Radiation Units Exposure & Dose

Exposure: $X = \frac{Q}{m}\ \left(\frac{C}{kg}\right)$, $X = \frac{A\Gamma t}{d^2}$ Brachy Dose: $D = \frac{A\Gamma t}{d^2}$

Exposure rate: $\dot{X} = \frac{A\Gamma}{d^2}$ Brachy Dose Rate: $\dot{D} = \frac{A\Gamma t}{d^2}$

Air Kerma Strength: $S_k = A\Gamma f\ \left(\frac{Cgy\ cm^2}{hr}\right)$, $f_{air} = 0.876\frac{cGy}{R}$, $f_{water} = 0.97\frac{cGy}{R}$

$1\,rad = 100\frac{erg}{g}$, $1\,rad = 1\,cGy$, $100\,rad = 1\,Gy$, $Gy = \frac{J}{kg}$, $100\,rem = 1\,Sv\,(w_t)$

$Dose\ to\ Ion\ Chamber$: $D = \frac{Q}{m}\left(34\frac{J}{C}\right)$, $34\frac{eV}{ion\ pair}$, $\rho_{air} = 1.3 * 10^{-3}\frac{g}{cm^3}$

Radioactive Activity: $A = \lambda N$, $\lambda = \frac{\ln ln\,(2)}{T_{\frac{1}{2}}}$, $N = N_o e^{-\lambda t}$, $A = A_o e^{-\lambda t}$,

$1\,Ci = 3.7 * 10^{10}\,Bq\left(\frac{dis}{sec}\right)$

Mean Life: $\tau = \frac{1}{\lambda} = 1.443T_{\frac{1}{2}}$, $A * \tau = Total\ Decays$, $D_{cumulative} = \dot{D}_0\tau$

Radium Equivalent: $X\ mCi * \frac{\Gamma_0}{\Gamma_{ra226}} = Y\ mg\ Ra\ eq$

Half Life: $T_{\frac{1}{2}} = \frac{\ln ln\,(2)}{\lambda}$, $\frac{A}{A_0} = \frac{N}{N_0} = \frac{1}{2^n}$, $Effective\ Half\ life$: $\frac{1}{T_{\frac{1}{2}eff}} = \frac{1}{T_{\frac{1}{2}bio}} + \frac{1}{T_{\frac{1}{2}phy}}$, $t_e = \frac{t_{\frac{1}{2}}t_b}{t_{\frac{1}{2}}+t_b}$

Equilibrium: $Transient\ eq \rightarrow T_{\frac{1}{2}parent} > T_{\frac{1}{2}daughter}$, $Secular\ eq \rightarrow T_{\frac{1}{2}parent} >> T_{\frac{1}{2}daughter}$

$Transient$: $A_d = A_p * \left(\frac{\lambda_d}{\lambda_d - \lambda_p}\right) * e^{-\lambda_p t}$, $Secular$: $A_d = A_p * \left(1 - e^{-\lambda_d t}\right)$

Time of Max Daughter Activity: $t_{max} = \frac{\ln ln\left(\frac{\lambda_d}{\lambda_p}\right)}{\lambda_d - \lambda_p}$

Required # of TVL's : n=-log(B)

GENERAL

Energy of a charged particle: 1 amu = 931.5 MeV = 1.602176634×10⁻¹⁹ J

- Energy, $E = \frac{hc}{\lambda}$, $c = v\lambda$,
  - hc = 1.986*10^-25 (J/m) = 1.2398 eV/um
  - λ = wavelength
  - v = frequency (hz)
- Kinetic Energy Relativistic, $KE_{rel} = (\gamma - 1)m_0c^2$
  - $m_0c^2 = 0.511\ MeV$ is the rest mass of an electron.

- $\gamma = \frac{1}{\sqrt{1-\frac{v^2}{c^2}}}$, relativistic mass is γm

Dead Time(Resolving time)

Paralyzable: $m = ne^{-n\tau}$, m is measured count rate, n is true count rate, τ resolving time.

- $m_{max} = \frac{1}{\tau e}\ when\ n = \frac{1}{\tau}$

Non-paralyzable: $n = \frac{m}{1-m\tau}$, m is measured count rate, n is true count rate, τ resolving time.

Atoms per gram:

$N = \frac{N_a}{A_w}$, $N_a is\ avagadro's\ number\ 6.02 * 10^{23} \frac{Atoms}{mol}$, $A_w is\ atomic\ weight\ in \frac{g}{mol}$

Specific Activity: $SA = \lambda N$

TG-51 Photon Calculations(Reference is 10cm depth with a 10x10cm^2)

$$M = M_{raw} P_{ion} P_{Pol} P_{TP} P_{Elec} P_{leak} P_{rp}$$

$$P_{Ion} = \frac{1-\frac{V_H}{V_L}}{\frac{M_H}{M_L}-\frac{V_H}{V_L}} \quad , \quad P_{Pol} = abs\left(\frac{M^+ - M^-}{2M}\right) \ , \quad P_{TP} = \frac{273.2\ K+T}{295.2\ K}\left(\frac{760\ mmHg}{P}\right)$$

Pelec is measurement error in electrometer, and correction is given by ADCL calibration.

Pleak accounts for leakage current, ref class ion chambers may be taken as 1.000.

Prp is taken as 1 for flattened fields(radiation profile correction).

Dose at 10 cm depth: $D_w^Q = Mk_Q N_{D,w}^{Co-60}$

TG-51 Electron Calculations

$k_Q = P_{gr}^Q * k_{R50}^1 * k_{ecal}$, PQgr used to correct for effective point of measurement being upstream, kR50 is used to account for variations in readings due to electron beam quality, kecal functions to take your chambers calibration factor which is for photons(chamber specific value).

$M_{corr} = P_{ion} * P_{TP} * P_{elec} * P_{Pol} * M_{raw}$

$D_W^Q = M_{corr} * P_{gr}^Q * k_{R50}^1 * k_{ecal} * N_{D,W}^{Co-60}$

Radiation Interactions

Bremsstrahlung(Braking) Radiation $efficiency = 9 * 10^{-10} * ZE$, $Probability \sim Z^2E$

Photoelectric $E_{electron} = E_{photon} - E_{binding}$, $PE\ Probability \sim \frac{Z^3}{E^3}$

- Competes with Auger Electrons, where a higher energy electron emits a characteristic x-ray to lose the energy needed to drop levels, but that x-ray interacts with a subsequent outer shell electron to eject it from the atom.

Compton Scattering $E' = \frac{E_0}{1+\frac{E_0}{0.511}*(1-\cos cos\,(\phi))}$, $Compton\ Probability \sim \frac{1}{E}$

- Dependent on electron density. Hydrogen being double that of others.
- Forward scatter no energy loss, 90 degree retains 0.511 MeV, 180-degree back scatter 0.255 MeV
- Dominate interaction from 25 keV to 25 MeV roughly.

Pair Production Minimum of 1.022 MeV to happen, $PP\ Probability \sim Zln(E)$

- Triplet production happens with an orbital electron and requires 2.044 MeV, normally PP happens with an interaction w/ the nucleus.

Photo Disintegration atomic nucleus absorbs high-energy photon to spring protons, neutrons, and alpha particles.

- Minimum Energy is 8-10 MeV

Charged Particle interactions: Continuously slowing down approximation details how interactions happen during the entire track of the particle. $Energy\ loss \sim \frac{C^2}{v^2}$, $C = Charge,\ v = speed$

- Follows Coulomb's Law $F = k_e \frac{q_1 q_2}{r^2}$, $k_e = 9 * 10^9 \frac{Nm^2}{C^2}$, $q = charge,\ r = distance$
- Stopping Power is energy loss per unit path length $S(E) = -\frac{dE}{dx}$, mass stopping power is the ratio of stopping power to the material density.
  - Energy dependent $\frac{dE}{dx} \sim \frac{z^2}{v^2}$, z = charge & v = speed.
- Charge particle range: $R = \frac{E}{-\frac{dE}{dx}} = \frac{E}{S(E)}$
- Protons tend to move along a straight line due to their large mass, electrons are more easily scattered.
- Protons also exhibit a Bragg peak at the end of their range.

Neutron interactions: Recoil interactions when hitting a charged nucleus, thermal(slow) neutrons can be absorbed by the nucleus and cause spontaneous decay.

Radioactive Decay Modes: $x_Z^A,\ A = atomic\ mass,\ Z = atomic\ number(protons)$

IsotoPes: Atoms that have the same number of Protons, different number of neutrons.

IsobArs: Atoms have the same mass number(A)

IsotoNes: Same number of neutrons but different number of protons.

IsoMers: Metastable atom with same Z & A, excited state.

Alpha Decay: emission of a Helium atom from the nucleus.

- Primarily occurs in heavy atoms (Z>82)
- Discrete energies.

$$X_Z^A \rightarrow Y_{Z-2}^{A-4} + He_2^4 + Q$$

Beta Decay: Emission of an electron or positron w/ accompanying neutrino (or anti), also a neutron becomes a proton or vice versa.

- Beta(-) Decay: Converts a neutron into a proton, creating an electron and anti-neutrino.
  - Occurs in radionuclides with a high N/P ratio
  - Continuous energy spectrum

$$X_Z^A \rightarrow Y_{Z+1}^A + \beta^- + \hat{\nu} + Q$$

- Beta(+) Decay: Converts a proton to a neutron, creating a positron and neutrino.
  - Occurs in radionuclides with a low N/P ratio.
  - Continuous energy spectrum

$$X_Z^A \rightarrow Y_{Z-1}^A + \beta^+ + \nu + Q$$

Electron Capture: Electron is captured by the nucleus which converts a proton into a neutron.

- Occurs in radionuclides with a low N/P ratio.
- Captured electron is usually from an inner shell and leaves a vacancy.
  - Causes characteristic x-rays and/or auger electron emission.

$$X_Z^A + \beta^- \rightarrow Y_{Z-1}^A + \nu + Q$$

Internal Conversion: Excited atom lowers its energy to ground state via gamma or electron emission.

- If an electron is emitted, characteristic x-rays and/or Auger electrons may be emitted.

$$X_Z^A \rightarrow X_Z^A + Q$$

Photodisintegration: High energy photon interacts with an atomic nucleus and imparts enough energy to free one or more nucleons.

- Threshold energy is around 10 MeV for most nuclei.

$$X_Z^A + \gamma \rightarrow \text{Nuclear Decay}$$

Source	Half-Life
Radium-226	1622 years
Iridium-192	73.8 days
Cobalt-60	5.3 years
Cesium-137	30 years
Gold-198	2.7 days
Iridium-192	73.83 days
Iodine-125	59.4 days
Palladium-103	16.99 days

Exposure rate constants for various radionuclides

R•cm^2 / hr•mCi

Radionuclide	Exposure rate constant
cobalt-60	12.838
molybdenum-99	1.03
technetium-99m (6 hour)	0.720
palladium-103 (unfiltered)	1.48[6]
silver-110m (250 day)	14.9
caesium-137	3.400
iodine-125 (unfiltered)	1.46[6]
iridium-192 (unfiltered)	4.69[6]
radium-226	8.25

Isotope	*Energy (keV)*	*Γ (R cm^{2}mCi^{-1}hr^{-1})*	*Half-Life*	HVL Lead
Co-60	1250	13.07	5.3 years	1 cm
Ra-226	830	8.25	1600 years	0.7 cm
Cs-137	660	3.26	30.2 years	0.55 cm
Au-198	420	2.38	2.7 days	0.30 cm
Ir-192	380	4.69	73.83 days	0.25 cm
Cs^{131}	29	1.4	9.7 days	0.02 mm
I-125	28	1.45	59.6 days	0.02 mm
Pd-103	21	1.48	17 days	0.01 mm

Quantity	Unit	Symbol	Derivation	Year	SI equivalence
Activity (*A*)	becquerel	Bq	s^{-1}	1974	SI unit
	curie	Ci	3.7×10^{10} s^{-1}	1953	3.7×10^{10} Bq
	rutherford	Rd	10^6 s^{-1}	1946	1,000,000 Bq
Exposure (*X*)	coulomb per kilogram	C/kg	C·kg^{-1} of air	1974	SI unit
	röntgen	R	esu / 0.001293 g of air	1928	2.58×10^{-4} C/kg
Absorbed dose (*D*)	gray	Gy	J·kg^{-1}	1974	SI unit
	erg per gram	erg/g	erg·g^{-1}	1950	1.0×10^{-4} Gy
	rad	rad	100 erg·g^{-1}	1953	0.010 Gy
Equivalent dose (*H*)	sievert	Sv	J·kg^{-1} × W_R	1977	SI unit
	röntgen equivalent man	rem	100 erg·g^{-1} x W_R	1971	0.010 Sv
Effective dose (*E*)	sievert	Sv	J·kg^{-1} × W_R × W_T	1977	SI unit
	röntgen equivalent man	rem	100 erg·g^{-1} × W_R × W_T	1971	0.010 Sv

Counting statistics:

Binomial Distribution: applicable for all distributions with constant probabilities. Not commonly used.

- Mean = probability of success * number of trials, $\bar{x}$ = pn
- Standard deviation is $\sqrt{(\bar{x}(1 - p))}$

Poisson Distribution: Mathematical simplification of the Binomial distribution.

- Probability of exactly 'x' successes: $P(x) = \frac{\lambda^{x} * e^{-\lambda}}{x!}$
- Mean, $\bar{x}$ = pn
- Standard Deviation, $\sigma = \sqrt{\bar{x}}$, can approx. for normal if mean is high.
  - $1\sigma = 68.2\%$
  - $2\sigma = 95.4\%$
  - $3\sigma = 99.6\%$

Accuracy: how closely a measurement matches the true value.

Precision: ranges of measured values, closely related to deviation & standard deviation.

Measurement Error: difference between measured value and true population value.

Measurement Uncertainty: Interval around a measured value such that any repeated measurement will produce a result within that interval.

Confidence Interval: the percent certainty that the true value lies within a given range of values.

Sources of Error:

- Random error: statistical error, electronic noise, daily pt alignment errors.
- Systematic error: measurement device error, atmospherically induced errors, pt simulation errors.
- Combined errors, $\sigma_C = \sqrt{\sigma_x^2 + \sigma_y^2}$
- Standard error, $\sigma_{\bar{x}} = \frac{s}{\sqrt{N}}$

Test Quality Metrics:

- Sensitivity, True Positive Fraction = $\frac{TP}{TP+FN}$
- Specificity, Fraction of normal cases diagnoses correctly = $\frac{TN}{TN+FP}$
- False Positive Fraction = $\frac{FP}{FP+TN}$
- Accuracy, Correct diagnosis = $\frac{TP+TN}{TP+TN+FP+TP}$
- Positive Predictive Value = $\frac{TP}{TP+FP}$
- Negative Predictive Value = $\frac{TN}{TN+FN}$

Informatics:

Binary System is a base-2 system in which numbers are expressed as a combination of 0's and 1's.

- 1024, 512, 256, 128, 64, 32, 16, 8, 4, 2, 1
- Read from right to left, 1010 is the sum of 8+2 = 10.

Units of Storage

- Bit(b) is single piece of binary information(1,0)
- Nibble is a unit consisting of 4 bits(1011)
- Byte(B) is a group of 8 bits, commonly used unit for computer memory. Useful for storing information such as an ASCII code.
- Kilobyte(KB) is 1024 bytes, equal to $8 * 2^{10}$ bits
- Megabyte(MB) is 1,048,576 bytes, equal to $8 * 2^{20}$bits
- Gigabyte(GB) equal to $8 * 2^{30}$bits
- Terabyte(TB) equal to $8 * 2^{40}$bits

ASCII Format: The American Standard Code for Information Interchange (ASCII) is a 7 or 8-bit (1 byte) format used to encode letters and other characters. Most modern character-encoding formats, such as unicode and ISO/IEC 8859, are based on the ASCII system.

Memory

- Volatile memory: Requires constant power to maintain the stored information but can be written and read very quickly. Makes this memory ideal for storing intermediate system states and is present in RAM(random access memory).
- Non-volatile memory: stores memory even without power and used to store data for long periods of time. Utilized in hard drives, and includes flash & ROM.

Unit (Symbol)	Definition	Units
Exposure (X) $X = \frac{dQ}{dm}$	Amount of ionization per mass of air due to X-rays and gamma rays.	Roentgen (R) $1R = 2.58 \times 10^{-4} \frac{C}{kg}$
Absorbed Dose (D) $D = \frac{\text{Energy Deposited}}{\text{Mass}}$	Amount of energy imparted by radiation per unit mass.	SI: Gray (Gy) $1Gy = 1\frac{J}{kg}$ Traditional: rad 1 rad = 0.01Gy
Kerma (K) $K = \frac{dE_{tr}}{dm} = \psi(\frac{\mu_{tr}}{\rho})$	**K**inetic **E**nergy **R**eleased per unit **MA**ss Kinetic energy transferred by uncharged particles to charged particles per unit mass.	SI: Gray (Gy) $1Gy = 1\frac{J}{kg}$ Traditional: rad 1 rad = 0.01Gy
Equivalent Dose (H_T) $H_T = w_R D$	A measure of absorbed dose weighted for the biological effectiveness of the types(s) of radiation (relative to the low LET photons and electrons) to produce stochastic health effects in humans.	SI: Sievert (Sv) $1Sv = 1\frac{J}{kg}$ Traditional: rem 1 rem = 0.01Sv
Dose Equivalent (H) $H = QD$	A measure of absorbed dose weighted for the biological effectiveness of the types of radiation (relative to low LET photons and electrons) to produce stochastic health effects in humans.	Sievert (Sv) $1Sv = 1\frac{J}{kg}$ Traditional: rem 1 rem = 0.01Sv
Effective Dose (E) $E = \sum(W_T H_T)$	A measure of equivalent dose, weighted for the biological sensitivity of the exposed tissues and organs (relative to whole body exposure) to stochastic health effects in humans.	Sievert (Sv) $1Sv = 1\frac{J}{kg}$ Traditional: rem 1 rem = 0.01Sv
Effective Dose Equivalent (HE)	A measure of equivalent dose, weighted for the biological sensitivity of the exposed tissues and organs (relative to whole body exposure) to	Sievert (Sv) $1Sv = 1\frac{J}{kg}$

Quantifying Radiation:

Exposure(X) - the quotient of dQ by dm where dQ is the absolute value of the total charge of the ions of one sign **produced in air** when all electrons **liberated by photons** in air of mass dm are completely stopped in air. Exposure is only defined up to about 3MeV because of the size of a free air ionization chamber required for higher energy measurements.

Fluence(φ) - measure of total number of quanta that enter an area.

Flux(ϕ) – measure of the rate at which quanta enter an area per unit time.

Linear Attenuation Coefficient(μ) – fraction of photon beam which is attenuated per unit path length(units: 1/cm)

- $\frac{I(x)}{I(0)} = e^{-\mu x}, \ \mu = \ln ln\left(\frac{\frac{I(x)}{I(0)}}{x}\right)$

Absorbed Dose(D) – energy deposited per unit mass to a medium by radiation.

- $D_{air} = f_{air}X = 0.876\left(\frac{cGy}{R}\right) * X(R)$
- $D_{medium} = f_{med}\left[\frac{\left(\frac{\mu_{en}}{\rho}\right)_{med}}{\left(\frac{\mu_{en}}{\rho}\right)_{air}}\right]\left(\frac{\psi_{med}}{\psi_{air}}\right)X = 0.876\left(\frac{cGy}{R}\right) * X(R)$

KERMA(K) – Kinetic Energy Released per unit Mass, quotient of the initial kinetic energies of all charged ionizing particles liberated by unchanged particles(photons) per mass of material.

- K = $\frac{dE_{tr}}{dm} = \psi\left(\frac{\mu_{tr}}{\rho}\right)$
- Kerma is divided into two components:
    - Collision Kerma is the amount of energy released by collision type interactions, under CPE Kcoll is equal to absorbed dose. Under TCPE it is approximately equal to absorbed dose.
    - Radiative Kerma is the amount of energy released in radiative type interactions.

$$K = K_{coll} + K_{rad} = \psi\left(\frac{\mu_{en}}{\rho}\right) + \psi\left(\frac{\mu_{rad}}{\rho}\right)$$

TERMA – Total energy released per unit mass, similar to kerma but includes energy losses due to coherent scattering, for MV beams this is negligible.

Equivalent Dose(H) – represents the stochastic health effects of low levels of ionizing radiation to the human body.

- $H_T = \sum_R \left(W_R * D_{T,R} \right)$, W_R is the radiation weight factor.

Effective Dose(E) – uses tissue weighting factors to compare the risk of stochastic effects from a specific distribution of doses to organs with the risk of stochastic effects of a uniform whole body dose.

- $E = \sum_T \left(w_T * H_T \right)$, w_T is a tissue weighting factor.

Radiation Weighting Factors

Radiation Type	W_R
X-rays, gamma rays, electrons and muons	1
Neutrons	• <10keV: 5 • 10-100keV: 10 • 100keV - 2MeV: 20 • 2MeV-20MeV: 10 • >20Mev: 5
Protons > 2MeV	2
Alpha particles, fission fragments and non-relativistic heavy nuclei	20

Source: ICRP 60

Tissue Weighting Factors (W_T)

Organ	W_T
Gonads	0.20
Red bone marrow Colon Lung Stomach	0.12
Bladder Breast Liver Esophagus Thyroid	0.05
Skin Bone surface	0.1
Remainder	0.05
Total	**1.0**

Data from Table 1.1 of NCRP 116

Radionuclide identification process:

Differences in attenuation properties require multiple readings through various materials.

- A sheet of paper would likely stop alphas
- Betas & keV x-rays will likely make it through paper but not penetrate aluminum sheet
- High MeV gamma rays will make it through the aluminum, but will be attenuated heavily by lead shielding.
- Neutrons will likely not be stopped by any of the barriers unless it includes boron or high neutron cross section.

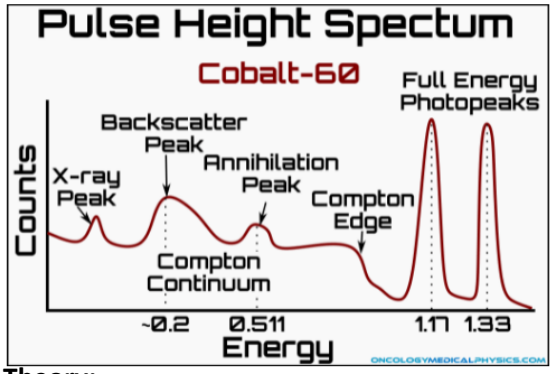
Magnetic Analysis can also be performed as moving charged particles will be deflected in a magnetic field by the Lorentz force.

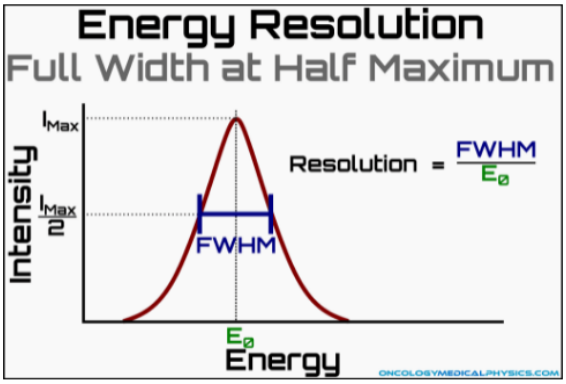
- $F = qv \times B$
  - F is the Lorentz force
  - Q is the charge of the particle
  - V is the particle velocity, found by v/c
  - B is the strength of the magnetic field($Tesla = \frac{Newton*second}{coulomb*meter}$, 1T = 10000 Gauss)
- $r = \frac{mv}{qB}$
  - r is the radius of particle deflection
  - m is the particle's mass
  - mv is the momentum of the particle.

Pulse Height Spectrum

- Can determine the radionuclide by analysis of energy spectrum pulse height analysis.
  - Germanium detectors, scintillation detectors, gas filled proportional counters.
- Full energy photopeak: full energy represents counts where all of the energy of the decay product is fully captured.
- X-ray peak: photoelectric event occurs in the detector, the absorber may release a characteristic x-ray. Most of these peaks occur at k-shell binding energy of the absorber atom.

- Backscatter peak: caused by backscattered photons after Compton interaction w/ detector.
- Annihilation Peak: present if there are positrons emitted, 2 .511 MeV photons emitted.
- Compton Edge. Maximum energy transferable to a compton scatter electron by photon.





Cavity Theory:

Bragg-Gray Cavity Theory – relates dose to medium, Dmed, to dose to the cavity filled with gas, Dgas, via ***mass collisional stopping powers*** between the two mediums.

- $\frac{D_{med}}{D_{gas}} = \left(\frac{S}{\rho}\right)^{med}_{gas}$
- Assumptions:
  - CPE or TCPE exist
  - All electrons causing ionization in the cavity arise from phantom material
  - Secondary electron spectrum is unperturbed by the cavity
  - Energy of secondary electrons created inside the cavity are deposited locally.
    - Neglects delta rays generated within the cavity as a result of interactions with scattered electrons.

Spencer-Attix Cavity Theory – resolves issues with BG and applies to small cavities using ***restricted mass collisional stopping power*** ratios.

- $\frac{D_{med}}{D_{gas}} = \left(\frac{\bar{L}}{\rho}\right)^{med}_{gas}$
  - Uses cutoff energy, Δ, removes the requirements to deposit their energy locally. Usually 10-20 keV, below that they are assumed to deposit their energy where created and above dissipate their energy through CSDA.
- Assumptions/Requirements:
  - CPE or TCPE
  - All electrons causing ionization in the cavity arise from the phantom material.
  - Secondary electron spectrum is unperturbed by the cavity.

Burlin Cavity Theory – Generalized cavity theory for large and small cavities, also uses ***RMCSP***

$$\frac{D_{med}}{D_{gas}} = d\left(\frac{\bar{L}}{\rho}\right)^{med}_{gas} + (1-d)\left(\frac{\mu_{en}}{\rho}\right)^{med}_{gas}$$

- d is a parameter related to cavity size, d = 1 for small cavities and approaches 0 for large.
- CPE must exist in the medium and cavity.

Detection Efficiency and Dead Time:

Geometric Efficiency: isotropic emission of radiation means the geometric efficiency is approx equal to the area of the detector divided by the surface area of a sphere with an equal radius to the SDD.

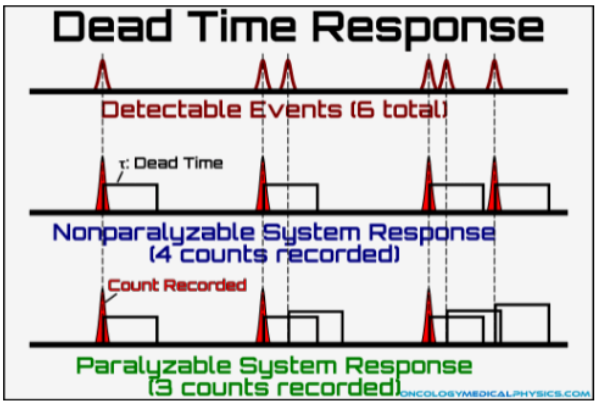
- $\epsilon_{geometric} = \frac{r^2}{4d^2}$, r is detector radius & d is source to detector distance.(Point source)
- True geometric efficiency is $\epsilon_{geometric} = \frac{\Omega_{detector}}{4\pi}$, Ω is solid angle in steradians.

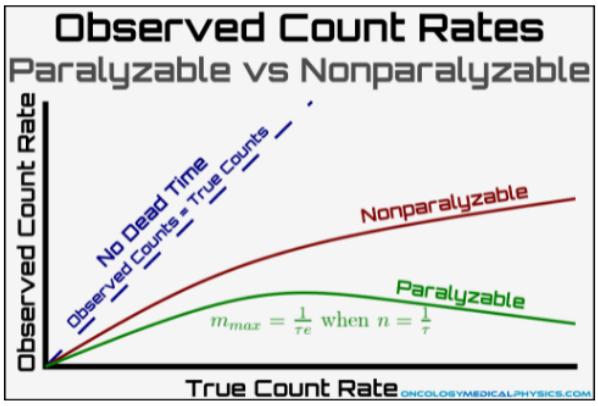
Intrinsic Efficiency: ratio of number of counts recorded by a detector to the total number of particles incident on the detector.

- $\epsilon_{intrinsic} = \frac{Counts\ recorded}{\#\ of\ particles\ incident\ upon\ detector}$
- Influenced by detector properties and incident particles.

Dead Time: minimum time between events. Depends on length of post count dead time(τ), rate of countable events at detector, and whether the detector is paralyzable/nonparalyzable.

- Paralyzable: events occuring during previous events dead time increase the deadtime length but are uncounted.
  - $m = ne^{-n\tau}$, m is measured count rate, n is true count rate.
  - $m_{max} = \frac{1}{\tau e}\ when\ n = \frac{1}{\tau}$
- Non-paralyzable: counting systems are those in which events dead time do not add additional dead time and are not counted.
  - $n = \frac{m}{1-m\tau}$, m is measured count rate, n is true count rate.

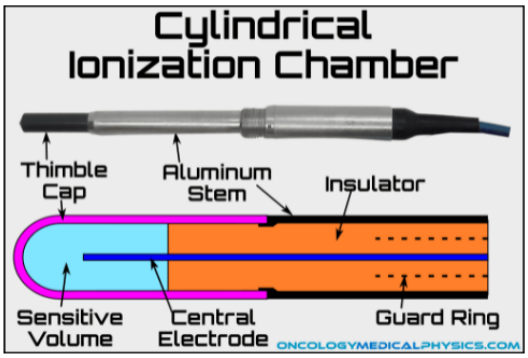


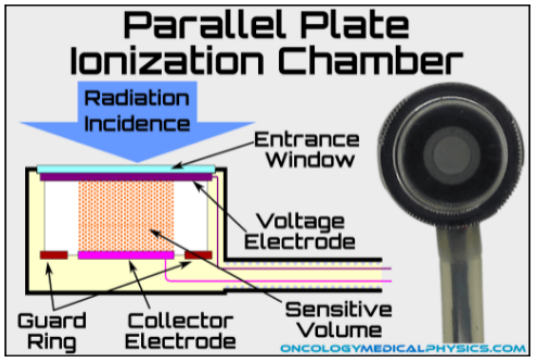


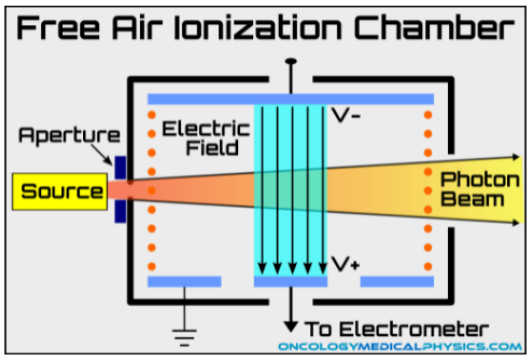
Common Dosimeters:

Ionization chambers are gas filled cavity surrounded by two electrodes of opposite polarity and an electrometer. Electric field between electrodes accelerates the radiation produced ions and gets collected by said electrodes. Charge collected can be read out and converted to absorbed dose.

- Cylindrical Chambers: Used for photons and electrons above 6 MV, due to axial design point of measurement upstream of the central axis by $0.6r_{cav}\,(photons)\ and\ 0.5r_{cav}(electrons)$
- Plane Parallel Chambers: Used for low every electron dosimetry, and used for precise measurement location accuracy for PDD's. Point of effective measurement is the most upstream plane of the chamber.
- Free Air Chambers: instrument to measure Roentgen, used as reference dosimeter by ADCL's but due to large size they are not suitable for clinical applications.





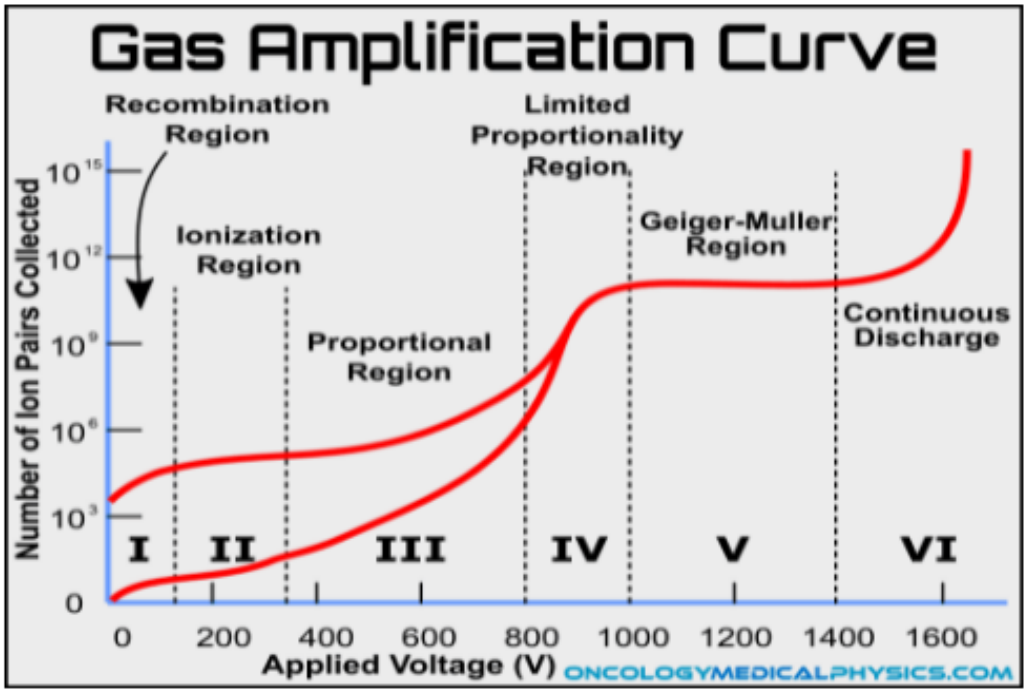


Farmer Chambers: Thimble ion chambers for reference dosimetry.

- Volume: 0.6cc, approx cylindrical 0.3 cm radius, 2 cm length.
- Response 20 nC/Gy
- Point of measurement upstream by $0.6r_{cav}\,(photons)\ and\ 0.5r_{cav}(electrons)$

Gas Amplification Curve:

- Region 1: **Recombination**, of ions prior to collection
- Region 2: **Ionization region**, where the voltage is not strong enough to produce secondary ionizations but not weak enough to allow recombination before collection.
- Region 3: **Proportional region**, response is proportional to energy collected and to the applied voltage
- Region 4: **Limited Proportionality region**, response collected diminishes when applied voltage increases.
- Region 5: **Geiger-Muller region**, Townsend avalanche creates a large number of secondary avalanches.
- Region 6: **Continuous discharge region**, continually arcs due to excess voltages.



Comparison of Common Dosimeters

Device -Accuracy	Common Uses	Advantages	Limitations
Ion Chambers ±1%	-Reference Dosimetry -Percent Depth Dose Distributions	-Best understood -Sub 1% accuracy possible -Low energy dependence	-Size limitations -ADCL calibration required
Diode Detectors ±2-3%	-Small field dosimetry -Array devices -Electron PDD	-Small volume -Rapid readout -No external bias	-Temperature dependence (0.5%/C) -Dose rate dependence -Energy dependence
Film ±2-5%	-Planar dose distributions -Electron PDD	-Best spatial resolution (μm) -Large area measurement -Persistent dose record -Tissue equivalent (radiochromic only)	-Delayed readout -Batch-to-batch variation -Chemical development (radiographic only)
Luminescent Dosimeters ±3%	-In Vivo Dosimetry -Personnel dosimeters -End-to-end testing (IROC)	-Small size -Low MV energy dependence	-Delayed readout -Signal loss over time -Supralinear response with accumulated dose
MOSFET Detectors	-In vivo dosimetry -Small Field Dosimetry -Surface dose	-Extremely small effective volume -Permanent dose record -Instant readout	-Finite life (~100Gy) -Energy Dependence -Temperature Dependence -Sensitivity changes with accumulated dose
Plastic Scintillators	-Small Field Dosimetry -Array Measurements -Electron measurements	-Small volume -Near water equivalent -Dose and rate independent	-Noise, especially Cherenkov Radiation -Sensitivity change with plastic yellowing -New technology, few vendors

Important facts about dosimeters:

Geiger Counters – survey detectors to determine presence of radiation and abundance of radiation.

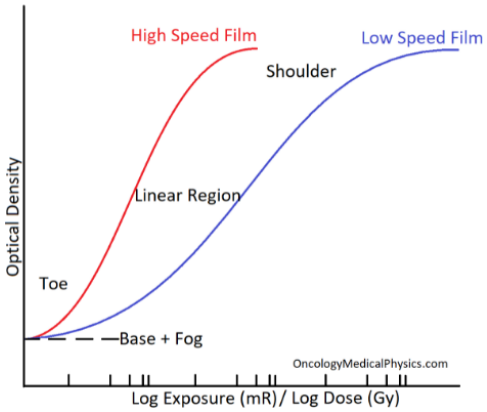
- Uses: Detect presence and relative abundance of radiation, Accuracy +/- 10%-20% , radiation surveys and leak tests.
- Advantage: Highly sensitive even at low levels of radiation, collected charge count is high
- Disadvantages: Dead time accumulation(10-100ms), can't determine energy.
- Townsend avalanche refers only to the cascading ionizations caused in electron induced ionizations.
- Geiger counter quench gasses exist to supply electrons to the fill gas and disassociate, thereby preventing additional Geiger discharges as the fill gas returns to ground state.

Diode Detectors – small volume solid state detectors used in small field dosimetry, array measurements and in-vivo dosimetry.

- Exhibit temperature, energy, dose rate, and orientation dependence.
- Consists of buildup cap, metallic leads, and Diode(Die) p-type or n-type semiconductor junction. Doping introduces impurities to produce p-type(electron acceptor) or n-type(electron donor) semi conductors.
    - Doped with silicon or germanium.
- The depletion zone is the active region of a diode detector. Because this region is only a small fraction of the die, volume averaging of a diode detector is very small.

Film Dosimeters: can be used for radiation measurement and relative dose distributions. Accuracy is limited to 2-5% because of compounding sources of error. Radiochromic film doesn't require chemicals and develops via polymerization reaction.

- Optical Density: log base 10 of fraction of light transmitted through unexposed film to the light transmitted after exposure.
  - $OD = log\left(\frac{I_0}{I}\right) = -\log log\,(T)$, $Transmittance = T = \frac{I}{I_0}$
- H & D curves to relate exposure/dose to optical density.
  - Film speed, fast film will have a greater increase in OD with dose, but have a smaller range of OD. Slower film provides better range and dose resolution. Fast film is good for dose reduction.
  - Gradient, slope of the curve in linear region(fast means higher gradient)
  - Latitude, range of exposures possible
  - Linear region, proportional OD to log of exposure
  - Toe, Shoulder, relationship non-linear
  - Fog, darkening of film due to background light/radiation
  - Base, natural attenuation of the film without exposure or fog



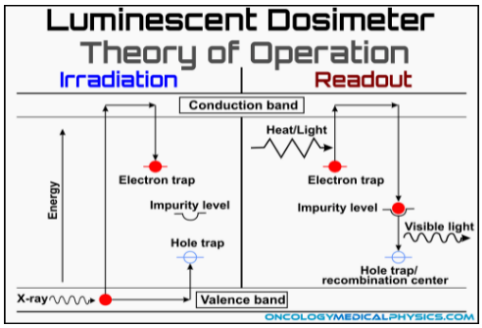
	Dose Accuracy	Advantages	Disadvantages

Radiographic	3-5%	• Highest spatial resolution (μm)
		• Energy independent in MV region only
		• Lower dose threshold
			• Requires chemical processing
			  ◦ Increased cost of development
			  ◦ Increased measurement error
			• Not tissue equivalent
			• Fogging in visible light
			• Strong energy dependence in kV region

Radiochromic	2-3%	• Tissue equivalent (Z = 6-6.5) • Energy independent • No chemical processing • Very high spatial resolution (sub-mm)	• Requires higher minimum dose (~0.1Gy)

- Types of film:
  - Radiographic: highest spatial resolution(um), energy independent in MV region, lower dose threshold, requires chemical processing, not tissue equiv.
    - Electrons aggregate around impurities to form negative charge, this charge attracts Ag+ ions leaving behind silver leaving a latent image. Developer amplifies the amount of metallic silver.
  - Radiochromic: Tissue equivalent(Z=6-6.65) Energy independent, no chemicals, high spatial resolution(<mm), requires more dose.
    - Radiation incident polymerizes the active layer, fully develops after 24 hours

	Accuracy	Chemical Composition	Advantages	Disadvantages
Thermoluminescent Dosimeters (TLD)	~3%	LiF:Mg, TI	• Energy independent >100keV • Neutron sensitive models available (TLD-600) • Reusable	• Can only be read out once • Requires significant time to read out • Supralinear response with reuse
Optically Stimulated Luminescent Dosimeters (OSLD)	~3%	Al_2O_3:C	• Multiple readouts possible • Rapid readout • Reusable • Persistent dose record	• Higher effective Z than TLDs • Stronger energy dependence than TLDs



Luminescent Dosimeters: crystals trap and store energy when irradiated, energy is then released via luminescence in the form of visible light. Measurement of light may be used to determine the dose delivered to the dosimeter.

- Dose measurement
  - Ionizing radiation creates electron-hole pair in crystal structure.
  - Liberated electron promoted to conduction band, and migrates to electron trap. The hole migrates along the valence band to a hole trap.
    - Trapped in imperfections labeled as Trap centers
  - Energy is imparted to an electron and hole allowing them to escape the traps. Causing the electron-hole pair to recombine and release light.
    - Ion pair meet in an imperfection called Luminescent Centers
  - Emitted light is measured by a photomultiplier tube or camera(CCD or CMOS). Light is used to determine absorbed dose.
  - Competitive Centers trap charge carries but don't contribute to luminescence
- Doping will add imperfections to the crystals, allowing creation of traps, luminescent centers, and competitive centers.
  - Common Dopants are Mg, Ti and C.
- TLDs are subject to precise heating to release the energy from the traps, and need to be annealed to remove any remaining trapped charges and redistribute traps.
  - Response of TLD's increases with repeated use due to decrease in availability and efficiency of competitive centers.
- OSLD's are exposed to lasers, LEDs or fluorescent lamps for measurement. This process only takes a second and only released 0.05% of stored luminescence, for multiple readings.
  - Filter to remove the wavelength of the stimulated light source.
  - Bleaching with halogen lamp, fluorescent lamp or green LED to empty trap centers for reuse of the OSLD. Not all deep traps are emptied and cause sensitivity change.

Activation Foils:

Materials that become radioactive when exposed to neutrons, determine neutron absorbed dose is made by measuring decay products.

- Dose measurement
  - Activation foil is placed in an area with neutron presence for a time period.
  - After exposure the foil is taken to a sensitive detector to read out, thin window geiger counter(betas) or NaI Crystal or Gi(Li) detector for gammas.
  - Measured foil activity is used to determine neutron fluence or dose.

MOSFET Detectors:

Metal oxide semiconductor field effect transistor detectors are semiconductor based radiation detector use for small field dosimetry, in-vivo dosimetry and profile measurements. Very small collection volume, but devices accumulate damage with exposure.

- Dose measurement
  - MOSFET is irradiated created electron-hole pairs in SiO2 layer.
  - Holes then migrate to interface between SiO2 and n-type substrate.
  - Increase in holes in the interface region retard e-field by applying negative voltage to gate. This requires a larger voltage to induce current flow between the source/drain
  - Change in threshold voltage is proportional to the change in the number of trapped holes, corresponding to absorbed dose.
- Unlike TLDs or OSLDs which are able to empty their traps, MOSFET detectors accumulate trap filling, reducing their effective usable life to ~100Gy. This property also allows them to act as a persistent dose record.

Bubble Dosimeters:

Clear plastic tube filled with a gel polymer. Fast neutrons are incident upon the device superheats the polymer creating bubbles. The number of bubbles correspond to the neutron fluence or absorbed dose.

- Dose measurement
  - Bubble detector cap is unscrewed reducing the chamber pressure and allowing measurement.
  - Detector is exposed to neutrons, and forms bubbles.
  - Bubbles are counted by eye or optically, and number correlates to fluence/dose.
  - Reset the detector by screwing down the cap causing internal pressure to increase.

Plastic Scintillators:

Scintillation dosimeters emit light when irradiated, and read out by a photodetector and correlated to absorbed dose. Plastic scintillators are near tissue equivalent, small in size and able to detect small amounts of light and energy independence in the MV range.

- Dose measurement
  - Radiation induces scintillation in the sensitive volume
  - Scintillation light is transmitted by light pipe to photo detector, background can be transmitted through second light pipe
  - Photodetector converts light to electrical signal
  - Readout device removes the background signal and determines dose.
- Low SNR for these detectors.
  - Cherenkov radiation arises in inactive regions of the detector may account for as much as 3% of signal for photons or 12 for electrons.
    - Use Dual pipe design, optical filtration, temporal discrimination.
- Other materials are NaI(tl) for low energy(<360 keV photons), ZnS for alpha particles.

Radiation dose distributions: Photons

Photon beam quality refers to the spectrum of energy present in a beam that impacts the penetration.

- HVL describes beam quality by the thickness given to reduce an incident beams fluence by 50%. TVL is more appropriate for shielding calculations.
- MV beam quality is measured by attenuation in water. An approach used is ratio of maximum dose to dose at a depth of 10cm. This is labeled PDD(10cm) and used in TG-51.

Percent depth dose(%DD) gives the dose, relative to max dose, as a function of depth.

- Skin sparing: MV photons do not reach maximum dose at the surface/skin, and gradually build up to maximum dose.
- Depth of dmax: influenced by energy, field size, SSD, and beam profile.
- Dose fall off, gradual for photon beams and slope of fall off decreased with increased energy
- Inverse Square Law = $\frac{1}{r^2}$

Attenuation is the process by which photons interact with matter and deposit some/all of their energy.

- Mass Attenuation Coefficient: fraction of photons removed from the beam per unit distance over medium density.
- Mass Energy Transfer Coefficient: fraction of primary beam energy transferred to charged particles in the medium per unit distance over medium density.
  - $KERMA = \psi * \frac{\mu_{tr}}{\rho}$
- Mass Energy Absorption Coefficient: fraction of primary beam energy by the medium per unit distance over medium density.
  - $Dose = \psi * \frac{\mu_{en}}{\rho}$

Buildup occurs because much of the energy transferred from the photon beam is initially absorbed by electrons which travel some distance. The energy is eventually deposited as a dose at a deeper depth.

- Build up region ends when energy taken downstream by liberated electrons is equal to the energy deposited at the location by electrons liberated upstream.

Isodose Charts are 2D map of a dose distribution, normalized to max dose or prescription dose.

- Central Axis dose highest along the central axis, unless a flattening filter is placed in the beam. Isodose with a FF in the superficial region will experience “horns” , to produce a flat dose distribution at a depth.
- Field Edge: dose near field edges fall off rapidly, caused by penumbra & lateral scatter.

Energy Impact on %DD

- Increased photon energy increases dmax, and decreases the slope of dose fall off.

Photon Energy (MV)	d_{max} (cm)	%DD at 10cm depth	TMR at 10cm depth	Dose Fall-off beyond d_{max} (%/cm)	Attenuation rate Calculated & (Rule-of-thumb)
Co-60 SSD/SAD = 80cm	0.5 cm	56%	0.68	3.7%/cm	3.4%/cm
6MV	1.5 cm	67%	0.77	3.3%/cm	2.7%/cm (3%/cm)
10MV	2.5 cm	73%	0.84	3.1%/cm	2.1%/cm (2.5%/cm)
18MV	3.2 cm	80%	0.91	2.8%/cm	1.3%/cm (2%/cm)

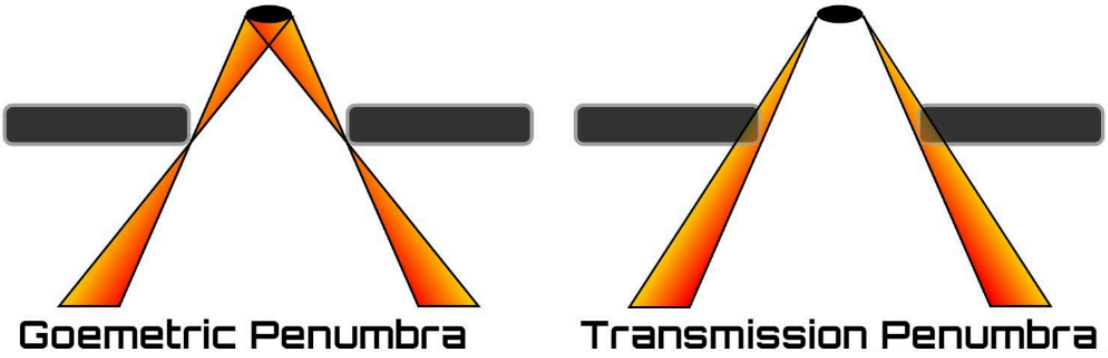
Energy Impact of Scatter Angle: Higher energy photons will be scattered in a more forward direction than lower energy photons.

Beam Profile: PDD increases with increasing field size, surface dose also increases with increased field size due to lateral scatter.

- Equivalent Square: A rectangular field has a %DD curve equivalent to a square field of the same area to perimeter
  - $Side\ of\ Square\ Field = 4\left(\frac{A}{P}\right)$

Penumbra is a partially shaded region on the border of an open photon field and a block.

- Geometric Penumbra: Increasing source size, Increasing SSD, Decreasing field size, Decreasing source to collimator distance.
  - $GP = \frac{S(SSD+d-SCD)}{SCD}$, S is source size, SSD, d is depth, SCD is source to collimator
- Transmission Penumbra: result of incomplete attenuation of the beam at the edges of the collimator.
- Physical Penumbra: defined as lateral distance between the 80% and 20% isodose lines.



Flatness and Symmetry: An unflattened MV photon beam will have its highest dose rate along the central axis of the beam. Flattening filters normalize the dose distribution laterally, leaving a uniform dose distribution at a given depth(usually 10 cm). Shallower than that depth there are prominent horns on either side of the center axis.

- Flatness: ratio of maximum to the minimum dose inside the central 80% of the field width at a given depth. +/- 3%
  - $Flatness = \frac{D_{max}-D_{min}}{D_{max}+D_{min}} * 100\%$
- Symmetry: maximum deviation of the left side dose from the right side over the central 80% of FWHM. +/- 2%

Source-to-surface distance(SSD)

- Dose rate decreases with increased SSD, but %DD increases.

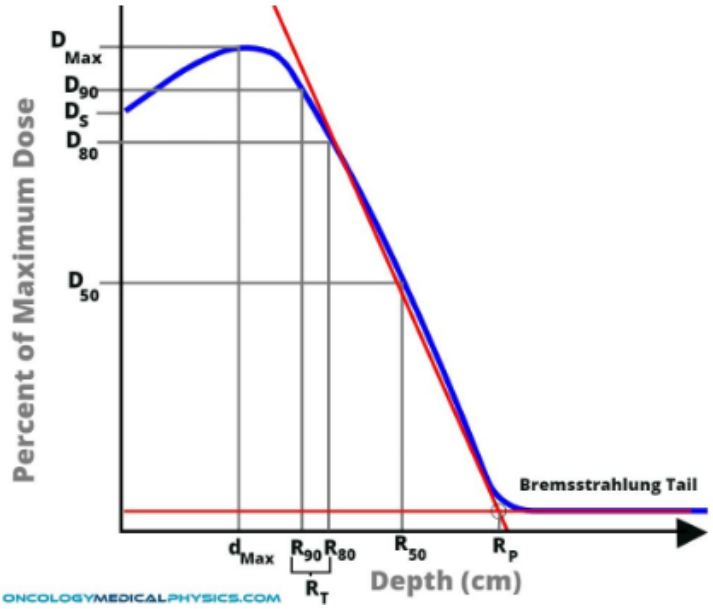
Inhomogeneities: regions of non-uniform electron density

- Compton interaction is the dominant form of interaction between clinical photon beams and body tissues. Most tissues have similar Z/A ratio(0.5), hydrogen(1) being an exception.
- Low Density(Air/Lung) : 1/3 to ¼ density of soft tissue,
  - Dose is reduced upstream the tissue lung interface
  - Dose in lung is lower at the beginning, then is higher in medium due to less beam attenuation.
  - Penumbra within the lung will be greater because of increased electron range. Effect is greatly increased for higher energies.
  - Exiting the lung tissue the dose will be depressed until build up, and will be increased on going.
- High Density(Bone): increased density and increases contributions of PP and PE in higher atomic number elements in bone.
  - Dose upstream of tissue/bone interface increases(due to backscatter 5-10%)
  - Within bone, dose to bone decreases, due to lower electron density.
  - Soft tissue embedded in bone, the dose will increase. Mass attenuation coeff is higher for bone causing more electrons to be created, mass stopping power of soft tissue is higher which means the electrons distribute their dose more.
  - Beyond the bone dose will be reduced due to attenuation.
- Estimating dose beyond inhomogeneity
  - $Effective\ Path\ Length = L * \left(\frac{\rho_{tissue}}{\rho_{water}}\right)$

Material	Physical Density	CT Number Range
Air	0.0012 g/cm^3	-1000
Lung	0.33 g/cm^3	-500 to -900
Water	1.0 g/cm^3	0 (by definition)
Soft Tissue	1.06 g/cm^3	10-50
Bone	1.8 g/cm^3	500-1000

Electron Dose Distributions:

- Dmax, max dose depth
- Ds, surface dose
- Dx, Dose at x percent of max dose
- dMax, depth max dose
- Rx, depth of x percent of max dose
- Rt, therapeutic range R90 or R80
  - R90(cm) ~ E/4
  - R80(cm) ~ E/3
  - R50(cm) ~ E/2.33
  - Rp(cm) ~ E/2
- Rp, practical range linear extrapolation
- Bremsstrahlung Tail, 5-15% of total dose



Energy	Surface Dose	d_{max}	$R_{90\%}$	$R_{50\%}$
6MeV	78%	1.2cm	1.7cm	2.3cm
9MeV	81%	2.0cm	2.7cm	3.5cm
12MeV	86%	2.8cm	3.9cm	5.0cm
15MeV	91%	3.2cm	4.9cm	6.3cm
20MeV	95%	3.5cm	6cm	8.5cm

Electron Energy(AAPM TG-25)

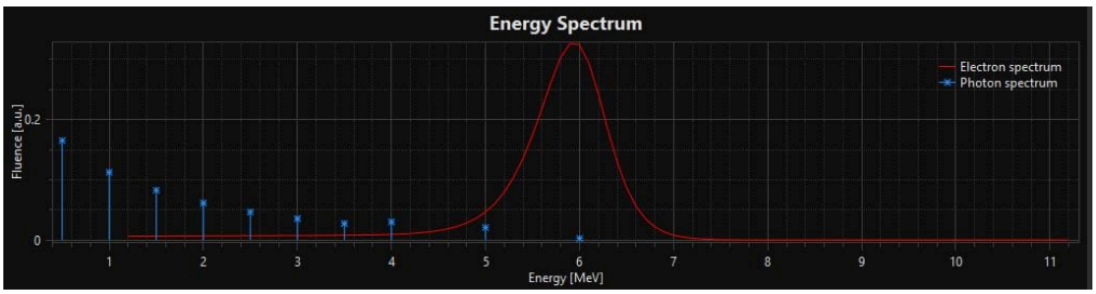
- $Mean\ energy\ incident,\ \overline{E}_0 = 2.33 * R_{50}$
- $Most\ probable\ energy\ at\ surface,\ E_{p,0} = 0.22 + 1.98 * R_p + 0.0025 * R_p^2$
- $Mean\ energy\ at\ depth\ d,\ \overline{E}_d = \overline{E}_0\left(1 - \frac{d}{R_p}\right)$

Energy and PDD: Increase in energy

- Increases skin dose, depth of dmax, range straggling, bremsstrahlung x-ray tail.
- Decrease sharpness of dose fall off.
- High dose isodose lines contract slightly, Low dose isodose lines expand laterally.

Linear Accelerator Energy Spectrum

- Electron spectrum is gaussian distribution centered about an energy just below accelerating energy. Results from the energy selectors and attenuation staggling.
- Photon component is predominantly low energy and strongly skewed right(Tail on Right).



Electron Buildup: Scattered in tissue leads up to TCPE at depth, with higher energies this scatter is more forward peaked meaning that build up is more gradual with a deeper dmax.

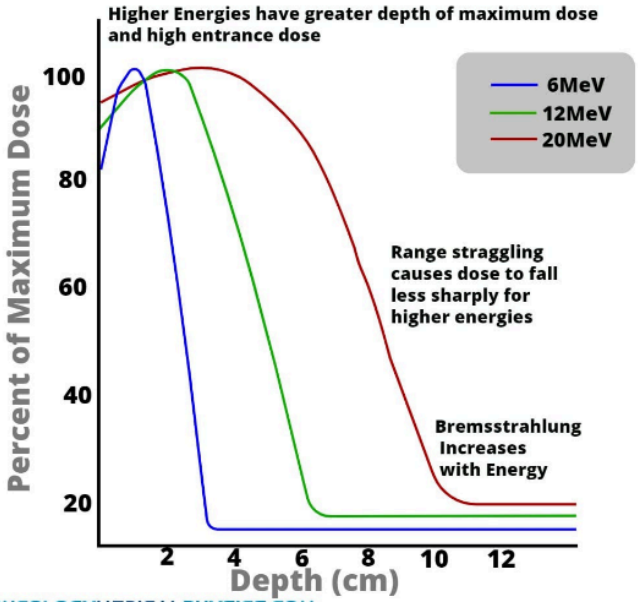
Field Size: for field sizes smaller than practical range, lateral CPE is lost.

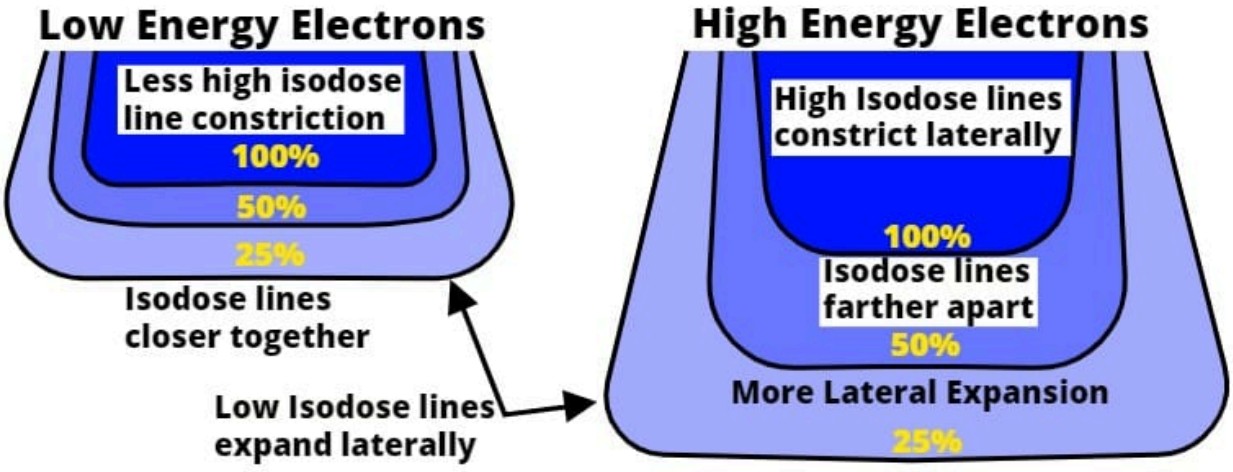
- Decreased field size reduces depth of maximum dose
- Decreased field size increases relative skin dose
- Decreased field size increases range straggling and decreases sharpness of dose fall off.
- Interpolation PDD with field size, $PDD^{m,b} = \sqrt{(PDD^{m,m})(PDD^{b,b})}$

Angle of Incidence: Increased angle of obliquity has the following effects:

- Shift the depth of max dose, and R80 toward the surface
- Increase Dmax, surface dose, and practicle electron range.

Bremsstrahlung contamination: production in clinical electron beams is primarily forward peaked and result in a bulge along central axis visualized in very low isodose lines(0-4%).





Radiobiology

Direct Raidation Damage: DNA single and double strand breaks.

Indirect Radiation Damage: Indirect damage via free radicals which intern damage DNA.

- A free radical is an atom/molecule carrying an unpaired electron in an outer shell.
- Hydroxyls are the most common, and is produced with reaction to water.
  - $H_2O + \gamma \rightarrow H^+ + OH^-$

DNA

- Backbone comprised of deoyribose and phosphate group bound together by covalent bonds
- Base pairs encode genetic information, 4 Bases are: Cytosine, Guanin, Adenine, Thymine.
  - Pairs are C-G, and A-T.
- Single Strand breaks are considered sub-lethal and generally repaired. Multiple single strand breaks can be lethal.
- Double Strand breaks when both strands are broken within the same area, generally non-repairable and result in cell death, sterilization, or mutation

Law of Bergonie and Tribondeau: The radiosensitivity of a cell is directly proportional t reproductive rate and is inversely proportional to its degree of differentation.

- Radiosensitivity increases with: increased rate of cell division, low degree of specialization, high metabolic rate, increased oxygenation, increased length of time they are proliferating.

Cell Cycle State

- G1: First growth phase
- S: Synthesis phase, cell replicates DNA and is least radiosensitive phase as the cell contains 2 sets of DNA.
- G2: Second growth phase
- M: Mitosis phase, cell divides into two cells and is most radiosensitive to disruption also well oxygenated.

Biological Responses

Stochastic effects: probability of effect increases with dose and effect is binary. No threshold.(Cancer)

Non-Stochastic effect: magnitude of effect increases with increased dose, has minimum dose, and a threshold dose.(erythema, cataracts, etc.)

Total Body Exposures

- Prodromal phase(minutes to days): marked by the emergence of initial symptoms.
- Latent Phase(days to weeks): symptoms are reduced or eliminated.
- Manifest illness(days to weeks): Primary symptoms
- Death or recovery.

Primary Responses

- Hematopoietic Syndrome(1-6 Gy): Drop in blood cell count, infections, poor wound healing, moderate fecer, headache, cognitive impairment, alopecia(3 Gy)
  - Survival 50-95%, Recovery in 6-8 weeks
- Gastrointestinal Syndrome(6-30 Gy): Above symptoms, nausea, vomiting, loss of appetite, abdominal pain, death is common without extreme treatments.
  - Survival 0-50%, >99% fatality above 8 Gy, Infection cause of death, Survival time is 2 days to 2 weeks.
- Neurovascular Syndrome(>30 Gy): Above symptoms, dizziness, headache, seizures, tremors, ataxia, loss of consciousness.
  - 0% survival, survival time 1-2 days.

Serial organs are organs in which disabling any subunit causes entire organ failure. (spinal cord, brain stem, optic structures.)

Parallel organs are organs in which subunits can be disabled without entire organ failure. (Lungs, Liver, Kidneys)

Organ Specfic:

- Skin: Epilation(3 Gy), Erythema(6 Gy), Dry Desquamation(12 Gy), Wet Desquamation(25 Gy), Radionecrosis(50 Gy)
- Eyes: Cataracts(5 Gy)
- Whole Body: Death(3-5 Gy)

Therapeutic Ration is an indication of how successful radiotherapy incorporating probability of cure and adverse complications.

- Equal Dose: $Therapeutic\ Ratio$: $\frac{Damage\ to\ tumor\ cells}{Damage\ to\ normal\ cells}$
- Equal Effect: $Therapeutic\ Ratio$: $\frac{Damage\ to\ normal\ cells}{Damage\ to\ tumor\ cells}$

Linear Quadratic Model: $S(D) = e^{-\alpha D - \beta D^2}$

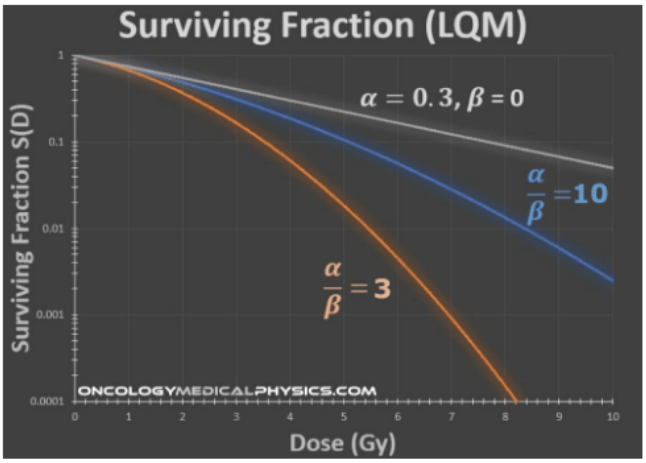
- D is the dose administered.
- S(D) is the fraction of cells to survive a given dose.
- αD probability of cell death from single “double hit” producing a double strand break.
- βD^2 probability of cell death arising from multiple “single hits”, generating single strand breaks close enough to cause double strand breaks.
- Assumptions: DNA hits are random probabilty proportional to dose, double strand breaks are required for cell sterilization, 2 methods of double strand break(multiple single, double)

Alpha/Beta Ratio: indicates how resistant a cell is to radiation damage.

- High alpha/beta ratio(~10) indicate single hit damage does not readily accumulate to lethal effects with little increase w/ dose. More linear graph
- Low alpha/beta ratio(1-3) accumulation of multiple single hits produces increased lethality.

- High alpha/beta ratio tumors benefit from fracionation, to reduce normal tissue toxicity.

Tissue Type	Alpha/Beta
Tumor and Early Effects	10
Late Complications	3
Late CNS Effects	2
Prostate	1.5



Fractionation is division of treatment dose into several dicrete treatments.

- Standard Fractionation, 1.8-2 Gy fraction, 1 treatment per day M-F, 2 Gy is expected to kill 50% of tumor cells, 30 fractions would reduce tumor cells to < 1%
- BID(Hyperfractionation), 2 fractions a day seperated by 6 hours, 1.2 Gy per fraction, less late effects to CNS as early effects are unchanged.
- Hypofractionation, >2 Gy per fraction, up to 1 fraction a day, increases late effects, decreases early effects.
- Radiosurgery, 1-5 fractions with doses form 8-90 Gy per fraction, not well understood.

Five R's

- Repair: repair rate of tumor and normal tissue.
- Repopulation: Cell division and population growth.
- Reoxygenation: Tumors are anoxic as cells die off, more oxygen for remaining tumor cells.
- Redistribute: distribution of cells in a cell cycle.
- Radiosensitivity

Biologically Effective Dose(BED) find an equivalent fractionation scheme if BEDs equal.

- $BED = nd\left(1 + \left(\frac{d}{\frac{\alpha}{\beta}}\right)\right)$, n is # of fractions, d is dose per fraction.

Equivalent Dose(EQD2) is standard for 2 Gy per fraction.

- $EQD_x = nd\left(\frac{d+\frac{\alpha}{\beta}}{x+\frac{\alpha}{\beta}}\right)$, x is set to 2 for EQD2

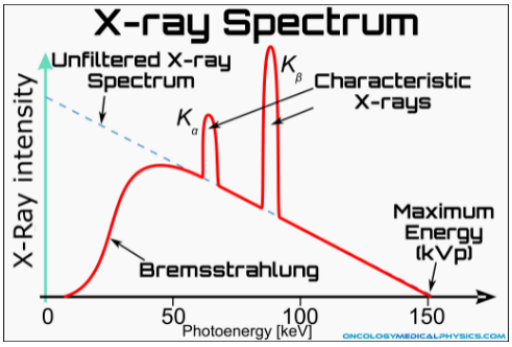
Relative biological effectiveness(RBE) ratio of absorbed dose required to produce an effect under reference conditions to absorbed dose required to produce the same effect under other conditions.

- $RBE = \frac{D_{ref}}{D_{eval}}$, *positive RBE indicates that the dose evaluated is more effective.

Oxygen Enhancement Ratio(OER) is a radiosensitizer which improves the RBE compared to anoxic cond.

- $OER = \frac{D_{anoxic}}{D_{oxic}}$, OER for photons is typically 2.5 and 3, neutrons is about 1.5.

Medical imaging



Theory of Operation

- Cathode heated filament emits elecrons via thermonic emission
- High voltage is applied between the cathode and anode which accelerates the electrons toward the anode.
- Bremsstruahlung x-rays are produces when accelerated electrons interact with the nucleus in the anode/target.

X-ray Spectrum: primarily comprised of bremmsstahlung with a < 1% efficiency.

- Proportinal to ZE, Z is the atomic number of target, E is the accelerating potential in kV.
- Mean energy is approx ½- 1/3 of max energy.
- Filtration attenuates the low energy photons, causing a hump shape.
- Characteristic X-rays emitted when photon interacts with an inner shell electron is ejected.
  - $E_{Xray} = E_{vacant\ shell} - E_{outer\ shell}$

X-ray Tube Design

- Anode(+) target for accelerated electrons, high atomic number material to raise efficiency and heat build up. Made of Tungsten is superior because of high melting point.
- Cathode(-) wire filament and negative charged focusing cup which concentrates electrons.
- Glass Envelope maintains vacuum within the x-ray tube.

Power Generation

- Power is supplied with DC voltage, need a full bridge rectifier and transformer to convert 120V AC to kV DC.
- Transformers use electromagnetic induction to transfer energy. Can be step-down, step-up, or remain unchanged.
  - Transformer Law: $\frac{V_s}{V_p} = \frac{N_s}{N_p}$, V(s,p) voltages sec/pri, N(s,p) number of windings.

Heat management: Target nees to have a high melting point, target anode should rotate to spread heat(3000-10000 rpm), Oil encasing the tube carry the heat away.

Factors impacting photon fluence

- kVp is the maximum voltage across the x-ray tube
- mA/mAs is the tube current, $Power = kVp * mA$, $Energy = kVp * mAs$
- Heel effect: Lower intensity of x-ray beam toward anode side due to self attenuation by the target material.

Image Quality Metrics

Noise(Quantum Mottle) is image noise fluctuation in pixel intensity due to statistical uncertainty. Typically measured as the standard deviation of pixel intensity over a uniform area.

- $\sigma = \sqrt{\frac{\sum_{i=1}^{N}\left(\bar{x}-x_i\right)^2}{N}}$, σ is standard dev, xi is an event, xbar is average number of events, n is numebr of measurements.
- Statistics can be modeled as a gaussian distribution, but for a large number of information carriers may be approximated Poisson Distribution.
  - $P(x) = \frac{m^x}{x!}e^{-m}$
  - $\sigma = \sqrt{\bar{x}}$, $SNR_{poisson} = \frac{\bar{x}}{\sigma} = \sqrt{\bar{x}}$, Rose criteria SNR 5 or greater for reliable detection

Contrast is the difference in intensity between two objects in an image. Contrast = Ia-Ib

- $CNR = \frac{\overline{N_S} - \overline{N_{BG}}}{\sigma_{BG}}$, Mean # of x rays per pixel for structure and BKG, standard dev BKG
- Inherent contrast between subject and background
- Display contrast Window/Level
- Physical perturbations, fogging, scatter, and non-uniform detector efficiency

Spatial resolution is minimum separation between 2 high contrast objects such that they appear as 2 separate objects, often measured in line-pairs per millimeter(lp/mm).

- Affected by Geometric penumbra, subject unsharpness, detector unsharpness, motion.

Nyquist Frequency(Fn) is the minimum sampling frequency required for an object to be distinguished.

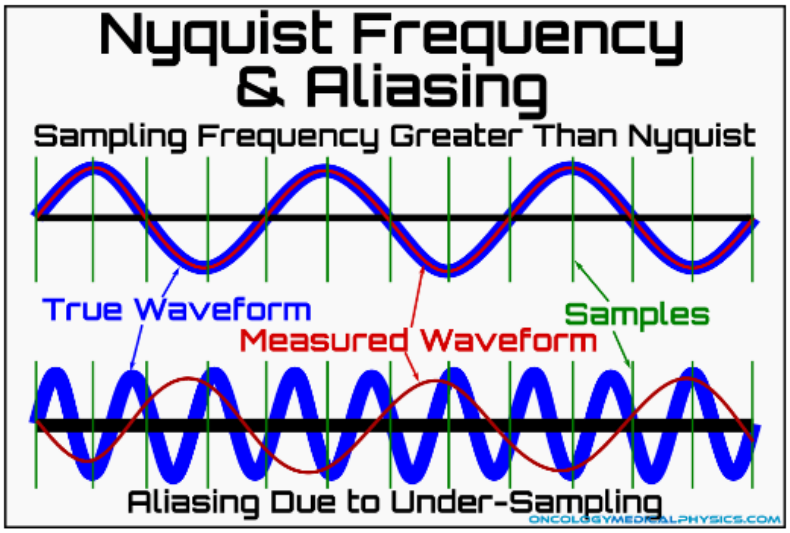
- $F_N\left(\frac{lp}{mm}\right) = \frac{1}{2R(mm)}$, smaller objects have a higher freq in the freq domain.

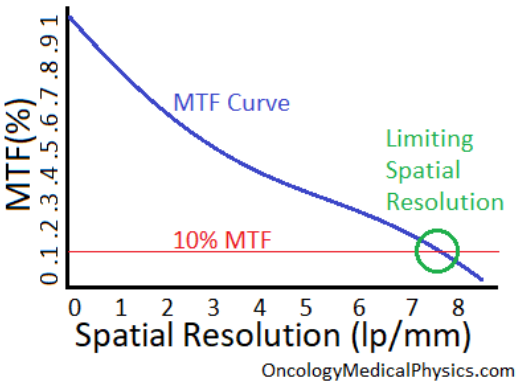
Quantitative Image Analysis

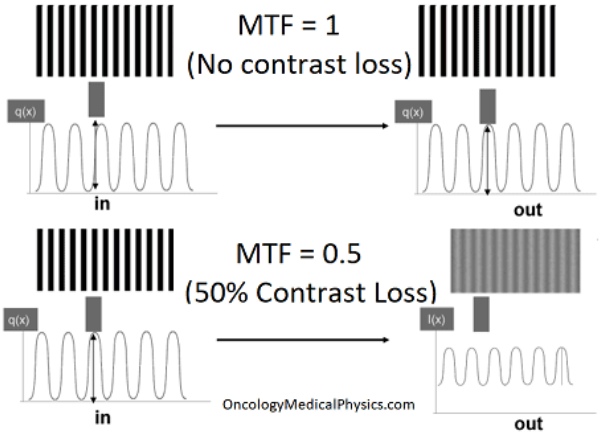
- $CNR = \frac{I_A - I_B}{\sigma}$
- Noise Power Spectrum is the magnitude of image noise variance with spatial freq.
- MTF is a measure of a detector systems ability to render contrast as a function of spatial resolution. MTF = 0.1 is generally taken to be the limit of detectability for humans
  - $MTF(f) = \frac{Contrast_{out}}{Contrast_{in}}$
- Detective Quantum Efficiency(DQE) is a measure of the imaging system's efficiency in converting data carriers into an image. Influences by every component of the imaging system and a good metric of overall dose efficiency.
  - $DQE = \frac{SNR_{Out}^2}{SNR_{In}^2}$

- Quantum Detection Efficiency(QDE) is a measure of how efficient the detection system is at collecting information carriers.
  - $QDE = \frac{n_{collected\ information\ carriers}}{n_{incident\ information\ carriers}}$

Modality	Spatial Resolution (mm)
Screen Film Radiography	0.08
Digital Radiography	0.2
Fluoroscopy	0.125
Screen Film Mammography	0.03
Digital Mammography	0.05-0.1
CT	0.3
SPECT	7
PET	5
MRI	1
Ultrasound (5MHz)	0.3







Shielding for Imaging Systems

Room Type	Typical Shielding
High Energy Linear Accelerator Primary Barrier	• 1.5-3m concrete
High Energy Linear Accelerator Secondary Barrier	• 1-1.5m concrete

High Energy Linear Accelerator Door	• With maze
	  ◦ 0.5 - 2cm inner lead layer
	  ◦ 2-4cm BPE
	  ◦ 1cm outer lead layer
	• No Maze
	  ◦ 5-9cm inner lead layer
	  ◦ 15-25cm BPE
	  ◦ 1cm outer lead layer

Ir-192 HDR Suite	• ~50cm concrete
PET/CT Room	• 1-2cm lead • 15-20cm concrete
CT Room	• 1/16" to 1/8"Pb walls • 1/32" to 1/8" Pb ceiling and (possibly) floor
Radiographic Suite	• Primary Barriers: 1/16"to 1/8"Pb • Secondary Barriers: 1/23"to 1/16" Pb

CT shielding(NCRP 147)

- $B = \frac{Pd^2}{TK^1N}$
  - B is maximum shielding transmission factor
  - P are shielding goals,
    - Controlled 0.1 mGy/wk, 5 mGy/yr
    - Uncontrolled 0.02 mGy/wk, 1 mGy/yr
  - d is the distance from the bore center to POI
  - T is the occupancy factor
  - K1 is the air kerma per pt at a distance of 1 meter.* can use CDTI, DLP, Isodose map
  - N is the number of patients per week
- No use factors due to no primary barriers, actively shielded beam stop.
- $Barrier\ Thickness = \frac{1}{\alpha\gamma}\ln ln\left(\frac{\left(\frac{NTK^1}{Pd^2}\right)^{\gamma}+\left(\frac{\beta}{\alpha}\right)}{1+\frac{\beta}{\alpha}}\right)$, alpha/beta/gamma are fitting parameters given for broad beam x-ray source attenuation of a given material (NCRP-147)
- CTDI100 method uses measured values with a pencil ion chamber 10 cm long.
  - $K^1 = \kappa\frac{L}{p}CTDI_{100}$, p is pitch, L is scan length, k is scatter fraction per cm
    - $k_{head} = 9 * 10^{-5}cm^{-1}$
    - $k_{body} = 3 * 10^{-4}cm^{-1}$
- Dose length product(DLP)
  - $K^1 = \kappa * DLP$
    - $k_{head} = 9 * 10^{-5}cm^{-1}$

- $k_{body} = 3 * 10^{-4} cm^{-1}$

- Isodose Map Method is a map provided by the vendor.

PET CT Shielding (TG-108)

Shielding for PET/CT focuses on F-18 based PET scanning because it is the most common radionuclide, and has a long half-life(110 mins). Dominated by 511 keV photons.

- $B_{Scanner\ room} = \frac{12.8\ P\ d^2}{T\ N_w A_0 F_u t_l R_{tI}}$, $B_{Uptake\ room} = \frac{10.9\ P\ d^2}{T\ N_w A_0 t_u R_{tU}}$
- $R_t = 1.443 * \left(\frac{T_{\frac{1}{2}}}{t}\right) * \left(1 - e^{\frac{-0.693t}{T_{\frac{1}{2}}}}\right)$, $F_U = e^{\frac{-0.693t}{T_{\frac{1}{2}}}}$
  - B is max shielding transmission factor
  - P are shielding goals,
    - Controlled 0.1 mGy/wk, 5 mGy/yr
    - Uncontrolled 0.02 mGy/wk, 1 mGy/yr
  - d is the distance from the bore center to POI
  - T is the occupancy factor
  - Nw is the number of patients per week
  - A0 is administered activity(MBq)
  - tU and tI are the time in uptake room and time in imaging room(hr)
  - FU is uptake decay factor
  - Rtu and Rti are dose reduction over uptake and imaging times.
    - Rt=0.91 for 30 mins
    - Rt=0.83 for 60 mins
    - Rt=0.76 for 90 mins

CT Theory of Operation

Basic Imaging Process

1. X-ray tube emits an x-ray beam at a given gantry angle.
2. The beam passes through the patient and is intercepted by an imaging detector element.
3. The detector element emits a scintillation photon which is detected by a photon detector and converted into an electronic signal.
4. The electronic signal of each element at each gantry position is assembled in a computer and a sinogram is generated.
5. The sinogram is converted into a CT image using either filtered back projection or iterative reconstruction.

Ray: single transmission measurement made by a detector element at a given gantry position.

Projection: The sum of all rays at a given gantry angle is referred to as a projection.

Sinogram: A sinogram is a map of projections as a function of gantry orientation.

Hounsfield units: Maps of the linear attenuation coefficient(μ) of each voxel.

- $HU_{x,y,z} = 1000\frac{(\mu_{x,y,z}-\mu_{water})}{\mu_{water}}$

Linear Attenuation Coefficient(μ) is the percentage of a beam attenuated per unit path length.

- $\mu = \frac{1}{x} * \ln ln\left(\frac{I_0}{I_x}\right)$, I0 is initial intensity, Ix is intensity after x distance of material.
- Compton interaction is proportional to Z/A

Scanner Hardware: Gantry, Bore(70cm-85cm), Set up lasers, Couch, Control Panel.

Internal Components:

- X-ray source
  - Tungsten allow target, Oil circulated for heat dissipation,
  - Focal spot size: 0.6-1.2mm, smaller is less penumbra but more heat build up
  - Operating voltage: 80-140 kVp
  - Beam filtration: 5-10 mmAl
- Detector Array
  - Multiple rows of scintillation detectors read out by photodiodes and converted to a digital signal. (Gd2O2S)
  - Collect 64+ slices simultaneously
- Bow tie filter attenuated lateral portions to make more uniform field(less tissue)

Image Acquisition Parameters

- Energy 80-140 kVp, increasing kVp decreases pt dose and decreases soft tissue contrast
- Current 100-600 mA, increasing mA increases pt dose, reduced image noise.
- Automatic Exposure Control: modulated kVp or mAs to normalize detector fluence, yielding a roughly uniform SNR while minimizing dose.

Spatial Resolution

- Image grid: number of voxels in each slice typically 512x512
- Field of View: typically 50 cm in diameter, with 512x512 grid that's roughly 1mm per voxel for that FOV.
- Slice thickness can be anywhere from 0.6 to 5 mm per slice

Axial/Helical

- $Pitch = \frac{table\ movement\ per\ rotation}{beam\ width}$
  - Increase in pitch, decreases scan time, pt dose, and image resolution. Increases image noise. Pitch=1 no gaps, Pitch=0.5 overlap, Pitch=2 gaps.

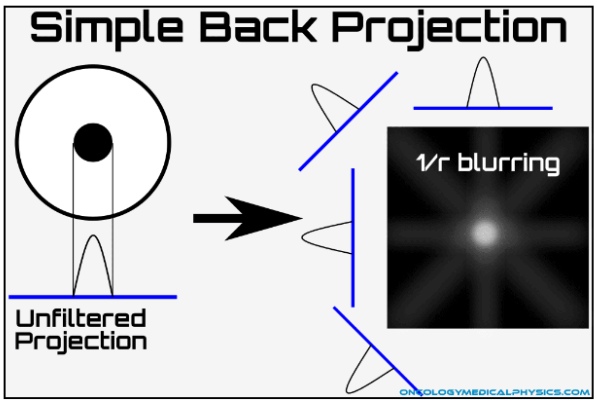
Factors impacting image noise:

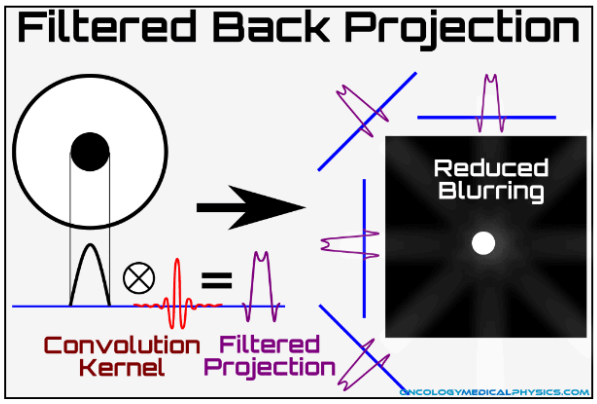
- $Noise\ \sigma = \left(\frac{B}{V*mAs}\right)^{\frac{1}{2}}$
  - B = pt attenuation
  - V = voxel size
- To reduce image noise by 50% the number of photons per voxel must be quadrupled.

Contrast Agents: take advantage of k-edge increase in photon attenuation via PE effect

- Iodine: k-edge around 33.2 keV, used for vascular studies.
- Barium: k-edge around 37.4 keV, used for gastrointestinal studies.

Reconstruction Algorithms:





- Many different kernels to improve soft tissue contrast, increase sharpness, or artifact reduction.
- Ramp filter applied to k-space image to eliminate 1/r blurring, by suppressing low spatial frequencies.

Iterative CT: high SNR w/ same dose. Seed image bias final image, possible false details.

- 1. Initial seed image generated

- 2. Computer determines what projections would have been if seed was true image.
- 3. Difference between calc proj and measured is computed to update seed image,
- 4. Repeat process until the difference between calc and measured in low enough.

CT Dosimetry:

- CT Dose phantoms are used to measure CTDI, a pencil chamber in a 16 or 32 cm diameter PMMA cylindrical phantom.

Computed Tomography Dose Index(CTDI) is the integral of the dose profile along the z-axis of phantom divided by nominal beam width.

- $CTDI = \frac{1}{NT}\int_{-\infty}^{\infty} D(z)dz$, N is number of slices, T is the slice width.
- $CTDI_{100}$ is the cumulative dose at the center of a 100mm(10 cm) axial scan
  - $CTDI_{100} = \frac{1}{NT}\int_{-50\,mm}^{50\,mm} (C * f * R(z) * dz)$
    - C is ion chamber calibration factor ~ 1
    - f is the exposure-to-dose conversion factor ~0.87 cGy/R
    - R is the chamber reading
- $CTDI_w$ is the weighted average of CTDI100 values measured at the center of the phantom and at the edge of the phantom.
  - $CTDI_w = \frac{1}{3}CTDI_{100,center} + \frac{2}{3}CTDI_{100,edge}$
- $CTDI_{vol}$ normalizes dose from a helical scan with an arbitrary pitch of 1.
  - $CTDI_{vol} = \frac{1}{Pitch}CTDI_w = \frac{N*T}{I} * CTDI_w$
    - I is table motion per rotation. (mm/rot)

Dose Length Product(DLP) accounts for differences in scan length.**(AAPM report No. 96)**

- $DLP = CTDI_{vol} * scan\ length$
- Effective Dose used to estimate the radiation risk from various radiological exams. This uses a weighting factor which takes into account risk to an organ relative to a whole body exposure.
  - Effective Dose = k*DLP

Size specific dose estimate attempts to make CT dose estimates more applicable to individual patients by applying a conversion factor (f_size) to CTDI_vol. F is found in a lookup table in TG-204. Believed to be accurate to within +/- 20%.

- $SSDE = f_{size} * CTDI_{vol}$
- $Effective\ diameter = \sqrt{AP\ separation * LAT\ separation}$

CT Artifacts:

Physics based artifacts: artifacts that arise from physical processes involved in image acquisition.

- Beam hardening: results in cupping effect or streaking effect between two dense objects
- Photon Starvation(Noise): insufficient # of photons reaching the detector, resulting in streaks along high attenuation beam path.

Patient base artifacts: caused by factors related to the patient during scan.

- Metal Artifact: streaking artifacts, also scanner maximum CT number may be reached meaning heavy metals may have same CT # as hard bone, causing dosimetric error due to difference in attenuation. Can be reduced by increasing kVp with MVCBCT or reconstruction algorithms.
- Motion Artifact: movement causes directional shading and streaking.

Scanner based artifacts: resulting from imperfections in scanners function.

- Partial Volume artifact: averaging of CT number over the volume of the voxel. Example, pixel containing bone(800 HU) and lung(-800 HU) would display a CT # of 0.
- Projection/View Aliasing(under sampling): too few projections in imaging reconstruction results in aliasing of high frequency objects. Projecting from hard, high contrast edges within the image.
- Ring Artifact: caused by one or more detectors out of calibration, manifests as ring superimposed on the image.
- Cone Beam Artifact: more slices acquired per rotation increases; the beam becomes con-shaped rather than fan. Beam divergence of the cone causes under sampling for objects farther away the central axis.

MRI Design and Operation:

Quick Reference

- Signal source results from atoms with unpaired protons.
  - Primarily Hydrogen
  - Also: O, F, Na, and K
- Magnetic Field Strength: 0.1T - 7T (earth's magnetic field is 6.5×10^{-5}T)
  - Increased field strength increases signal-to-noise ratio but also increases geometric distortions.
- The precession frequency is given by the Larmor equation

$$\omega_{Larmor} = \gamma B$$
$$\gamma_{hydrogen} = 42.58 MHz/T$$

- Gradient coils are responsible for the loud noise an MRI makes.
- Signal strength (S) is proportional to square of the magnetic field (B).

$$S = \frac{N\gamma^3\hbar^2B^2}{4kT}$$

Contrast Weighting

	Short TR (TR = T1)	Long TR (TR > T1)
Long TE (TE = T2)	Not clinically useful	T2 weighted image
Short TE (TE < T2)	T1 weighted image	Proton density image

- T2 is much smaller than T1 (milliseconds versus seconds).
- A reliable way to distinguish T1 and T2 weighting is to looks at fluid-filled structures which will appear dark on T1 weighted images and bright on T2 weighted images.

Theory of Operation

Basic Steps in Imaging:

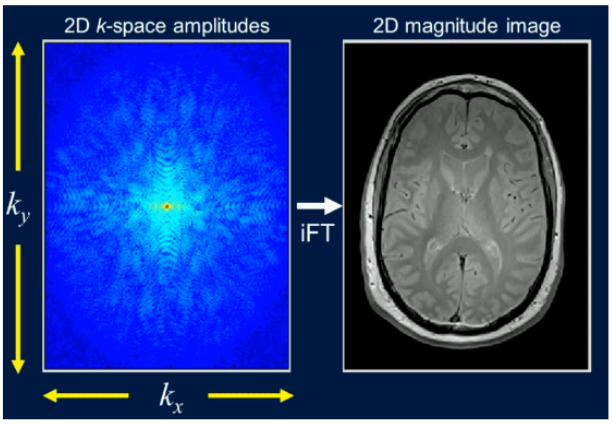
1. A strong magnetic field(0.1T-7T) is applied to the pt aligning hydrogen atoms(parallel and antiparallel) with the field.
   a. Parallel alignment is only slightly preferred(~3ppm/T).
2. Once aligned, the protons precess about their poles at a frequency given by larmor freq.
   a. $\omega_{Larmor} = \gamma B$, $\gamma_{hydrogen} = 42.58 \frac{MHz}{T}$
3. A resonant radio frequency(RF) pulse of Larmor Freq is used to excite the aligned atoms.
   a. Excitation causes the amplitude of their precession to increase.
4. After the pulse the excited atoms return to their lower energy state by emitting RF waves at their Larmor frequency.
5. Receiving coils detect the emitted RF waves, recording their signal.

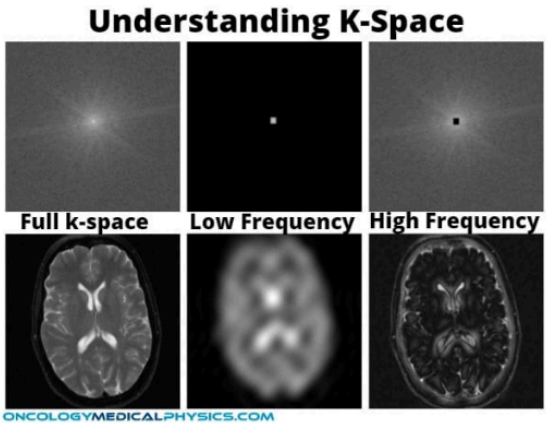
Voxel Encoding: signal is just intensity and frequency of RF return, signal is then encoded in 3 cartesian axes.

- Slice Selection: first method, selectively excite only 1 slice at a time. Gradient mag field is applied in addition to uniform magnetic field. This causes the larmor freq to vary as a function of position along the direction of the field gradient.
  - Slices of variable width may be selected by changing the RF pulse freq & bandwidth.
- Phase Encoding: temporary small gradient briefly superimposed in a direction perpendicular to the slice selection direction. This speeds up atoms on one side of the slice and slowing on the other. When shut off the original precession freq are now at different phases.
- Frequency Encoding: Final step uses gradient coils to create a magnetic field gradient along the axis perp to prior 2. This gradient is left on during readout.

Image Reconstruction: inverse Fourier transform in k-space.

- K-space is an image mapping frequency rather than x/y coordinates. Pixels near the center correspond to low frequency data while data near the edge are high frequency.





Source of Tissue Contrast:

T1 (Longitudinal Relaxation, Spin-Lattice relaxation): This is the time it takes for bulk magnetization to regrow to 63% of its equilibrium value. It is the time that it takes for the protons to realign with the magnetic field.

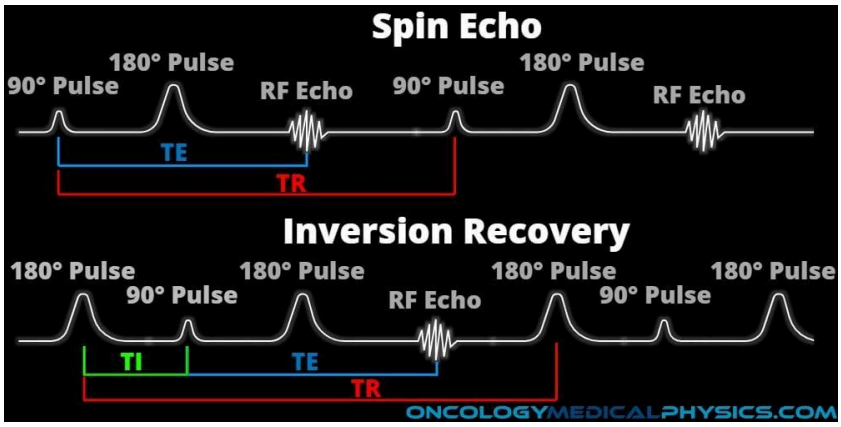
T2 (Transverse Relaxation, Spin-Spin relaxation): Orthogonal to the direction of the magnetic field. Normally the phases of precession are randomly distributed and the transverse mag is zero. After excitation the phases are aligned and signal can be detected. T2 is the time required after excitation for transverse mag to decay to 37% of its maximum value.

- Random magnetic disturbances rapidly decay this signal in a precess called Free Induction Decay(FID); the rate of FID is characterized by the T2 time.

Scan Parameters: Choice of TR and TE is driven by the T1 & T2 times of the tissues being distinguished. Timing is chosen to maximize the difference in signal intensity.

- Repetition time(TR): the time between excitation RF pulses.
  - Controls the T1 weighting
- Echo time(TE): The time between excitation and signal acquisition.
  - Echo in MRI is similar to a sound echo is that it is a return of the FID signal at some time after the initial FID. Echo is generated by using a 90 degree flip angle then at time TE/2 employing a 180 degree flip angle. Echo imaging allows for rapid acquisition.
  - Controls the T2 weighting
- Flip angle: The angle by which the net magnetization is directed away from the mag field. 90 or 180 degrees, used to adjust contrast and create echo time weighting.
- Time of Inversion(TI): inversion recovery sequences operate by initially inverting the magnetization with a 180-degree pulse, then allowing some degree of T1 recovery prior

to emitting a 90-degree pulse to yield contrast. TI is the time between this initial 180 degree pulse and the subsequent 90-degree pulse.



Scanner Components:

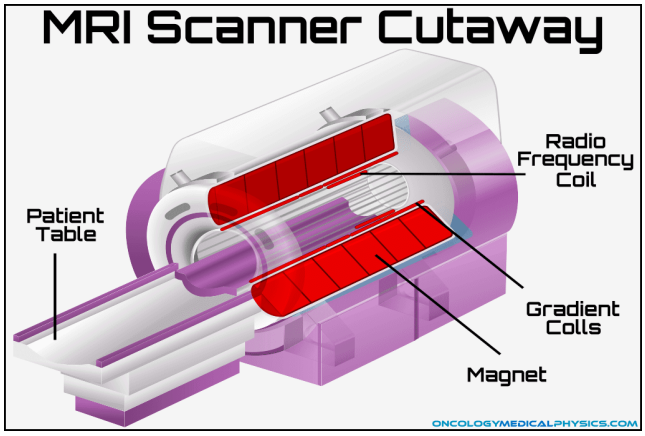
Primary magnet- creates the static magnetic field(0.1-7T)

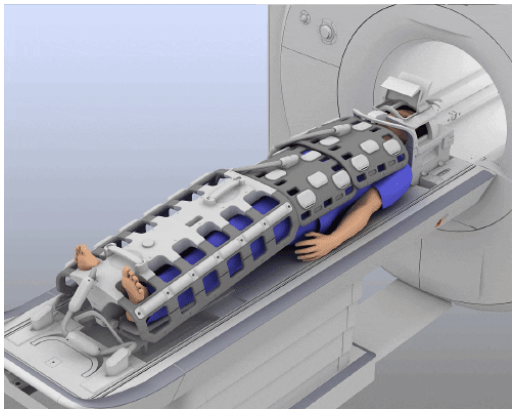
Gradient Coils- creates a linearly varying magnetic field(mT) used for spatial encoding. Used to produce contrast in diffusion/flow imaging. Also responsible for the noise in the MRI.

RF coils- Sends and receives the radio frequency signals(uT) used to excite hydrogen atoms to their Larmor frequency.

Shim Coils- Correct for magnetic field inhomogeneities from primary magnet.

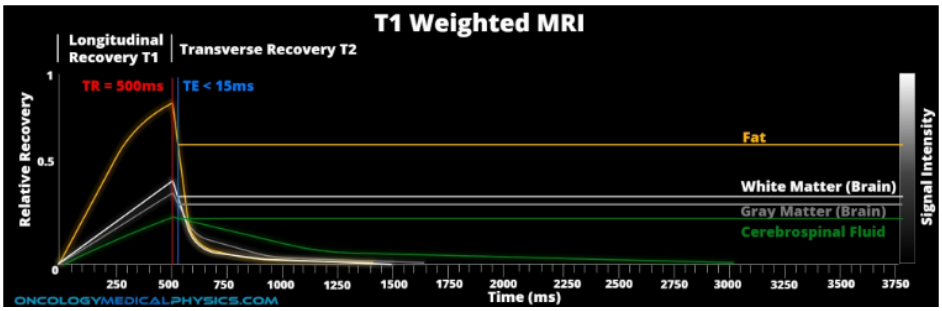
Helium Cooling System- the magnets in a modern MRI scanner must be superconducting and require extremely low temps(~3-4 Kelvin). Liquid Helium is used to achieve these temps.

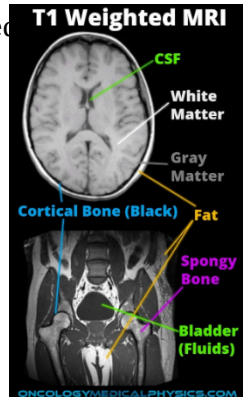




Contrast and Weighting Schemes:

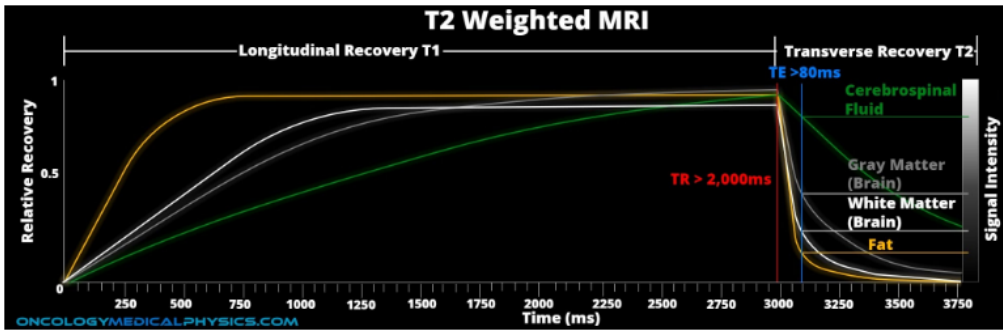
T1 Weighting(SHORT TR and SHORT TE): attempts to maximize differences in T1 relaxation while minimizing the impact of T2 relaxation. TR is approximately equal to T1(500ms) while TE is shorter than T2(<15ms). T1 weighting offers good tissue contrast but sees less use in RT than T2 due to limited tumor contrast. T1 is used to assess nodal invasion, the most common sequence is spoiled gradient echo.

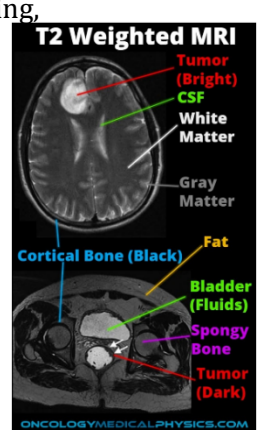




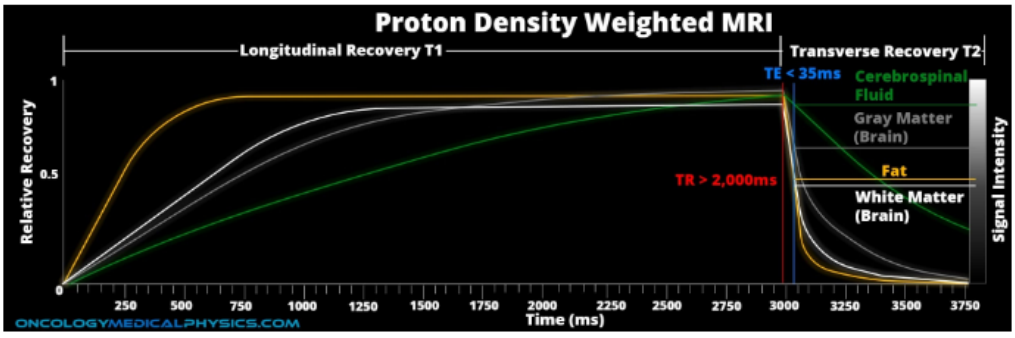
T2 Weighting(LONG TR and LONG TE): T2 contrast offers higher soft tissue contrast. Most widely used weighting scheme in radiation therapy planning. Long TR(>2000ms) allows near full T1 recovery while long TE(=T2) creates contrast. Most common is Fast spin echo.

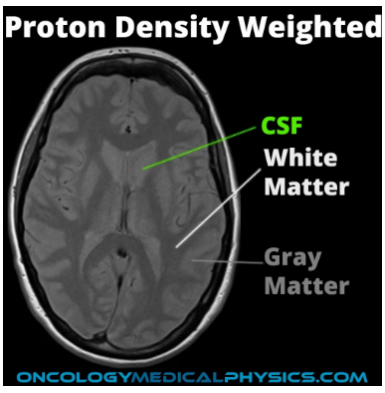
- Cancerous tissue tends to have longer T2, which makes them bright in this weighting, especially when surrounded by edema.



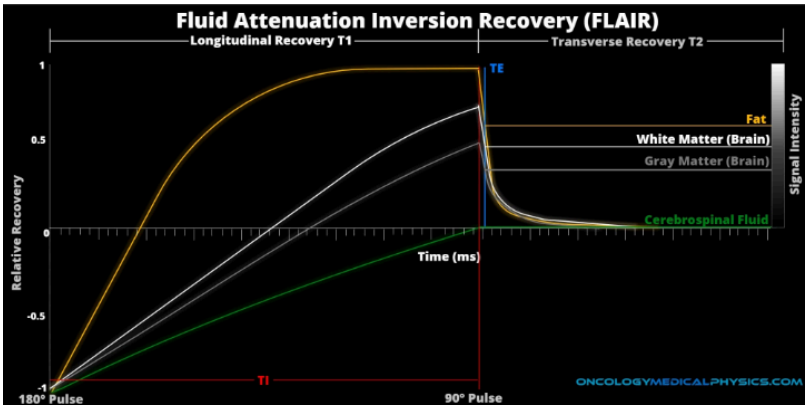


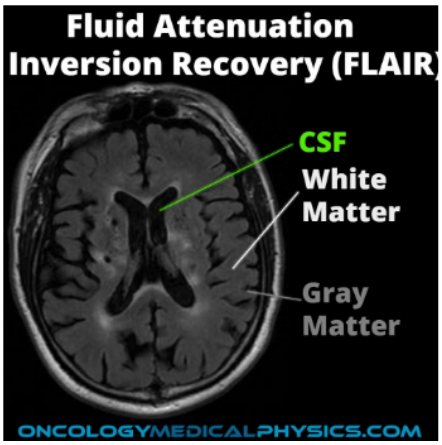
Proton Density Weighting(LONG TR and SHORT TE): derived form differences in the number of protons in a voxel. Used for low contrast imaging techniques for menisci and brain structures. Long TR (>2000ms) to allow for T1 recovery and short TE times(<TE) to minimize T2 contrast.





Fluid Attenuated Inversion Recovery(FLAIR): begins with a 180 degree RF pulse to invert the magnetic moment of the affected hydrogen. The 90-degree excitation pulse is timed to coincide with fluid T1 recovery crossing 0 magnetic moment thereby suppressing signal. Most commonly used to suppress CSF in brain scans, to separate edema from CSF.

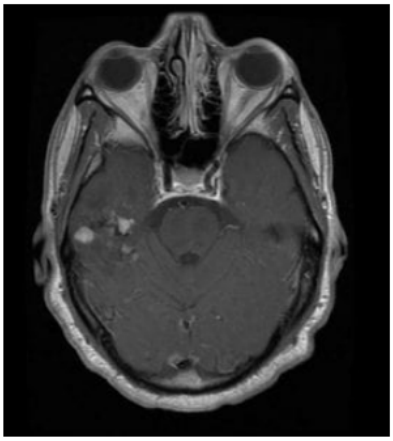




Diffusion Weighting: register Brownian motion of individual water molecules. Useful to detect necrotic regions of a tumor as necrosis appears brighter than the surrounding tissue.

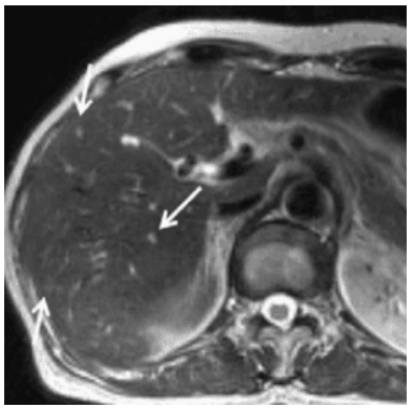
Contrast agents:

- Gadolinium: most common contrast works by shortening the T1 time in tissues that absorb it. This causes tissues that absorb Gd to appear bright in T1 images.



Gadolinium contrast enhances glioma in brain T1 weighted MRI imaging.

- Paramagnetic Iron Oxide: primarily impacts T2 times and produces a signal drop in T2* weighted images. A normal liver will absorb SPIO while a metastasized liver will not.



SPIO darkens normal liver but leaves metastasis bright in T2 weighted image.

MRI Artifacts:
Aliasing(Wrap Around)

- Appearance: Anatomy outside of the FOV is superimposed on the opposite side of the image. Most commonly in the phase encoding direction.
- Causes: Sampling at below the Nyquist frequency(< twice the RF freq of the voxel) causes the apparent freq to be lowered in a process known as aliasing.
- Artifact Reduction: Automatic wrap around removal by oversampling or by applying a low pass filter prior to ADC, change phase encode direction, increase FOV.

Chemical Shift

- Appearance: Dark or light edges on borders of structures with different chemical composition.
- Causes: All chemical shift artifacts occur because of differences in the intrinsic shielding of bodily chemical structures. Causes apparent displacement in either the freq or phase encoding direction.
    - Freq encoding: occurs as a direct result of small changes in precession freq resulting in the signal being mapped in a different location.
        - Appears as displacement with a light edge on a side and the other dark
    - Destructive phase shifts: occurs in gradient echo imaging as the result of constructive/destructive interference in dephasing and rephasing the echo.
        - Appears as a dark border around the entire structure

Geometric Distortion

- Appearance: physical distortion of the image, may not be apparent in tissue but grid phantom will show distortion.
- Causes: Result of non-uniformity of the magnetic field leading to inaccurate encoding the spatial position, or caused by the machine itself or by areas of variable susceptibility within a patient(implanted devices).
- Artifact reduction: use lower field strength scanner, place regions of interest in the center of field where there is the lowest distortion, regular QC to assure proper shimming.

Herringbone Artifact

- Appearance: A repeating pattern superimposed over the image.
- Cause: Caused by a bad pixel in k-space due to hardware fault.
- Artifact Reduction: Repeat scan, scanner repair needed.

Moire Fringes Artifact

- Appearance: Repeated irregular bands of light/dark
- Causes: Signals form different phases are superimposed, creating constructive and destructive interference of signal. Non-uniform magnetic field.
- Artifact reduction: Improve field uniformity by shimming

Magnetic Susceptibility Artifact

- Appearance: Appears as darkening/signal drop in part of image, or spatial distortion.
- Causes: Results from drastic changes in local mag field from magnetic materials in scan, ferromagnetic materials cause strong signal drop, para/diamagnetic cause lesser drop.
- Artifact Reduction: Remove metal from scan, use short echo time sequences.

Motion artifact

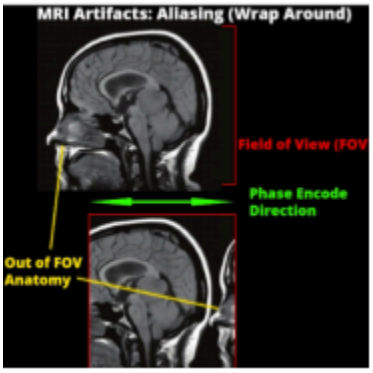
- Appearance: faint copies of the image, ghost images in phase encoding direction
- Causes: Motion, including involuntary motion and blood flow
- Artifact Reduction: Reduce patient motion, choose faster sequences

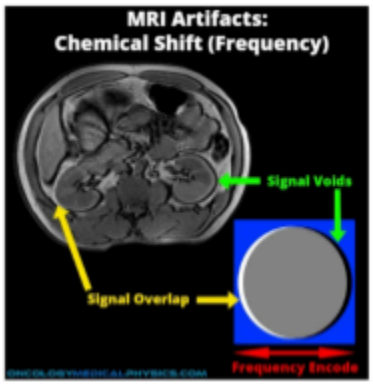
Truncation(Ringing) Artifact

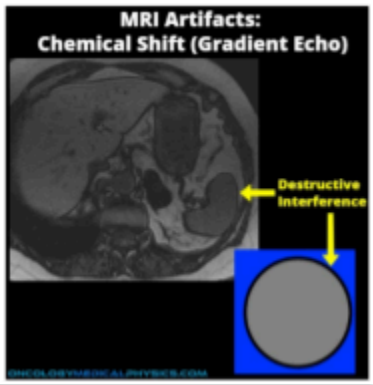
- Appearance: Fine parallel lines adjacent to a high contrast surface
- Causes: Inverse Fourier transform used to convert k-space map into image. A step gradient requires a very large amount of frequency sin waves to construct, this is cut short and results in an oscillation present in the image.
- Artifact Reduction: K-space post-processing can reduce these artifacts.

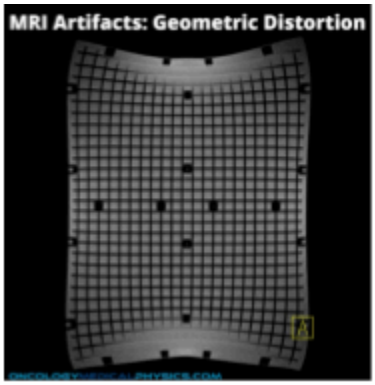
Zipper Artifact

- Appearance: Interference across one or more rows, usually in phase encoding direction.
- Causes: RF signal is picked up by receiver system(phase encoding), may arise from stimulated echoes resulting from poor slice selection profiles(Frequency encode direction)
- Artifact reduction: Assure room door is closed for RF shielding, move patient monitoring system away from scanner.

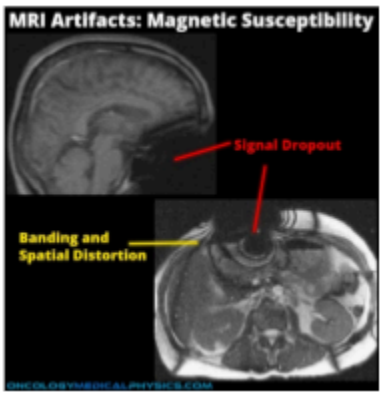

MRI Artifacts: Aliasing (Wrap Around)
Field of View (FOV
Phase Encode Direction
Out of FOV Anatomy

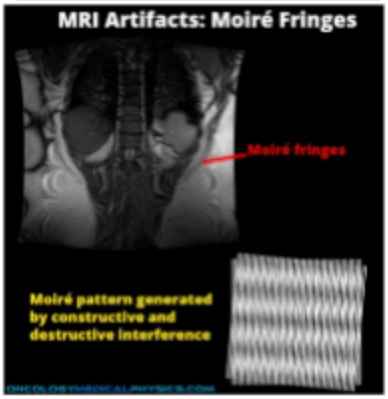

MRI Artifacts:
Chemical Shift (Frequency)
Signal Voids
Signal Overlap
Frequency Encode


MRI Artifacts:
Chemical Shift (Gradient Echo)
Destructive Interference

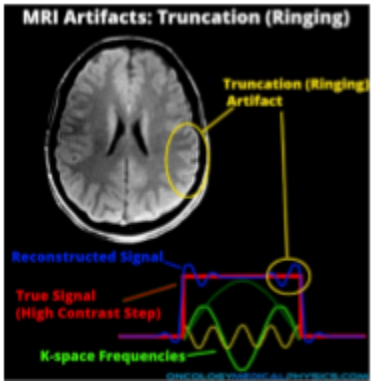

MRI Artifacts: Geometric Distortion

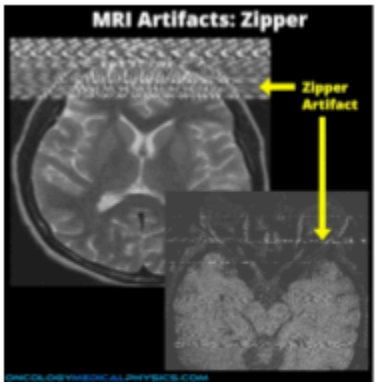

MRI Artifacts: Herringbone
Repeating pattern superimposed on image
Caused by k-space error


MRI Artifacts: Magnetic Susceptibility
Signal Dropout
Banding and Spatial Distortion


MRI Artifacts: Moiré Fringes
Moiré fringes
Moiré pattern generated by constructive and destructive interference


MRI Artifacts: Motion


MRI Artifacts: Truncation (Ringing)
Truncation (Ringing) Artifact
Reconstructed Signal
True Signal (High Contrast Step)
K-space Frequencies


MRI Artifacts: Zipper
Zipper Artifact

Nuclear Imaging

Radiopharmaceuticals: radioactive compounds used for both diagnostic and therapeutic purposes.

- Therapeutic Radiopharma
  - Intended to deliver a sufficiently large dose of the radiopharmaceutical to the target tissue, while minimizing the dose to non-target tissues.
  - To achieve target dose and minimize OAR dose, short range emissions are used, either alphas or beta emitters.
- Diagnostic Radiopharma
  - Used to determine the presence and position of biological processes or features.
    - Low radiation dose but high enough energy photon for detection
    - High radiation detection efficiency, energy exiting pt should be low enough energy to be detected
    - High specificity, should accumulate in tissue of interest and not at others
    - Safety, chemically non-toxic and safe to administer
    - Cost Effective
  - Most nuc med imaging except for pet scanners are optimized for collecting photons of about 140 keV.
- Active Transport
  - I-123 commonly used to assess the structure & function of the thyroid gland, Iodine is actively transported to the thyroid.
  - F-18 FDG is a glucose analog that concentrates in highly active cells using glucose for energy. Useful for imaging areas with cancer growth.
- Compartment Localization
  - Xe-133 gas which is inhaled into the lungs
  - Tc-99m labeled red blood cells which are introduced into the circulatory system.
- Receptor binding
  - In-11-octreotide is used which binds with somatostatin receptor sites.
- Physiochemical absorption
  - Ra-223 mimics calcium allowing it to be incorporated into mineral structure of bone. Ra-223 is an alpha emitted and is used for treatment of prostate cancer that has invaded boney anatomy.

Brachytherapy Sources

Nuclide	Half-life	Photon Energy (MeV)	HVL (mm Lead)	Exposure Rate Constant (R-cm^2/mCi-hr)
^{192}Ir	73.8 days	0.38 average	2.5	4.69
^{226}Ra	1,600 years	0.83 average	12.0	8.25
^{222}Rn	3.83 days	0.83 average	12.0	10.15
^{60}Co	5.26 years	1.17, 1.33	11.0	13.07
^{137}Cs	30.0 years	0.662	5.5	3.26
^{198}Au	2.7 days	0.412	2.5	2.38
^{125}I	59.4 days	0.028 average	0.025	1.46
^{103}Pd	17.0 days	0.021 average	0.008	1.48

PET Nuclides

Nuclide	Half-life (minutes)	Positron Maximum Energy (MeV)	Dose rate constant (μSv m^2/MBq)
^{18}F	109.8	0.63	0.143
^{11}C	20.4	0.96	0.148
^{13}N	10.0	1.19	0.148
^{15}O	2.0	1.72	0.148
^{64}Cu	762	0.65	0.029
^{68}Ga	68.3	1.9	0.134
^{82}Rb	1.27	3.35	0.159
^{124}I	6048	1.54, 2.17	0.185

Data taken from AAPM TG-108

Medical Nuclides

Iridium-192, used for HDR brachytherapy.

- Man-made isotope by neutron bombardment of Iridium 191.
- Beta decay 0.136-1.06 MeV, Avg energy 0.38 MeV
- Half life: 73.83 days, activity loss estimate is 1% per day
- Air Kerma rate constant $(\Gamma_{\delta})\kappa = 4.11\frac{cGy\ cm^2}{mCi\ h}$

Cesium-137, used in both interstitial and intracavitary brachytherapy.

- Beta decay 0.514 MeV w/ Gamma 0.662 MeV
- Half life: 30.07 years , Activity loss per year ~2%

Cobalt-60 used in external beam radiation therapy & brachytherapy.

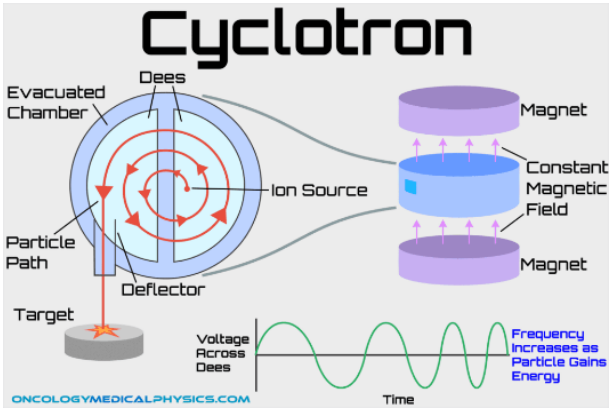
- Beta decay resulting in gamma decay of 1.18 MeV & 1.33 MeV, Mean gamma 1.25 MeV
- Half-life 5.2714 years

Iodine-125 used in LDR brachytherapy in seeds.

- Low energy 28 keV photon decay mode
- Air kerma rate $(\Gamma_{\delta})\kappa = 1.32\frac{cGy\ cm^2}{mCi\ h}$

Palladium-103 is also used in LDR brachytherapy seeds.

- Low energy 21 keV photon decay mode
- Half life 16.99 days
- Air kerma rate $(\Gamma_{\delta})\kappa = 1.296\frac{cGy\ cm^2}{mCi\ h}$



Methods of Radionuclide Production:

Nuclear Transformation Notation

- $T(P,E)R$
  - T is target material
  - P is accelerated particle type
  - E is the number & type of emitted particles
  - R is the resulting radionuclide

Cyclotrons: charged particle accelerator consisting of an evacuated cylinder divided into 2 sections referred to as “Dees”.

- Charged particles are injected into the center of the cylinder and an oscillating electric field is applied to the Dees(NUTZ). The electric field causes the charged particles to accelerate and the constant mag field causes that acceleration to have circular trajectory. Frequency of electric field oscillation must increase until desired energy is reached.
- The accelerated particles are then directed at a target to create nuclides, most radionuclides created are neutron deficient and decay via EC or positron emission.
- Most widely used radionuclides produced by cyclotrons are for PET.

Nuclear reactions

- Nuclear fission is the splitting of a heavy atom into smaller nuclei. Decay products resulting can be used as medical radionuclides.
- Nuclear Activation via neutron bombardment form nuclear reactors on stable nuclei. When a stable nucleus absorbs a neutron it produces an isotope with moderate half life.

Radionuclide Generators

- Produce short lived medical radionuclides by decay of a longer lived radionuclide.
  - Example: Molybdenum-99/Technitium-99m referred as Moly cows

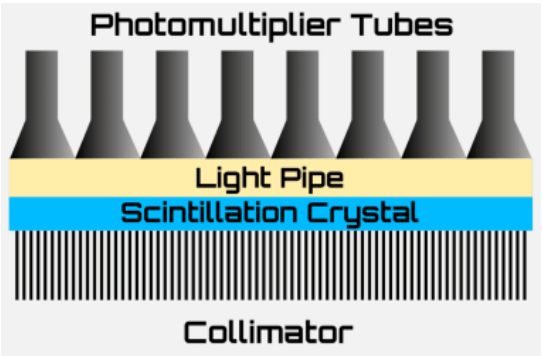
MIRD: Medical Internal Radiation Dosimetry

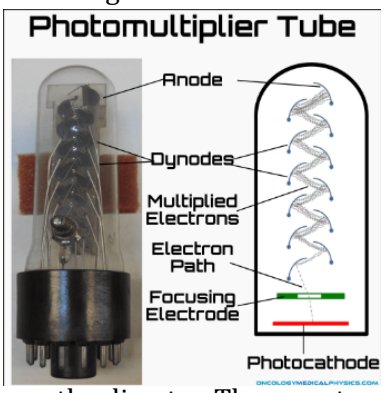
- Uses a model of the human body and considers organs as source organs which contain the radiopharmaceuticals, and target organs which absorbed dose is calculated. Organs can be both source and target.
- Computer accumulated activity:
  - $A_s(t) = A_0 F_s e^{-\lambda_e t}$
    - A_s is accumulated source activity at time (t)
    - A_0 is the administered activity
    - F_s is the fraction of pharmaceutical which is the fraction of radiation accumulated in the organ
    - λ_e is the effective decay constant
  - $Effective\ Half\ life$, $\frac{1}{t_{\frac{1}{2}effective}} = \frac{1}{t_{\frac{1}{2}physical}} + \frac{1}{t_{\frac{1}{2}biological}}$, $\lambda_{eff} = \lambda_{phys} + \lambda_{bio}$
- Determine S-factor which is the mean dose per unit of activity and has units of Gy/Bq*s.
  - Tabulated s-factors are determined by monte carlo simulation for an assumed 70 kg mean man phantom.
- Compute dose to the target organ, $\overline{D} = A_s * S$
- Computer effective dose to the whole body, $E = \sum W_t D_t$
  - Wt is the tissue weighting factor.

Scintillation (Gamma) Camera Design and Operation

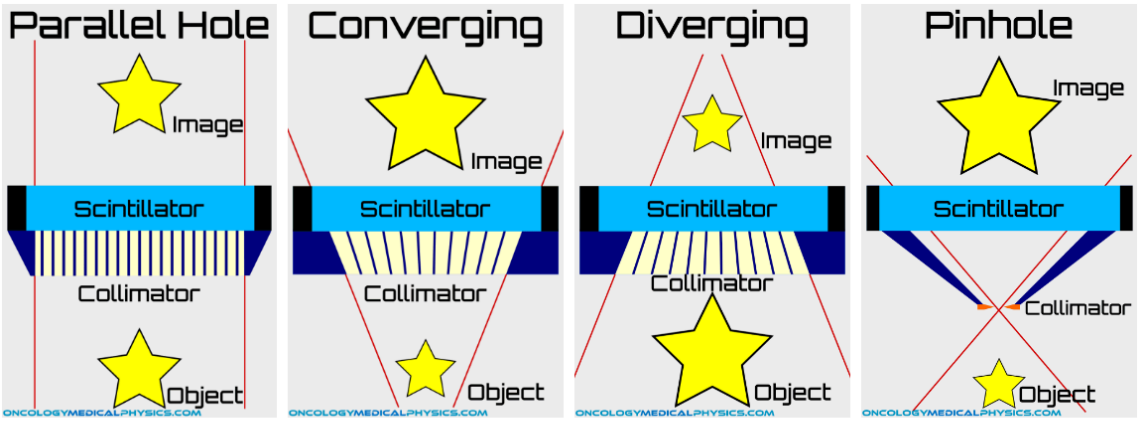
- Anger Scintillation Camera, has a single scintillation crystal coupled to multiple photomultiplier tubes which are able to encode the location of interaction.
- Scintillation Crystal emits visible or UV light after interacting with ionizing radiation.
  - High conversion factor(>10%) meaning large fraction of radiation energy deposited is converted to light,
  - Short decay time with limits the delay between irradiation and scintillation
  - Material should be transparent to its own wavelength.
  - Scintillation wavelength should be readily detectable by the readout device.
  - A large attenuation coefficient for the radiation being measured.
- NaI(TI) Scintillation Crystals
  - Advantages: high conversion efficiency(13%), High atten coeff for photons (70-365 keV) Prompt photon emission with decay constant of 250 ns, High light yield.
  - Disadvantages: Attenuation coeff is too low to efficiently collect 511 keV, crystal is hygroscopic meaning it must be enclosed to prevent water absorption. Crystals are fragile and are susceptible to accumulated radiation damage.

Photomultiplier tubes are used to convert scintillation light into electrical signals.





Collimators are used to restrict the angle of incidence of radiation upon the director. They must attenuate photons in the kV range, so they are typically constructed of materials with high atomic number, such as lead. Lead walls are **septa**, and radiation openings are **apertures**.



Parallel Hole

The most common type of collimator is the parallel hole

Converging

Converging collimators are used to produce magnified

Diverging

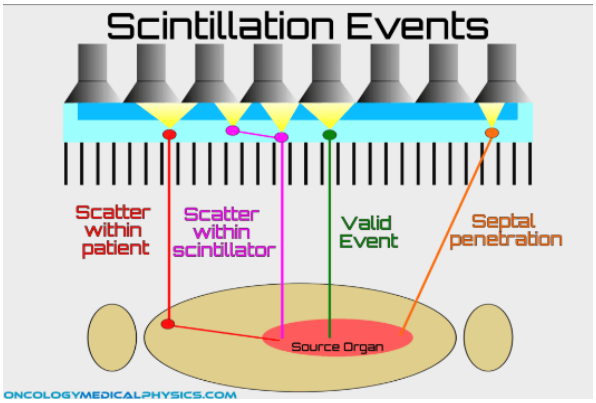
Diverging collimators are not in common use today, but were

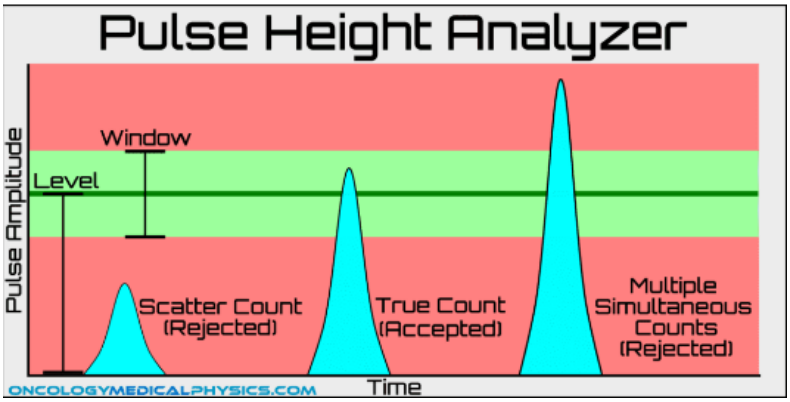
Pinhole

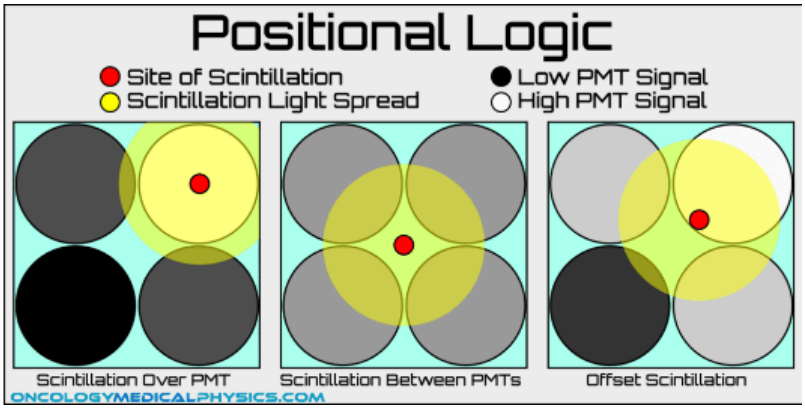
Pinhole collimators feature a single aperture approximately

Image Formation Process

1. Radiopharmaceuticals are administered to the pt and differentially accumulate based on biological activity
2. Gamma photons are generated by radioactive decay.
3. Gamma photons exit the patient and are incident upon the collimator.
4. Remaining photons not absorbed by the collimator interact with the scintillation crystal, producing scintillation photons.
    a. Valid Events
    b. Septal penetration events
    c. Object scatter events: scattered within the body and counted.
    d. Detector scatter events: scattered with the scintillation crystal.
5. Scintillation photons are converted to electronic signals by a PMT.
6. Energy discrimination circuit rejects the scattered signals.
    a. Pulse Height Analyzer used.
7. A position logic circuit determines the origin location of each signal on an x, y plane. Each localized signal is known as a count.
8. Image is formed by accumulation of these localized counts.







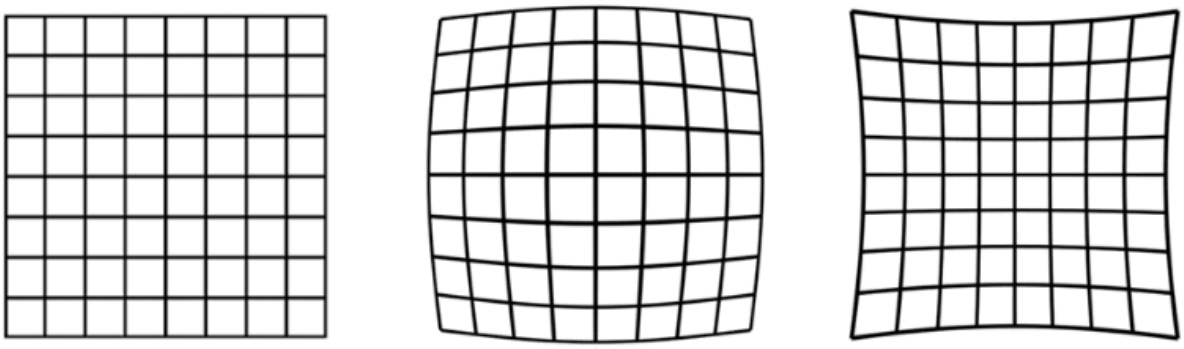
Clinical Gamma Camera can have single or double head, allowing adjustments for imaging angle.

Imaging Parameters

- Frame Mode: counts are digitized into appropriate image matric bin after detection. The counts are accumulated in the image matrix over a given amount of time building an image directly.
- List Mode: individual x,y coordinates of each count are stored along with a time marker, this allows retrospective framing of the data after acquisition.

Artifacts:

- Barrel is bowing out of straight lines and regular repeating distortion lines in flood field.
- Pincushion is bowing in of straight lines.
- Cause is imperfect mapping of scintillation location within the positional logic.
- Resolution is to perform linearity calibration, correcting for imperfect positional mapping.



Object Barrel Distortion Pincushion Distortion

	SPECT	PET
Radiation Detectors	Planar Anger Camera	Full 360 degree scintillators
Scintillator	NaI(Tl)	LSO, GSO, LYSO
Most Common Nuclide	Tc-99m	F-18 (FDG)
Attenuation Correction	Patient thickness Attenuation scans CT	CT
Photon Vector Determination	Collimator	Coincident photon detection
Spatial Resolution	~10mm FWHM Dependent on orbit and collimator choice	4-5mm

Nuclear Tomographic Imaging (PET and SPECT) creation of 3-D images via computed tomo reconstruction. Similarly reconstructed to CT with filtered back projection or iterative. It is different from CT in a few key areas:

1. Nuclear tomo imaging produces a map of biological activity identified by radiopharmaceutical markers. CT produces a map of linear atten coeff and density.
2. Nuclear tomo imaging relies on nuclides within the body to generate information.
3. Nuclear imaging the photon source is in the body, resulting photon attenuation depends strongly on the sources path length to detector. This requires an attenuation correction to prevent sources deep appearing darker than superficial ones.

SPECT

- Consists of 1 to 3 gamma cameras with dual image design being the most common.
- Image array ,
  - 64x64 or 128x128
- Collimation
  - Parallel collimators are most common, or Fan beam collimators for brain scans.
- Data collection modes
  - Step and shoot collection at discrete angles(64-128 positions), continuous collection
- Orbit
  - Circular which has a low collection efficiency and low SNR
  - Elliptical/Contoured imager follows closely to patients' body to increase collection efficiency and SNR
- Degrees for rotation: 360 is most common, 180 common for cardiac imaging
- Typical Imaging Dose = 1-5 mSv

Attenuation Correction

- Thickness based estimates: corrected using standard model and measure of pt thickness
- Attenuation Scans: taken using scanners imager using an external photon source, to create attenuation maps.
- CT imaging

Common SPECT Nuclide Tc-99m

- Used for Bone, Renal, Cardiac, Breast, Cerebral perfusion, Hepatic function.
- Gamma energy 140.5 keV
- Half-life = 6 hours
  - Impractical to transport, so generator stored on site(Moly Cow)

PET

Design

- Several rings of gantry-mounted detectors which surround the patient during imaging. Scintillation crystals in PET are different from Gamma/SPECT.
  - Crystal is usually LSO, GSO, and LYSO,
  - These crystals are more efficient at collecting the 511 keV photons.
  - Higher densities requiring smaller crystals and less self attenuation
  - Superior spatial resolution
- Coincidence Detection of the positron annihilation emitting two photons in opposite directions. This creates 2 new forms of noise
  - Scatter coincidence occurs when one of the photons is scattered prior to detection.
  - Random coincidence occurs when 2 photons arising from separate annihilations are detected simultaneously.
- Time of Flight takes advantage of the fact that photons travel at the speed of light from the point of origin to the detector. This allows PET scanners to determine the approximate point of origin along the photons path.
- PET/CT uses a coupled CT scanner for attenuation correction, taken together.
- Fluorine-18
  - Used for metabolic activity
  - Half life 109.8 minutes
  - Emission Energy: Positron 0.633 MeV (approx. range 2.4 mm in tissue)
  - Production is cyclotron bombardment of O-18
  - Typical dose is 7 mSv for adult scan

Limits on Spatial Resolution(few mm)

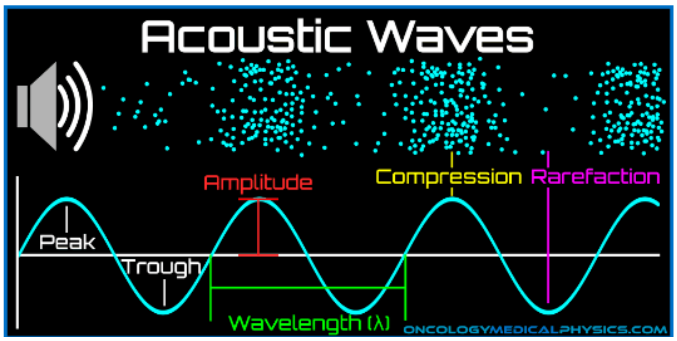
1. Positrons travel a small but non-zero distance between emission and annihilation. For F-18 the distance is approx 2.4 mm.
2. The 511 keV annihilation photons do not travel in exactly opposing directions.
3. TOF calcs are dependent upon the time required for the scintillator to emit scintillation photons after an interaction. This causes determination of point of origin to within a few centimeters for LSO or LYSO.

Ultrasound

The Physics of Sound

Speed of sound

- Speed of sound, $c = \lambda f$
  - c is the speed of sound in the medium (distance/time)
  - λ is the wavelength
  - f is the frequency of oscillation
- $c = \sqrt{\frac{B}{\rho}}$
  - B is the bulk modulus of the medium(resistance to compression)
  - P is the density of the medium



Material	Density (kg/m^3)	c (m/s)	Acoustic Impedance (Rayls) Z=ρc
Air	1.2	330	0.0004×10^6
Lung	300	600	0.180×10^6
Soft Tissue	1,050	1,540	1.617×10^6
Bone	1,912	4,080	7.801×10^6
PZT	7,500	4,000	30.0×10^6

Acoustic rarefaction and compression cause changes in local pressure of a medium. The pressure amplitude is defined as the difference between max and min pressure, and the average pressure of the medium in absence of a wave.

- SI unit is Pascal, 1 Pa = 1 kg/m
  - Typically, around 1 MPa which is 10 times atmospheric pressure
- Acoustic intensity is represented with units of mW/cm^2
- Relative Intensity(dB scale)
  - $dB = 10 \log log \left(\frac{I_2}{I_1}\right)$

Acoustic wave interactions

- Constructive interference: have same frequency and phase results in net increase in amplitude.
- Destructive interference: same freq but phase shift of 180 degrees reducing amplitude
- Complex interference: different freq and phases of waves to mix both constructive and destructive interferences.

Acoustic Impedance: similar to stiffness of a spring

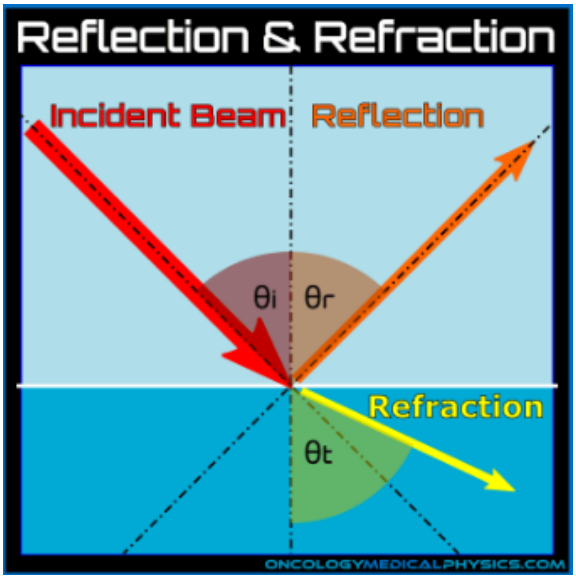
- Acoustic impedance, $Z = \rho c$
  - ρ is density of medium
  - c is speed of sound in medium

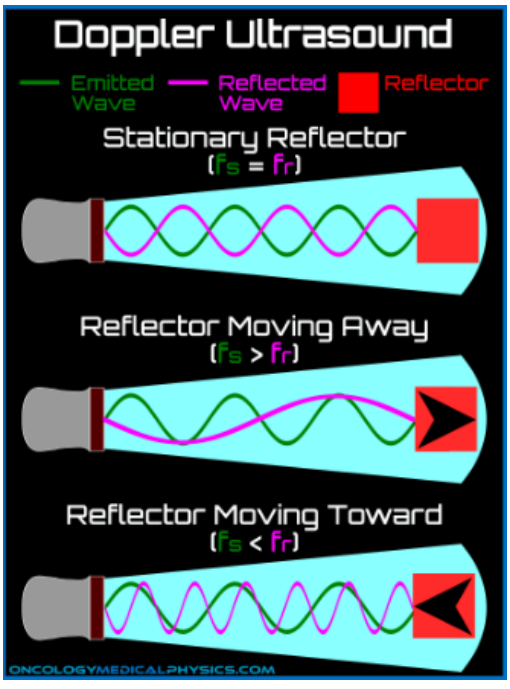
Wave-Medium Interactions: Reflection, refraction, and attenuation at medium change boundaries.

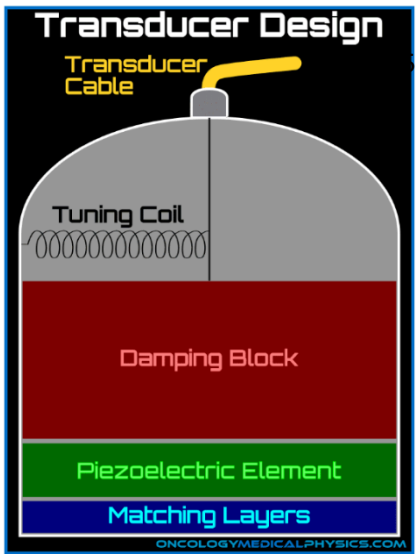
- Reflection is redirection of acoustic energy propagation which occurs at the interface of materials with different acoustic impedances.
  - $\theta_i = \theta_r$, incident angle and reflected angle are equal
  - $Fraction\ of\ acoustic\ energy$, $\frac{I_r}{I_i} = (\frac{Z_2 - Z_1}{Z_2 + Z_1})^2$
- Refraction is the change in direction of the transmitted portion of an acoustic wave incident upon an interface.
  - $\frac{\sin\sin(\theta_t)}{\sin\sin(\theta_i)} = \frac{c_2}{c_1}$
    - Θt angle of transmitted energy
    - Θi angle of incidence
    - (c1,c2) speed of sound in both mediums
- Attenuation is loss of acoustic wave intensity due to interactions between the wave and the medium.
  - Scattering occurs because of small non-uniformities within a medium
  - Absorption and Attenuation converts acoustic energy to heat.
    - Attenuation in soft tissue $= 0.5$ dB/(cm/MHz)

Doppler Frequency Shift: change in frequency which occurs when an ultrasound wave travels through and is reflected by objects moving axially relative to the transducer.

- $f_r = f_s \frac{c \pm v_r}{c \pm v_s}$
  - fr is receiver frequency
  - fs is source frequency
  - vr is velocity of the medium relative to the receiver
  - vs is velocity of the medium relative to the source







Ultrasound System Design and Operation

- dB=10log(I/Io)
- Speed of sound in soft tissue = 1540 m/s
- Speed of ultrasound in bone = 4080 m/s
- Attenuation rate in tissue = 0.5 dB/(cm/MHz)
  - Higher freq means higher attenuation
- Typical ultrasound frequencies around 1-20 MHz
- Typical ultrasound pressure is about 1 MPa

Basic Operation

1. The transducer emits an ultrasound beam which travels through the body.
2. Ultrasound beam encounters an interface between tissues of different acoustic impedance, causing some ultrasound energy to be reflected to the transducer.
3. Transducer detects the reflected ultrasound energy.
4. A computer uses the time between beam emission and reflection detection to determine distance, and also maps its intensity.

Transducers: piezoelectric element, a damping block and a matching layer.

- Piezoelectric element: electrical signal applied deforms the crystal, and if compressed the crystal produces an electrical signal.
  - Lead-zirconate-titanate(PZT) most common
  - Effectiveness of detection increased for wavelengths in multiples of ½ the material thickness. For those wavelengths the material is able to resonate at its natural frequency.
  - Ring Down is vibration in the PZT which continues after electronic signal has ended. Long ring down time leads to lower bandwidth but increases the spatial pulse length which negatively impacts axial resolution.
- Damping block absorbs the backward directed ultrasound energy and reduces ring down time.
  - Q-factor is the ratio of center freq to bandwidth, $Q = \frac{f_0}{bandwidth}$
  - High Q factor indicates a narrow bandwidth and long SPL, Low Q factor indicates broad bandwidth and short SPL.
    - High Q transducers are commonly used in Doppler ultrasound, narrow bandwidth is needed to accurately quantify flow rate.
    - Low Q for most other applications for higher resolution along beam axis
- Matching layer provides interface between PZT and the surface of pt. Has an acoustic impedance that is between the acoustic impedance of the element and that of the pt surface. This bridging improves the efficiency of acoustic energy transmission from the transducer to the pt and back again by reducing the amount of energy reflected.
- Transducer arrays: 128 to 512 rectangular elements organized in an array
  - Linear array: small group of elements used to produce ultrasound waves, but whole array is used to detect echoes.
  - Phased array: fire all elements of the transducer in rapid succession, allowing beam steering without physically moving transducer.

The Ultrasound Beam

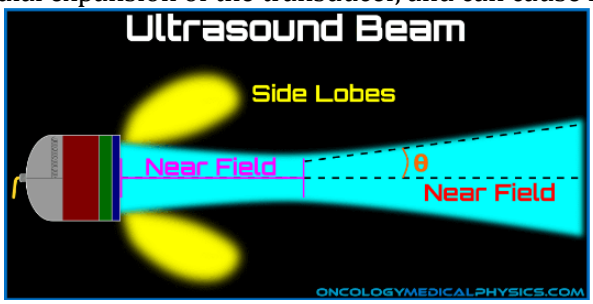
Near field: also called the Fresnel Zone, converges because of wave interference at the periphery of the beam.

- Near field length $= \frac{r^2_{transducer}}{\lambda}$
  - Soft tissue $= \frac{r^2_{transducer}}{1.54mm}$

Far field is called the Fraunhofer zone and is the diverging region of the acoustic beam.

- Far field divergence angle $\theta = (1.22\frac{2*\lambda}{r})$

Side and Grating lobes are unwanted ultrasound energy emitted laterally to the main beam. Side lobes are caused by the radial expansion of the transducer, and can cause false echoes.



Ultrasound Imaging Modes (Echo Display Modes)

- A-mode(amplitude) displays echo amplitude as a function of distance. Historically used to determine the distance to objects of different intensities.
- B-mode(Brightness) converts echo amplitude into grayscale to produce a 2D image in near real time.
- M-Mode(Motion) gray scale information produced in B-mode to produce a graph of object motion over time. Commonly used in cardiac imaging.

Doppler Mode displays fluid motion using doppler shift induced by moving reflectors, such as red blood cells.

- $v_{flow} = \frac{c*f_d}{2*f_i*\cos\cos(\theta)}$
  - v_{flow} is the fluid flow velocity
  - f_i is the emitted frequency
  - f_d is the Doppler shifted frequency
  - Θ is the angle between the direction of wave propagation and fluid flow

Resonant Frequency of crystal

- $f_{res} = \frac{c}{2*t}$, $c\ is\ speed\ of\ sound\ in\ crystal,\ t\ is\ thickness\ of\ crystal$

Spatial Resolution

Axial Resolution is the ability to distinguish closely spaced objects along the direction of beam propagation minimal object spacing is ½ of spatial pulse length

- $SPL = n_{waves} * \lambda + ringdown$, Highest is achieved Low Q-factor, Low wavelength/High frequency

Lateral Resolution is called azimuthal resolution is the ability to discern 2 closely spaced objects orient perpendicular to the beam.

- Highest later resolution is achieved when objects are located at the end of the near field and small groups of elements firing.

Slice thickness, sometimes called elevational resolution, is the width of the imaging plane.

Contrast Resolution

- Contrast resolution is highest for:
  - Variables which maximize the strength of an acoustic echo
    - Objects with very different acoustic impedance to their surroundings.
  - Variables which reduce partial volume averaging
    - Higher spatial resolution, Higher temporal resolution
  - Variables which maximize SNR
    - Use of higher ultrasound power requiring less electronic signal amplification
    - Superficial objects subject to less attenuation and returning scatter
  - Rose criteria states an SNR of 5 or more is needed to identify image features.

Temporal Resolution (Frame rate)

- $Frame\ Rate = \frac{c}{N_{A-lines} * 2 * D(cm)}$
- $Frame\ Rate(soft\ tissue) = \frac{1}{N_{A-lines} * 13\left(\frac{us}{cm}\right) * D(cm)}$
  - c is speed of sound in tissue
  - $N_{(A-lines)}$ are the number of A-lines required to produce a B-mode 2D image
  - D is the depth of scan, 2D is used for depth of travel to and from the transducer.
- Highest resolution achieved with low number of A-lines and low depth of scan.

Ultrasound Artifacts

Shadowing

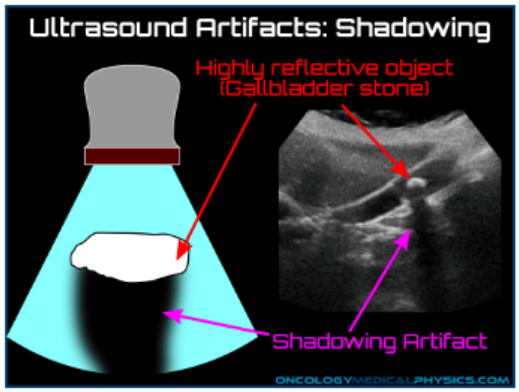
- Appearance: a region of hypo-intense signal distal to high atten objects(bone)
- Cause: Attenuation by objects superficial to artifact.

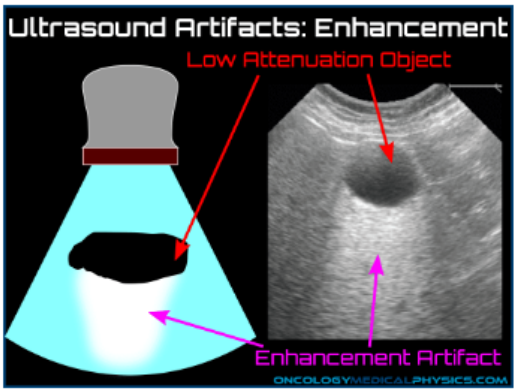
Enhancement

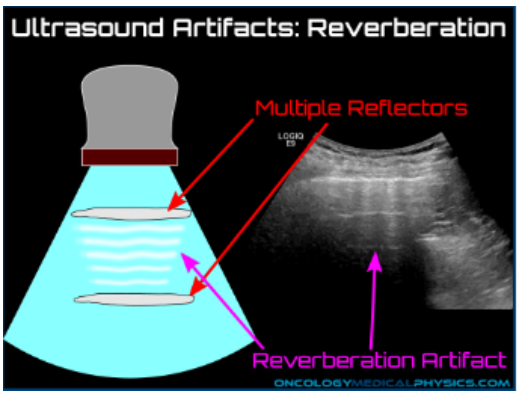
- Appearance: A region of hyper-intense signal distal to low atten objects(bladder)
- Cause: Lack of attenuation by objects superficial to the artifact.

Reverberation

- Appearance: Hyper-intense repeating signal.
- Cause: Repeated reflections between two closely spaced objects.







Mis-mapping

- Appearance: The image displays an incorrect location of an object.
- Cause: refraction due to variations in the speed of sound at the interface of the tissues causes displacement of the returning echoes.

Side-lobe energy emission

- Appearance: Manifests itself as an apparent signal which disappears when the transducer orientation is rotated.
- Cause: Anatomy outside of the beam is mapped into the main beam.

Speed Artifact

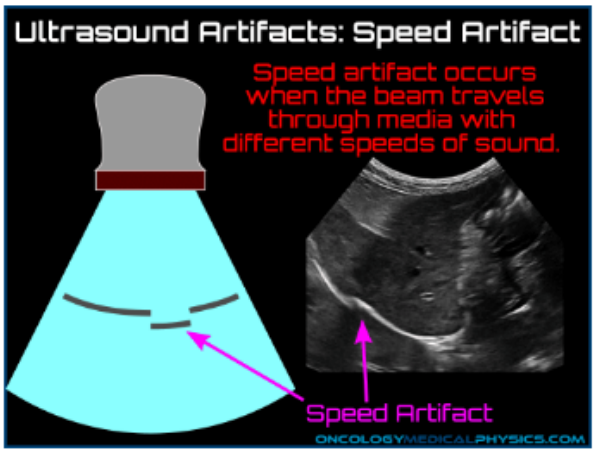
- Appearance: Abrupt mis-mapping of an object along direction of beam axis.
- Cause: Variations of the speed of sound between beam projections.

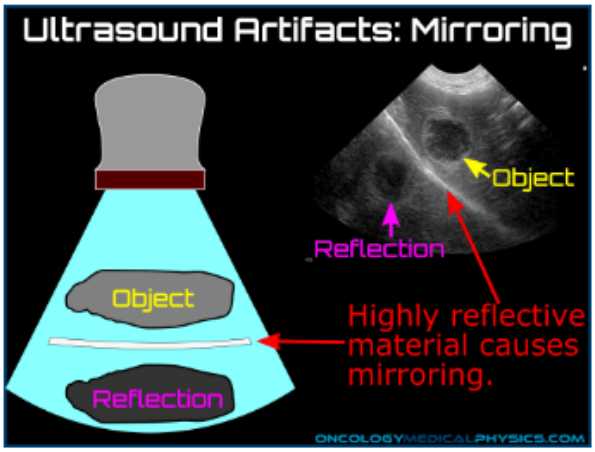
Mirroring

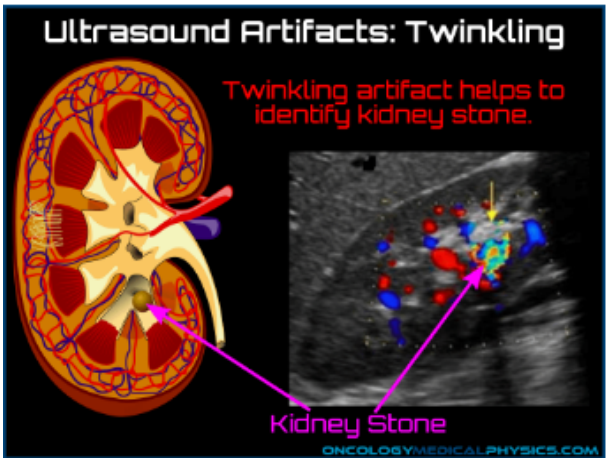
- Appearance: A second inverted object appears beyond a highly reflective surface.
- Causes: Multiple beam reflections between the object and highly reflective surface.

Twinkling Artifact

- Appearance: in color Doppler imaging mode, a region appears as a rapid changing mix of colors.
- Cause: Multiple beam reflections between the object and the highly reflective surface.







Radiation Exposure Limits(NCRP-116)

Deterministic Effect: a somatic effect which increases in severity with increasing radiation dose above a threshold dose.

Stochastic Effects: health effects which occur randomly and for which the probability of the effect, rather than the effect's severity, is related to dose.

As Low As Reasonably Achievable (ALARA): making every reasonable effort to maintain exposures to radiation as far below the dose limits as is practical

Radiation Exposure Rick Estimates

- Workers: Fatal Cancer 4% per Sv, Non-fatal/Genetic Effects 0.8% per Sv
- GP: Fatal Cancer 5% per Sv, Non-fatal/Genetic effects 1% per Sv
- Fetus/Embryo: ~10% per Sv

Radiation Worker Exposure Limits

- Stochastic Effects: 50 mSv annual eff dose, 10 mSv * age(y) cumulative eff dose
- Deterministic Effects: 150 mSv annual eq dose limit to lens, 500 mSv annual eq dose to skin and extremities

Public Exposure Limits

- Stochastic Effects: 1 mSv annual continuous exp, 5 mSv annual infreq exp
- Deterministic Effects: 50 mSv annual eq dose limit to lens, skin and extremities.
- Embryo Fetus: 0.5 mSv limit per month(Once pregnancy is declared)

NRC 10 CFR Part 20: Standards for Protection Against Radiation

Radiation Area: radiation levels could result in an individual receiving *dose eq. in excess of 0.005 rem (0.05 mSv) in 1 hour at 30 cm from source protection surface.*

High Radiation Area: radiation levels from radiation sources external to the body could result in an individual receiving a *dose eq. in excess of 0.1 rem (1 mSv) in 1 hour at 30 cm from protection.*

Very High Radiation Area: radiation levels from radiation sources external to the body could result in an individual receiving a *dose eq. in excess of 500 rads (5 Gy) in 1 hour 1 meter from radiation source or protection.*

Dose Limits:

- Total Effective Dose: Annual Occupation Limit(adult) 5 rem (50 mSv), Annual Occupation Limit(minor) 0.5 rem (5 mSv), General Public Annual 0.1 rem (1 mSv), General Public per hour in unrestricted areas 0.002 rem (0.02 mSv), General Public Infrequent Exposure 0.5 rem (5 mSv).
- Individual Organ(excluding Lens): Adults 50 rem (500 mSv), Minors 5 rem (50 mSv)
- Lens: Adults 15 rem (150 mSv), Minors 1.5 rem (15 mSv)
- Skin: Adults 50 rem (500 mSv), Minors 5 rem (50 rem)

MU hand calcs

Photon Field MU Calc

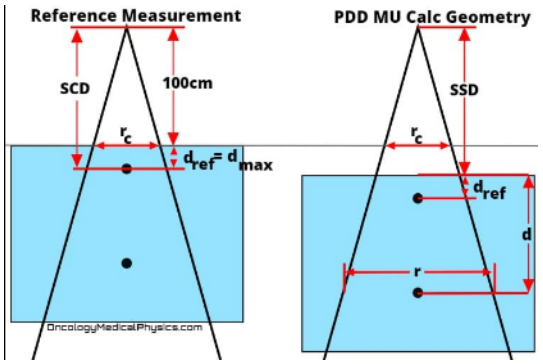
- $MU = \frac{Rx}{O*PDD(or\ TMR)*S_c(r_c)*S_p(r)*ISC*TT*WF*TF*OAF}$
  - MU number of monitor units to be delivered.
  - Rx is the prescription dose at the depth of interest(cGy)
  - O is the output factor in units of cGy/MU
  - PDD is percent depth dose, used in non-isocentric techniques.
  - TMR is tissue maximum ratio, used for isocentric techniques.
  - Sc and Sp are the collimator and phantom factors.
  - ISC is inverse square correction
    - $ISC_{PDD} = \left(\frac{Source\ to\ Calibration\ Depth}{SSD+d}\right)^2$
    - $ISC_{TMR} = \left(\frac{Source\ to\ Calibration\ Depth}{SAD}\right)^2$
  - TT is treat to level, if an isodose line is not specified.
  - WF is wedge factor
  - TF is transmission Factor or Tray Factor
  - OAF is off axis factor

Electron Field MU Calc

- $MU = \frac{Rx}{O*PDD*AF*CF*TT*ISC}$
  - MU number of monitor units to be delivered.
  - Rx is the prescription dose at the depth of interest(cGy)
  - O is the output factor in units of cGy/MU
  - PDD is percent depth dose.
  - AF is the applicator factor
    - $AF = \frac{Output\ with\ applicator}{Output\ without\ reference\ applicator}$
  - CF is the cutout factor
    - $CF = \frac{Output\ with\ cutout}{Output\ without\ cutout}$
  - TT is to treat level
  - ISC is inverse square correction
    - $ISC = \left(\frac{VSD+d_{max}}{VSD+g+d_{max}}\right)^2$

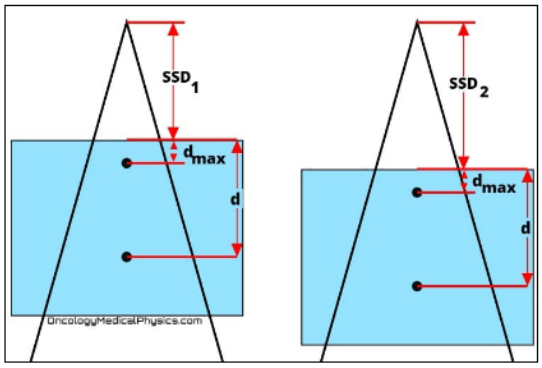
PDD, Percent Depth Dose, ratio of dose at a given depth to max dose.

- $PDD\left(SSD, d, r_c\right) = \frac{dose\ at\ depth\ d}{dose\ at\ depth\ d_{max}}$
  - SSD is source to surface distance
  - d is the depth of interest
  - rc is field size at iso
  - dmax is depth of max dose



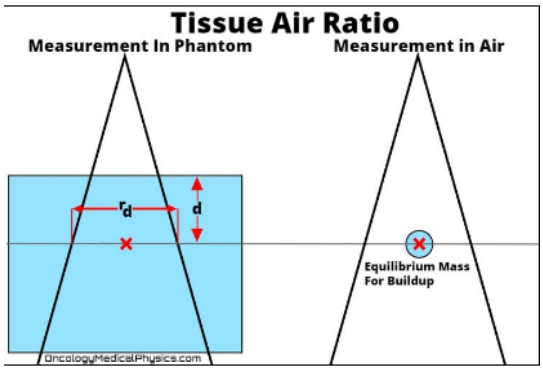
Mayneord F Factor: ratio of inverse square component of PDD from reference SSD to another SSD

- $F = \frac{ISC\left(SSD_2\right)}{ISC\left(SSD_1\right)}$
  - $PDD\left(SSD_2, d\right) = PDD\left(SSD_1, d\right) * F$



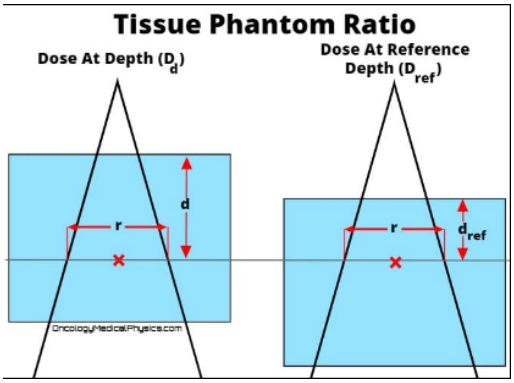
Tissue Air Ratio(TAR) is the ratio of dose at a point in the phantom, to that point in free space.

- $TAR\left(s, r_d\right) = \frac{D_d}{D_{fs}}$



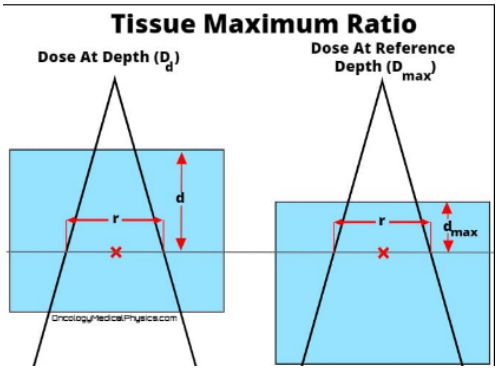
Tissue Phantom Ratio(TPR) is the ratio of dose at a given point and depth in phantom to dose at the same point at a ref depth in a phantom

- $TPR(d,r) = \frac{D_s}{D_{dref}}$



Tissue Maximum Ratio(TMR) is the ratio of dose at a given point and depth in phantom to dose at the same point at the depth of maximum dose in a phantom. Special case of TPR.

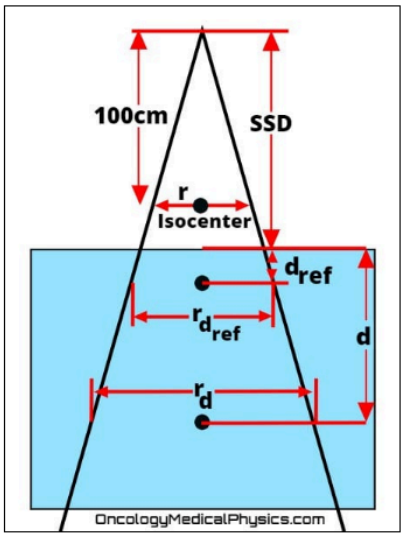
- $TPR(d,r) = \frac{D_d}{D_{dmax}}$



PDD and TMR

$$\text{TMR} = \text{PDD}(\frac{SSD+d}{SSD+d_{ref}})^2(\frac{S_p(r_{d_{ref}})}{S_p(r_d)})$$

$$PDD = TMR(\frac{SSD+d_{max}}{SSD+d})^2(\frac{S_p(r_d)}{S_p(r_{dmax})})$$



TMR and TAR

$$TMR = \frac{TAR}{BSF}$$

TMR and TPR

$$TPR = \frac{TMR(d)}{TMR(d_{ref})}$$

Back Scatter Factor(BSF) is the ratio of dose at depth of max dose in the phantom to dose at the same point in free space with full build-up.

- BSF is a special case of TAR evaluated at depth of max dose(dref=dmax)
- BSF is the highests for low energies and large field sizes.
  - $BSF\left(r_{dmax}\right) = \frac{D_{max}}{D_{fs}}$

Collimator Scatter Factor(Sc) accounts only for scatter from treatment head, defined as the ratio of output in air for a given field to that of a reference field.

- Measurement using no phantom, best measured using ion chamber with a build-up cap and measured at isocenter.

Phantom Scatter Factor(Sp) accounts for scatter arising from phantom, defined as the ratio of dose rate for a given field size at a ref depth to the dose rate at the same depth using a reference field size with the same collimator opening.

- Measurement, have to determine from $S_p = \frac{S_{c,p}}{S_c}$

Total Scatter Factor(Sc,p) accounts for scatter in phantom and head of the machine, defined as the ratio of dose rate at the ref depth for a given field size to a ref field size.