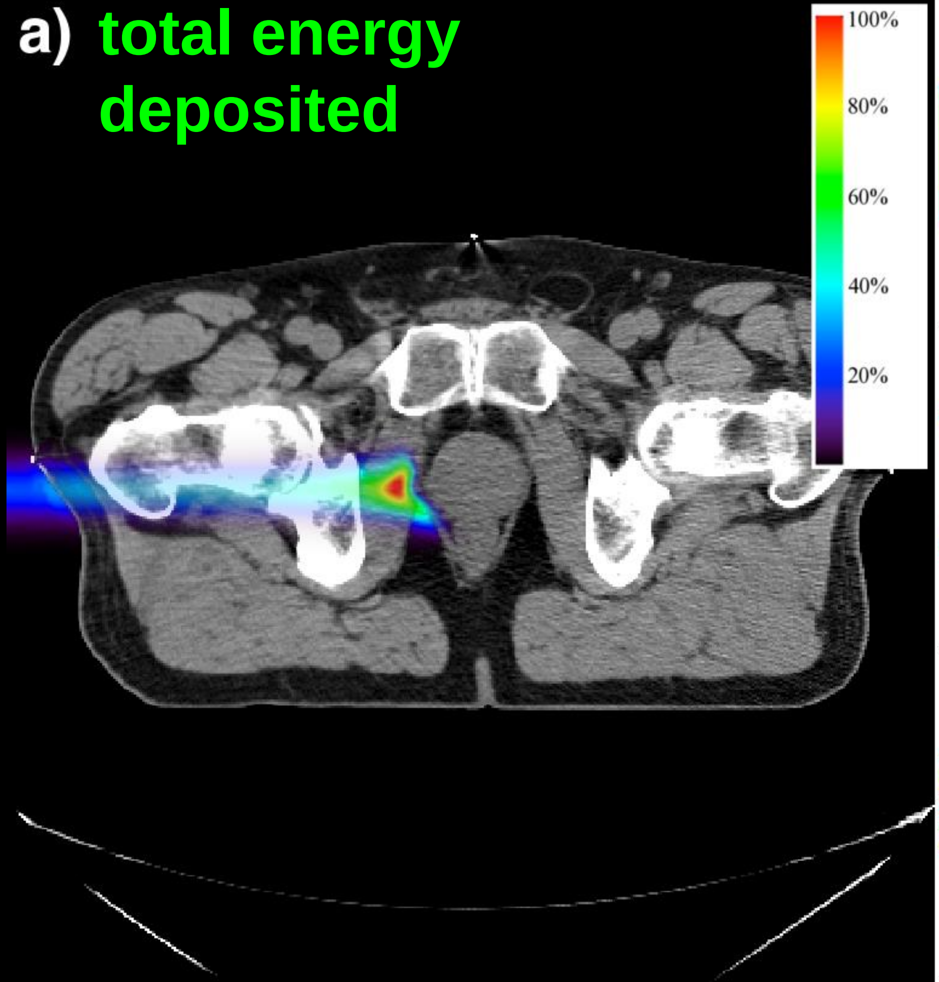


Nuclear interaction

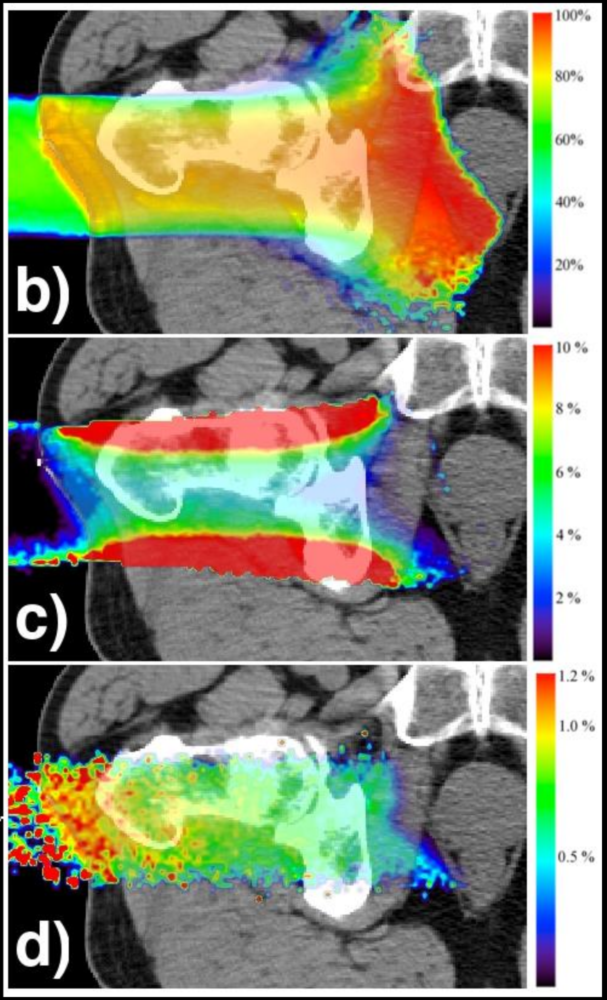
- A certain fraction of protons have nuclear interactions in tissue (about 1% of all protons per cm of penetration)
- Mostly with oxygen and carbon nucleus
- Nuclear interactions cause a decrease in primary proton fluence
- Nuclear interactions lead to secondary particles and thus to local and non-local dose depositions (neutrons!)
- The dose from nuclear interactions is negligible in the Bragg peak
- Target fragments have high LET and therefore high relative biological effectiveness (RBE) and can cause normal tissue complications

Spatial Distribution

a) total energy deposited



Contribution in %

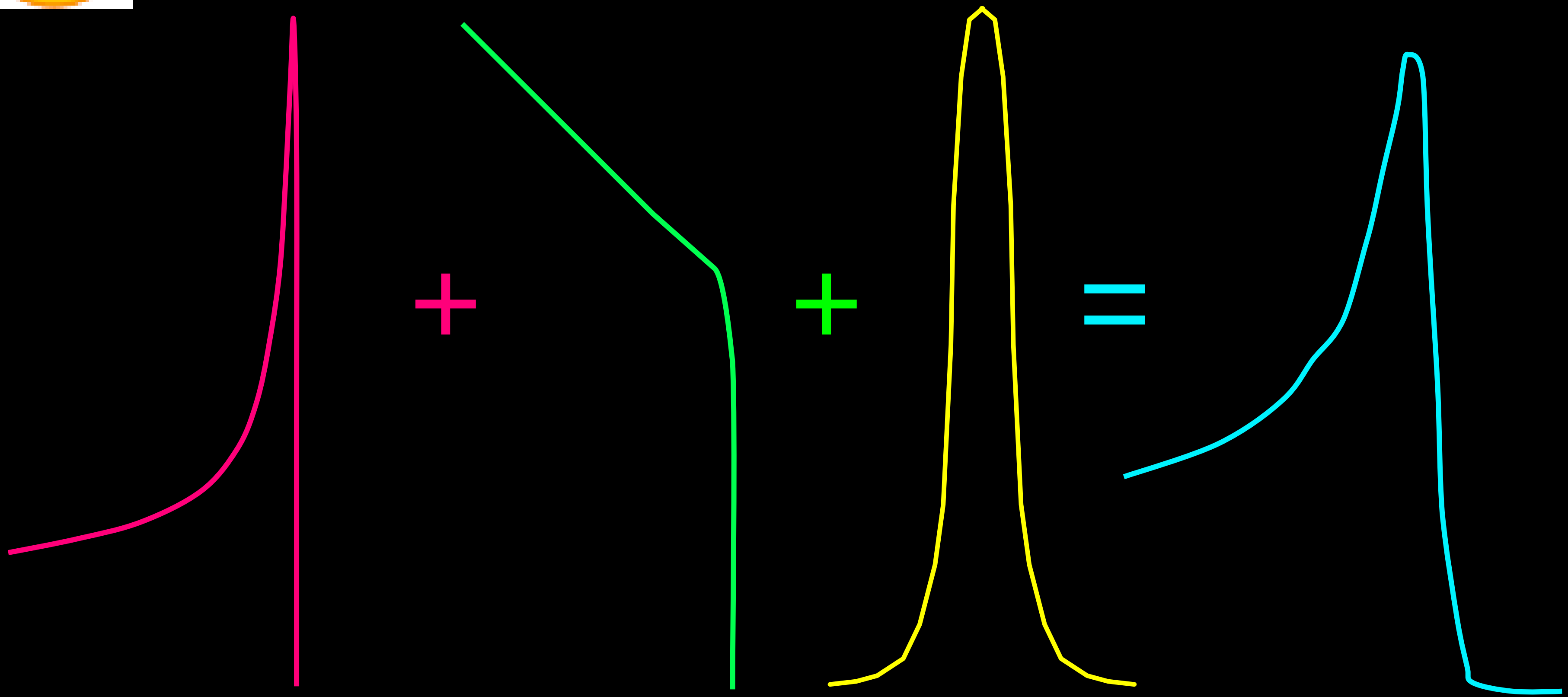


primary protons

secondary protons

alphas
&
recoils

This makes the 1D Bragg Curve!



Nuclear Reaction

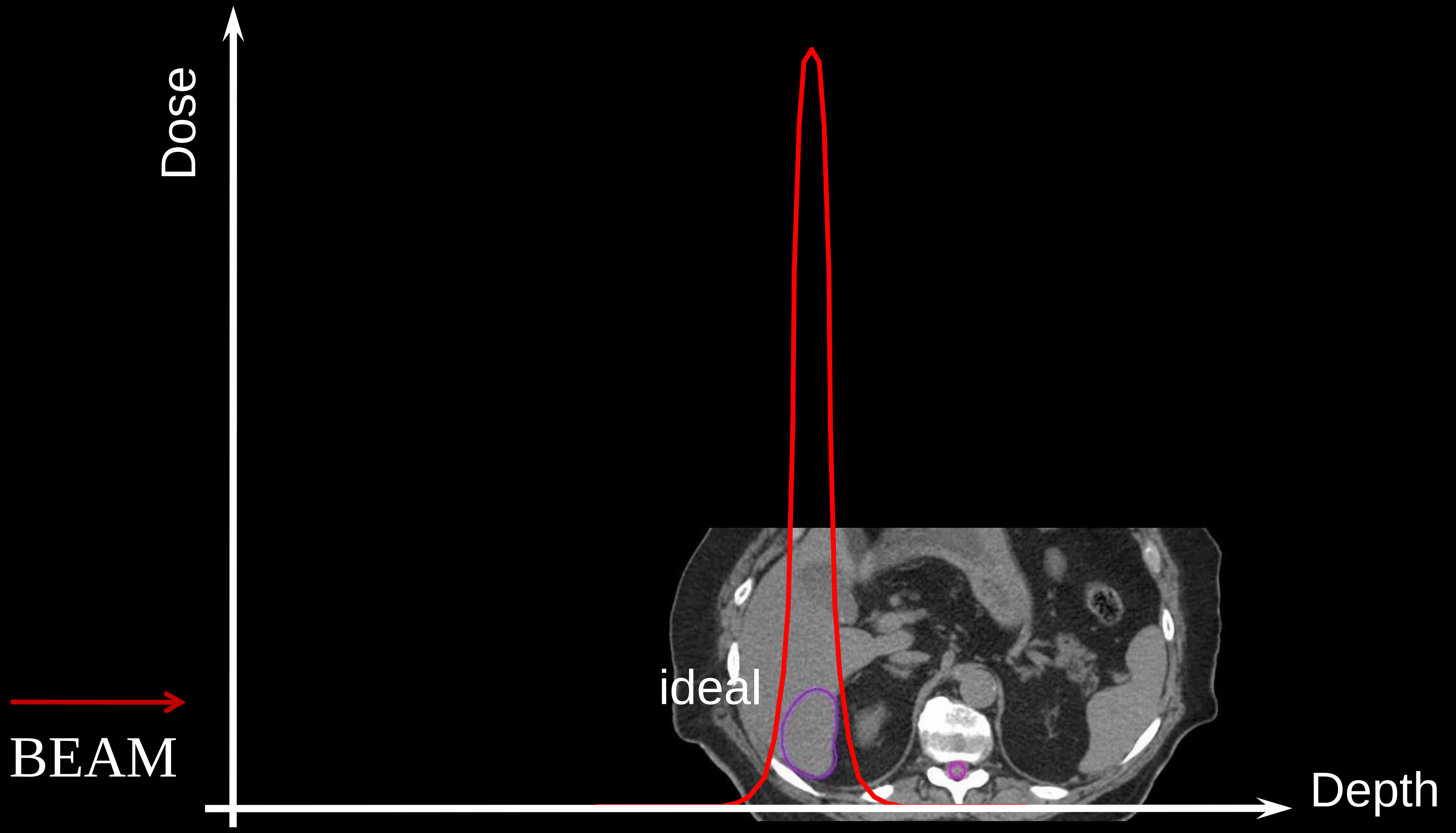
Bragg Curve

+ + =

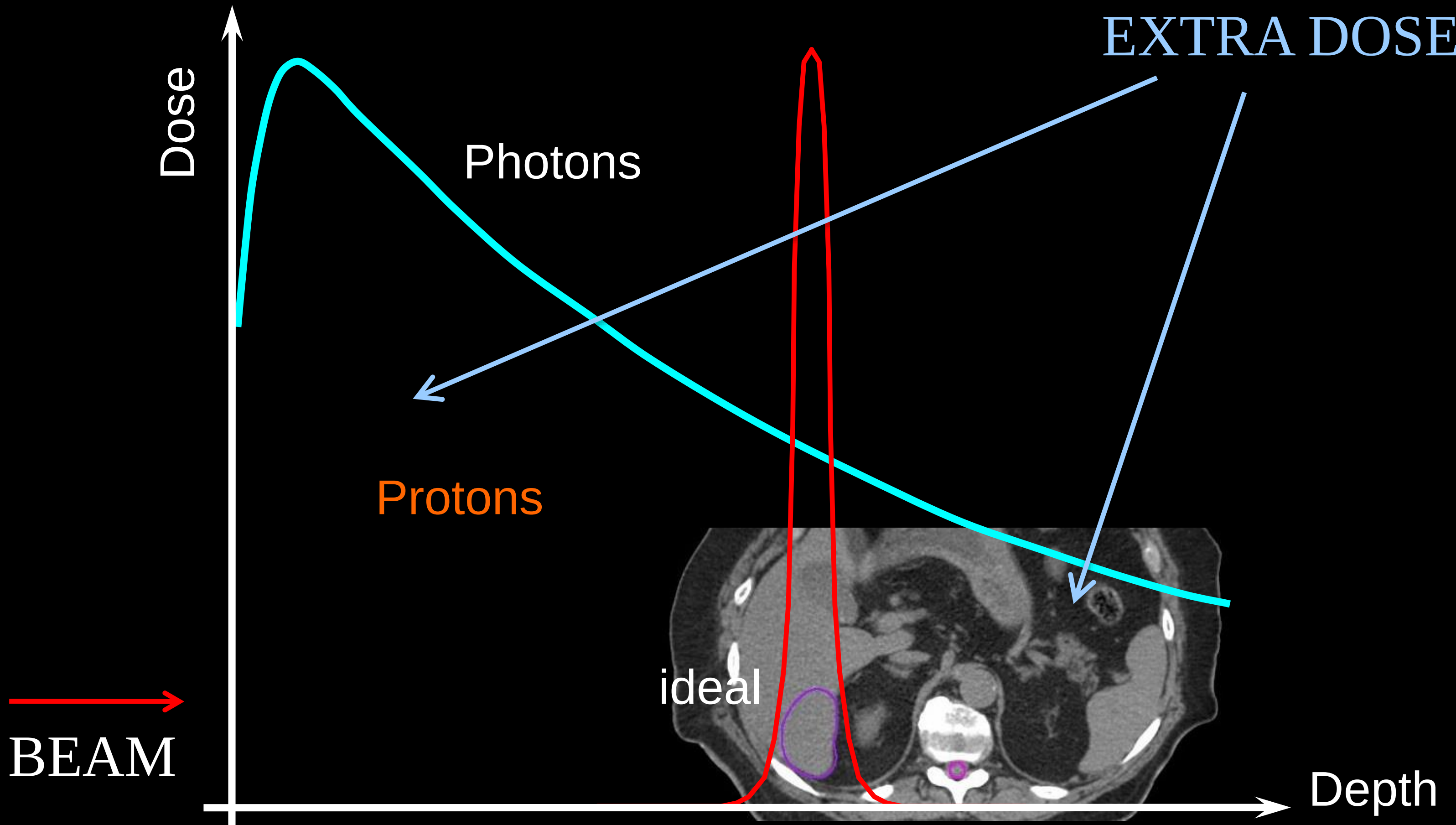
Coulomb Interaction

Range Straggling

External Beam Therapy



External Beam Therapy



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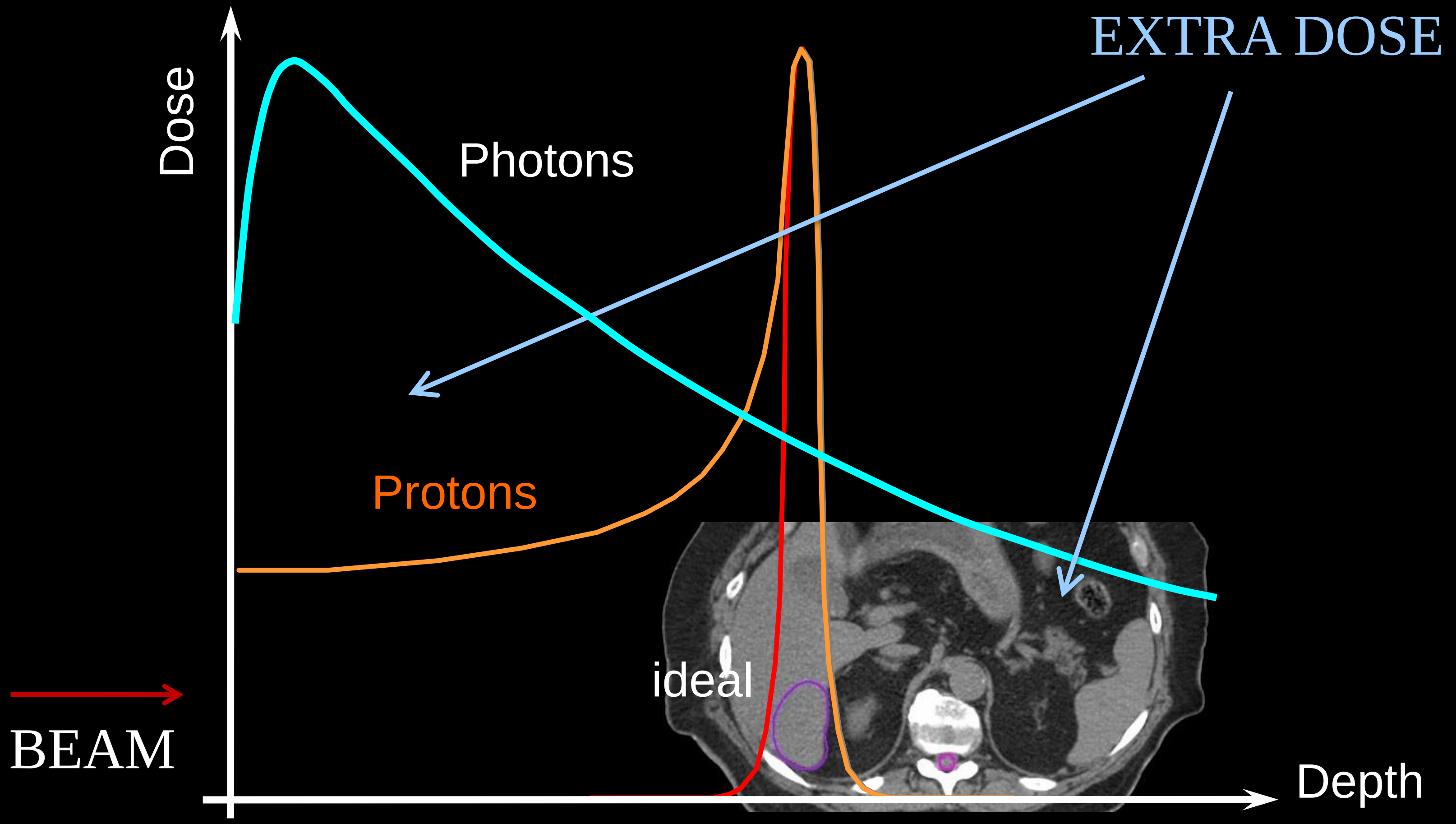
External Beam Therapy



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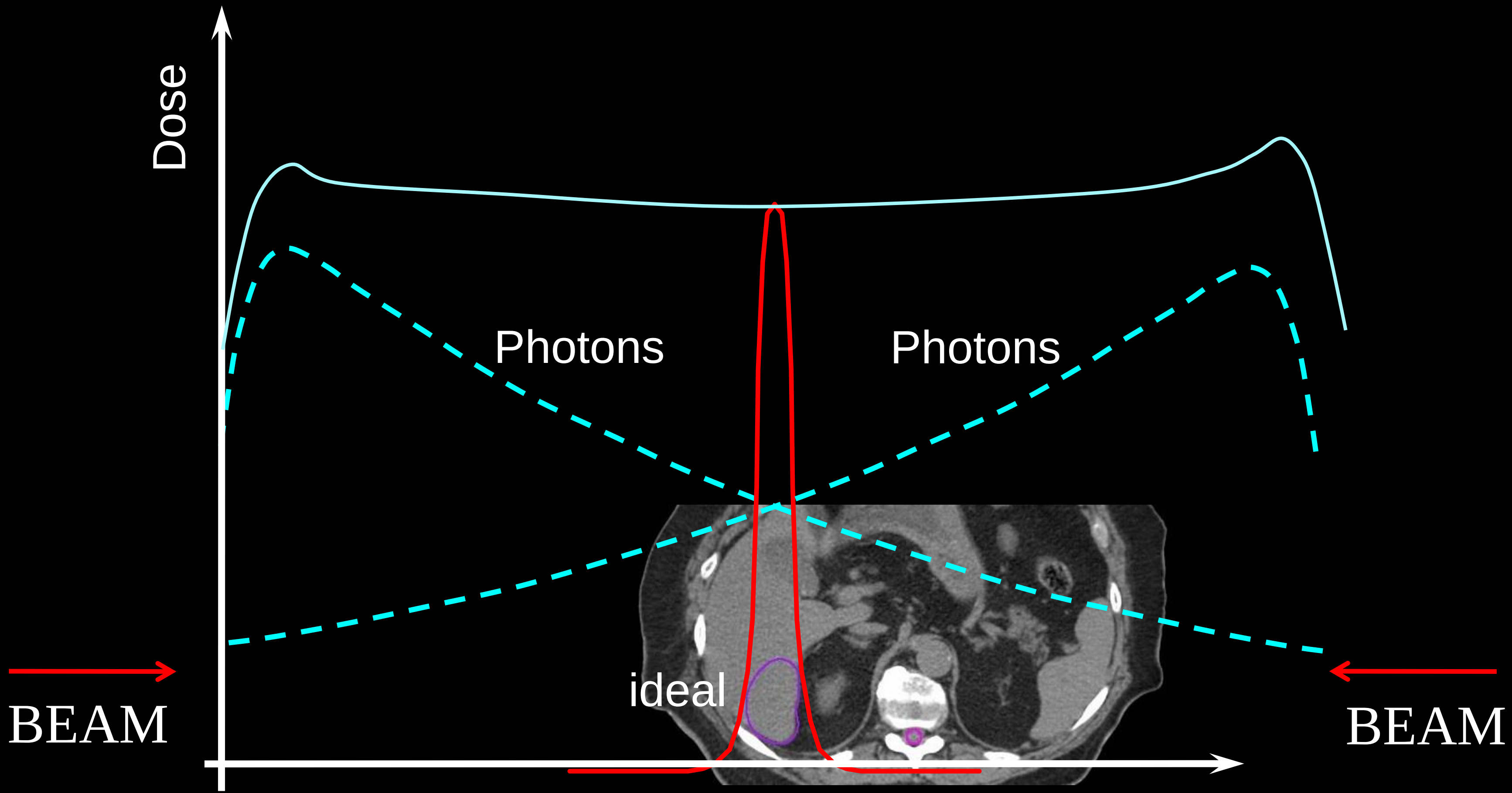
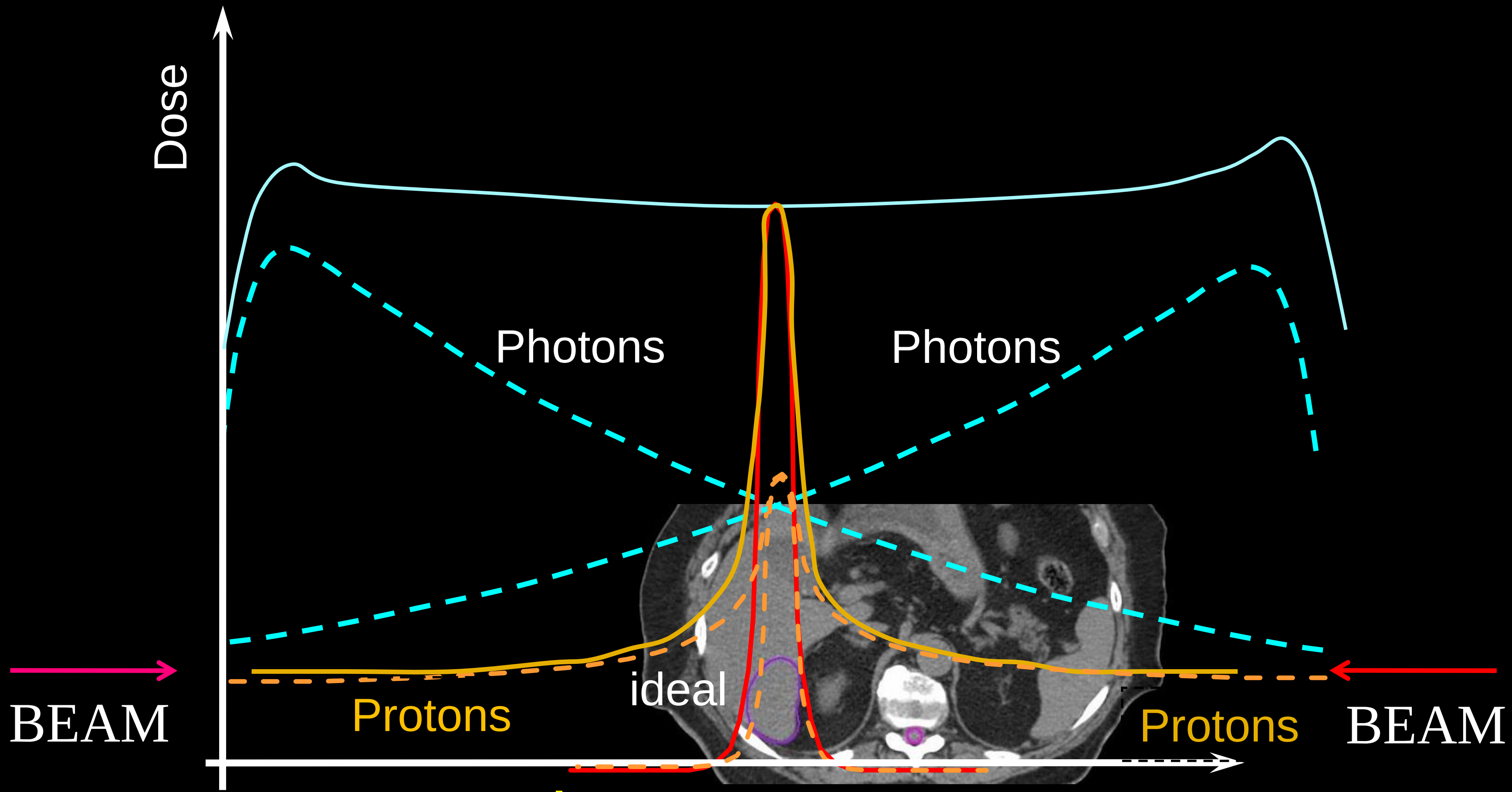
Proton Beam Advantage



Beam for beam you can always do a better job with particles (except at the surface). *Flanz '09*

Proton Beam Advantage



Photon vs Proton

Beam for beam you can always do a better job with particles (except at the surface).

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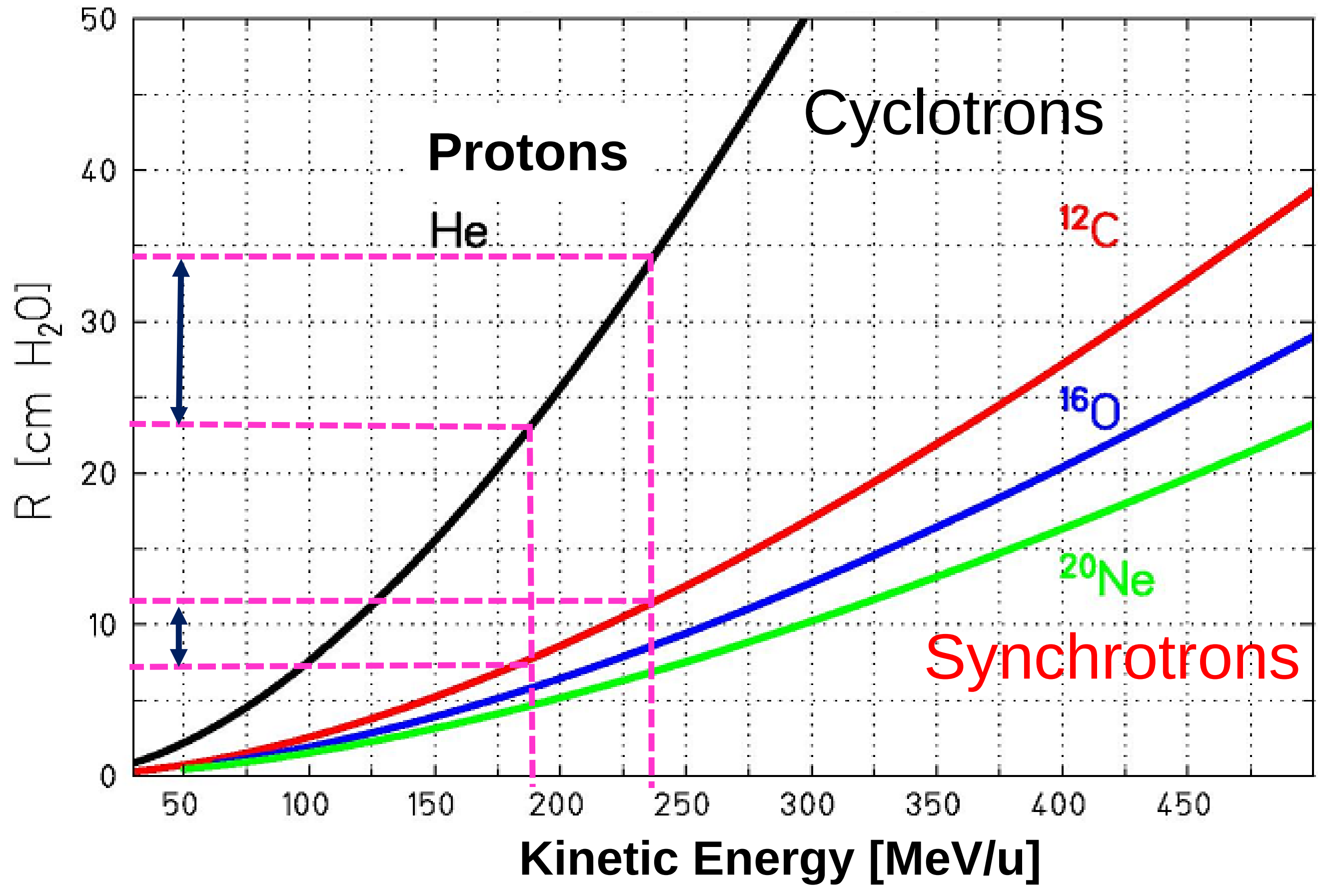
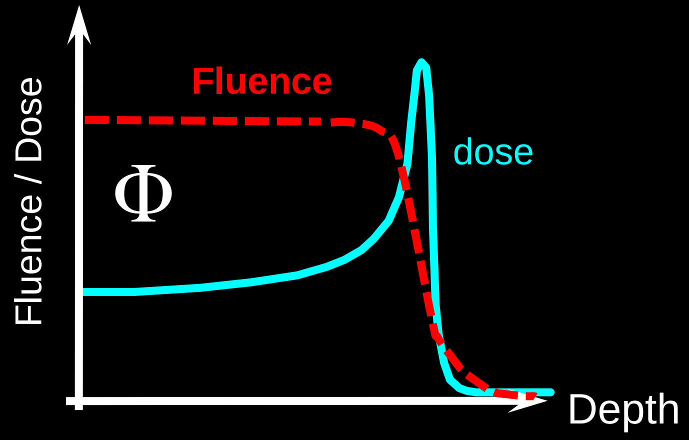
Particle Range vs Energy



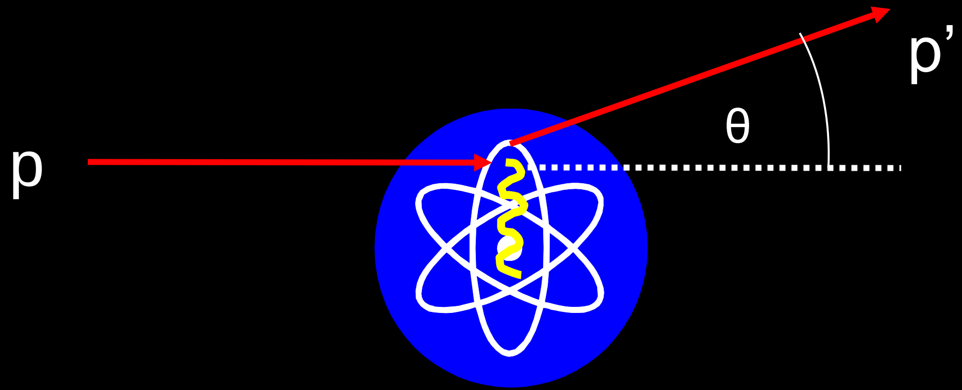
$$R(E) = \int_0^E \left(-\frac{dE}{dx}\right)^{-1} \cdot dE$$

Dose Dependence on Depth

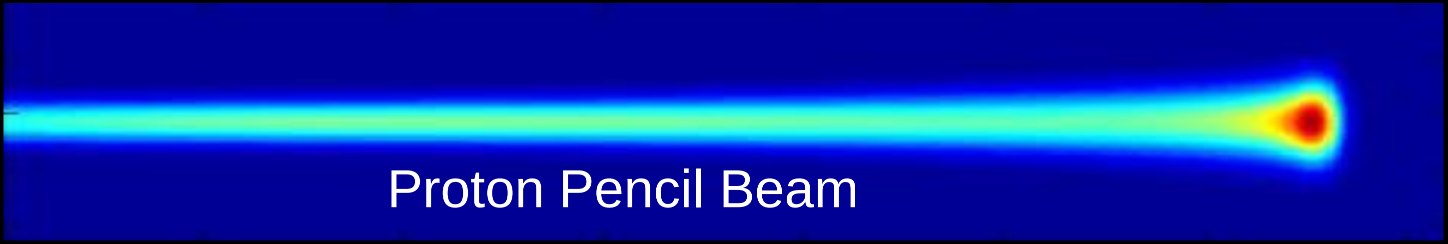


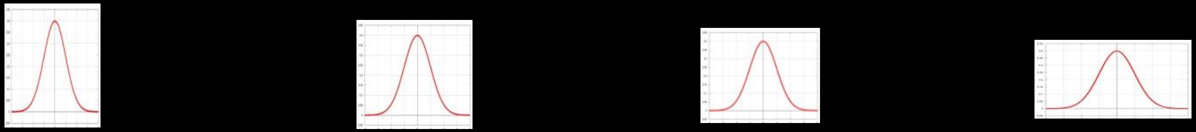
- Nearly no attenuation, fluence stays constant except near the end of the ranges of the ions
- Ions lose energy gradually
- Energy loss per ion pair stays same
- Ion pairs per unit length increases
- Increase in LET, and in the ratio between biological dose and physical dose, i.e., increase in RBE, at the end
- Electron energy low → no build up

Multiple Coulomb Scattering (MSC)



- Ions are deflected in the electric field of the nuclei
- In general, multiple deflections will occur for each ion
- Play key role in determining lateral dose distribution





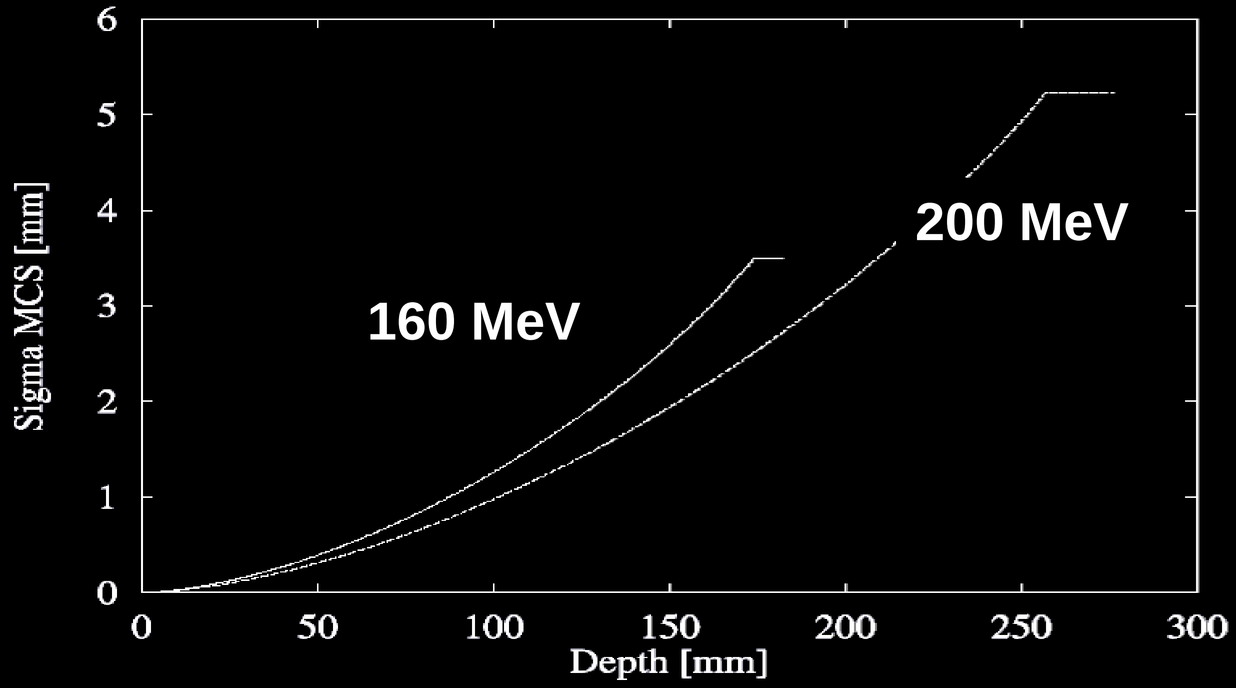
Multiple Coulomb Scattering (MSC)

Moliere Theory

- The definitive theory of multiple Coulomb scattering was published by Molière in 1947
  - It has *no* empirical parameters and covers *arbitrarily thick* scatterers
  - The angular distribution at large angles falls off roughly as $1/\theta^4$ but is nearly Gaussian for small angles

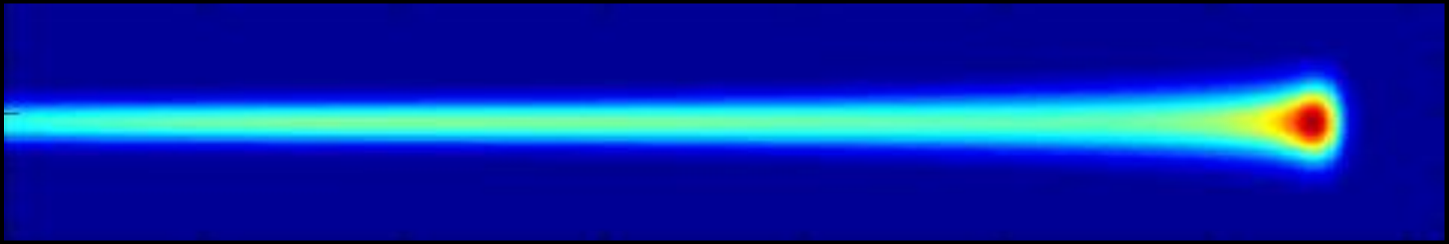
MCS Dependence on Beam Energy

Protons



Lateral Penumbra Changes in Depth

- Ions experiences MCS
- Each time deflected by a small angle, but the particle stays in the beam
- Effect of deflection accumulates
- Ions spread out laterally Gaussian flattens out
- Beam penumbra increases
- At the same time, ion energy decreases and deflection angle increases for each interaction
- Beam penumbra increases faster near the end of beam range



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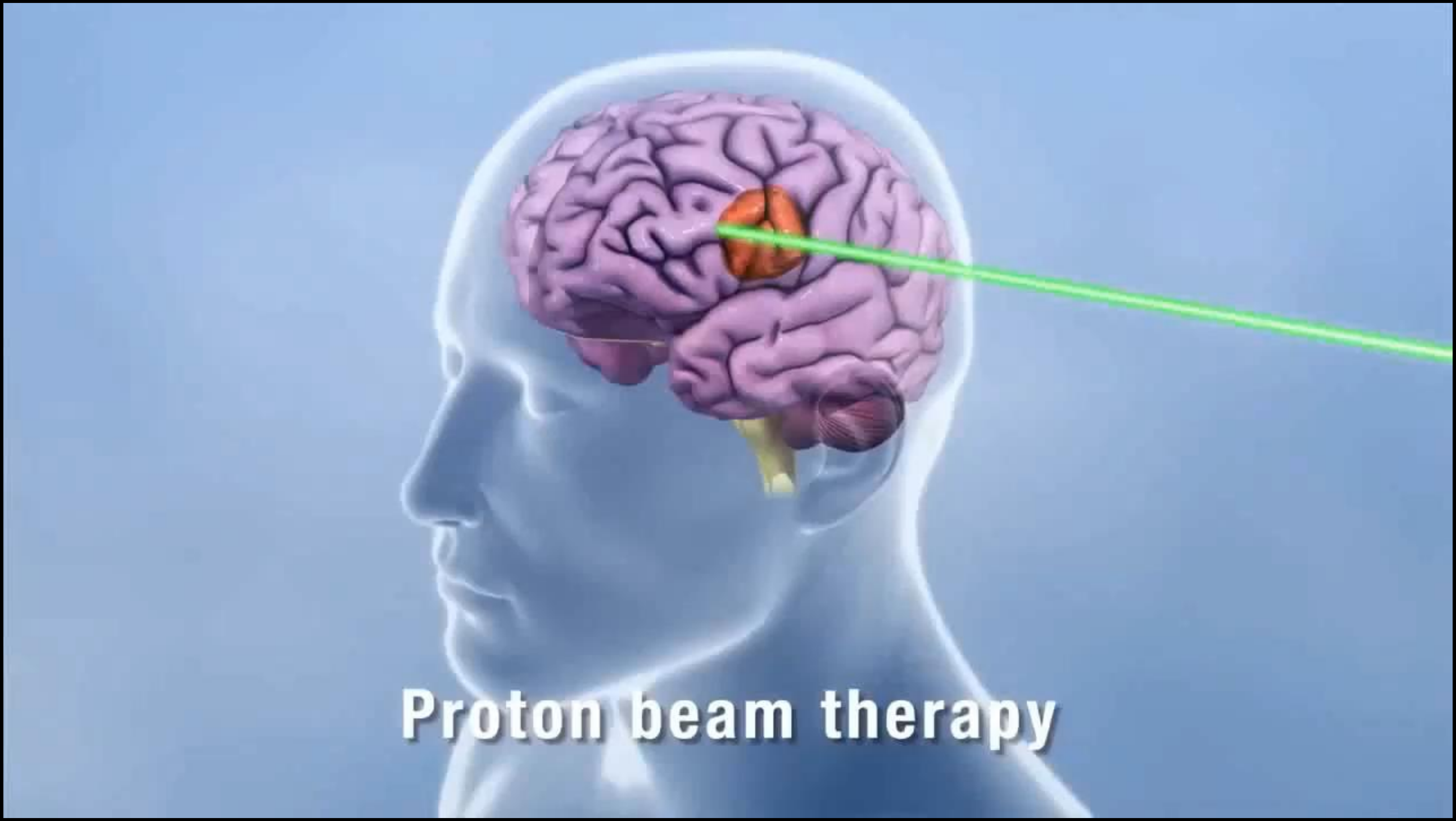
Proton Pencil Beam Scanning



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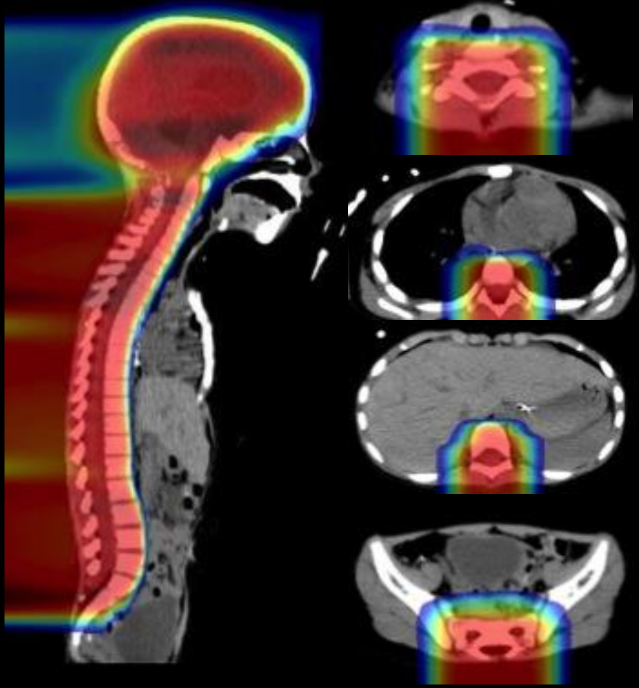
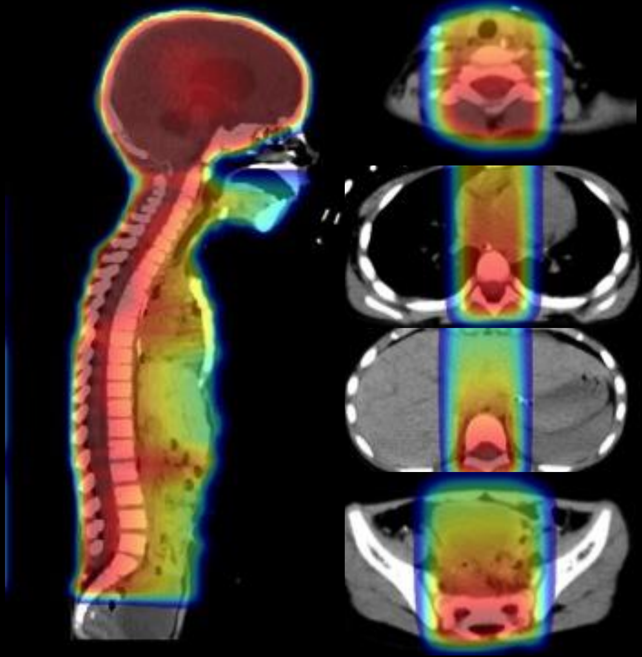
Pediatric cancer (e.g. medulloblastoma) - representative for the superior dose distribution using particles

Proton beam



X-ray beam



Absolute risk of secondary cancer

Method	Risk	X-ray/Proton
Proton	0.05	1.0
X-ray (Standard)	0.75	15.0
X-ray (IMRT)	0.43	8.6

Intention to treat spinal cord and brain only!

Miralbell R et al., Int J Radiat Oncol Biol Phys 54: 824-829, (2002)

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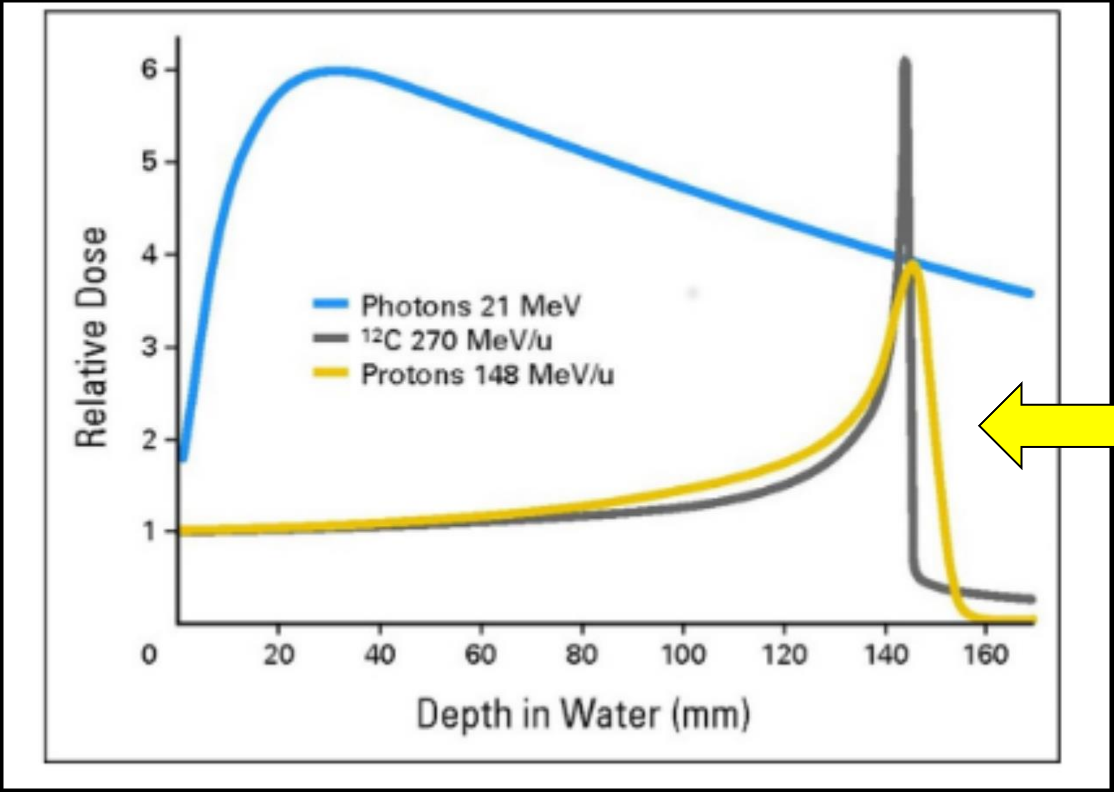
Why Carbon Therapy?

Physical Advantages of Carbon Beams



Jäkel O. Physical advantages of particles: protons and light ions.
Br J Radiol 2020; 93: 20190428.

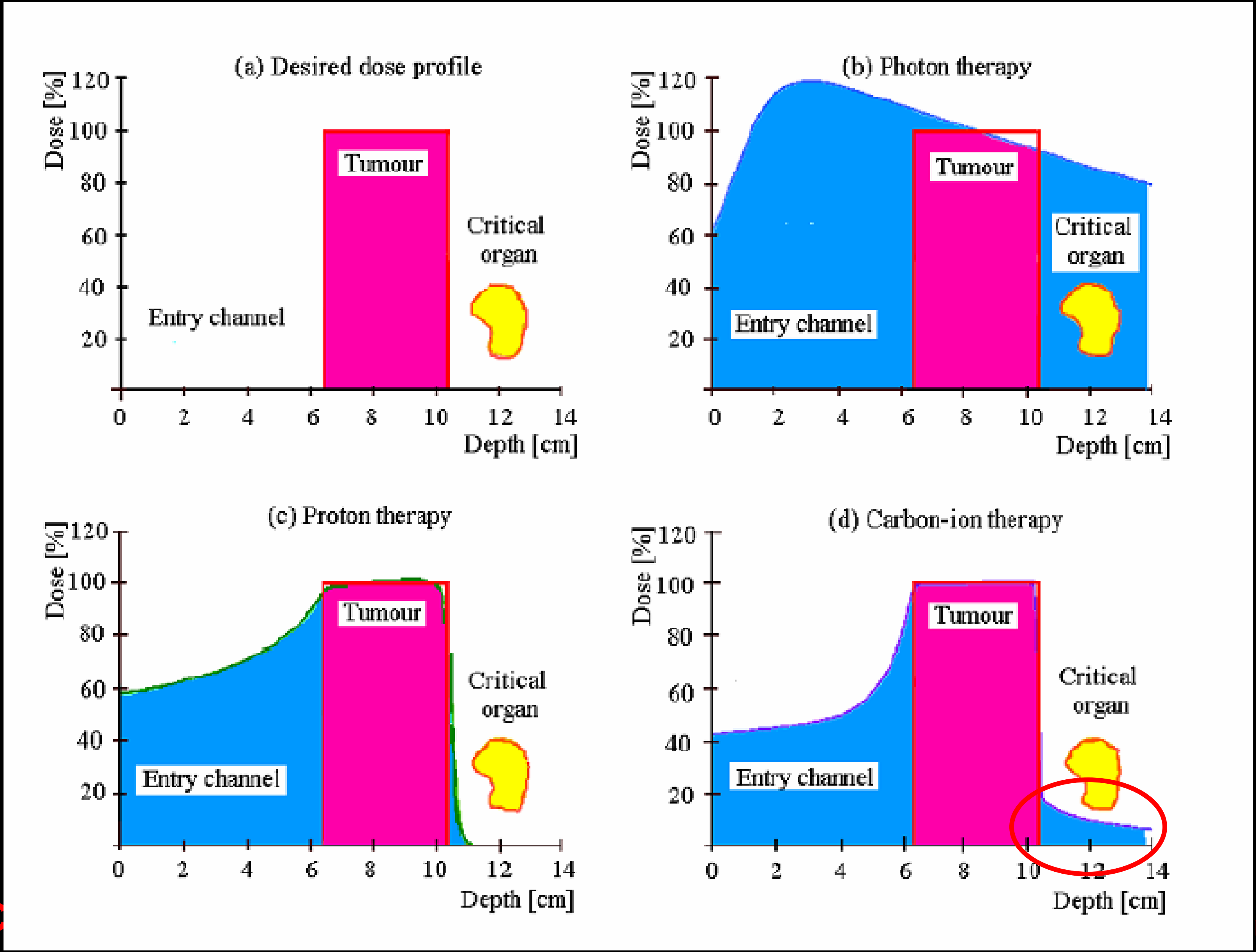
The variance of the range Straggling, σ_R^2, is related to the energy losses, σ_E^2.

The with of the range straggling can be expressed by:

$$\sigma_R = \frac{R}{\sqrt{m}} f\left(\frac{E}{mc^2}\right),$$

where m and E are the projectile's mass and Energy. The $1/\sqrt{m}$ dependence causes protons to have a higher straggling than light ions, by a factor of 3.5 with respect to carbon ions.

Physical Advantages of Ion Beams

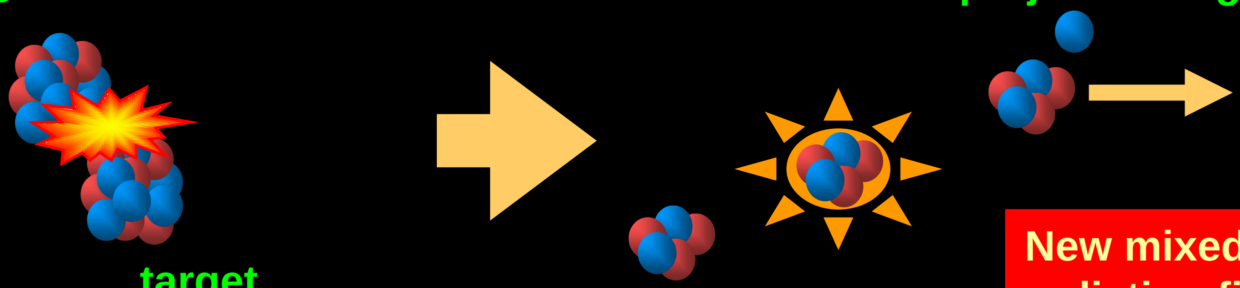


Nuclear Reactions

Carbon ion therapy: 120 - 400 MeV/u

For the therapy we have to know all interaction events, i.e. *particles (all generations) fluences vs. energies, etc.*

Nuclear Reactions



projectile

target

target fragments

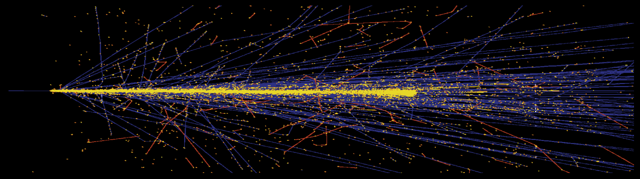
projectile fragments

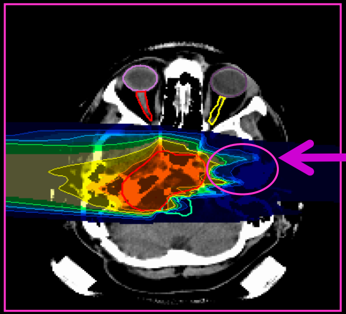
New mixed radiation field !

Interaction of the radiation with the tissue and organs in the body...

Target Fragments	*Projectile fragments*
… lower charges than target	… lower charges than primaries
… high LETs	… mixed LETs
… short ranges	… long ranges

High-energy carbon beam stopping in water





Caused by projectile fragments

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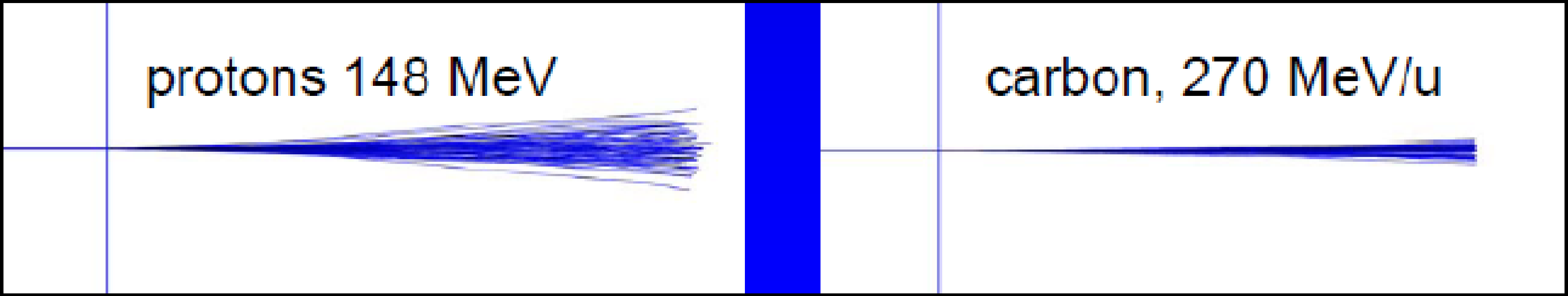
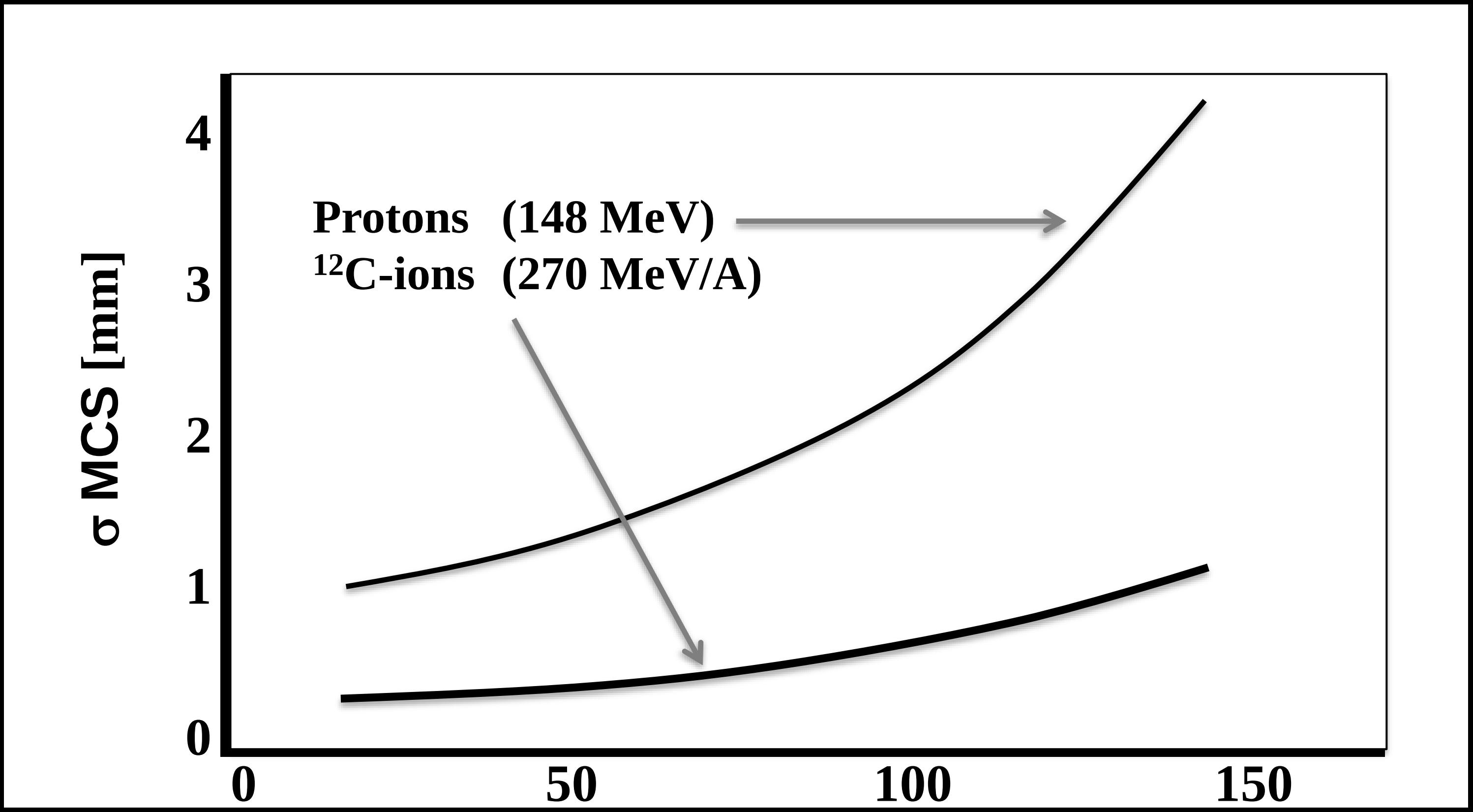
Mixed radiation field with charged particles, neutrons and gamma rays is a challenge for the dosimetry!

Multiple Coulomb Scattering (MSC)

Heavy ions exhibit more precise physical dose distributions than protons because angular and range scattering are inversely proportional to the square of the atomic number.



MCS Dependence on Particle Charge



Depth [mm]

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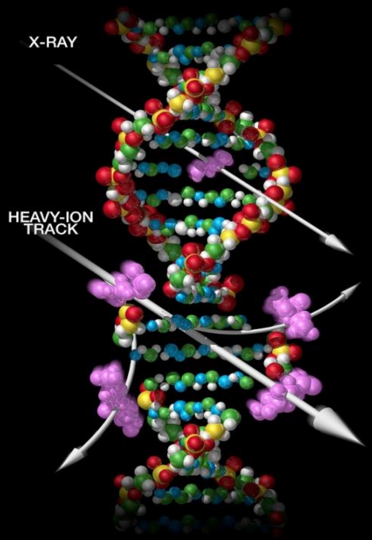
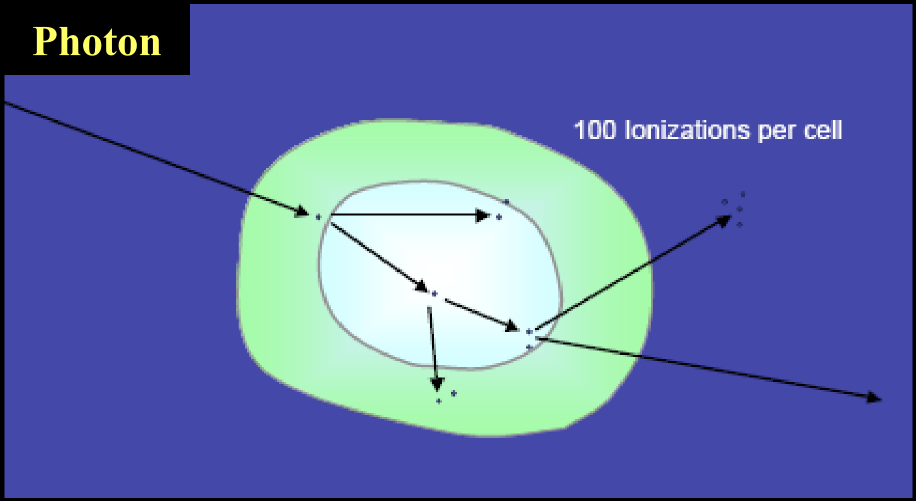
Biological effects are controlled by the differences in physics and chemistry of different ionizing radiation!

Distribution of Ionizing Events



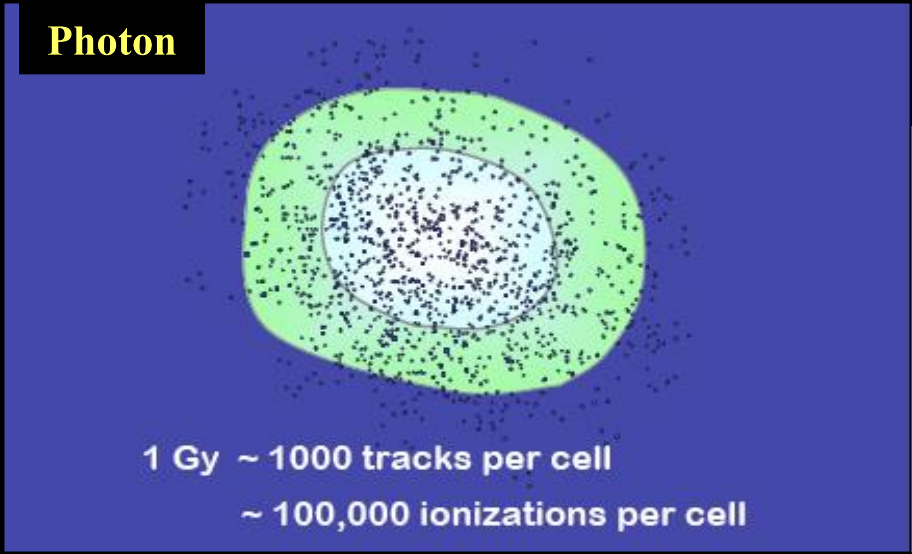
Low LET radiation

Photon

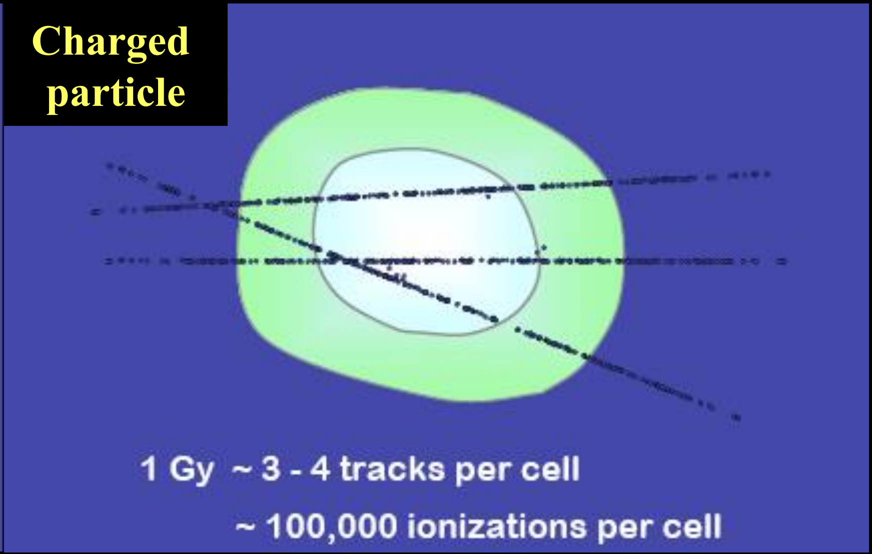


High LET radiation

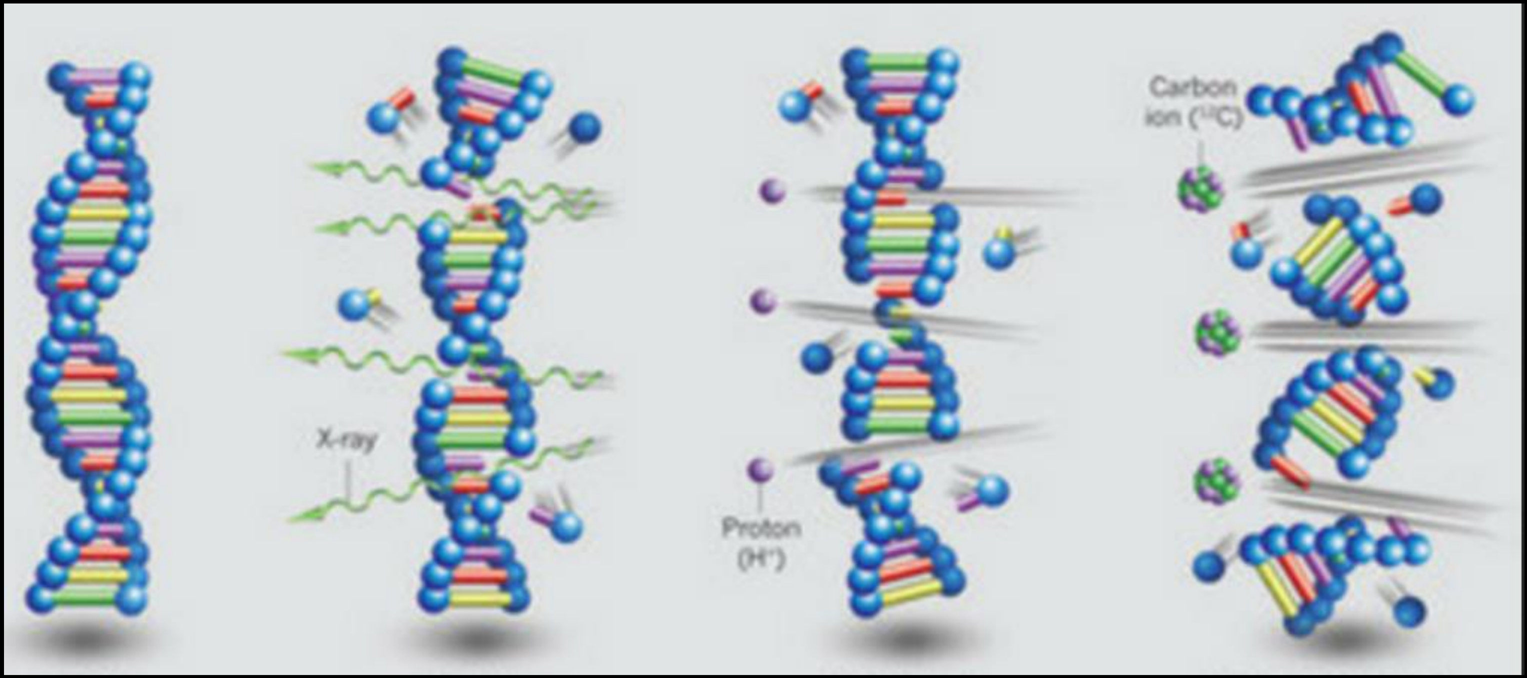
Photon



Charged particle



Radiation Types and their Damage to DNA



DNA

X-ray

Proton Beam

Carbon Ion Beam

Low-LET radiation:
Repairable single/double strand breaks

High-LET radiation:
Complex DNA lesions
Multiple DNA pathways
More difficult to repair
Enhances cell death

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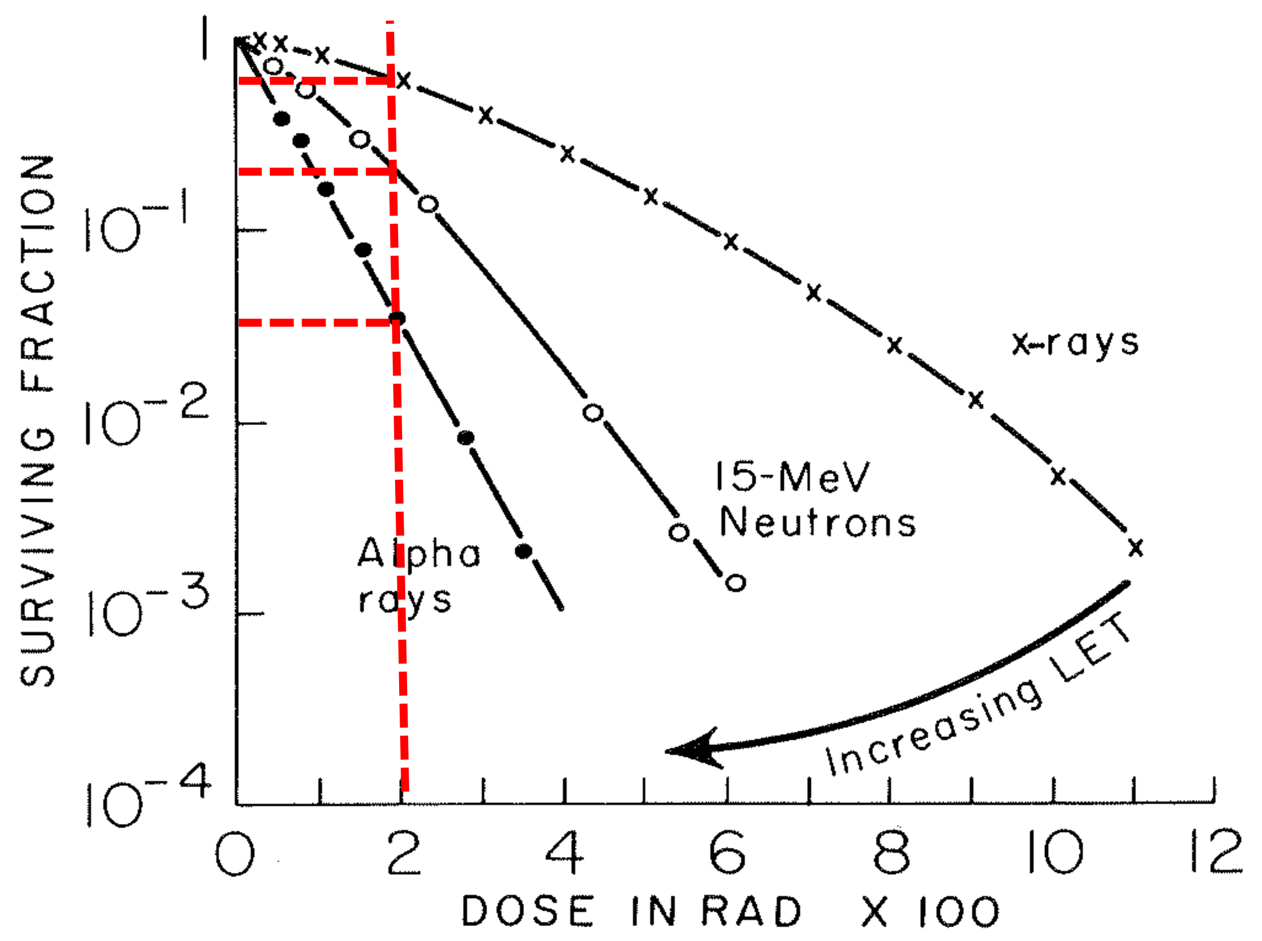
Cell Survival and LET



1 rad = 0.01 Gy = 0.01 J/kg 1 rem = 0.01 Sv

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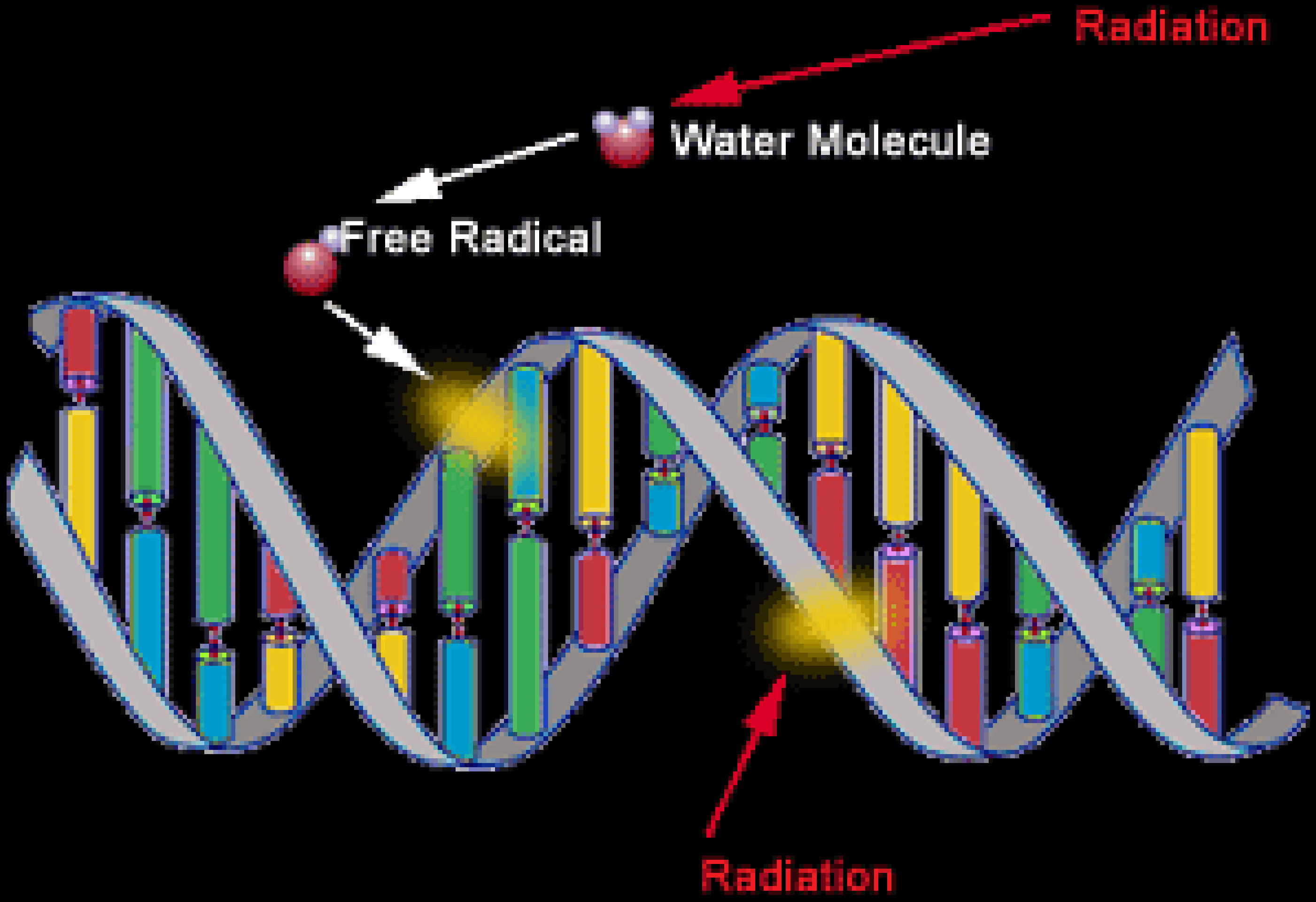
Indirect effects



Direct effects

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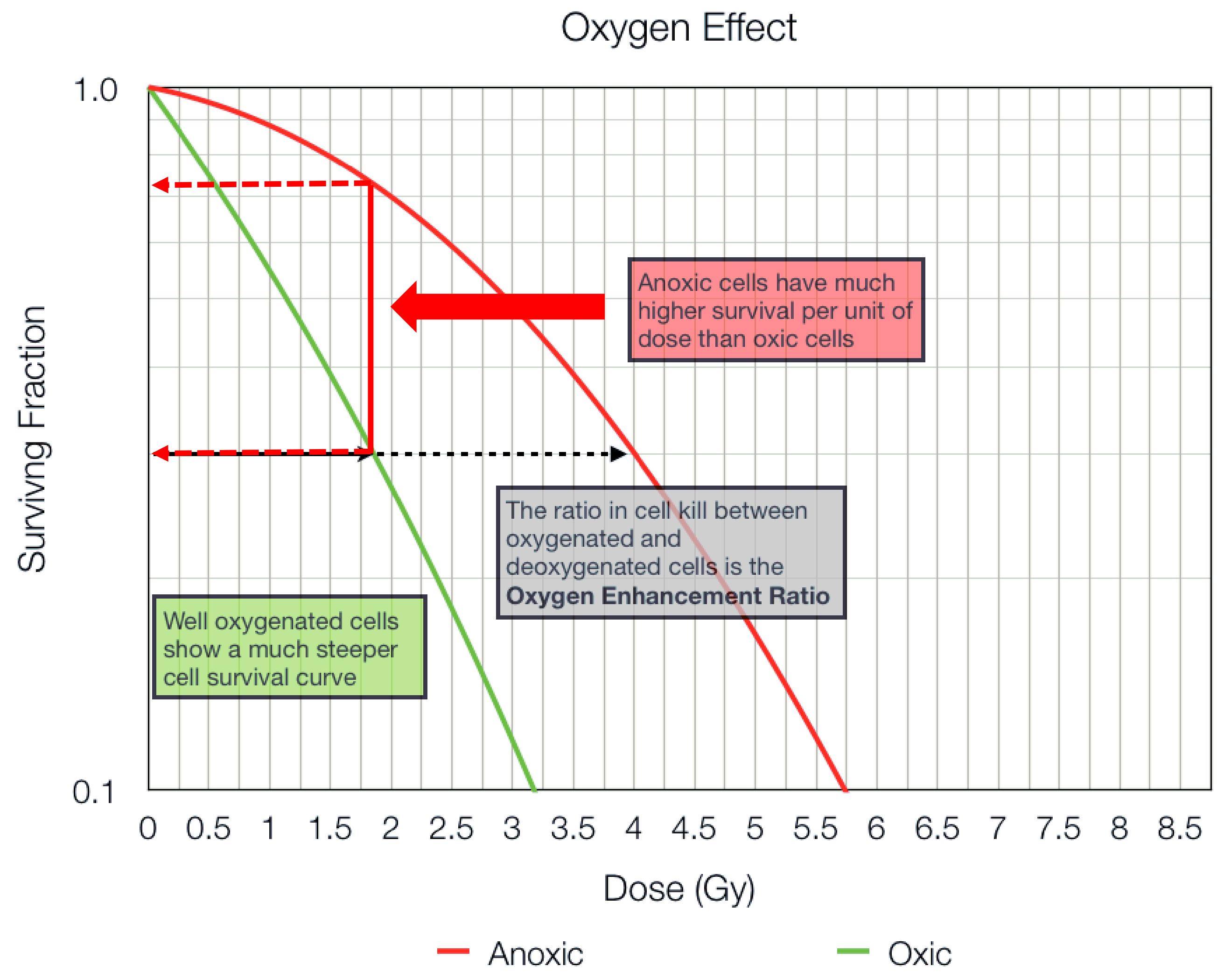
Oxygen Enhancement Ratio (OER) for low LET radiation (γ and e^-)



Anoxic cells have much higher survival per unit of dose than oxic cells

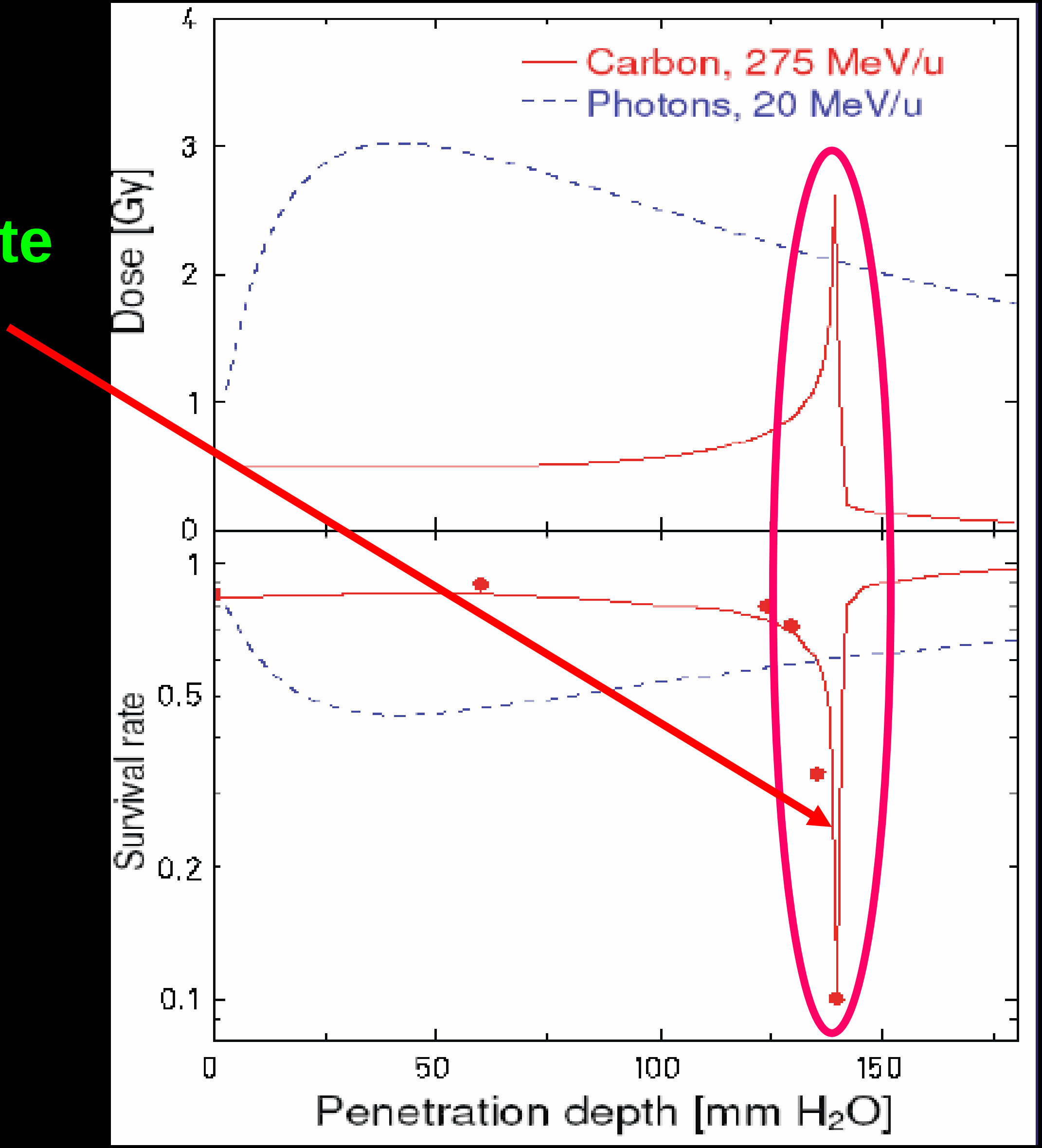
The ratio in cell kill between oxygenated and deoxygenated cells is the **Oxygen Enhancement Ratio**

Well oxygenated cells show a much steeper cell survival curve

$$\text{OER} = \frac{\text{Dose to produce a certain effect under hypoxic condition}}{\text{Dose to produce the same effect under oxic condition}}$$

Cell Survival

Low cell survival rate at Bragg Peak



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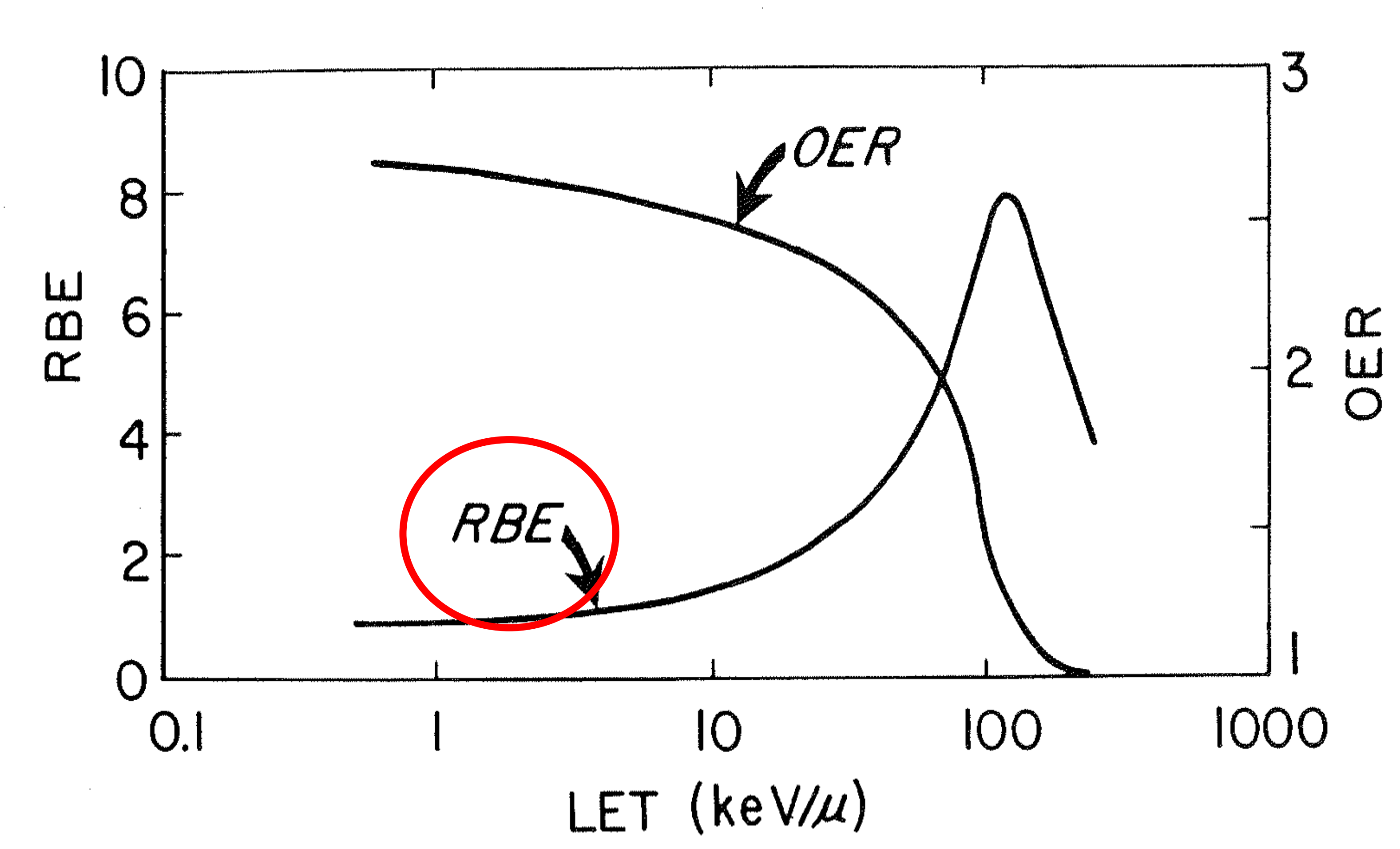
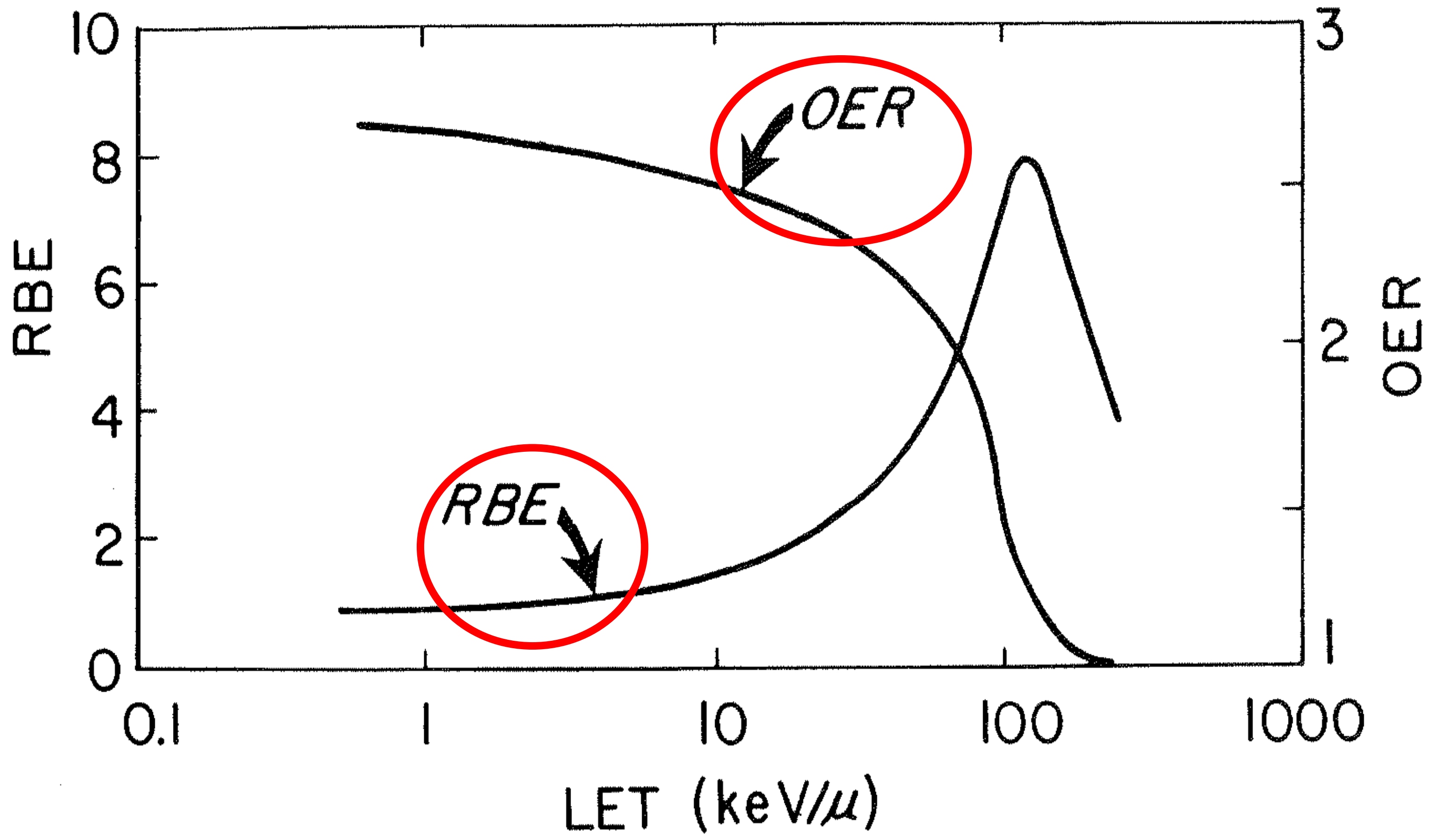
Relative Biological Effectiveness (RBE) vs LET



$$RBE = \left[\frac{D(X-rays/250kV)}{D_T(test-radiation)}\right]_{iso\,effect}$$

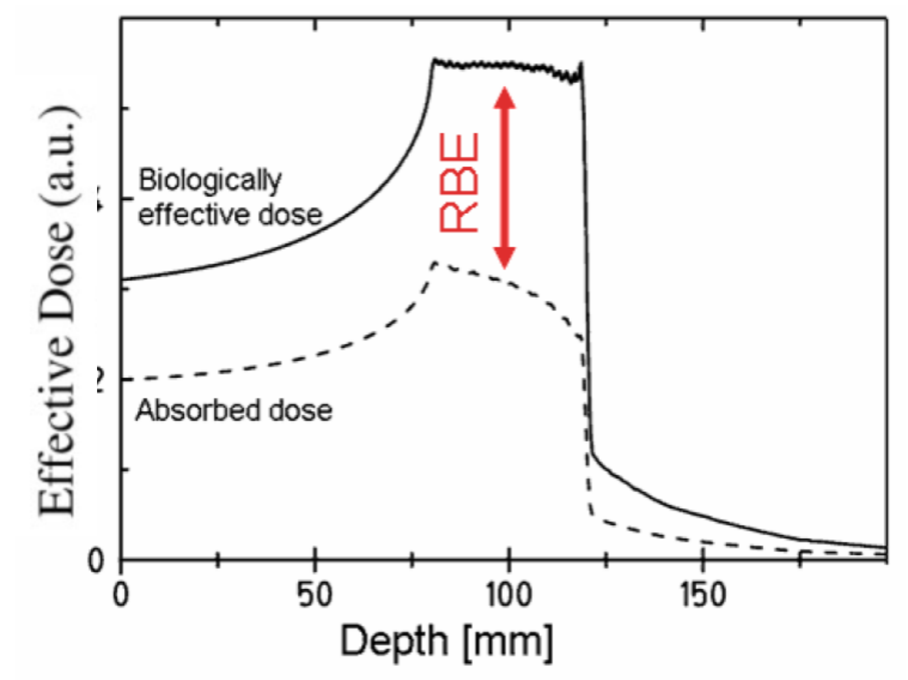
Oxygen Enhancement Ration (OER) vs LET



$$RBE = \left[\frac{D(X - rays / 250kV)}{D_T(test - radiation)} \right]_{isoeffect}$$

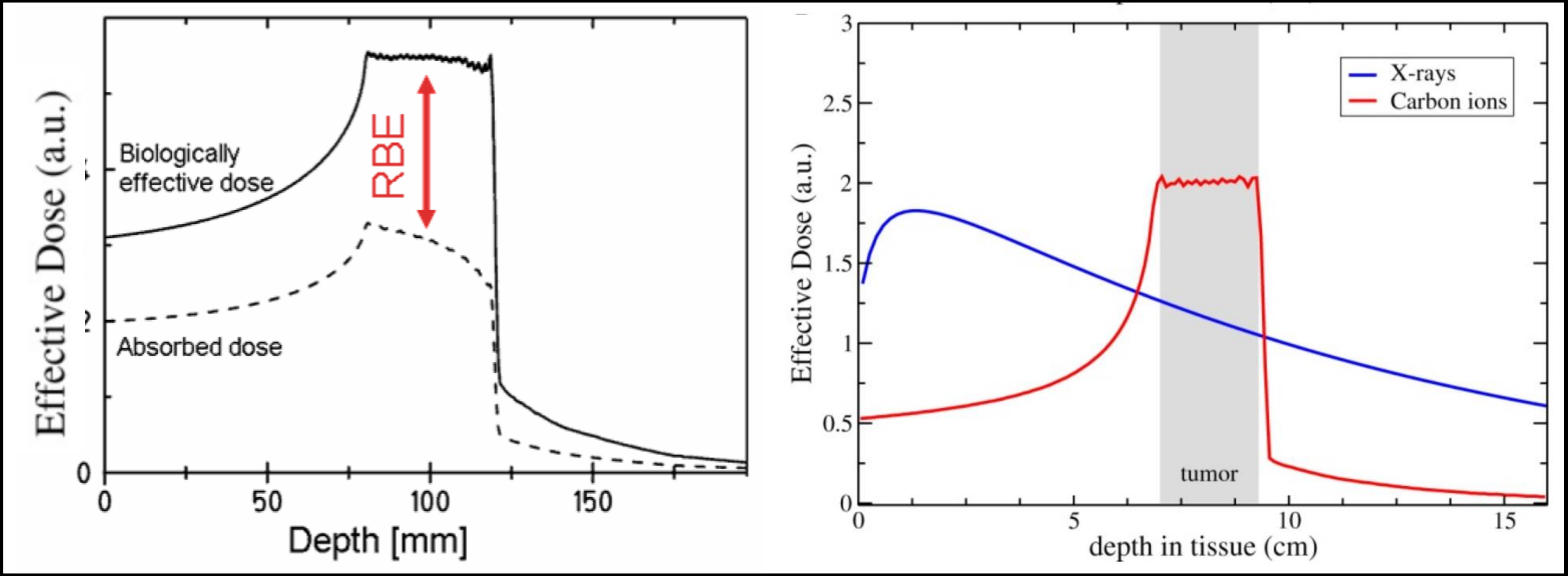
Biologically Effective Dose

Biologically Effective Dose = Physical Dose × RBE



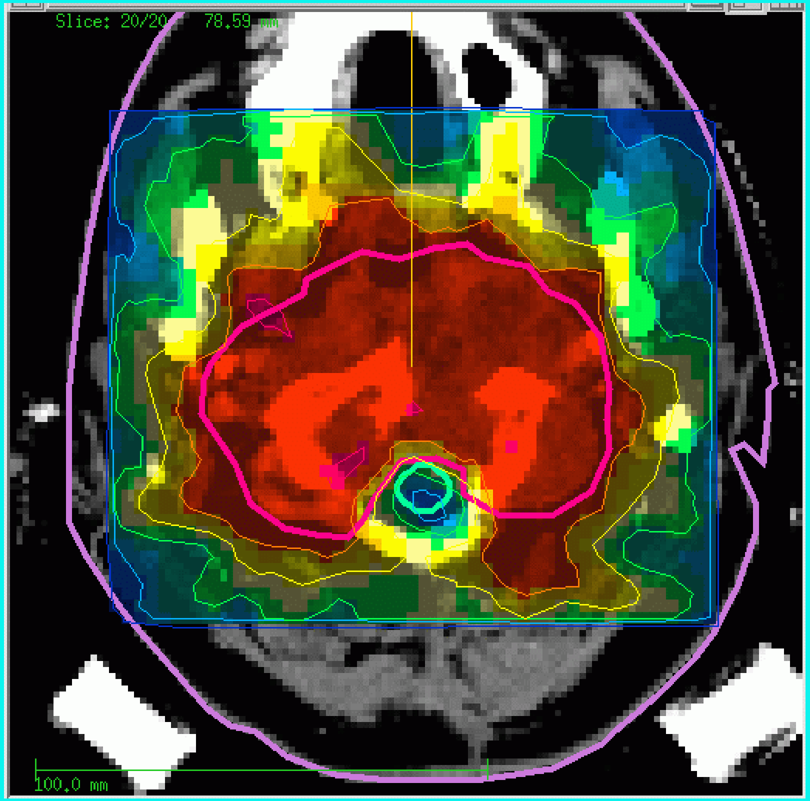
Biologically Effective Dose

Biologically Effectvie Dose = Physical Dose × RBE

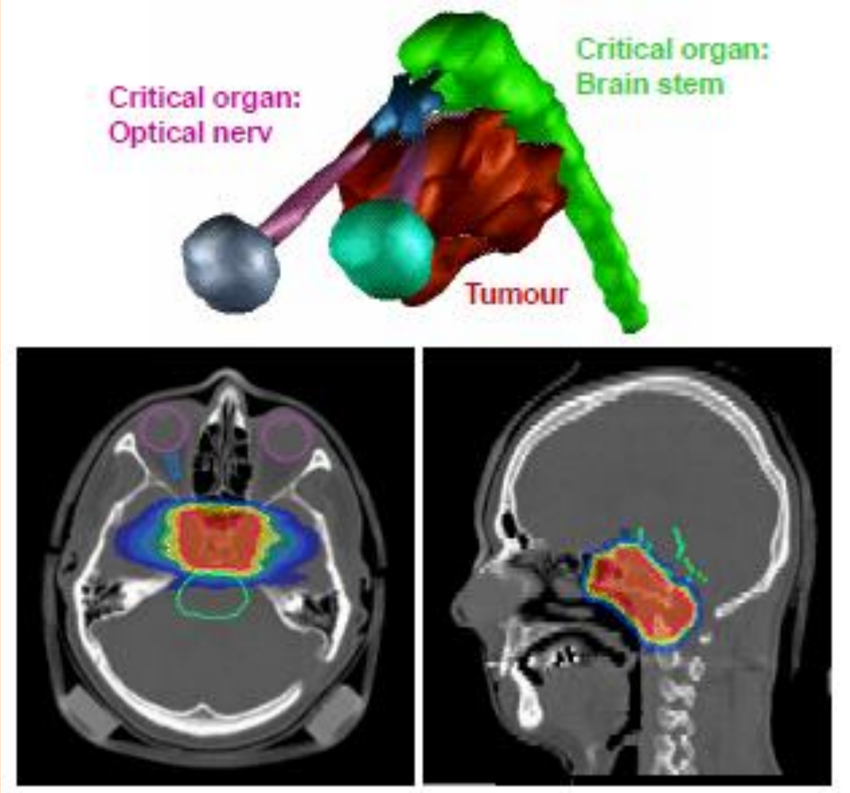


The superior dose-conformality of ion beams
Biological Dose

Photon IMRT



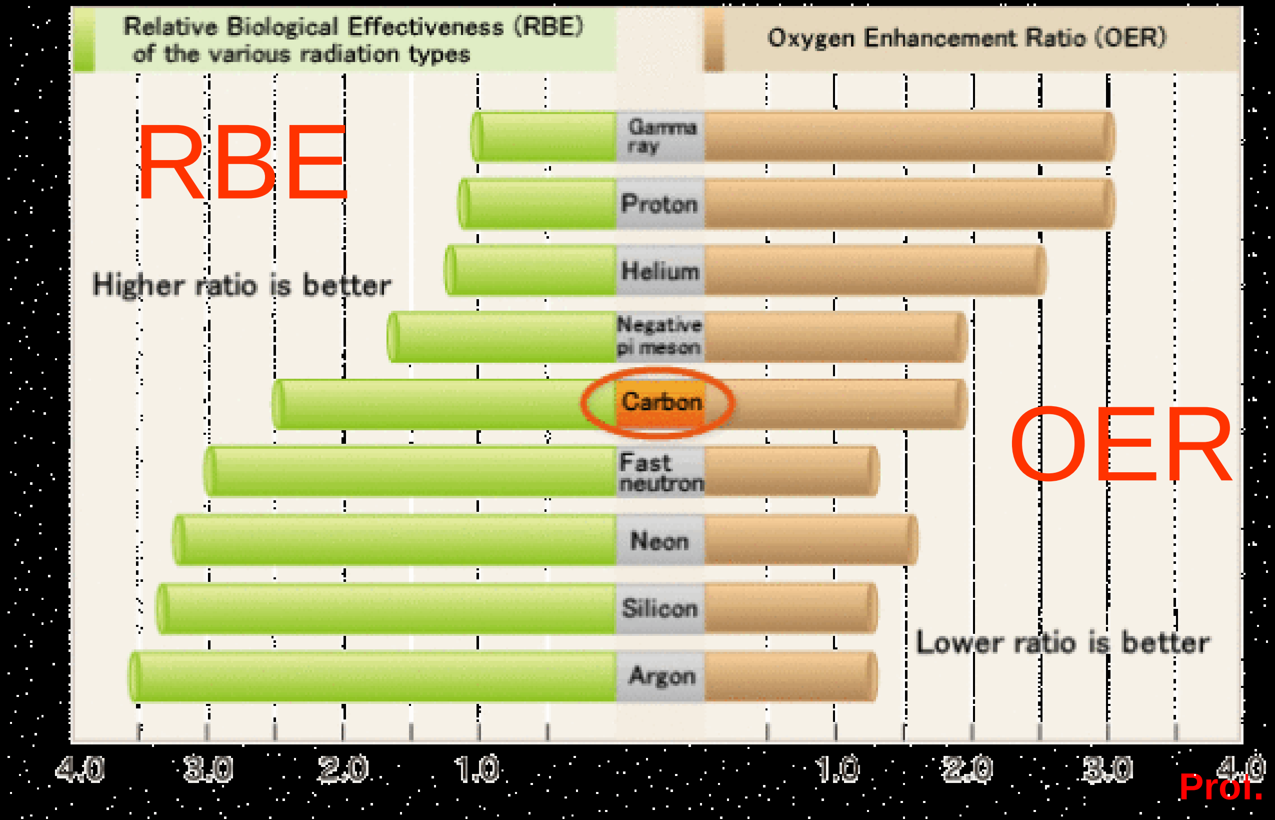
Carbon ions: 2 Fields



T. Bortfeld, S. Nill, U. Oelfke, O. Jäkel, DKFZ Heidelberg

Why use carbon beams for radiotherapy ?

- High Relative Biological Effectiveness (RBE)
- Low Oxygen Enhancement Ratio (OER)
    - Effective against hypoxic ("lack of oxygen") tumor cells
- Carbon ions have max in RBE close to the Bragg Peak
- Limited amount of projectile and target fragmentation, etc…



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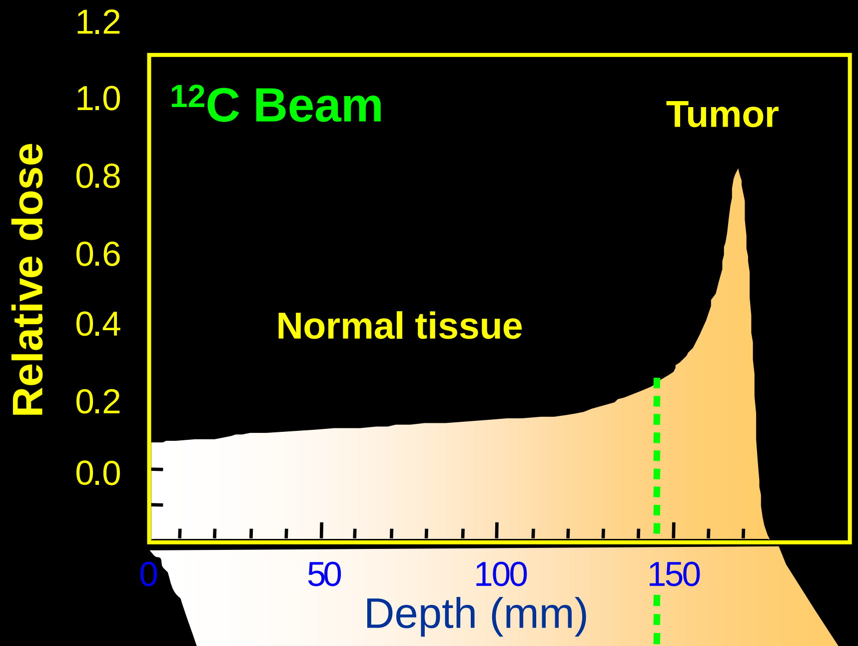
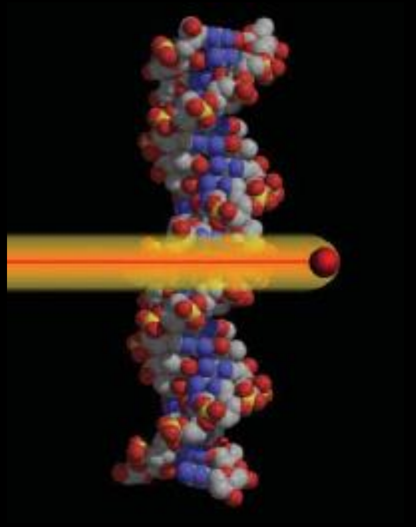
Durante & Loeffler,
Nat Rev Clin Oncol 2010



Energy	high	low
LET	low	high
Dose	low	high
RBE	≈ 1	> 1
OER	≈ 3	< 3
Fractionation dependence	high	low
Cell-cycle dependence	high	low
Angiogenesis (development of new blood vessels.)	↑	↓
Cell migration	↑	↓



Potential advantages

High tumor dose, normal tissue sparing

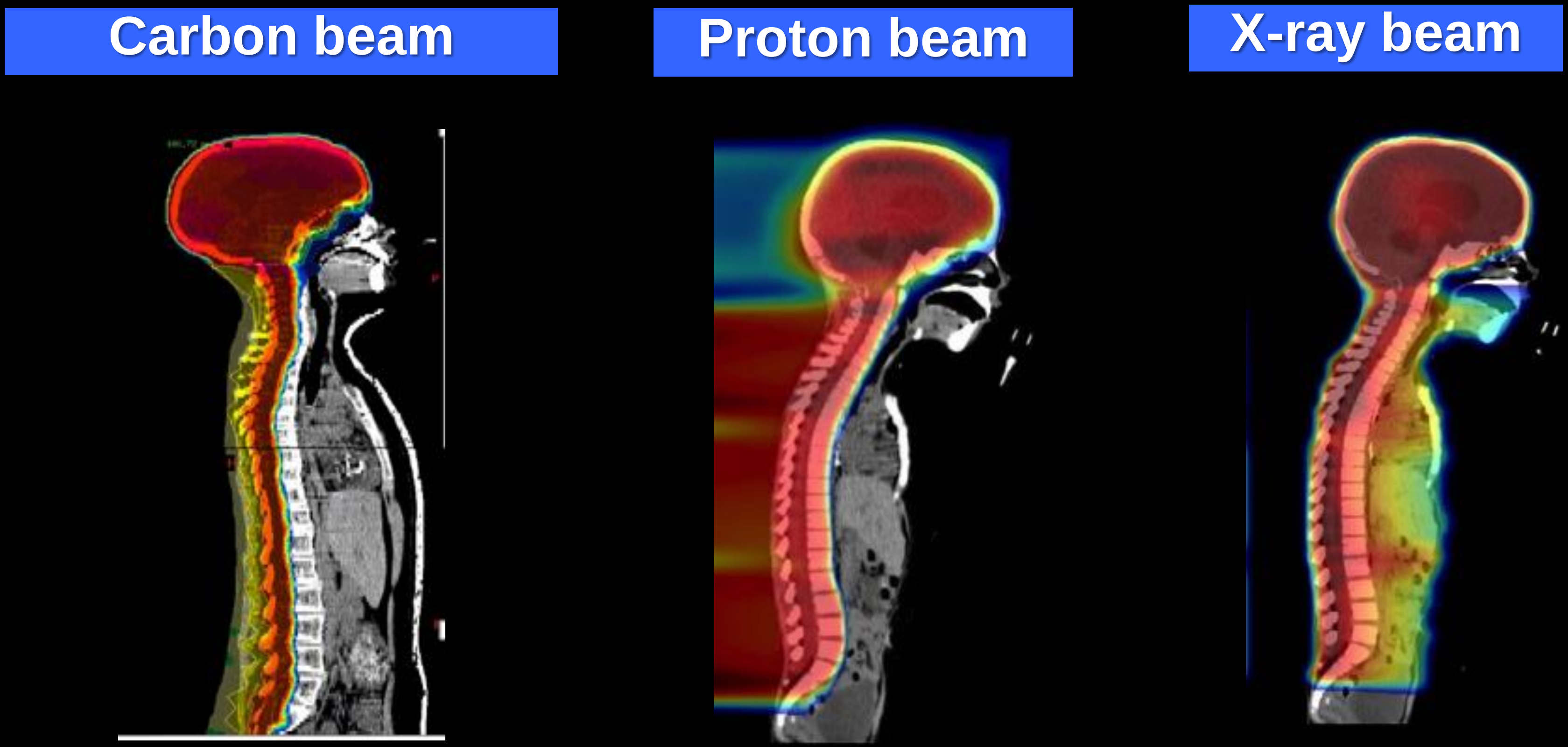
Effective for radioresistant tumors

Effective against hypoxic tumor cells

Fractionation spares normal tissue more than tumor

Increased lethality in the target because cells in radioresistant (S) phase are sensitized

Pediatric cancer (e.g. medulloblastoma) - representative for the superior dose distribution using particles



Carbon beam

Proton beam

X-ray beam

Intention to treat spinal cord and brain only!

Miralbell R et al., Int J Radiat Oncol Biol Phys 54: 824-829, 2002)

But we need to be sure that we can measure the absorbed dose correctly!

Properties of an Ideal Dosimeter

- Linear over whole dose range
- Dose rate independence (non-linear effects at higher dose rates, e.g. recombination effects, no fading for passive detectors)
- No energy/particle dependent response
- No directional dependence
- High spatial resolution (image, small effective volume)
- Online active readout
- Easy to use, easy to set up, reliable, …

Commonly used Dosimeters

- Calorimeter (real time, absolute dosimetry)
- Fricke dosimeter (passive detector)
- Ionization chambers (active detector)
- Radiochromic films (passive detector for relative profile and homogeneity control)
- TLDs / OSL (passive detectors)
- Si diodes / Diamonds ...(active detectors)
- Scintillators (active detectors)
- …

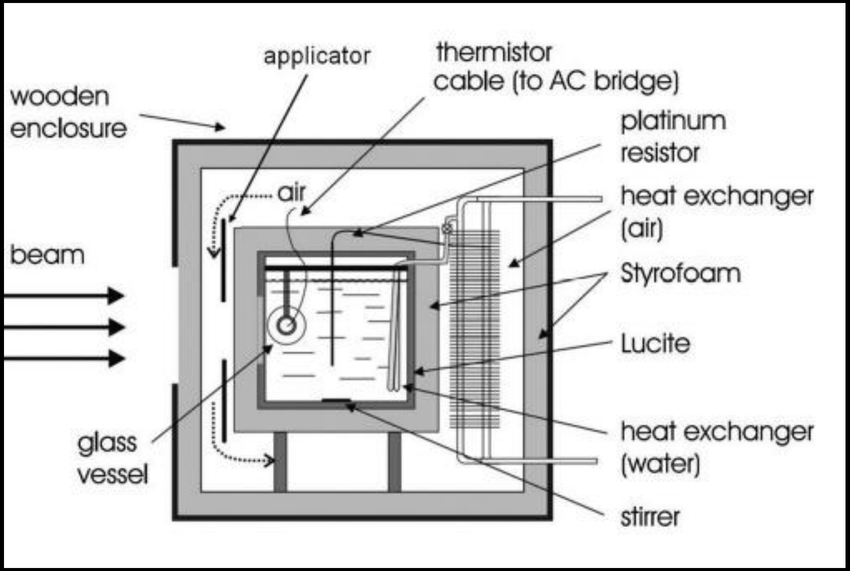
Dosimetry

Standards to measure absorbed dose to water

- The only solid state dosimeters used for absolute dosimetry in radiotherapy
    - Calorimeter
    - Fricke chemical detector
        - Fricke solution – (1mM $FeSO_4$ or $Fe(NH_4)_2(SO_4)_2$ + 0.8N H_2SO_4 air saturated + 1mM NaCl)
        - Fe^{2+} oxidizes to Fe^{3+}, when irradiated, which absorbs at 304 nm
- Ionization chamber

Calorimeter

- **Water calorimeter**
  - **In a stagnant water calorimeter, absorbed dose at a point can be directly measured because of the relatively low thermal diffusivity of water: $D_w = c_w \Delta T_w$**
    ΔT_w = temperature increase in undisturbed water solely due to radiation,
    c_w = the specific heat capacity of water (in J/kg,K)



  - **Not practical for clinical use**

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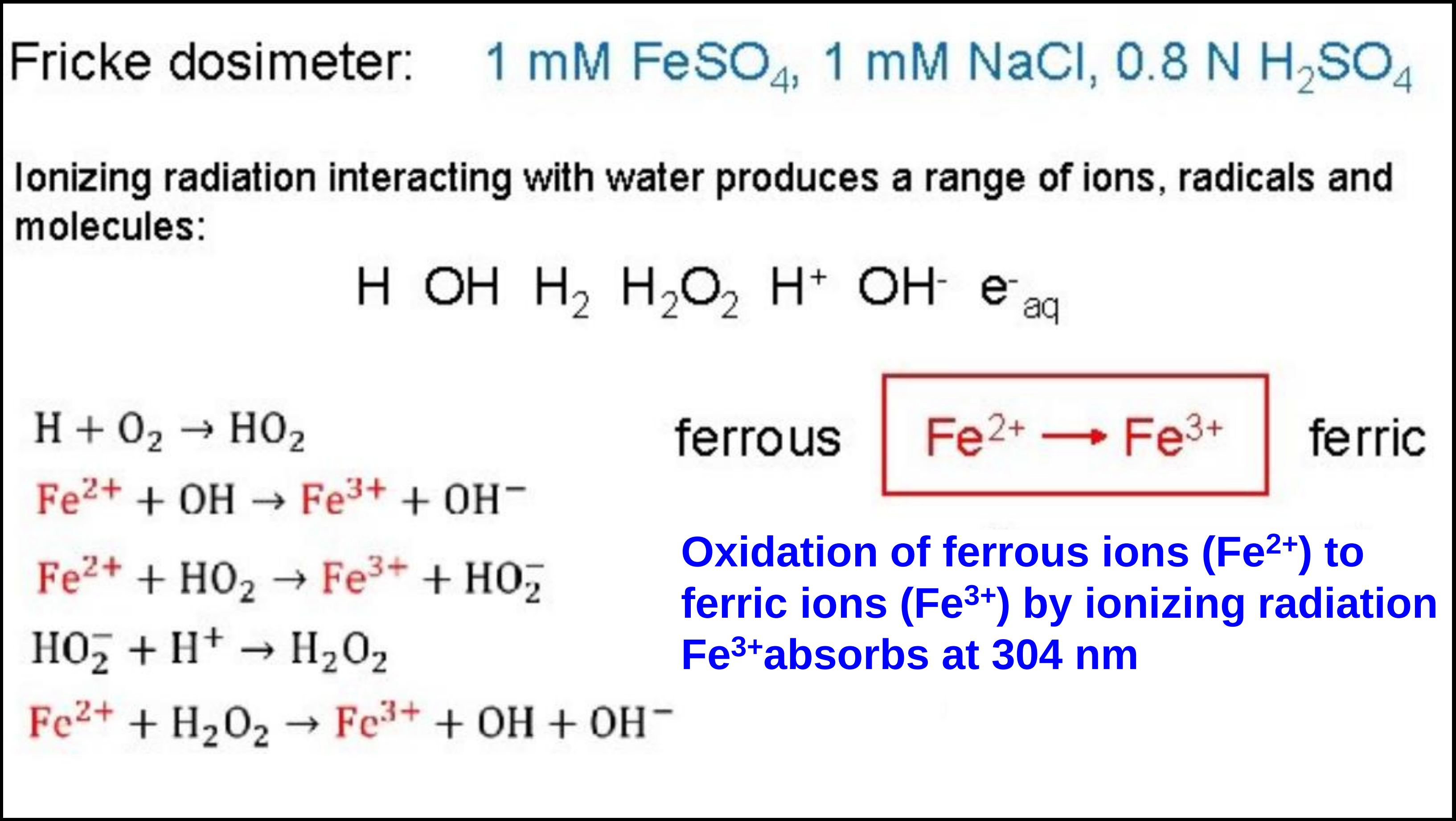
Fricke Dosimeter



Fricke dosimeter: 1 mM $FeSO_4$, 1 mM NaCl, 0.8 N H_2SO_4

Ionizing radiation interacting with water produces a range of ions, radicals and molecules:

$$H \quad OH \quad H_2 \quad H_2O_2 \quad H^+ \quad OH^- \quad e^-_{aq}$$

$$H + O_2 \rightarrow HO_2$$

$$Fe^{2+} + OH \rightarrow Fe^{3+} + OH^-$$

$$Fe^{2+} + HO_2 \rightarrow Fe^{3+} + HO_2^-$$

$$HO_2^- + H^+ \rightarrow H_2O_2$$

$$Fe^{2+} + H_2O_2 \rightarrow Fe^{3+} + OH + OH^-$$

ferrous $\boxed{Fe^{2+} \longrightarrow Fe^{3+}}$ ferric

Oxidation of ferrous ions (Fe^{2+}) to ferric ions (Fe^{3+}) by ionizing radiation
Fe^{3+}absorbs at 304 nm

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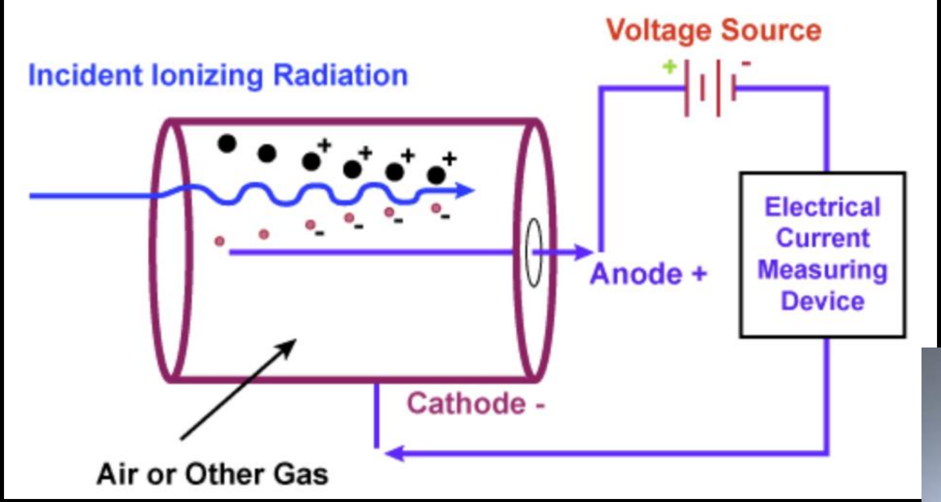
Ionization Chamber (IC)

- The standard is an air filled ionization chamber.



- Ionization of the molecules of the gas occur.
- Within a high voltage field positive ions will be attracted to the negative side of the detector (the cathode) and the free electrons will travel to the positive side (the anode). These charges are measured.

Dose in Water

The dose in water from charged particles:

$$D_w = \int_0^{E0} \phi_E \left(\frac{dE}{\rho dx}\right)_w dE \quad [Gy]$$

Ionization Chamber (IC)

The dose from charged particles:

$$D_w = \cancel{\int_0^{E0} \phi_E^e \left(\frac{dE}{\rho dx}\right)_w dE}$$

- **Not applicable in clinics, since fluence spectra are unknown**
- **We want dose to the tissue equivalent material (water), but IC is normally filled with air!**

$$D_w = \frac{QW_{air}}{\rho V}$$

Ionization Chamber (IC)

The dose from charged particles:

$$D_w = \xcancel{\int_0^{E0} \phi_E^e \left(\frac{dE}{\rho dx}\right)_w} dE$$

- **Not applicable in clinics, since fluence spectra are unknown**
- **We want dose to the tissue equivalent material (water), but IC is normally filled with air!**

$$D_w = \frac{QW_{air}}{\rho \boxed{V}} \longrightarrow$$

The volume of the gas filled cavity is not know with sufficient accuracy!

Calibrartion of Ionization Chamber (IC)

- When an IC is used in a beam of quality Q different from that used for its calibration, Q_o, the absorbed dose to water is given by:

$$D_{w,Q} = M_Q \, N_{D,w,Q_o} \, k_{Q,Q_o}$$

Absorbed dose in water

Measured signal

Calibration factor at reference condition

Beam quality correction factor of the chamber to differenciate between the reference beam quality Q_o and the actual treatment beam quality Q

Calibrartion of Ionization Chamber (IC)

- **When an IC is used in a beam of quality Q different from that used for its calibration, Q_o, the absorbed dose to water is given by:**

$$D_{w,Q} = M_Q \, N_{D,w,Q_o} \, k_{Q,Q_o}$$

↓

$$k_{Q,Q_o} = \frac{N_{D,w,Q}}{N_{D,w,Q_o}} = \frac{D_{w,Q} / M_Q}{D_{w,Q_o} / M_{Q_o}}$$

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Calibrartion of Ionization Chamber (IC)

A general expression o $k_{Q,Qo}$ is:

$$k_{Q,Q_o} = \frac{\left(s_{w,air}\right)_Q}{\left(s_{w,air}\right)_{Q_o}} \frac{\left(W_{air}\right)_Q}{\left(W_{air}\right)_{Q_o}} \frac{p_Q}{p_{Q_o}}$$

Stopping-power ratios signal water-to-air

Mean energies to produce an electron-ion-pair in air

Overall perturbation factors, including all departures from the ideal Bragg-Gray detector conditions, i.e., p_{wall}, p_{cav}, p_{cel}, $p_{fluence}$, p_{dis}, …

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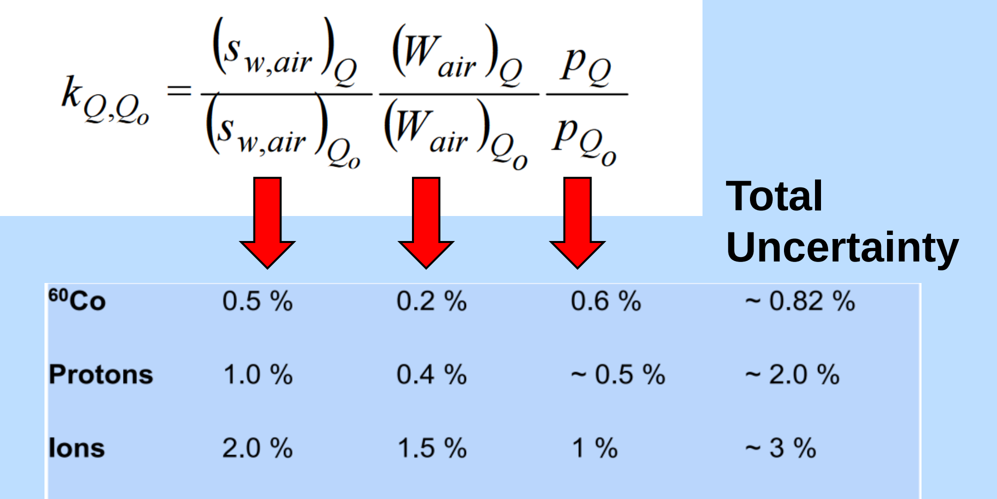
Uncertainty in the Measured Dose

- It has been estimated* that the overall uncertainty is

  ≈ 2% of the dose for proton beams
  ≈ 3% for carbon ion beams

  which are high compared to the 1% of photon beams

- In particular, the largest contribution comes from the uncertainty in the ratio of stopping powers between water and air ($s_{w,air}$)



$$k_{Q,Q_o} = \frac{(s_{w,air})_Q}{(s_{w,air})_{Q_o}} \frac{(W_{air})_Q}{(W_{air})_{Q_o}} \frac{p_Q}{p_{Q_o}}$$

				Total Uncertainty
^{60}Co	0.5 %	0.2 %	0.6 %	~ 0.82 %
Protons	1.0 %	0.4 %	~ 0.5 %	~ 2.0 %
Ions	2.0 %	1.5 %	1 %	~ 3 %

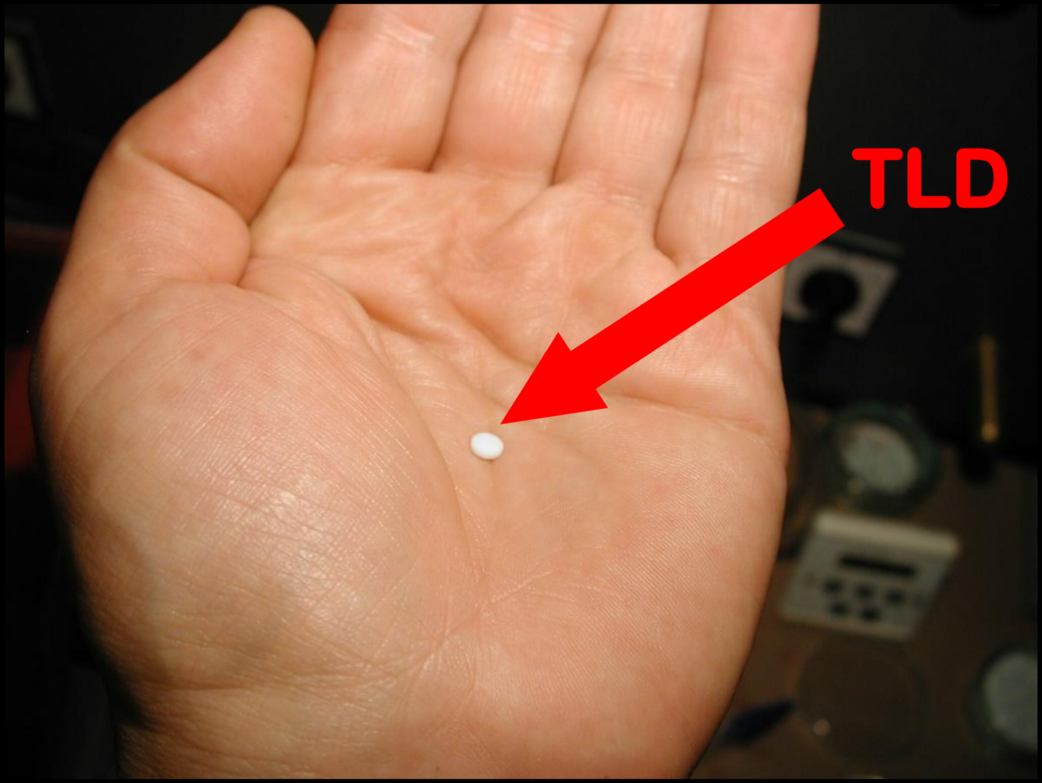
* IAEA 2000. Technical Report Series TRS-398

Uncertainty in the Measured Dose

- Unknown electronic stopping powers remains as the primary source of uncertainty for IC dosimetry
- Uncertainties in the mean excitation (I) values for water and compounds are important

$$-\left(\frac{dE}{dx}\right)_{coll} = 2\pi N_A r_e^2 m_e c^2 \rho \frac{Z}{A}\frac{z^2}{\beta^2}\left[\ln\left(\frac{2m_e\gamma^2 v^2 W_{\max}}{\boxed{I^2}}\right) - 2\beta^2 - \delta - 2\frac{C}{Z}\right]$$

Thermoluminescence Detectors (TLD)



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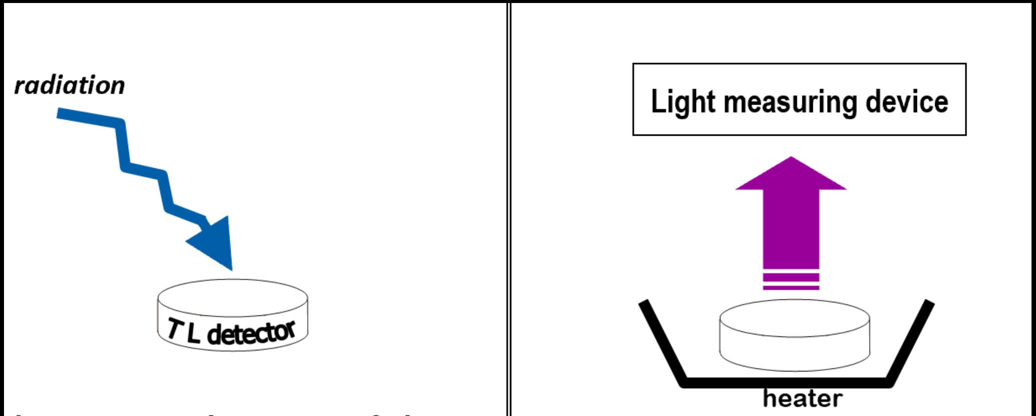
Thermoluminescence Detectors (TLD)

Typical TL materials

Lithium fluoride
Calcium sulphate
Calcium fluoride
Aluminium oxide

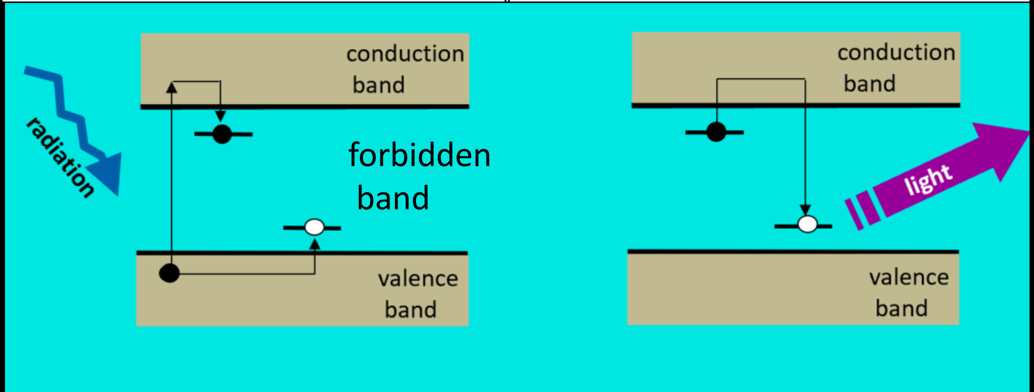
Thermoluminescence Detectors (TLD)

Thermoluminescence (TL) is a 2-stage process



Ionisation and trapping of charge carriers in trapping centers

After heating – recombination with light emission

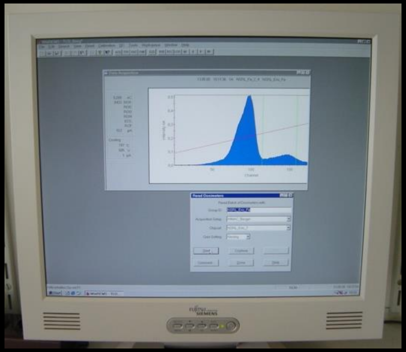


Ionisation and trapping of charge carriers in trapping centers

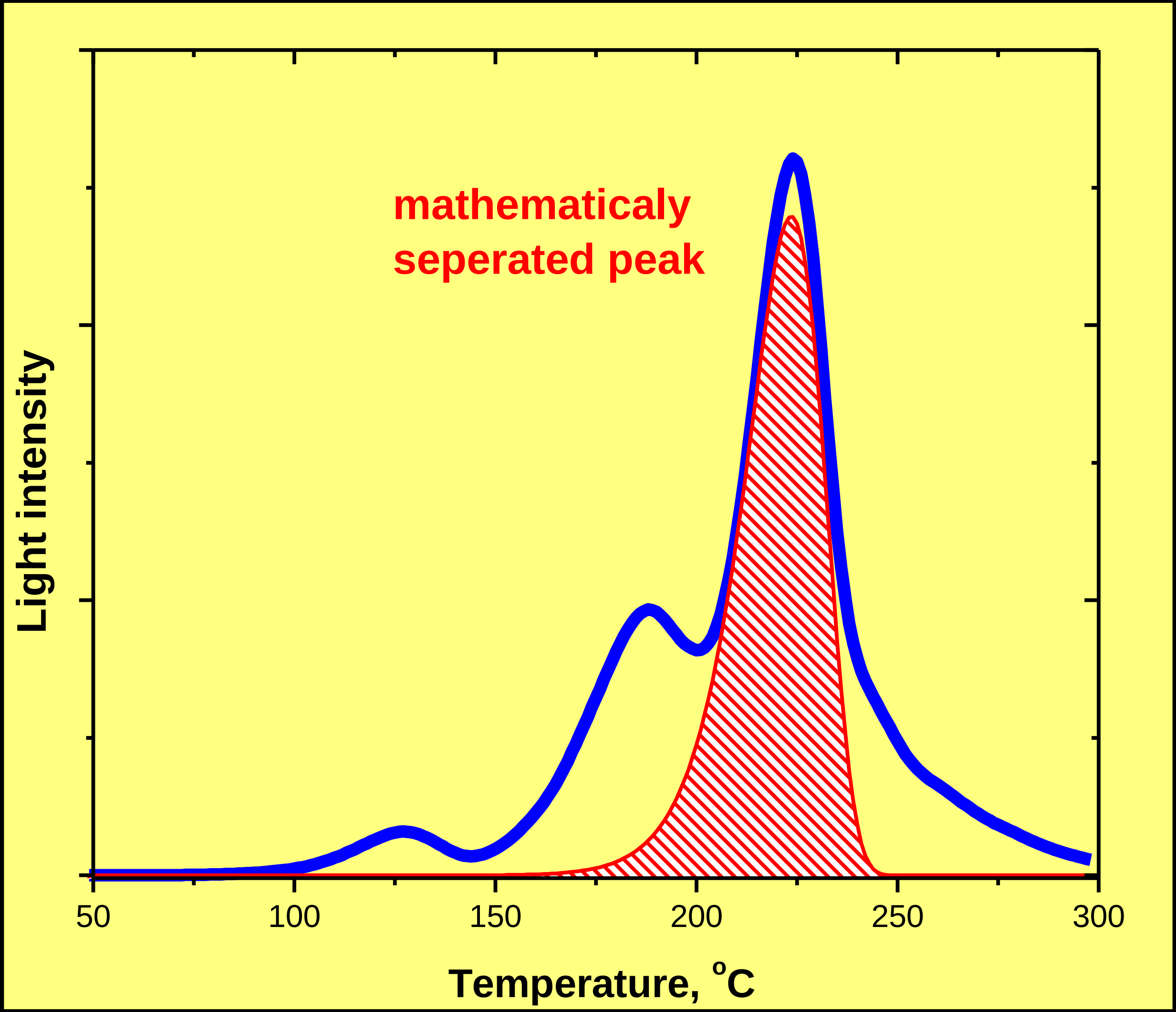
After heating – recombination with light emission

Thermoluminescence Detectors (TLD)

- **TL – detector is place on a heating plate**
- **TL – detector is heated**
- **The light emission is measured depending on the temperature with a photomultiplier**
- **This leads to the so called “Glow Curve“**



TL Glow Curve



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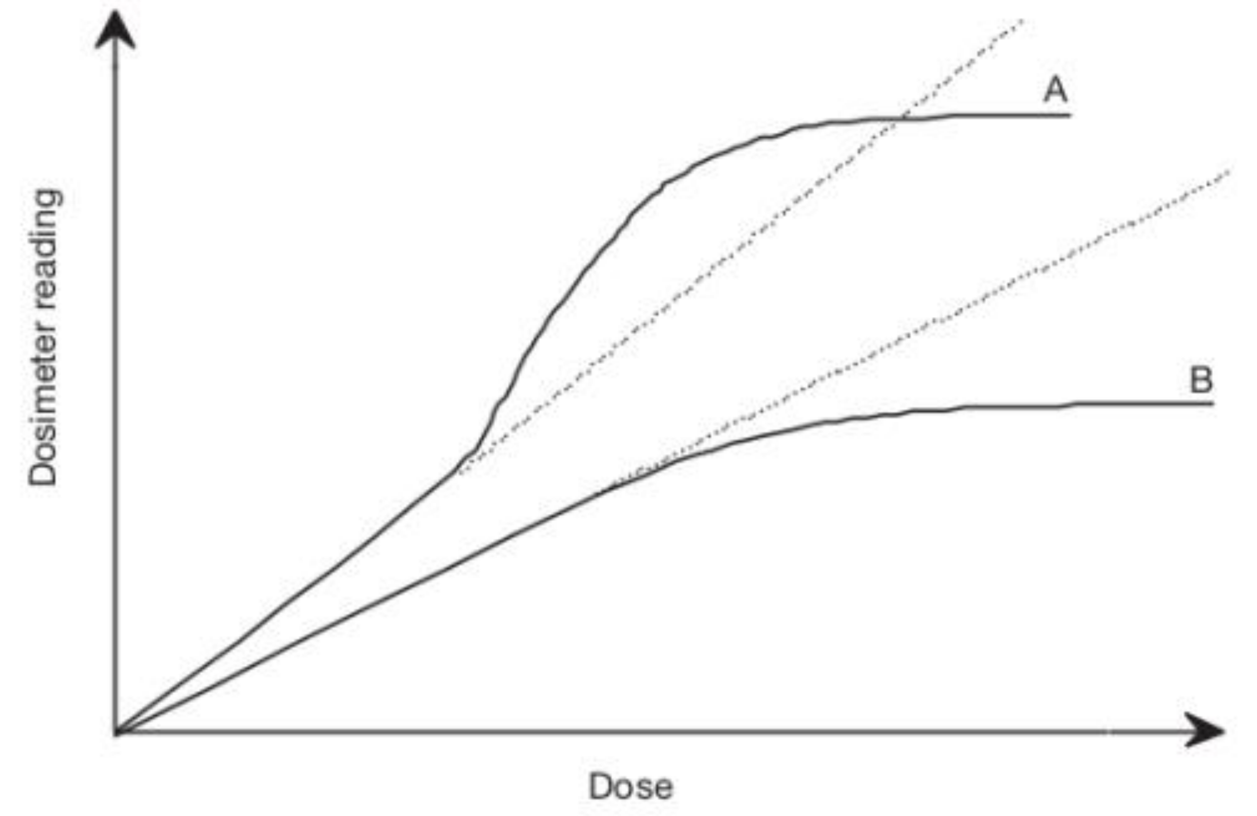
Linearity of a Dosimeter



A: Supralinearity and saturation
B: Sublinearity / saturation

Relative Effectiveness

- **Different types of radiation produce different effects in a dosimeter for the same endpoint**

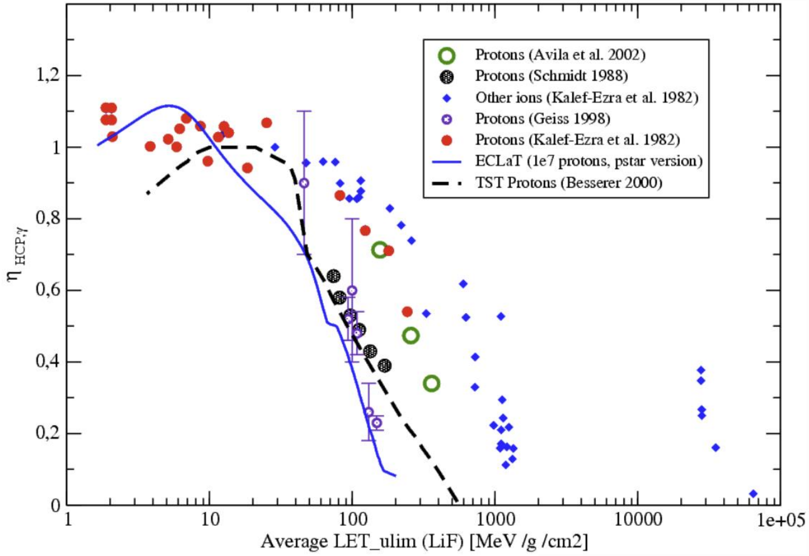
- ***"Relative effectiveness" (RE)*** **characterizes this effect and is usually defined as**

$$RE = D_{\gamma,isoeffect} / D_{ion,isoeffect}$$

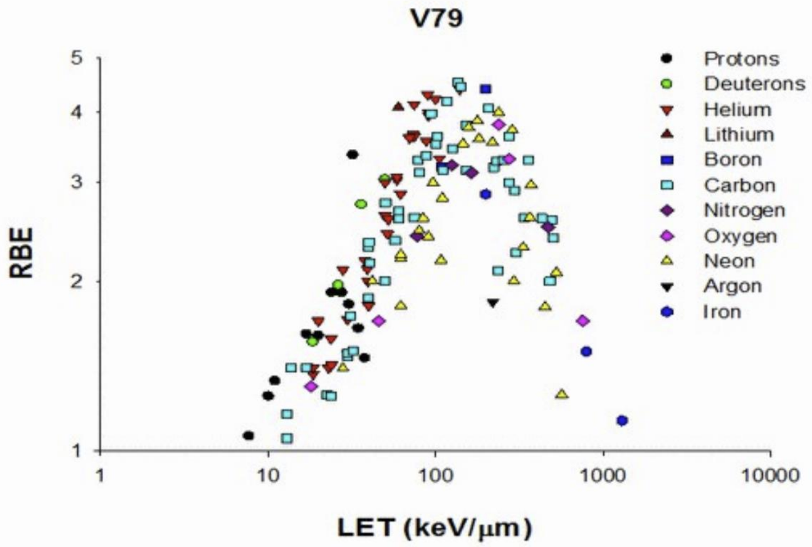
Dosimetry

Compare!

Detectors



Radiobiology

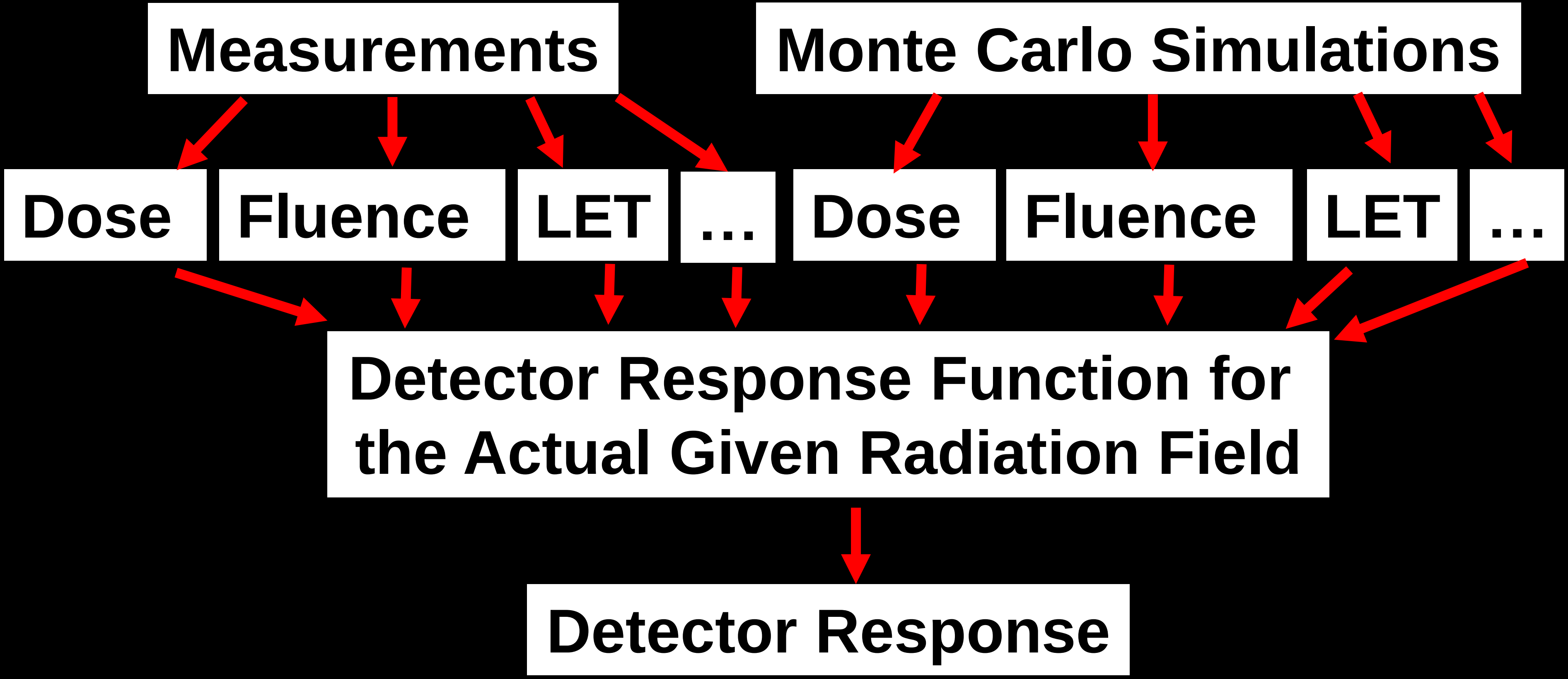


$$\eta_{Q_0,Q} = \frac{D_{Q_0}}{D_Q}\Big|_{\mathrm{iso-effect}}$$

$$\mathrm{RBE} = \frac{D_{Q_0}}{D_Q}\Big|_{\mathrm{iso-effect}}$$

Modelling of Detector Response Function

$$D_w = \int_0^{E0} \emptyset_E^e \left(\frac{dE}{\rho dx}\right)_w dE$$



Measurements

Monte Carlo Simulations

Dose

Fluence

LET

...

Dose

Fluence

LET

...

Detector Response Function for the Actual Given Radiation Field

Detector Response

MC Modelling k_Q factor of an IC using Response Factor

Calculation of the response factor (f_Q) of an IC:

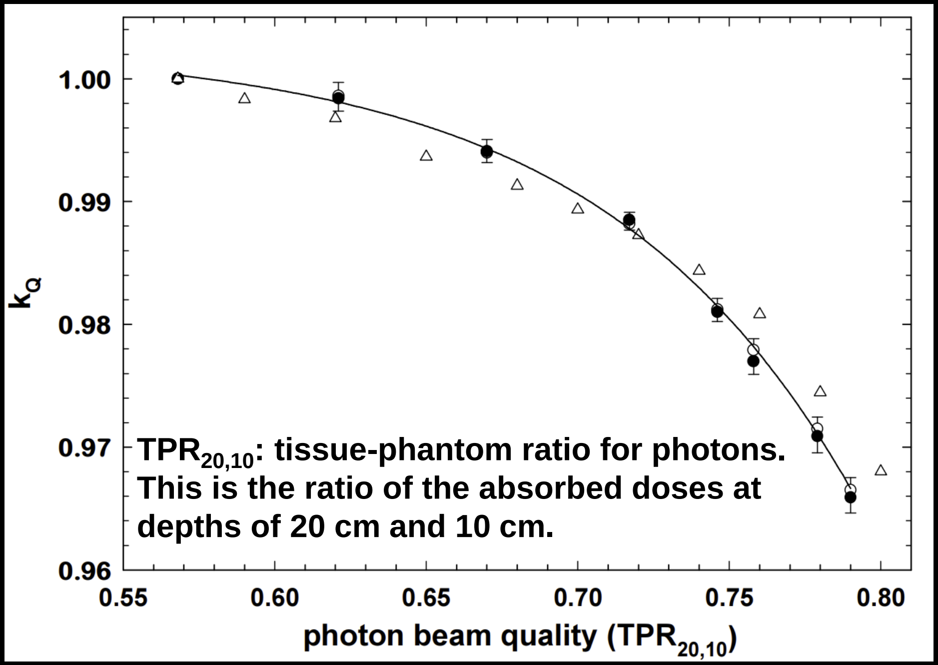
$$f_Q = \frac{D_w}{D_{det}} = (s_{w,air})\, p_Q$$

p_Q: perturbation factor

Calculation of beam quality correction factor k_Q:

$$k_Q = \frac{f_Q}{f_{Q0}} = \frac{(W_{air})_Q}{(W_{air})_{Q0}}$$

- For clinical photon beams, there is a good agreement between measures and Monte Carlo calculated values for k_Q, but there is a lack of data for protons and heavier ions



$TPR_{20,10}$: tissue-phantom ratio for photons. This is the ratio of the absorbed doses at depths of 20 cm and 10 cm.

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Needed Dosimetry Improvements

- Improved accuracy of the ionization potential/range in water and other tissue/bone equivalent materials
- Micro- and nanodosimetric ionization density (ID)/track structure measurements
- On-line detection of dose and LET/ID
- For non-gaseous detectors, the sensitivity depends on the ID so when ID increase at the Bragg peak, the sensitivity decreases (quenching of the response)
- Dosimetry in strong magnetic fields during combined MRI and RT
- Dosimetry for FLASH radiotherapy (RT)
  - ultra-high dose rate (UHDR), ≥ 40 Gy/s

….

Schematic Comparison of different Types of Radiations (Raju, 1974)

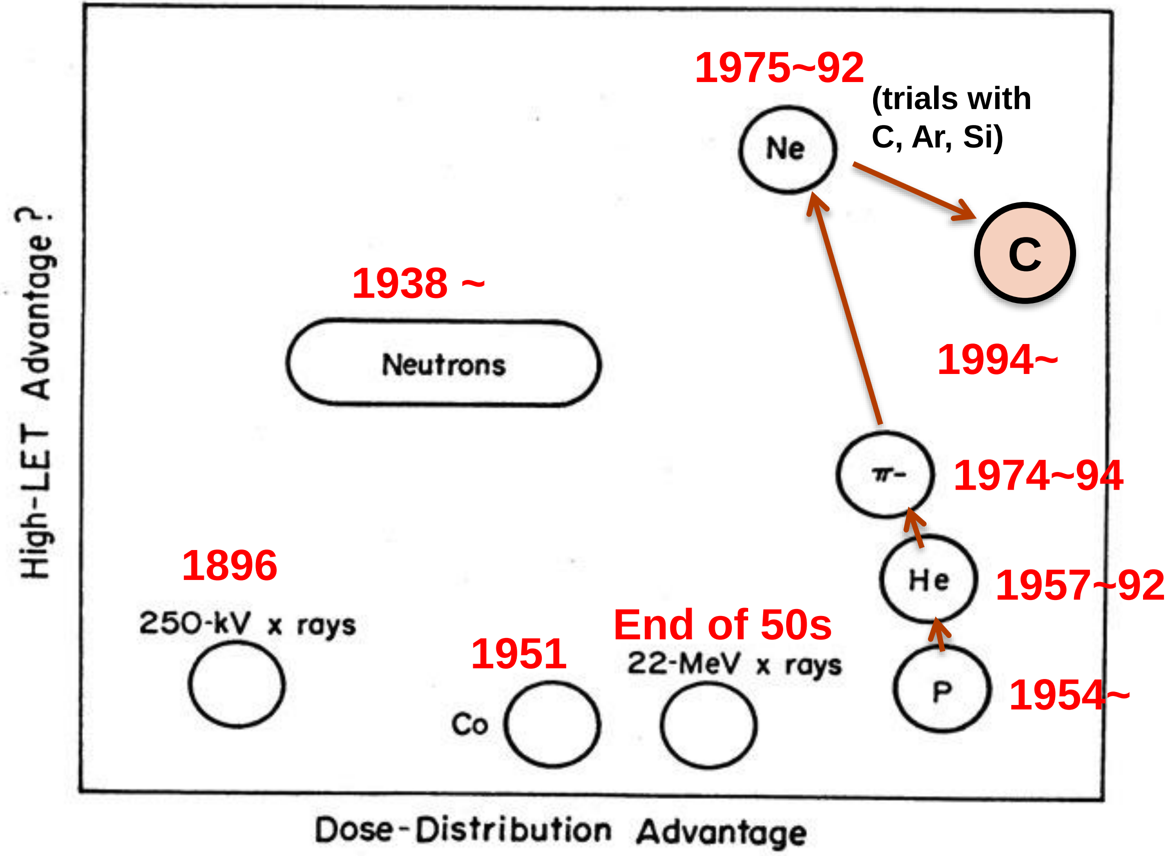
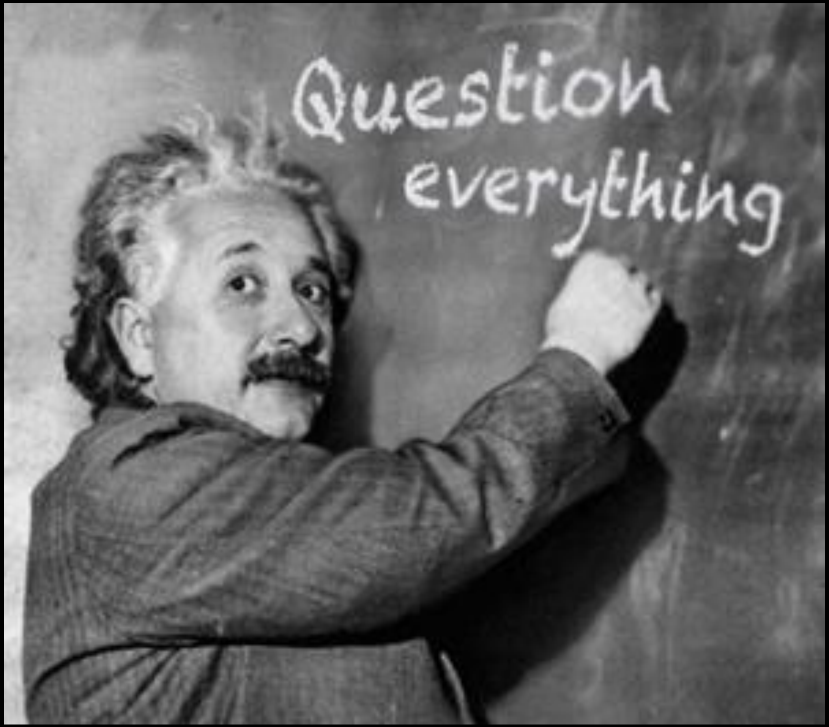


Fig. 17. Schematic comparison of different types of radiations of interest in radiotherapy. Although based on experimental results, this comparison is an oversimplification. It is recognized that the relative positions of the particles in this schematic diagram can be argued and should be used for clarity purposes only. RBE is not taken into consideration in the dose-distribution advantage (abscissa). This idea of representing the relative merits of particles was suggested by Mr. A. M. Koehler (Raju *et al.*, 1974b).

- 1946 Ion beam therapy proposal by R.Wilson
- 1968 CT invented
- ?? FLASH RT

Thank you very much for your kind attention!!

Questions?



Comments?

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